

EFFECTS OF TORSO FLEXION ON FATIGUE FAILURE  
OF THE HUMAN LUMBOSACRAL SPINE

DISSERTATION

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By

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## ABSTRACT

Recent reviews of the epidemiology literature have concluded that a number of work-related physical factors show consistent and positive associations with development of low back disorders. Among the workplace factors associated with back disorders are jobs involving manual materials handling activities, jobs involving frequent or prolonged bending of the torso, and jobs requiring frequent lifting of loads. A common theme underlying these workplace factors is that all involve situations where the tissues of the lumbosacral spine will be repeatedly subjected to elevated levels of compression and shear forces. It is known that repeated application of sizeable loads to biological materials can and will result in the development of fatigue failure; however, relatively few studies have examined this failure mechanism using lumbosacral motion segments. None have attempted to simulate repeated loading of lumbosacral motion segments associated with lifting moderate weights in various angles of torso flexion. Accordingly, the purpose of this study was to carefully simulate spinal postures and loads associated with lifting a 9 kg weight in three torso flexion angles (0 degrees, 22.5 degrees, and 45 degrees torso flexion).

Twelve human cadaver spines (average age  $81 \pm 8$  years) each were dissected into three motion segments (L1-L2, L3-L4, and L5-S1) which were then randomly assigned to a loading condition representative of each of the three torso flexion angles. Care was taken to reproduce the posture of the motion segments to those observed *in vivo* via use

of multiple radiographs during the fixation period. Tests were performed in a humidified environmental chamber at a temperature of 37 degrees C. Creep loading was performed for 15 minutes to condition the specimen, and then cyclic loads at 0.33 Hz were imposed until failure or until 10020 cycles were completed. Failure was taken as a displacement of 10 mm after the termination of the creep loading period.

Simulated torso flexion angles had a dramatic impact on the number of cycles to failure of lumbosacral motion segments. Lifting the load in a simulated neutral trunk posture resulted in failure after an average of 8253 cycles ( $\pm 2895$ ), while the 22.5 degree torso flexion simulation resulted in an average of 3257 cycles ( $\pm 4443$ ), and motion segments experiencing the 45 degree condition averaged 263 ( $\pm 646$ ) cycles to failure. Torso flexion accounted for slightly over 50% of the total variance in fatigue life. Results suggest repetitive lifting in flexed torso postures may greatly increase the rapidity of fatigue failure in lumbosacral tissues.

Dedicated to my patient, thoughtful, and loving wife, Nancie.

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## FIELD OF STUDY

Major Field: Industrial and Systems Engineering

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# **CHAPTER 1**

## **INTRODUCTION**

Low back disorders (LBDs) represent one of the most frequent health-related complaints in industrialized countries, and are a major cause of morbidity, disability, activity limitation, and economic loss (Garg and Moore 1992, Kelsey and Golden 1988, Webster and Snook 1994). Estimates of lifetime prevalence indicate that a large majority of people will experience pain emanating from the low back region at sometime during their lives, with most estimates in the range of 60-80% (Andersson 1981, Heliövaara et al. 1991, Hult 1954). Approximations of the economic costs associated with low back disorders range up to 100 billion dollars annually (Cat-Baril and Frymoyer 1991), and continue to escalate. However, the economic costs of low back disorders are overshadowed by the pain and disabilities suffered by those afflicted, many of whom are forced to change their occupation, retire prematurely, and/or live for extended periods on worker compensation payments (Andersson 1981, Taylor 1976). Unfortunately, the pathophysiology of these disorders is not well understood (Garg and Moore 1992), and patients rarely receive a specific diagnosis to which the pain may be ascribed (Deyo 1998).

While the exact causes are often indeterminate, most authorities believe that mechanical factors are an important piece of the idiopathic low back pain puzzle. Specific types of mechanical loading comprise the greatest known risk factors for low back pain in general (Marras et al. 1993, Marras et al. 2000, Norman et al. 1998) as well as for specific injuries involving disc herniation (Kelsey et al. 1984). Furthermore, lumbar structures observed to fail in laboratory tests of mechanical tolerance to applied loads seem to exhibit similar failure patterns as those observed in cadaver specimens (Vernon-Roberts and Pirie 1973). These observations provide support for the notion that mechanical overload to the lumbar spine occurs in vivo (Adams and Dolan 1995, McGill 1990).

For many people, loads on the lumbar spine will be maximal during manual lifting activities. It is not surprising to discover, therefore, that lifting has regularly been associated with low back pain (Andersson 1999, Chaffin and Park 1973, Garg and Moore 1992, Hult 1954, Kelsey et al. 1984, National Academy of Science 2001, NIOSH 1997, Troup et al. 1981). While numerous lifting actions have been implicated, two actions most commonly associated with low back pain are bending (Bergquist-Ullman and Larson 1977, Cust et al. 1972, Klein et al. 1984, Lawrence 1955, Marras et al. 1993, Punnett et al. 1991, Snook 1988) and frequent lifting of moderate to heavy loads (Bergquist-Ullman and Larson 1977, Frymoyer et al. 1983, Magora 1973, Venning et al. 1987). Frymoyer et al. (1983) identified repetitive lifting of 20 kg or more as the factor most prognostic of low back pain.

Bending (torso flexion) and lifting are almost inextricably intertwined, yet most epidemiologic studies have examined these as separate factors. However, some have suggested that the combination may be the most common precipitating factor for low-back pain (Garg and Moore 1992). There is a clear biomechanical rationale for such a relationship, as the forces experienced by the lumbar spine in torso flexion may be double or triple that experienced in the neutral posture (Shultz et al. 1982). While these loads may not necessarily be sufficient to cause failure to lumbosacral tissues in one loading cycle, if one experiences frequent high magnitude spine loadings, it is quite conceivable that the process of fatigue failure in tissues of the lumbar spine will be initiated and will continue to propagate until failure does occur (Litsky and Spector 1999).

Despite the strong probability that fatigue failure occurs in structures of the lumbar spine, few studies have studied this mechanism. Those that have suggest that fatigue failure can occur fairly quickly when lumbosacral motion segments are exposed to loads within the physiologic range (Brinckmann et al. 1988, Hannson et al. 1987, Liu et al. 1983). Studies examining fatigue failure of lumbosacral motion segments to this point have primarily investigated responses to straight axial loading of the segments. Since the majority of lifting tasks involve partial or full flexion of the spine, a combination of axial compression and anterior-posterior shear forces would be anticipated, not simple axial compression. Furthermore, fatigue studies performed to this point have not considered the impact of flexion of the motion segment on fatigue response, and have not considered the differences in load rate that

would be experienced when lifting a load at various angles of torso flexion, an important issue given the viscoelastic nature of spine tissues.

As a result of these issues, it is difficult to generalize the results of previous fatigue failure studies of lumbar motion segments to situations where the spine is loaded in common real-life lifting situations involving torso flexion. Since these situations happen to correspond to known risk factors for low back disorders, it would seem incumbent upon us to examine the role of spines loads experienced when lifting in torso flexion and its possible impact on the development of fatigue failure of the lumbar spine.

Accordingly, the goal of this dissertation is to develop an improved understanding of the relationship between spinal loads experienced when lifting a load at different torso flexion angles and the development of fatigue failure in lumbosacral motion segments. To achieve this goal, the following specific aims will need to be accomplished:

- 1.) An evaluative review of existing literature related to mechanical testing of lumbosacral motion segments will be conducted, particularly with respect to this literature's relevance to dynamic lifting tasks involving torso flexion.
- 2.) A dynamic biomechanical model will be used to estimate loads on the lumbosacral spine when lifting a 9 kg load in different torso flexion angles (0 degrees, 22.5 degrees, and 45 degrees), and results of radiographic studies of spine flexion will be analyzed to achieve appropriate orientations of the motion segments in these postures.

- 3.) An *in vitro* study using lumbosacral spine motion segments obtained from human cadavers will be performed to examine the response of these motion segments to cyclic loading according to the postures and loads determined above.
- 4.) The influence of covariates such as bone mineral content and motion segment morphological characteristics on fatigue failure will be analyzed to enhance understanding of the relationship of these factors to fatigue failure of the lumbosacral spine.

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## **CHAPTER 2**

# **THE RELEVANCE OF MECHANICAL TESTS OF CADAVERIC LUMBAR MOTION SEGMENTS TO SAGITTAL PLANE LIFTING TASKS: A REVIEW**

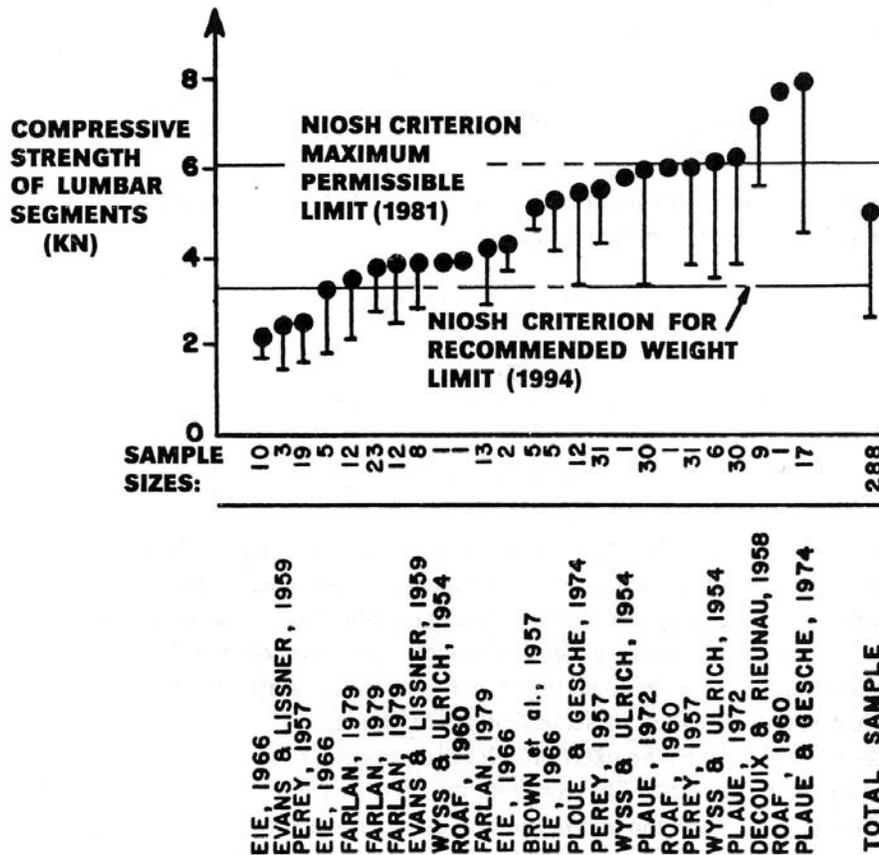
### **2.1 Introduction**

It is likely that the greatest lumbar spine loads experienced by many people will be those associated with heavy manual lifting tasks. It is not surprising to discover, therefore, that lifting has regularly been associated with low back pain (Andersson 1999, NIOSH 1997). As a result, a great deal of interest has been devoted to characterizing spinal loads during lifting tasks and attempting to relate these loads to available spine tolerance data in an attempt to develop load lifting recommendations (e.g., Waters et al. 1993). However, while numerous tolerance studies have been performed to date and some important insights have been achieved (Adams and Dolan 1995), our understanding of the stress response of the lumbar spine to loads experienced during lifting tasks remains far from complete. A particular need is to better understand the effects of repetitive loading on the spine, which is thought to

cause a degradation of the load tolerance of tissues of the low back through an accumulation of trauma resulting from repeated application of low to moderate loads (McGill 1997). Damage resulting from such a process will lead to a loss of stiffness in the spine, which may lead to lumbosacral instability, a loss of structural integrity, an up-regulation of pro-inflammatory mediators, and ultimately the development of low back pain (McGill et al. 2003). The purpose of this review is to examine current knowledge of the spine loads experienced during occupational lifting activities, and to evaluate the available research on failure of lumbar motion segments in the context of what is known regarding spine postures and loads encountered in sagittal plane lifting tasks.

## **2.2 Strength of motion segments in previous studies**

Figure 2.1 provides a summary of the results from a meta-analysis of studies that have examined the ultimate strength of human lumbar motion segments, as reported by Jager (1987) and Jager and Luttmann (1991). One can see from this figure that estimates of the average ultimate strength of vertebral motion segments vary from approximately 2 kN (e.g., Eie 1966) to 8 kN (e.g., Roaf 1960). The great disparity in these estimates may be due to a great many factors. Some variability is related to characteristics of the specimens themselves (e.g., age, gender, bone mineral content, lumbar level); however, a great deal of the variability in these strength estimates is due to the manner in which the specimens were tested (e.g., load rate, type of loading imposed, orientation of the specimen, etc.). If one considers the tolerance of lumbar



**Figure 2.1:** Results of cadaver studies examining compressive strengths of lumbar vertebral segments (figure from Chaffin et al. 1999, based on data presented in Jager 1987, also reported in Jager and Luttmann 1991)

motion segments to stresses experienced in occupational lifting tasks, one would want to understand the co-variability of the former, while exercising rigorous control of the variability of the latter – specifically by reproducing the spinal loading conditions experienced in lifting with the greatest possible fidelity. Unfortunately, for reasons that will be discussed below, there is a dearth of studies that have attempted an

accurate reproduction of spinal loads experienced during dynamic lifting tasks, let alone perform a careful reproduction of the loads and motion segment orientations experienced in different torso flexion postures. However, in the discussion that follows, it will become apparent that the much of the variability in estimates of motion segment strength resulting from widely varying methodologies, and that it is difficult to relate the findings of many of these studies to occupational lifting tasks. In addition, it will be clear that further elucidation of the tolerance of lumbosacral tissues in occupational lifting tasks will require testing methods that reproduce as closely as possible the loads, postures, and loading rates experienced when lifting in various angles of torso flexion.

### **2.3 The impact of torso flexion on loads experienced during a lift**

The lumbar spine can move in forward flexion, extension, lateral bending, and axial twisting, to which can be added compression and distraction (Bogduk 1997). Furthermore, the spine can also move using complex combinations of these primary motions. As different movements are made (or as different postures are adopted), the loading patterns experienced by the lumbar spine will be altered correspondingly (Bogduk 1997, Adams et al. 1980, Adams and Hutton 1982, Dolan and Adams 1993). As the distribution of the load shifts, the proportion of the load experienced by various structures (or portions of structures) of the spine will also change (Adams 1995). One would certainly anticipate altered probabilities of failure for the various structures of

the lumbar spine as different postures are adopted, and as loadings are varied, an expectation supported by the literature (Bogduk 1997).

While the lumbar spine is capable of a wide variety of complex motions in lifting, the focus of the current review will be limited to sagittal plane tasks starting from a position of torso flexion and ending with the torso in an upright posture (with no twisting or lateral bending). Limiting the review to sagittal plane lifting tasks offers several benefits. Firstly, sagittal plane lifting is commonly seen in occupational settings, so it represents a common occurrence. Secondly, sagittal plane torso flexion is frequently associated with the development of low back disorders (Holmstrom et al. 1994, Punnett et al. 1991). Finally, while sagittal plane lifting represents a situation involving significant complexity, it is much more straightforward than lifting tasks involving off-sagittal plane motions. And though this represents one of the most basic lifting techniques, and is one many authors have attempted to address in their research (e.g., Adams et al. 1980), it is not clear how well existing research regarding failure of cadaveric lumbar motion segments have actually represent the loadings experienced by the lumbar spine when dynamic lifting tasks are performed. The following section discusses issues relevant to accurate modeling of the loads experienced during a lifting task initiated in a flexed torso position and completed in an upright torso posture. The current literature on mechanical testing of cadaveric lumbar motion segments will then be weighed against these criteria in a subsequent section.

## **2.4 Important issues in simulation of spine loads experienced in sagittal plane lifting tasks**

If we are to gain a fuller understanding of the tolerance of the lumbar spine to loads experienced during a lifting task, it is necessary to carefully consider a number of factors expected to influence the ability of the lumbar spine to withstand the loads placed upon it. The following sections describe several factors that the author considers essential to the development of an accurate assessment of the loads experienced by the lumbar spine in occupational lifting tasks, each having important consequences with respect to the failure of tissues comprising the lumbar spine.

### **2.4.1 Choice of appropriate failure model: Ultimate strength or fatigue failure?**

As discussed in greater detail below, the majority of studies examining failure of lumbar motion segments have used the ultimate strength of lumbar motion segments as the primary outcome measure (refer to Tables 2.2 and 2.3). Ultimate strength is a useful measure in biomaterials testing, and it is possible that some spines may fail when subjected to very large and quickly increasing loads (McGill 1997). However, there are several reasons to believe that fatigue failure of tissues of the lumbar spine may be a better model of the manner in which damage occurs to spinal tissue during occupational lifting tasks (Brinckmann et al. 1988).

Mechanical fatigue can and will occur in biological materials subjected to repeated cycles of loading and unloading, and this process is a likely pathway in the development of low back pain (Litsky and Spector 1999, McGill 1997).

Microdamage, usually manifest through the development of a small crack or cracks in

a material, will begin a cascade of accumulating material breakdown that eventually will result in failure of the material being loaded. The relevance of this failure mode should be quite evident when considering the significant and repetitive loadings that can be experienced by the lumbar spine, especially in many occupational settings. Epidemiologic evidence clearly associated heavy repetitive lifting tasks as being important risk factors for low back problems (Frymoyer et al. 1980, Andersson 1999, Sandover 1983, NIOSH 1997, National Research Council 2001). Evidence provided by authors who have examined this mode of failure suggest that it may dramatically reduce the loads at which failure occurs in lumbar motion segments, and may be an important mechanism for failure *in vivo* (Brinckmann et al. 1988, Hansson et al. 1987).

Given the fact that occupationally related low back pain often involves repetitive lifting and spine loading, and given the impact that fatigue loading of lumbar motion segments has on the loads at which damage may occur, it is argued that the effects of repetitive loading in occupational lifting is a critical aspect to model in analysis of failure of lumbosacral tissues.

#### **2.4.2 Accurate representation of the orientation of the lumbar motion segments in torso flexion**

Another key factor that needs to be modeled correctly if one is to accurately determine loads on the lumbosacral spine during a lifting task is the orientation of the vertebrae in this region as the torso moves from a flexed to a neutral posture. Changes in the orientation of motion segments has been shown to affect the distribution of load

experienced by the specimen (Adams and Dolan 1995), which will impact the regions or structures where failure would be expected to occur. Fortunately, several studies have helped elucidate the position of the lumbar vertebrae when performing trunk flexion and extension activities *in vivo* (Davis et al. 1965, Dvorak et al. 1991, Esola et al. 1996, Jorgensen 2001, Pearcy 1985, Chen 2000). Radiography or MRI studies are considered gold standards for evaluating the position of lumbar motion segments in various positions of torso flexion (Bogduk 1997). Studies using these procedures have generally found similar results. Lumbar motion segments from L5-S1 through L1-L2 each appear to flex approximately 10 degrees in the flexion movement (Pearcy 1985, Pearcy et al. 1984). The middle three segments (L2-L3 through L4-L5) rotate to a slightly greater extent (10-13 degrees, on average) compared to those on either extreme (8-9 degrees, on average). Cadaver motion segments, free from the constraints of muscle attachments, rotate a few degrees more than what is observed *in vivo*. More specifically, full flexion *in vivo* causes each motion segment to be flexed to about 75% of the range allowed by the ligaments of the neural arch (Adams and Dolan 1991). Positions intermediate between neutral and fully flexed postures have also been studied and regression equations have been developed to describe the motions of the motion segments throughout the range of motion (Chen 2000).

Lifting weights from the ground requires the lumbar spine to be flexed by approximately 70 to 100% of static full flexion depending on whether the knees are flexed or not (Dolan et al. 1994). Despite suggestions by some experts that persons try to maintain a lordosis in the spine during manual lifting from the floor, it has been

shown to be functionally impossible to do so. Subjects attempting to minimize lumbar flexion when lifting from the floor found it impossible to reduce lumbar flexion below 57% (Dolan et al. 1994). It appears that when using a squat lifting technique (i.e., with knees flexed), the pelvis rotates backwards and the lordosis is reduced or reversed.

Certain conditions have been shown to result in flexion that exceeds the statically determined 100% flexion value. For example, rapid trunk flexion can lead to up to 110% of the static full flexion value (Dolan and Adams 1993, Dolan et al. 1994). Fatigue of the back muscles may also result in increased flexion (Dolan and Adams 1998, Sparto et al. 1997). These higher flexion values probably result from a compromised ability of the back muscles to perform their characteristic reflex contraction to limit flexion (Dolan and Adams 2001). In the former case, it appears that the contraction occurs too late to limit the flexion, while in the latter case the fatigued muscles take longer to generate the necessary force. Creep stretching of non-contractile tissues may affect the spine's proprioceptive apparatus and inhibit the reflexive protection afforded by the back muscles (McGill and Brown 1992, Solomonow et al. 1999).

The evidence above indicates that the *in vivo* movements of lumbar spine motion segments have been studied by several investigators, each obtaining reasonably consistent results. Highly dynamic movements have been shown to exceed the range of static flexion movements, a finding whose implications are not fully understood with respect to low back injury risk. The relationship between *in vivo* and

*in vitro* motion segment movement has been characterized. In general, we have an adequate understanding of the movement of lumbar motion segments in lifting tasks to properly simulate motion segment positions in the torso flexion angles experienced in many lifting tasks. Proper simulation of the orientation of motion segments during lifting tasks is considered another essential ingredient in the realistic simulation of the failure mechanisms of the lumbar spine during occupational lifting tasks.

### **2.4.3 Accurate representation of the spinal loads experienced during sagittal plane lifting tasks**

#### **2.4.3.1 Static versus Dynamic Estimates of Spine Loading**

When injury mechanisms are being studied, it is extremely important that complex loads similar to those experienced in lifting be applied to the specimens. This is particularly true in light of the fact that one component of loading may affect the spine's response to other loading components (Adams 1995). A high compressive force, for example, will also increase the spine's resistance to bending (Adams and Dolan 1991). As a result, several authors have developed systems to apply loads to motion segments that simulate this physiologic complexity (Adams et al. 1980, Wilke et al. 1994). The device developed by Adams and colleagues and can be used to apply any combination of compression, shear and bending forces in the sagittal plane (Adams et al. 1980). Devices to simulate loading by individual muscles have been developed (Wilke et al. 1994); however, such devices are not required for sagittal plane tests as long as the body to which the loads are applied are assumed to be rigid (Adams 1995).

Historically, estimates of the loads acting upon lumbar motion segments were derived from relatively simple static analyses of lifting tasks (e.g., Adams et al. 1980). Recent years have seen the development of more sophisticated dynamic models of spine loading (Granata and Marras 1995a). Such models use electromyographic data from trunk muscles to estimate the loads on the spine during dynamic lifting activities, and can provide important information regarding the effects of co-activity of trunk muscles to improve the fidelity of loading estimates experienced by the lumbar spine. These more complex and physiologically sophisticated models have demonstrated significantly higher compressive loads compared to simple static analyses (Granata and Marras 1995a) and significantly increased estimates of the shear forces acting in the flexed position (Potvin et al. 1991). Given that the spine's response to loading in one direction is dependent on loads in other planes (Adams 1995), it would appear critical that models incorporating the dynamic aspects of spine loading be used when simulating spinal loads associated with lifting in occupational settings.

#### **2.4.3.2 Differences in the rates of spine loading at different angles of torso flexion**

Lumbar motion segments (like all biologic tissues) exhibit viscoelastic behavior; thus, their material behavior will change based upon the strain rate, or rate at which they are loaded (also known as the load rate). Typically, an increase in the rate of loading of a viscoelastic material will cause it to behave more as a brittle elastic solid, while slower loading rates will cause a material to flow as a viscous liquid (Litsky and Spector 1999). Ultimate compressive strength is increased when a

viscoelastic material is loaded at a high strain rate (Yingling et al. 1997, Hutton and Adams 1982).

One should be cognizant of the fact that torso posture and torso muscle activity during a lifting task dramatically influence the rate at which the spine will be loaded. Specifically, as the lumbar spine is flexed at the start of a lift, it will be subjected to a significantly higher load rate compared to when the spine is in a more neutral posture. Conversely, when the body is in more of an upright position the lumbar spine would be subjected to a much lower strain rate than that experienced in the extremely flexed position.

Few studies have examined the effects of load rate on the tolerance of lumbar spine motion segments. Evans and Lissner (1959) tested whole lumbar spines at two different load rates, both of which were slow ( $0.0475$  and  $0.07 \text{ in}\cdot\text{min}^{-1}$ ), and which had no apparent effect on the tolerance of these specimens. Kazarian and Graves (1977) examine effects of strain rate on thoracic specimens; however, the lack of preload on these specimens may limit the applicability of these data to *in vivo* conditions. Hutton and Adams (1982) examined the effects of strain rate on isolated vertebral bodies and found higher compressive loads at failure at the high strain rate ( $300 \text{ mm}\cdot\text{min}^{-1}$ ) compared to the slow rate ( $6 \text{ mm}\cdot\text{min}^{-1}$ ). Evidence from tests on porcine motion segments (Yingling et al. 1997) suggests that loading rate can affect the compressive load at failure, the displacement at failure, and stiffness (slope of the load deformation curve in the elastic region). In this study, ultimate compressive load to failure was increased by approximately 30% under dynamic load rates compared to

quasi-static loading. However, due to anatomic differences in human versus porcine specimens, it is unclear how these findings might relate to tests on human lumbar motion segments (Adams 1995). What is clear from the data discussed above is that the rate at which lumbar motion segments are loaded can significantly affect the response of the tissues and the load at which failure will occur. Furthermore, it is clear from biomechanical modeling that the rate at which the lumbar spine is loaded will change dramatically as a function of the torso postures adopted throughout a lift. If one is to accurately assess the failure tolerance of lumbar motion segment tissues at different torso flexion postures during the lifting process, it is important to ensure that the rate at which the motion segments are loaded is correctly modeled.

#### **2.4.3.3 Load control vs. displacement control**

Mechanical testing of lumbar motion segments can be carried out in either a load-controlled or displacement-controlled manner. While some controversy surrounds the appropriate method to use when testing the spine, many researchers prefer the load-controlled technique due to the ability of this technique to allow simulation of clinically relevant motions (Goel et al. 1995). Displacement control places complex loads of varying magnitude along the spine segment, which are difficult to quantify at the vertebral level (Wilder et al. 1988). The more realistic motions obtained by loading motion segments using load control, and the ambiguity of the loadings on the spine when using displacement control, suggest that simulation of lifting tasks be performed using the former rather than the latter.

## **2.5 Summary of factors important to the realistic modeling of lumbar motion segment failure associated with sagittal lifting tasks**

Based on the evidence presented above, there are several factors that should be considered essential to realistic modeling of the failure of lumbar spine experienced during occupational lifting activities. Table 2.1 presents a summary of these factors. The balance of this chapter will evaluate the current literature in terms of these factors.

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1. Repetitive loading of the lumbar spine (a common occurrence in occupational lifting tasks) should be studied.
  2. Spine geometry associated with torso flexion during lifting tasks should be accurately reflected.
  3. Dynamic loads on the spine associated with manual lifting should be accurately reflected.
  4. Load rates associated with different torso flexion angles should be accurately reproduced, so that the viscoelastic response of spinal structures is appropriate
  5. Realistic simulation of spinal motions should be permitted
- 

**Table 2.1:** Criteria for evaluating studies of motion segment failure and their relevance to occupational lifting tasks.

## **2.6 Existing Mechanical Tests of Lumbar Motion Segments and their Applicability to Lifting Tasks**

Tables 2.2 and 2.3 present information pertaining to studies that have examined the failure of complete human lumbosacral motion segments in tests of ultimate compressive strength or fatigue failure, respectively. Restricting the review to human specimens was due to the fact that mechanisms of structural failure depend upon size, shape, and material properties, which are characteristics that appear to be species specific (Adams and Dolan 1995). Moreover, stress distributions in intervertebral discs are affected by both age and posture of the specimen, making any similarities between species questionable and perhaps misleading (Adams and Dolan 1995). Studies were included if they used entire motion segments (two complete vertebrae and intervening disc) and examined failure of the specimen in situations involving sagittal plane loadings.

The tables contain details of the selected studies including the reported specimen orientation, loading specifics (and rationale), and the failure model. Comments on the relevance of the testing conditions to motion segment orientation and loads experienced during lifting tasks are also provided. The following sections detail some of the issues related to the testing conditions specified in these studies and the relevance of the results to conditions that occur in manual lifting tasks.

Authors	Specimen Orientation	Loading Conditions	Failure Model	Comments
Brown Hansen and Yorra (1957)	Neutral for 5 specimens tested in compression; One specimen flexed 5 degrees (for cyclic loading test)	Compression tests: Load incremented upwards 222.5 N every 2 minutes up to 1335 N, then increased by 445 N  Cyclic Loading: Load of 66.8 N at 1100 cycles/min	Compression Tests: Ultimate Compressive Load  Cyclic Loading: Fatigue failure	Neither compressive nor cyclic loads characteristic of those experienced in lifting tasks.
Perry (1957)	Neutral assumed	Static Compression: Pressure increased at a "constant" (but unspecified) rate for up to 2 minutes  Dynamic Compression: A weight of 15 or 18 kg was dropped 50 cm onto a specimen already under static load of 10-25 kg	Ultimate Compressive Load	Neither loading condition characteristic of loads experienced when lifting
Nachemson (1960)	Neutral	Compressive loading (no load rate reported)	Ultimate Compressive Load	Loading conditions unclear as is their relevance to lifting tasks
Eie (1966)	Neutral	30 seconds of incremented axial compression (Load rate not reported)	Ultimate Compressive Load	Loading conditions uncharacteristic of lifting tasks
Rockoff et al. (1969)	Neutral	Axial loading at a rate of 0.02 in/min (displacement control)	Ultimate Compressive Load	Rate of loading very slow compared to lifting tasks
Hutton and Adams (1981)	Slightly flexed (4 to 8 degrees)	Compressive loading at a rate of 3000 N/s	Ultimate Compressive Load	Loading rate the same at each flexion level

**Table 2.2:** Studies examining ultimate stress of lumbosacral motion segment (continued)

**Table 2.2:** Continued

Adams and Hutton, 1982	Specimens flexed an average of 15 degrees; laminae removed	Predominantly compressive load increasing at 3000 N/s	Ultimate compressive Load	Motion segments flexed beyond physiologic range
Adams et al. 1994	Neutral posture and 75% of the full range of motion segment flexion	Compression increased at a rate of 2.5 mm/s until failure	Ultimate compressive load	Motion segments flexed to physiologic limit. Mechanical testing used displacement control. Possible confounders limit interpretation of results.
Krypton et al. 1995	Not explicitly stated – neutral assumed	Two displacement controlled loading rates (0.5 and 50 mm/s) were used to impart a pure shear force on the lumbar motion segments	Ultimate shear load	Pure shear loads on lumbar spine not characteristic of loads experienced when performing lifting tasks
Osvalter et al. 1990	Superior endplate of inferior vertebral body horizontal, position of superior vertebra not specified	Incremented static loading in increments of 80 N up to 320 N and 40 N after that until failure occurred. Each level was maintained for 30 seconds	Ultimate compressive load	Slowly incremented static loads not applicable to dynamic loads experienced in lifting tasks
Yoganandan et al. 1994	Midplane of disc aligned parallel to proximal and distal fixations	Specimens loaded using displacement control at 2 mm/s until failure. Failed specimens were tested a second time to the displacement where failure occurred in the first test	Ultimate compressive load	Specimen orientation method could reflect wide range of in vivo torso flexion angles. Displacement control not as physiologic as load control.

(Continued)

**Table 2.2:** Continued

Neumann et al., 1992	Superior endplate of inferior vertebral body horizontal, position of superior vertebra not specified	Creep loading of 100 N for 4 minutes, followed by 4 minutes of unloading. Experimental load was initially set at 70% of estimated ultimate strength for flexion-shear (predicted via bone mineral content) for 4 minutes then unloaded for 4 minutes. Loads were increased by 20 N for each subsequent loading cycle with 4 minutes of unloading between each loading sequence.	Ultimate flexion-shear load	Static loading method not relevant to lifting tasks.
Adams et al. 1994 ( <i>Spine</i> )	Superior endplate of inferior vertebral body horizontal, Original position of superior vertebra unclear.	Specimens loaded in compression and bending so that full flexion of the motion segment would be reached when compressive force reached 1100 N. Loading cycles of increasing severity were applied until elastic limit of flexion reached.	Ultimate compressive/bending load	Compress (continued) be expected considerably higher when lifting in a flexed posture.
Adams et al. 1993 ( <i>Eur Spine Journal</i> )	Superior endplate of inferior vertebral body horizontal, Original position of superior vertebra unclear though 0 deg angulation stated	Specimens placed at 75% of flexion elastic limit and compressed to failure using a displacement controlled loading rate of 2.5 mm/s.	Ultimate compressive load	Specimen orientation method could reflect wide range of in vivo torso flexion angles. Displacement control not as physiologic as load control.

<b>Authors</b>	<b>Specimen Orientation</b>	<b>Loading</b>	<b>Comments</b>
Adams and Hutton, 1983	Specimens flexed 1 degree less than physiologic limit	Cyclic loading performed at 40 cycles/min (peak loads between 1500 and 6000 N dependent on age, sex and body mass of cadaver)	Determination of choice of cyclic load unspecified, as is load rate for cyclic tests  Shear forces must be present, but are unspecified.
Adams and Hutton 1985	Specimens flexed to physiologic limit of flexion, and then increased by one degree every hour until risk of damaging specimen from flexion was too high	Specimens were cyclically (sinusoidally) loaded at 40 cycles/min (peak loads between 1500 and 6000 N dependent on age, sex and body mass of cadaver). Load increased by 500 N every half hour until failure occurred.	Flexion of specimens at or beyond physiologic range.
Gordon et al. 1991	Superior and inferior endplates of the motion segment potted parallel to trays, then flexed 7 degrees and cyclically rotated 3 degrees or less. Disc centered in tray	1334 N of compression applied under load control and displacement noted. Specimen then cyclically loaded (40 cycles/min) using displacement control to this initial displacement for an average of 6.9 hours (range 3-13 hours). Load rate not reported.	Specimen orientation difficult to relate to specific postures adopted when lifting. Cobb angles of motion segments not reported. Load rate not reported but would vary with different specimens (due to difference in initial displacement).
Hansson et al. 1987	Not specified (at least in Hansson et al. 1988)	Specimens loaded cyclically (sinusoidally) at a rate of 0.5 Hz to a load determined by the ultimate compressive strength of the weakest segment of the specimen predicted by bone mineral content (measured using (dual photon absorptiometry) . Maximum number of cycles was 1000.	Load rates would differ according to determined ultimate compressive strength which would change loading response of viscoelastic motion segments. Specimen orientation not specified.

**Table 2.3:** Studies examining fatigue failure of lumbosacral motion segments (continued)

**Table 2.3:** Continued

Brinkmann et al. 1988	Mid-plane of disc and upper endplate aligned approximately parallel to plane of lower tray.	Specimens loaded cyclically at 0.5 Hz (1 second loading, 1 second unloading). Base load was 700 N, upper load was based on percentages of ultimate compressive load predicted from a test of ultimate compressive load one motion segment of spine tested. Load rate for ultimate compression test was 1 kN/s. Load rates for cyclic loading varied according to computed compressive strength and the percentage of this strength used in testing.	Load rates for fatigue tests varied from 500 N/s to 5.2 kN/s. (Ult comp strength tested at one load rate, fatigue tested at another) Specimen orientation method could reflect wide range of in vivo torso flexion angles.
Liu et al. 1983	Not reported.	Axial load equivalent to 37-80% of average ultimate compressive strength for a segment based on data by White and Panjabi (1978)	Load rates would vary according to selected load. Load rates not reported.

### **2.6.1 Failure Model (Ultimate compressive strength versus Fatigue failure)**

As can be seen in Tables 2.2 and 2.3, the vast majority of studies examining failure of lumbar motion segments have concentrated on establishing the ultimate compressive strength of lumbar motion segments subjected to various experimental conditions. Only a handful has examined fatigue failure. Thus, the most likely failure mechanism has been the least studied.

Studies of fatigue failure have revealed a dramatic impact on the loads at which lumbar motion segments fail (Hansson et al. 1987, Brinckmann et al. 1988). In Brinckmann et al.'s (1988) study, even at a load predicted to be 40-50% of the ultimate strength of a motion segment, greater than half of the motion segments tested failed after only 500 loading cycles. Despite certain methodological concerns (discussed in detail below), these findings would appear to have relevance to a possible etiology of low back disorders due to repetitive occupational lifting tasks. The relevance of repetitive lifting has been reinforced by recent studies that have documented a trend toward lighter loads, but more frequent lifting in occupational tasks (Dempsey 2003). It may be noted that *in vitro* studies do not allow biological repair mechanisms to operate on damaged tissue, a process that would occur in the physiologic system. However, while these repair mechanisms are known to exist, such mechanisms often require a significant amount of time to achieve success. For example, based on known healing times for bone, it may take weeks or months for repair of bone microfractures in the lumbar spine (Brinckmann et al. 1988). Thus, it

is quite conceivable that the rate of damage to lumbar tissues would outpace the remodeling process.

### **2.6.2 Motion Segment Orientation**

A review of the literature indicates a great amount of variability in the adequacy of the description of how specimens were oriented, and in the development of a rationale used to specify motion segment orientation. Furthermore, it will be seen that it is very difficult in many cases to relate the motion segment positioning scheme used in many studies to a specific torso postures.

In many cases, inspection of Tables 2.1 and 2.2 will show that a high percentage of studies, and particularly earlier studies, were performed with motion segments placed in a “neutral” position. However, even when authors state that a neutral position was used, it is not clear what is meant by the term (i.e., no operational definitions are provided). A common thread among studies that do provide a definition of a “neutral” posture was to describe a position where the superior endplate of the inferior vertebra (or a line bisecting the disc) was placed horizontal to the bottom specimen tray. The superior endplate of the upper vertebra was then potted so that it was parallel (or approximately parallel) to this line (e.g., Gordon et al. 1991) and the test was performed.

However, it must be realized that employing such a definition is problematic when trying to relate motion segment positions to the physiologic positioning of motion segments in different torso positions assumed in lifting tasks. For example,

the definition given above defines the positioning of the L1-L2 motion segment in an upright torso posture fairly well (superior endplates of both vertebrae are approximately parallel to one another). However, if one considers the L3-L4 segment, the Cobb angle measured in an in vivo upright posture shows that the superior endplates of this motion segment is extended by approximately 8 degrees (Chen 2000, Jorgensen 2001). Placing this motion segment in the “neutral” position described above actually corresponds to the position this motion segment would experience in extreme torso flexion. One can appreciate that the operational definition of a “neutral” posture for motion segments as described above makes interpretation of the results of these studies difficult when one is interested in comparing the mechanical responses of motion segments in conditions representative of different postures of the torso.

Adams and colleagues, in a series of studies in the 1980’s and 1990’s, utilized a different approach to define the degree of flexion experienced by a motion segment. These authors developed a procedure where a neutral posture for a motion segment was determined by moving the point of force application on a cadaveric motion segment anteriorly and posteriorly in the sagittal plane. The point at which no flexion or extension of the motion segment occurred when a moderate force was applied was operationally defined to be the “neutral” posture. From this point, the authors would flex or extend the motion segment and perform various mechanical tests. While this technique provides a thoughtful and useful method with which to standardize testing procedures, it is not clear that a neutral posture for a motion segment as established by

Adams and colleagues is equivalent to the orientation the lumbar motion segment would exhibit in a neutral torso posture.

An interest in the mechanics of disc prolapse led Adams and colleagues to a fascinating series of articles where they caused disc prolapse. However, while the results of these articles are intriguing, it must be noted that it was necessary for these authors to flex the motion segment beyond the range observed to occur physiologically. While these studies are some of the few that have ever been found to frequently create disc prolapse in cadaveric material, their relevance to lifting tasks is questionable due to the extreme amount of flexion placed on the lumbar motion segments.

Adams and Hutton (1983) have examined fatigue loading using motion segments flexed within the physiologic range of flexion (for example, flexed 4-8 degrees). However, results of these studies may also be difficult to relate to the failure tolerances of the lumbosacral spine in specific torso postures during lifting tasks. Furthermore, one would expect increasing load rates as the lumbar spine flexes forward in a lifting task; however, Adams and Hutton (1983) maintain a constant load rate at each angle of flexion.

Radiographic and MRI studies of the lumbar spine in different torso postures provide a reasonably clear and consistent picture of the positioning of the lumbar spine motion segments *in vivo*. However, mechanical tests involving cadaveric motion segments often have not often used the information provided in these studies to position specimens in a manner that reflect their position at various angles of torso

flexion. Commonly used definitions used to test motion segments in so-called “neutral” postures may correspond to either upright or fully flexed torso positions dependent on the motion segment, making interpretation of results difficult in physiologic terms. It is also not clear how the definition of a “neutral” posture defined by finding the “balance” point of the specimen relates to positions observed in *in vivo* studies. A better understanding of the effects of torso posture on lumbar spine tolerance due to lifting may be achieved by positioning cadaveric motion segments in accordance with the orientations observed to occur *in vivo*.

### **2.6.3 Loading Conditions Imposed on Motion Segments (and their relationship to lifting tasks)**

Placing a load on motion segments that accurately simulates that experienced by the spine during lifting tasks requires attention to several important factors, as outlined earlier. For example, the loading should simulate the proper time-frame of loading experienced in lifting tasks (usually a couple of seconds in duration), should provide physiologically-realistic (complex) loads, and should be sensitive to differences in load rates that would be expected to occur as different torso positions are adopted in the lifting process. Issues regarding whether specimens should be tested under load control versus displacement control should also be carefully considered.

Some of the studies in Tables 2.2, particularly some of the earlier cited studies, clearly did not provide loading conditions that simulate what one would anticipate during lifting tasks. For example Brown et al. (1957) loaded motion segments using a

load that was incremented upwards by 222.5 N every two minutes. These authors also performed a cyclic loading test of one motion segment, but at the unreasonably high frequency of 1100 cycles/sec. Perry (1957), in tests involving static compression increased the load at a constant rate for up to 2 minutes. Similarly Eie (1966) subjected segments to a 30 second period of incremented axial compression., while Rockoff et al. (1969) used a displacement controlled axial load at a rate of 0.02 inches per minute. All of these studies involved load rates that bear little correspondence to spinal loads encountered in lifting tasks.

Studies performed in more recent years have used loadings that are more in line with the loads expected in lifting tasks; however, the fidelity of the loading situations to conditions experienced during lifting tasks deserve some scrutiny. Issues that are of concern are use of static estimates of spine loads to estimate loads experienced when lifting, and issues related to the rate at which specimens are loaded.

As stated earlier, when examining the failure mechanisms of the lumbar spine, it is important that appropriate physiologically complex loadings be used. For example, a higher compressive load will increase resistance to bending of a motion segment. However, Adams and colleagues have typically based their loading schemes on static analyses of loads on the lumbar spine (e.g., Adams et al. 1980), which may significantly underestimate compressive loading experienced when performing dynamic lifts. Recent years have seen the development of models of increasing sophistication that take into account the increased loadings resulting from dynamic lifting tasks, including the effects of muscle co-activity during lifting tasks. Results of

dynamic biomechanical models suggest that compressive loads during lifting tasks may be 40% greater than that predicted using static analyses (Granata and Marras 1995b). Since this substantially increased compressive force will significantly impact the motion segment responses in other planes, it is not clear how realistic tests based on static analyses of loads on the spine when lifting will be.

Many studies in the literature do not appear to pay much attention to the impact that the loading rate may have on the results of their research. As a viscoelastic material, spinal motion segments will respond much differently according to the rate at which they are loaded. Low loading rates will tend to allow the motion segments to exhibit a viscous flow behavior, while higher loading rates will result in a much stiffer response, and a higher ultimate strength (Litsky and Spector 1999). For example, in the article addressing fatigue failure of lumbar motion segments by Brinckmann et al. (1988), predicted ultimate strengths for motion segments from a spine were based upon a test of ultimate compressive strength obtained from a randomly selected motion segment at a load rate of 1 kN/s. A correction factor for the strength of different spine levels was then applied according to data provided by Hansson et al. (1981). The adjacent motion segments were then tested at a certain percentage of the predicted strength based upon the results of this test. However, an issue that greatly complicates interpretation of the Brinckmann et al. (1988) data is that the procedure above, combined with a set duration of loading leads to specimens being tested at a wide variety of load rates, ranging from 0.5 kN/s to 6.4 kN/s. As an example, consider 2 specimens, both being tested at 50% of their ultimate compressive

strength. The estimated ultimate compressive strengths of the specimens are 2400N and 7400 N, and for each the minimum load will be 700N (per Brinckmann et al. 1988). Given the constant loading duration (2 seconds), while both are being tested at 50% of their ultimate strength, one will be tested at a load rate of 250 N/s while the other will be tested at 1500 N/s. Given the tremendous strain rate sensitivity of bone (Hawkins 2001), results of tests using such a procedure become very difficult to interpret.

As suggested above, accurate simulation of the loading of motion segments during lifting tasks requires an appreciation of the different rates at which the lumbar spine will be loaded when the torso is flexed as compared to when the torso is upright. An analysis of load rates using the Biodynamics EMG-assisted model indicates that the average peak loading rate when lifting a moderate load (approx 9 kg) in upright postures is 0.7 kN/s; whereas the average peak loading rate observed in a flexed torso posture using the same load is 4.8 kN/s (see Chapter 3). This difference may greatly impact the failure response of motion segments to loads experienced when lifting; however, no studies in the current literature have evaluated the impact of increased loading rates with trunk flexion during performance of sagittal plane lifting tasks.

#### **2.6.4 Load Control versus Displacement Control**

Most authors in the summary tables have opted to use load control, though some have used displacement control and some have used both. While displacement control can and has been used for motion segment studies, it seems that the ability to

simulate motions in a more physiologically accurate manner makes load control the preferred method for testing lumbar motion segments, especially when simulating lifting tasks.

## **2.7 Discussion**

Table 2.4 is a table that summarizes the available literature on tests of motion segment failure in relation to issues considered important in establishing relevance of the results of these studies to occupational lifting tasks. Issue such as failure model, motion segment orientation, and loadings placed on the motion segments are considered. As can be seen from this Table and as will be discussed below, there are many issues that seriously limit the applicability of the results of these studies to common occupational lifting tasks.

### **2.7.1 Studies involving fatigue failure**

It is surprising to find, given the interest in occupational lifting tasks expressed in many studies, that only a handful have investigated the effects of repetitive loading on the human vertebral column. There can be no doubt that exposures to repetitive lifting of heavy loads are extremely pervasive in industry. Due to the fact that studies have indicated that biological materials under repetitive loading fail at a level well beneath their ultimate load, one would think that this would be an area of research that would have been explored more thoroughly.

Authors	Failure Model	Motion Segment Orientation		Motion segment loads		Load rates		Mode of testing
	Fatigue Failure?	Motion segments flexed?	Related to specific <i>in vivo</i> torso flexion angles?	Complex Loads?	Based on dynamic biomechanical analysis?	In physiologic range for lifting tasks?	Adjusted for higher rates in flexed postures?	Load control?
Brown et al. (1957)							N/A	✓
Perry (1957)							N/A	✓
Nachemson (1960)						?	N/A	
Eie (1966)							N/A	✓
Lin et al. (1978)		✓		✓		?		
Hutton and Adams (1982)		✓		✓		✓		✓
Adams and Hutton (1982)		✓		✓		✓		✓
Adams and Hutton (1983)	✓	✓		✓		✓		✓
Adams and Hutton (1985)	✓	✓		✓		✓		✓
Adams et al. (1994b) (CB)		✓		✓				
Krypton et al. (1995)							N/A	
Liu et al. 1983	✓					✓		✓
Osvalter et al. (1990)		✓		✓				✓
Gordon et al. (1991)	✓	✓		✓		?	N/A	
Hansson et al. (1987)	✓						N/A	✓
Brinkmann et al. (1988)	✓					✓	N/A	✓
Yoganandan et al. (1994)							N/A	
Neumann et al. (1992)		✓		✓				✓
Adams et al. (1994) b		✓		✓				
Adams et al. (1993)		✓		✓		✓		

**Table 2.4:** Summary of mechanical tests of lumbosacral motion segments to lifts involving torso flexion. A check (✓) indicates an issue was appropriately addressed. A question mark (?) indicates the issue was not clearly addressed or was not reported. N/A indicates the issue was not applicable to the study.

If one takes a closer look at the few studies that have examined fatigue failure of lumbar vertebrae, a number of issues arise that lead one to question the applicability of these studies to loads actually experienced in lifting tasks. Of the six studies reviewed, only three investigated conditions involving flexion of the motion segments. Of these three, one involved extreme flexion at beyond the limits of the physiologic range (Adams and Hutton 1985), and in the remaining two, the methods used to determine the degree of flexion of the motion segments makes it difficult to relate to the loads and spine orientations to torso flexion postures characteristic of lifting tasks.

The study by Brinckmann et al. (1988) is instructive as to an apparently dramatic impact associated with fatigue failure, yet there are several difficulties in relating the results to loads experienced in lifting activities. One issue relates to the load rates imposed on the specimens and the relation of these load rates to the orientation of the motion segments. The loads chosen by Brinckmann et al. 1988 represent percentages of predicted ultimate strengths for the motion segments based on guidance on the strength differences for various lumbar levels provided by Hansson et al. (1987). In the first place, it would be expected that the variability in estimates of strength may be quite significant, even within segments from the same spinal column. One reason is that the bone mineral density may vary substantially from one lumbar level to the next. Furthermore, while some of the load rates and magnitude of loadings employed in this paper may reflect what might be experienced in a neutral position during a lifting task, many of the conditions greatly exceed the load and rate of loading that would be expected in during lifting tasks in a neutral trunk posture. Furthermore,

the axial loading would not reflect the physiologic loading in the neutral posture, which would involve a degree of shear loading as well. These factors make it difficult to relate the loadings imposed in this study to situations experienced in lifting.

The fatigue failure studies by Adams and colleagues (1983, 1985) examined the effects of cyclic loading on the intervertebral disc. Peak loads of between 1500 to 6000 N were used in a cyclic loading experiment, with the amount of loading determined according to the age, sex, and gender of the cadaver. Unfortunately, the manner in which this determination was made was not reported in the papers. Motion segments were flexed to one degree less than the physiologic limit (from 8.7 to 12.5 degrees). This represents an extreme amount of flexion, an amount that may not be experienced in routine lifting tasks (Chen 2000, Jorgenson 2001). It is apparent that an interest in disc prolapse mechanisms motivated these high levels of motion segment flexion; however, typical lifting tasks would involve flexion angles that are considerably less than those studied. It should also be noted that given the constant cycle time and varying peak loads applied in these experiments, that a wide range of load rates were used, which would change the response of the motion segments due to their viscoelastic nature. Some of these load rates may correspond to those anticipated in fully flexed positions; others would be considerably lower than what would be experienced in lifting tasks.

In this, as well as many other papers by Adams and colleagues, loads are applied to a flexed motion using a combination of compression and an anterior shear force (i.e., the superior vertebra of the pair is shearing anteriorly with respect to the lower motion

segment). This combination of loads is apparently based on static lifting analyses (e.g. Adams et al. 1980). However, models that incorporate the dynamic forces associated with lifting suggest that the shear forces experienced at the beginning of a lift are actually in the opposite direction. If one considers the initiation of a lifting task, with the lumbar spine flexed forward, it is apparent that the back muscles must create a force that will extend the motion segments of the lumbar spine, resulting in a posterior rotation and a posterior translation of superior vertebrae upon their respective inferior vertebrae (Bogduk,1997). Accomplishing this feat requires a posteriorly directed shear force (i.e. superior vertebra shearing posteriorly with respect to inferior vertebra), as amply demonstrated by models incorporating the geometry and action of the trunk musculature. Furthermore, as noted previously, dynamic models predict compressive loads that may be 40% higher than those obtained in static analyses. Thus, it appears that loadings on motion segments based on static modeling procedures may not realistically portray the actual compression forces, and may not even portray the correct direction of shear forces, as compared to those expected to be experienced during lifting tasks. Since it has been noted that loadings in one plane affect the responses of the motion segment to loadings in other directions, it is clearly important that the loadings imposed in mechanical tests involving failure use loading patterns that most closely resemble those experienced in lifting tasks.

### **2.7.2 Studies Examining Ultimate Strength of Lumbar Motion segments**

As discussed previously, studies investigating the ultimate compressive strength of lumbar motion segments, while useful from a biomaterials testing perspective, probably do not accurately reflect the most common mechanism of tissue failure resulting from occupational lifting tasks. However, in addition to this inherent deficit, there are additional issues that make results of these studies difficult to relate to occupational lifting tasks. These include issues related to motion segment orientation, loads on the segments and the load rates employed.

A number of early studies (those performed in the 1950's and 60's) examined pure axial compressive loading, with no flexion of the motion segments. In many cases, load rates for these studies were not in a range characteristic of lifting tasks, often involving slowly incremented loading regimens, sometimes involving minutes (Brown et al. 1997, Perry 1957, Eie 1966). Results of these studies obviously bear very limited relevance to occupational lifting tasks.

Research in the 1980's and 1990's, predominately by Adams and colleagues, began to examine the effects of ultimate compressive strength using complex loading and flexed motion segments (Adams and Hutton 1982, Hutton and Adams 1992, Adams et al. 1993, Adams et al. 1994 a,b). However, several issues remain that obscure the relevance of these studies to occupational lifting tasks (in addition to the choice of failure model common to all studies considered in this section).

An interest in possible mechanisms of disc herniation led Adams and colleagues to examine the effects of motion segment flexion using complex loads and load rates

close to those experienced in lifting tasks (Adams and Hutton 1982, Adams and Hutton 1983, Adams and Hutton 1985). Achievement of a disc herniation (annular protrusion or nuclear extrusion) was only achieved when motion segments were flexed beyond the normal physiologic range (Adams and Hutton 1982). Hutton and Adams (1981) and Adams et al. (1994a) also reported results of compressive strength for slightly to moderately flexed specimens, but used a constant load rate not matter the degree of motion segment flexion, limiting applicability of the results to lifting tasks.

Furthermore, as has been described elsewhere, the determination of the complex loads placed on the motion segments was based on a simple static analysis of load lifting (Adams et al. 1980). As stated above, such estimates may underpredict the loads on the spine, and may predict shear forces acting in a direction opposite that predicted by contemporary biomechanical models during lifting tasks.

Several other authors have performed studies examining the ultimate strength of lumbar motion segment – all involving experimental conditions that limit, to one degree or another, their applicability to occupational lifting tasks. Among these, some have overlooked the impact of motion segment flexion (Begeman et al., 1995, Yoganandan et al. 1994), complex loads (Begeman et al. 1995, Yoganandan et al. 1994), and/or have failed to use (or have not reported) load rates anticipated in dynamic lifting tasks (Neumann et al. 1992; Osvalter et al. 1990). Several have used displacement control as opposed to load control (Adams et al. 1993, Adams et al. 1994 a,b, Yoganandan et al. 1994), which is generally considered to lead to motions that are less physiologic in nature than load control (Wilder et al. 1988).

### **2.7.3 Confounding Variables and Statistical Treatments**

Several of the studies in Tables 2.2 and 2.3 suffer from poor control of possible confounding variables and questionable assumptions which may lead one to question the authors' interpretation of the findings. In some cases authors have attempted to develop schemes to estimate the strength of one motion segment from others within the same spine (Brinckmann et al. 1988, Adams et al. 1994, Hansson et al. 1987). For example, Hansson et al. (1987) used estimates of strength derived from bone mineral content measurements to establish a scheme devised to place proportional loads on motion segments. However, as Brinckmann et al. (1988) point out, the standard error of the estimate of the regression equations for strength is approximately 1 kN. Thus, there is considerable uncertainty regarding the accuracy of the predictions of ultimate strength upon which these estimates are based. Furthermore, since ultimate strength is itself dependent on the load rate at which a specimen is tested (Hutton and Adams 1982, Yingling et al. 1997), and a variety of load rates were used in this study, the percentage of ultimate strength assigned based on the regression equation (at one load rate), may not equate to the percentage of ultimate strength at the load rate tested.

Brinckmann et al. (1988) used an estimate of the ultimate compressive strength at different spinal levels from the data of Hansson and Roos (1980). The estimates suggest an increase in strength of 0.8 kN between adjacent complete motion segments. However, no estimate of the error variability of the strength estimate is provided. Furthermore, the estimate is based on a study involving a limited number of samples at each spine level (33 total specimens), and the precision of the estimates of the ultimate

strength at different levels is not clear. Furthermore, like the Hansson et al. (1987) study, Brinckmann et al. (1988) assign a proportional load based upon a compressive strength derived at one load rate (1 kN/s), then subject the segment to a load rate that may be dramatically different from the one the estimates rely upon (up to 3.2 kN/s) where the ultimate compressive strength is expected to be considerably greater. This causes further ambiguity in terms of the interpretation of the results of this study.

Adams et al. (1994b) used a similar estimation procedure to predict the strength of motion segments in “flexed” versus “lordotic” postures. In this case, the strength between adjacent motion segments was predicted to increase by 13% per level. These authors used a matched pairs approach to attempt to estimate the effects of motion segment posture; however, questions remain regarding the accuracy of the estimates of compressive strength at adjacent levels of the spine. As discussed previously, these estimates would be expected to be subject to considerable variability, and the possible confounding effects of variability of bone mineral content between adjacent levels is not considered. While these authors concluded that “posture” had no effect on the compressive strength of lumbar motion segments, the very small sample used (7 pairs of segments) makes one wonder if sufficient statistical power was available to detect differences if they existed. Furthermore, the questionable and unknown accuracy of the predictions of motion segment strength at different levels of the spine make one further question the authority of the conclusions.

Though the study is limited and the conclusions drawn by the authors are subject to question, the Adams et al. (1994b) study represents one of the few studies that have

made a comparison using a scheme attempting to control for biologic variability through the use of statistical controls (matching in this case). In fact, the literature shows scant use of techniques that could help control for the influence of confounding variables, and improve the estimates of effects authors purport to be interested in. For example, using regression techniques that control for the effects of differences in age, gender, bone mineral density and specimen size may greatly improve the estimates of, for example, the effects of posture on the fatigue failure or ultimate strength of lumbar motion segments. In studies involving fatigue failure, where some specimens may survive the entire range of cycles tested (i.e., censored observations), techniques such as Cox regression may improve estimates of the effects of interest. Given the considerable (and generally unwelcome) biologic variability inherent in mechanical tests of lumbar motion segments, it is somewhat surprising that available tools have not often been utilized to achieve a better understanding of the effects of manipulated variables.

## **2.7 Summary**

Though studies of mechanical tests of lumbar motion segments have achieved several important insights in terms of the structure and function of the lumbosacral spine, the available literature is quite difficult to relate to the strain experienced by the spine during occupational lifting tasks, and failure of spine tissues that may result from such loading. The following points summarize several issues that currently inhibit our understanding of these effects:

- 1.) Occupational lifting tasks involve repetitive loading on the lumbar spine, suggesting that fatigue failure is a more appropriate model than ultimate compressive strength. However, the majority of studies have examined the latter rather than the former. Of the handful that have examined fatigue failure, none have subjected motion segments to the combinations of compression and shear forces and load rates predicted to occur in lifting using contemporary biomechanical models.
- 2.) Though we possess a reasonably good understanding of the positioning of lumbar motion segments during trunk flexion *in vivo*, no studies have yet attempted to recreate the combination of position, load, and load rates predicted to occur in the positions of torso flexion common to occupational lifting tasks.
- 3.) Many studies in the literature seem insensitive to the effects of load rate on the response of lumbar motion segments. Load rates experienced in flexed postures during lifting tasks would be expected to be much higher (up to an order of magnitude higher) than those experienced in a neutral position. Thus, physiologic fidelity of lifting tasks would require flexed motion segments to be loaded at a higher rate than motion segments set in a neutral posture. No studies have yet examined the impact of higher load rates on flexed specimens.
- 4.) Many studies have based lifting simulations on static biomechanical models that have been shown to underpredict compressive loads and provide questionable estimates of shear forces acting on the motion segments in lifting. Since the responses of motion segments to loads in one direction are contingent upon

those experienced in another, use of biomechanical models that incorporate dynamic effects would permit improved simulations of spinal loads experienced in repetitive lifting tasks.

- 5.) Existing studies often appear to provide insufficient control of confounding variables, which cloud both the results of these studies and their interpretation. Use of more sophisticated statistical techniques may do much to clarify the true effects of the independent variables under study.

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## **CHAPTER 3**

# **DETERMINATION OF COMPRESSION AND SHEAR LOADS ON LUMBAR SPINE MOTION SEGMENTS IN NEUTRAL AND FLEXED POSTURES**

### **3.1 Introduction**

If one is interested in failure mechanisms of the lumbar spine resulting from mechanical overload experienced in lifting activities, it is important to understand how loads on the spine are distributed in the context of the posture adopted by the torso. The motion of torso flexion is accomplished through a change in the curvature of the lumbar spine. Specifically, the lordosis observed in the neutral posture is reduced at the lower levels of the lumbar spine, and may be reversed (i.e. a kyphosis may be achieved) at the upper levels of the spine (Bogduk 1997). This process is the result of anterior rotation and anterior translation of the lumbar vertebrae.

The change in lumbar curvature necessitates changes in the orientation of the paraspinal musculature as well. Specifically, the rotation and translation of the vertebrae of the lumbar spine result in an elongation of the extensor musculature (reducing force producing capabilities of the muscle via the length-strength

relationship of the muscle), a resultant decrease in the cross-sectional area a reduction in the effective moment arm of these muscles. In extreme torso flexion, the extensor musculature exhibits electrical silence and the load bearing responsibilities shift to the posterior ligaments (Basmajian and DeLuca 1985, Floyd and Silver 1955, Gracovetsky and Farfan 1986).

Since the purpose of this dissertation was to evaluate the fatigue failure of lumbar motion segments when performing lifting tasks in various degrees of trunk flexion representative of manual lifting activities, it was necessary develop the following procedures: (1) determine the compression and shear loads acting at L5-S1 using the OSU biodynamic model in neutral (0 degrees torso flexion), 22.5 degrees lumbar flexion, and 45 degrees of lumbar flexion, (2) To calculate the sagittal resultant force vector, (3) To determine the orientation of the superior endplates of the lumbar and sacral vertebral bodies *in vivo* in the three orientations described above, and (4) To translate the resultant force vector to the upper lumbar motion segments (L3-L4 and L1-L2) and determine the compressive and shear components of the resultant given the orientation of segments in the various postures. The following sections describe in detail the procedures used to develop these load estimates.

### **3.2 Development of Loading Profiles at L5-S1 at Torso Flexion Angles of 0, 22.5 and 45 Degrees**

Loading profiles for lifting tasks initiated at 0, 22.5 and 45 degrees of trunk flexion were estimated using an EMG-assisted biomechanical model which has been under development in the Ohio State University Biodynamics Laboratory over the past

18 years (Marras and Reilly 1988, Reilly and Marras 1989, Marras and Sommerich 1991a, b, Granata and Marras 1993, Granata and Marras 1995, Mirka and Marras 1993, Marras and Granata 1995, Marras and Granata 1997). This model provides estimates of spine loading parameters based upon measured activity of ten trunk muscles, from which estimates of muscle force and subsequent spine loading are determined (Marras and Granata 1997). Forces for each muscle are estimated using the following equation:

$$F_j = PCSA_j \times Gain \times \frac{EMG_j(t)}{EMG_{Max-j}} \times f(L-S) \times f(F-V) \quad (1)$$

where:

$F_j$  = predicted muscle force for muscle j;

$PCSA_j$  = physiological cross-sectional area for muscle j;

Gain = estimate of muscle stress (force/area);

$EMG_j(t)$  = instantaneous integrated EMG for muscle j at time t;

$EMG_{Max-j}$  = integrated EMG from maximum voluntary contraction for muscle j;

$f(L-S)$  = muscle length-strength relationship modulation factor;

$f(F-V)$  = muscle force-velocity relationship modulation factor.

The muscle generated internal moments about the axis of rotation are predicted from the sum vector products combining the tensile muscle force for each muscle and the moment arms of each respective muscle as given by equation (2):

$$M_i = \sum r_{ij} \times F_{ij} \quad (2)$$

where:

$M_i$  = predicted internal moment for the  $i^{\text{th}}$  plane;

$r_{ij}$  = moment-arm for muscle j in the  $i^{\text{th}}$  plane;

$F_{ij}$  = vector force component for muscle j in the  $i^{\text{th}}$  plane.

The model has been validated under forward trunk bending motions (Marras and Sommerich 1991a, b, Granata and Marras 1993, Granata and Marras 1995), trunk twisting motions (Marras and Granata 1995), lateral bending motions (Marras and Granata 1997), and lowering tasks (Davis et al. 1998).

A database of lifting tasks from a previous study (Marras et al. 2001) was used to obtain compression and shear estimates, along with rates of loading on the L5-S1 joint when performing lifts of 4.5, 6.8, 9.1, or 11.4 kg at trunk flexion angles of 0, 22.5, and 45 degrees using the Lumbar Motion Monitor (LMM). Sagittal flexion angles obtained from the LMM have demonstrated an excellent correspondence to torso flexion (Jorgenson 2001). A sample of 27-31 lifts starting at each of these trunk flexion angles was obtained and analyzed to establish appropriate compression, shear, and load rates using this model. Lifting trials were included in this analysis if they started within 2 degrees of the desired trunk flexion angle. The compression, shear and load rate values obtained in this analysis were taken at the point of peak sagittal plane resultant force acting on the spine during the lift. Results of this analysis are provided in Table 3.1.

Since different weights were lifted in this study, it was thought worthwhile to check to make sure that lifting trials sampled at for each torso flexion angle represented lifts of approximately the same average weight. In this process, a bias was discovered. File selected for analysis in the neutral (0 degree torso flexion) posture tended to be ones where less weight was lifted (6.1 kg on average) than in 22.5 degrees flexion (7.7 kg), which was lower still than the average weight lifted at 45 degrees (8.4 kg). It was surmised that heavier weights lifted in the more erect postures

Trunk Flexion Angle (degrees)	n	Averaged Peak Compression Force (N)	Averaged Peak A-P Shear Force (N)	Averaged Peak Resultant Force (N)	Angle of Peak Resultant (degrees)	Averaged Peak Loading Rate (N/s)	Average LMM Angle at Peak Resultant (degrees)
0	27	1019.7 (± 252.4)	425.8 (± 98.3)	1106.5 (± 268.9)	-22.7	653.4	0.5 (± 0.9)
22.5	29	2264.5 (± 891.6)	268.5 (± 116.9)	2284.6 (± 888.3)	-6.3	2087.1	22.3 (± 3.1)
45	31	3117.1 (± 1071.8)	-375.2 (± 222.3)	3141.4 (± 1089.3)	6.9	4656.4	44.1 (± 1.6)

**Table 3.1:** Results of analysis of compression, shear, load rates and resultant forces acting on L5-S1 discs estimated by the OSU EMG-assisted biomechanical model. Positive A-P shear reaction forces (and negative peak resultant angle values) represent a shear force where the superior vertebra is shearing anteriorly on the inferior vertebra

caused the torso to bend forward to a point where they were beyond the trunk flexion criterion. For this reason, regression analyses were used to estimate the influence of weight on the averaged peak resultant force and to make appropriate adjustments so that the resultant in each posture represented the lifting of an equivalent weight. Regression results indicated that each additional kilogram of weight lifted (when controlling for subject and torso angle effects) imposed an additional 52.3 N of resultant force. This regression result was used to adjust all resultant forces to that predicted by the regression when an equivalent load (established as 9.1 kg) was lifted in each posture. Results of this adjustment procedure produced the following resultant force estimates: 1279 N for 0 degrees trunk flexion, 2378 N in 22.5 degrees trunk flexion, and 3171.8 N for 45 degrees trunk flexion. To simplify subsequent analyses, these resultant force values were rounded to 1300 N, 2400 N, and 3150 N for 0 degrees, 22.5 degrees and 45 degrees of trunk flexion, respectively.

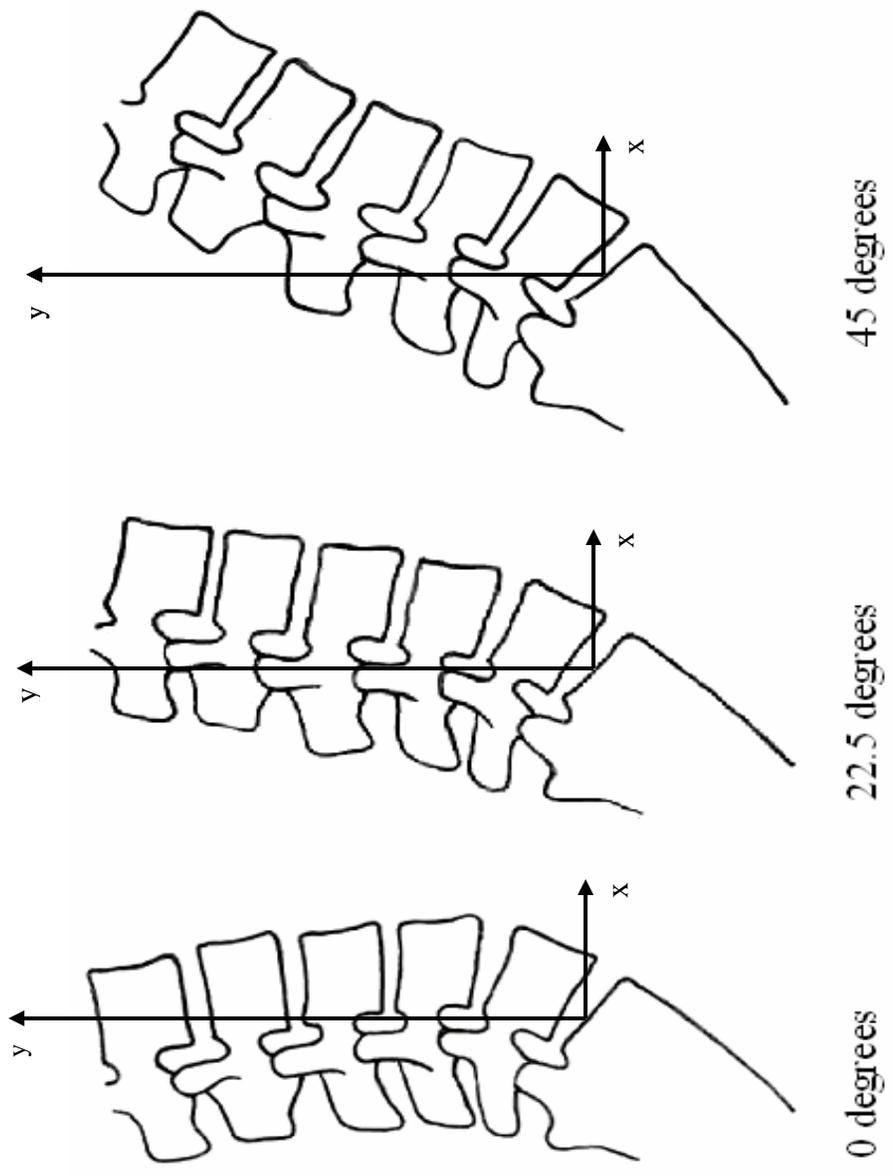
### **3.3 Determination of the Orientation of L2, L4, and S1 Superior Endplates in Neutral and Flexed Torso Postures**

The orientation of the lumbar and sacral vertebrae in various levels of trunk flexion was determined using data and regression equations provided by Chen (2000). This study utilized sixteen healthy male subjects who were radiographed using high quality film at torso flexion angles of 0, 30, 60, and 90 degrees in relation to a vertical orientation, as defined by a line running from the subject's hip joint to the shoulder joint. The orientation of the superior endplates of the lumbar vertebrae and sacrum

were obtained by digitizing the vertebrae and determining the angle of the line describing the superior endplate surface with respect to the horizontal.

Regression techniques (both linear and exponential equations) were utilized by Chen (2000) to predict superior endplate inclinations of the lumbosacral vertebrae as a function of torso flexion throughout the range studied. These prediction equations resulted in excellent fits to the data, with all six prediction equations having  $R^2$  values of greater than 0.95. It was found that the upper levels (L1-L3) were well described by linear equations, while the lower levels (L4-S1) were best described by exponential fits.

For the purposes of the current investigation, superior endplate angles were required for L2, L4, and S1 at torso flexion angles of 0 degrees, 22.5 degrees, and 45 degrees. Chen (2000) provide observed data on superior endplate angles derived from radiographic measurement at 0 degrees torso flexion, and these were used as provided by this author. Superior endplate angles for L2, L4, and S1 at 22.5 and 45 degrees of torso flexion were obtained through the use of the regression equations developed by Chen (2000). Results of these analyses are contained in Table 3.2. Figure 3.1 depicts graphically the orientation of the spinal column at the three levels of torso flexion simulated in this experiment.



**Figure 3.1:** Graphic representation of the predicted orientation of the lumbar spine in neutral and forward flexed postures

	Torso Flexion Angle		
	0 degrees	22.5 degrees	45 degrees
L2	10	- 9	-32
L4	-3	- 15	-25
S1	-38	-41	-49

**Table 3.2:** Superior endplate angles for L2, L4 and S1 vertebrae at three torso flexion angles determined from data and regression equations developed by Chen (2000). Negative values indicate an endplate sloping downward (back to front) with respect to the horizontal. Positive values indicates an endplate sloping upwards (back to front) compared to horizontal.

### **3.4 Procedure for Estimation of Compression and Shear Forces at L1-L2 and L3-L4 at 0, 22.5 and 45 degrees of Torso Flexion**

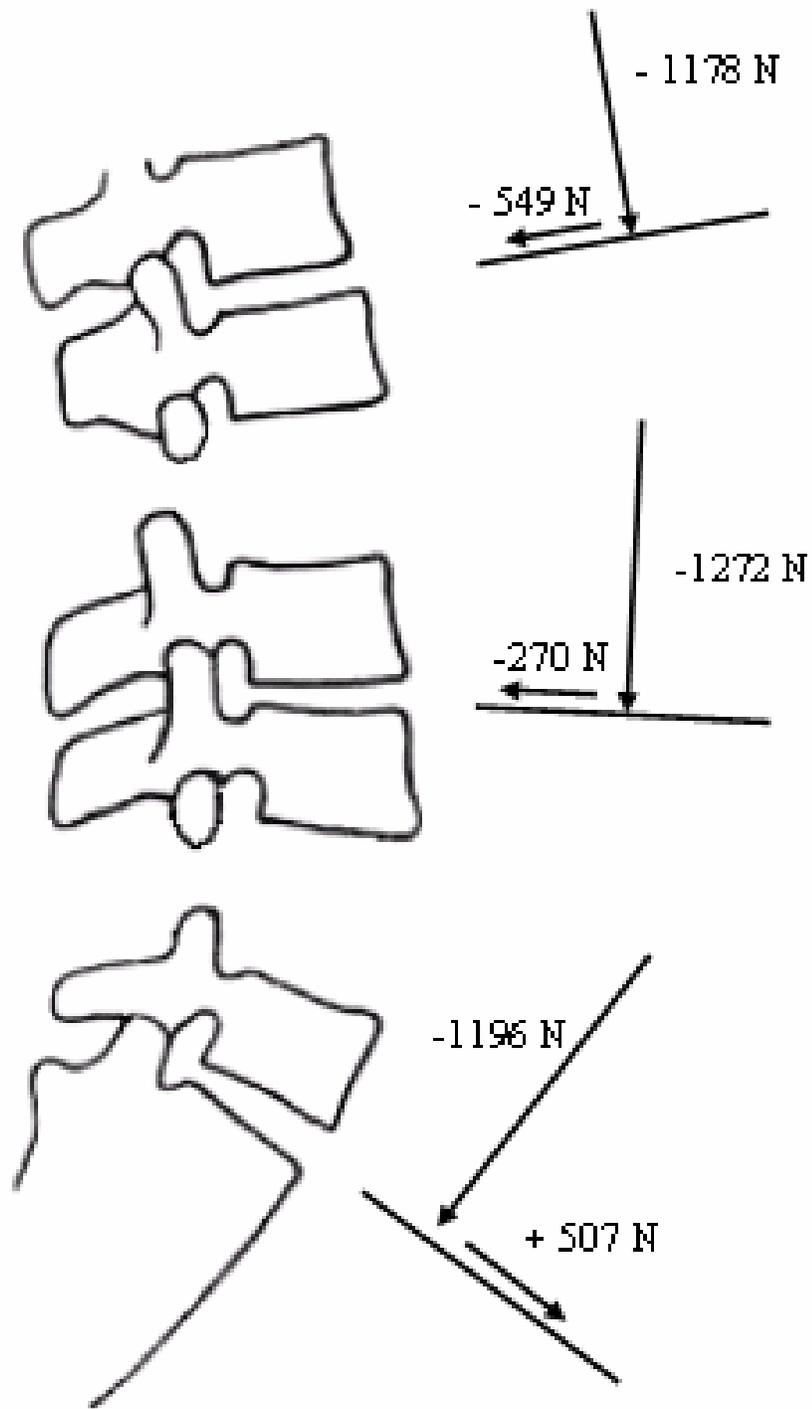
It should be noted that the OSU Biodynamics Model was designed to provide estimates of forces and moments at the lumbosacral junction (Granata and Marras 1993). However, for the current experiment, estimates of compression and shear forces were needed for levels L1-L2 and L3-L4, as well. Since no direct estimation could be made using this model, a simplifying assumption was made that the magnitude and direction of the loads and the rate at which loading was experienced at L1-L2 and L3-L4 were equivalent to that experienced at L5-S1. While it must be acknowledged that the loads on middle and upper levels of the lumbar spine are

somewhat less than those at the lumbosacral joint, biomechanical models typically suggest that the differences are relatively small (Daggfeldt 2002).

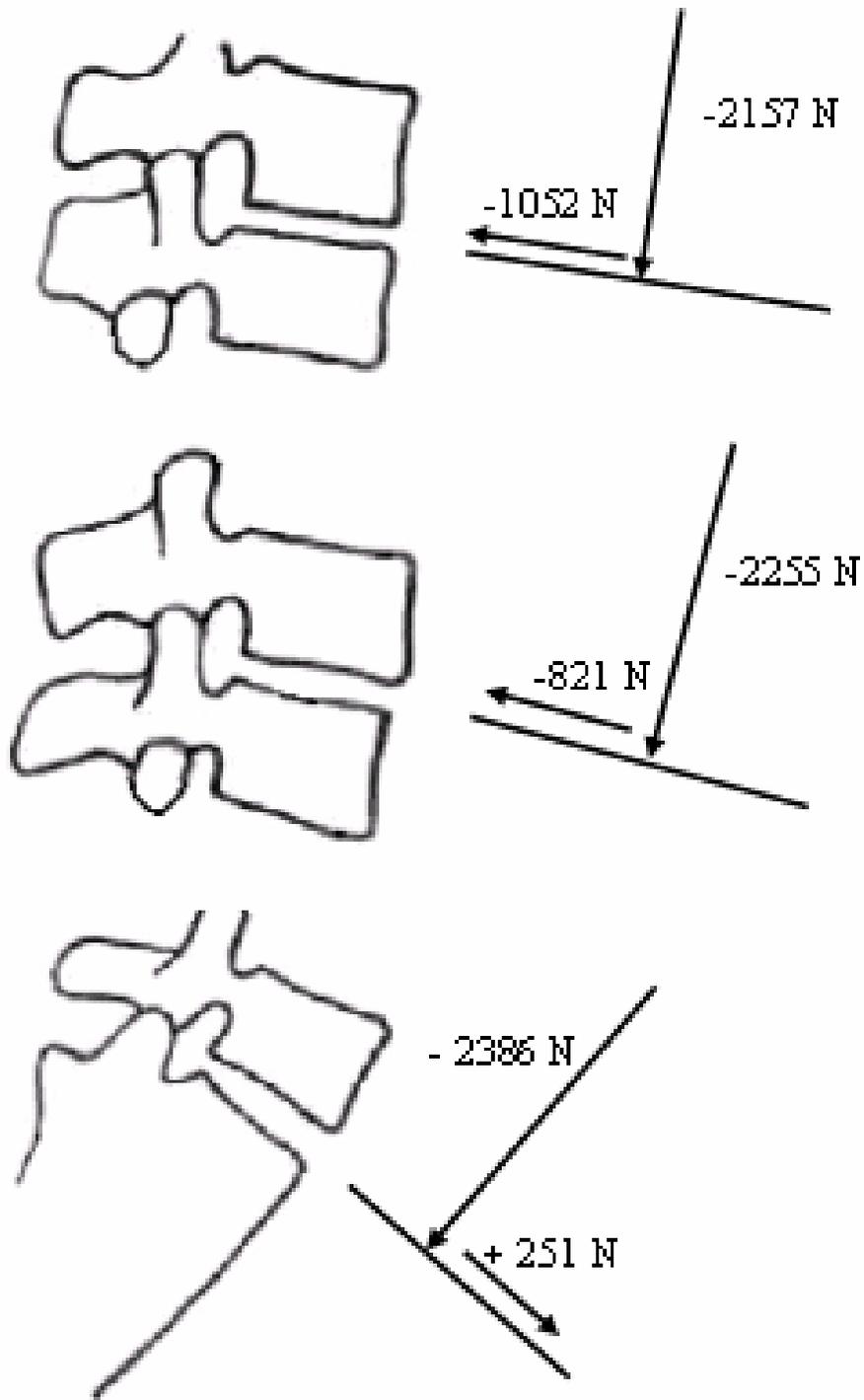
Once the compression and AP shear forces were calculated at L5 –S1, and a sagittal resultant vector magnitude and angle calculated, the vector was translated to the superior endplates of L2 and L4 to estimate the relative compression and shear forces acting on the upper lumbar levels. These calculations of compression and shear forces at the upper lumbar levels were based upon the magnitude and direction of the vector calculated at L5-S1, and the orientation of the vertebrae at the upper levels of the lumbar spine, as calculated from the data provided by Chen (2000).

### **3.5 Results of the Analysis of Loads on the Lumbar Spine at 0, 22.5 and 45 degrees of Torso Flexion**

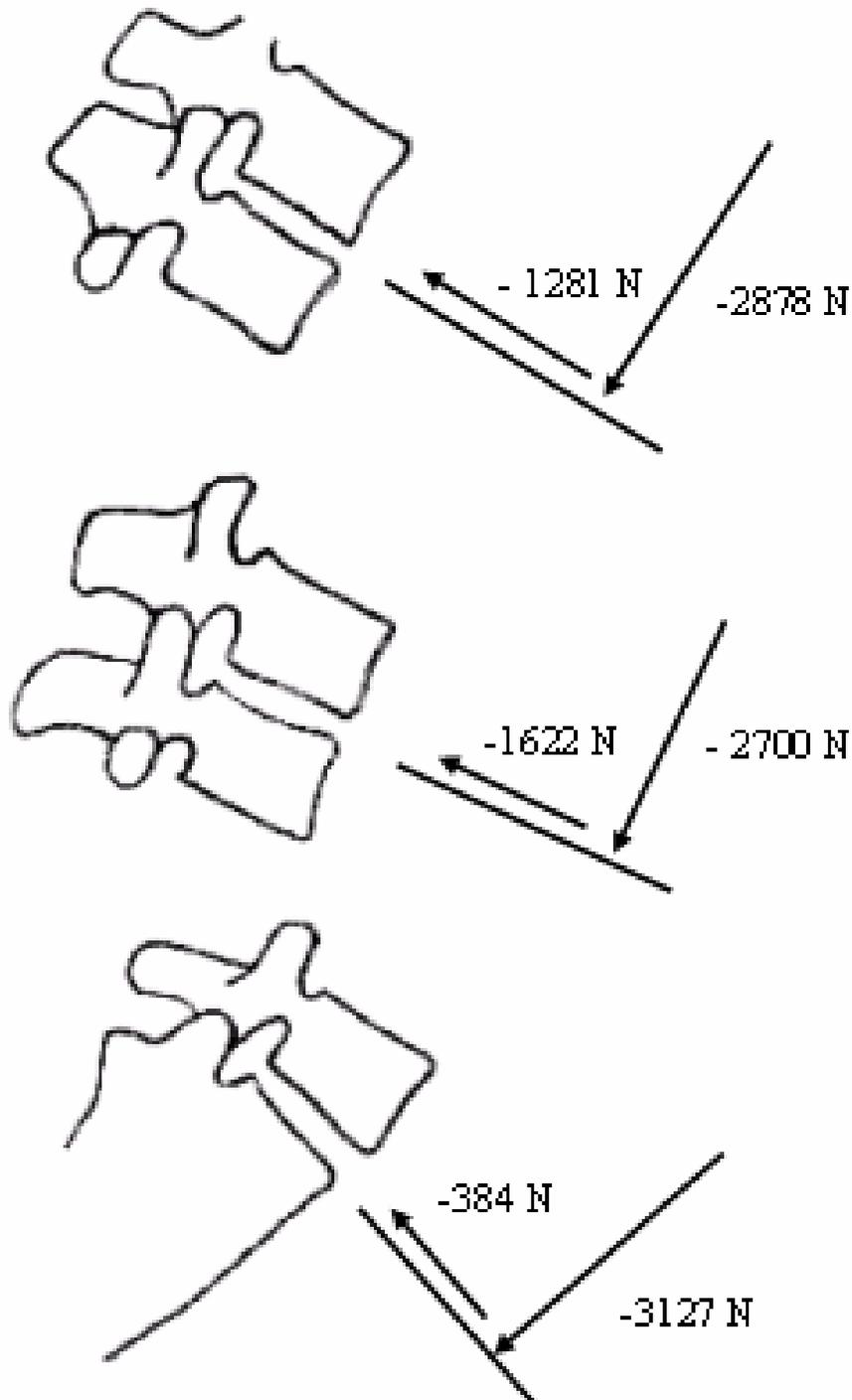
Figures 2.2 - 2.4 present the results of analyses of the compression and shear forces on L5-S1, L3-L4, and L1-L2 motion segments of the lumbar spine at 0, 22.5 and 45 degrees of torso flexion. Each figure illustrates the predicted orientation of each of the motion segments, and to the right of each motion segment graphic is a representation of the predicted compression and shear reaction forces acting on the superior endplate of the inferior vertebra of that motion segment. Compression reaction forces act in a negative direction in the z axis, which is considered orthogonal to the line describing the superior endplate of the inferior vertebra. Negative values for AP shear indicate a posterior reaction force and vice versa.



**Figure 3.2:** Compression and AP shear reaction forces acting on the superior endplate of the inferior vertebra of motion segments L1-L2, L3-L4, and L5-S1 with the torso in the neutral posture (0 degrees flexion)



**Figure 3.3:** Compression and AP shear reaction forces acting on the superior endplate of the inferior vertebra of motion segments L1-L2, L3-L4, and L5-S1 with the torso flexed forward 22.5 degrees



**Figure 3.4:** Compression and AP shear reaction forces acting on the superior endplate of the inferior vertebra of motion segments L1-L2, L3-L4, and L5-S1 with the torso flexed forward 45 degrees

### 3.6 Determining the degree of flexion of the motion segments at 0, 22.5 and 45 degrees of Torso Flexion

Torso flexion in the sagittal plane is accomplished in part by the anterior rotation and anterior translation of the lumbar vertebrae (Bogduk 1997, DeLisle et al. 1997, Frobin et al. 1996, Gracovetsky et al. 1987). In addition, to reorienting the angle of the endplates, this process causes a flexion of the lumbar motion segments. Each lumbar motion segment flexes approximately the same amount with only slightly greater flexion observed in the central motion segments of the lumbar spine (Pearcy et al. 1984, Pearcy and Tibrewal 1984). Establishing appropriate motion segment flexion values is necessary, therefore, to accurately reproduce the spinal loading incurred when the torso is in a forward flexed position.

Jorgensen (2001) provided data on *in vivo* motion segment flexion at for all lumbar motion segments during torso flexion obtained via magnetic resonance imaging. Data on flexion of L1-L2, L3-L4, and L5-S1 motion segments for both male and females were subjected to a regression analysis, which were then used to predict the amount of flexion from the neutral posture for each motion segment at torso flexion angles of 22.5 and 45 degrees. Table 3.3 provides a summary of predicted motion segment angles based on the following regression equation:

$$\begin{aligned} \text{Motion segment flexion angle (degrees)} = & 17.25 - 0.14 * \text{trunk flexion angle} \\ & (\text{degrees}) - 15.64 * L1 - 5.41 * L3 + 0.09 * \text{trunk flexion angle} * L1 - 0.02 * \text{trunk} \\ & \text{flexion angle} * L3 \quad (\text{Adjusted } R^2 = 0.97) \end{aligned} \quad (3)$$

	Trunk Flexion Angle		
	0 degrees	22.5 degrees	45 degrees
Predicted L1-L2 angle	1.6	0.4	-0.9
Degrees of flexion from neutral	0	1.2	2.5
Predicted L3-L4 angle	11.8	8.7	5.5
Degrees of flexion from neutral	0	3.1	6.2
Predicted L5-S1 angle	17.2	14.1	10.9
Degrees of flexion from neutral	0	3.1	6.3

**Table 3.3:** Predicted Cobb angles for *in vivo* lumbar motion segments based on regression analysis of data from Jorgenson (2001).

### 3.7 Discussion

Results of analyses of L5-S1 disc loads in neutral and forward flexed torso postures illustrate the striking influence that torso posture has on the strain experienced by the low back at the point of peak force during lifting tasks. Compared to the neutral posture, for example, adopting a 22.5 or 45 degrees torso flexed posture results in a doubling or tripling of the compressive load on the spine, as well as an increase in the variability of the loading. Changes in torso posture also resulted in changes in the magnitude and direction of the A-P shear forces. Shear reaction forces in the neutral posture and in partial trunk flexion were both directed anteriorly with less shear evident in partial flexion. In full flexion, however, the shear reaction force

was directed in a posterior direction. The magnitude of the resultant force, dominated by compression, closely followed that of the compressive component; however, the change in magnitude and direction of the A-P shear forces caused the angle at which the peak resultant was operating to swing nearly 30 degrees over the range of torso flexion angles studied.

While anterior shear forces were experienced at the lumbosacral joint in neutral and 22.5 degrees of torso flexion, when the resultant was translated to upper levels of the lumbar spine, shear forces acted in a posterior direction no matter the posture. These shear forces were fairly modest at the upper lumbar levels in the neutral posture, but were estimated to be quite substantial in full flexion. Shear values obtained in the most extreme torso flexion conditions were greater than 1200 N, approaching the loads (i.e. 1800-2800 N) at which shear failure of the motion segment may be observed (McGill 2003). Interestingly, the average shear loads calculated at the lumbosacral disc remained modest no matter which posture was used, being less than 430 N in all cases. It may be reasonable to speculate that the anatomy of the lumbar spine, in particular the lumbosacral junction, with its significant lordosis, may play an important role in limiting the amount of shear experienced at the lumbosacral joint, resulting in a predominately compression load without much shearing, throughout a wide range of torso flexion.

Results of the regression analysis of data regarding the flexion of motion segments from Jorgensen (2001) is in line with that of other researchers who have examined the flexion of motion segments in trunk flexion (Chen 2000, Pearcy et al. 1984, Pearcy and Tibrewal 1984). Results of this analysis suggest that the greatest

flexion occurs at the lower and middle levels of the lumbar spine with a lesser contribution at the L1-L2 motion segment. The amount of motion segment flexion observed at 45 degrees of trunk flexion suggests that motion segments at all levels are still capable of several additional degrees of flexion based on results of studies examining full range flexion of the spine (Chen 2000, Percy et al. 1984, Percy and Tibrewal 1984).

This analysis of the loads on the lumbar spine in neutral and flexed torso postures provides a great deal of information useful to the modeling of fatigue failure of motion segments of the lumbar spine. However, certain limitations must be acknowledged. For example, the simplifying assumption that loads at upper levels of the lumbar spine are equivalent to those at L5-S1, while considered to be a reasonable estimate, is probably not precisely correct. We would anticipate that the magnitude at the upper levels to be somewhat less than that at the lumbosacral junction. Furthermore, the effects of zygapophysial joint load-sharing and changes in ligament recruitment as a function of spine flexion are not taken into account by the current model. In addition, estimates of the endplate positions of the lumbar spine are based on a population of Chinese males and may not be the best representation the spines being tested in this study. However, the study was considered the best data set available describing the endplate angles of the lumbar spine throughout the range of torso flexion in a standing posture. Motion segment flexion angles were obtained from a study where subjects (both males and female) were situated in an MRI in a side-lying position, which may not perfectly represent the flexion of motion segments

in a standing posture. However, the values obtained in these analyses correspond well to one another, and the results are consistent with those obtained by other authors.

### **3.8 Conclusions**

Based on the analysis of loads acting on the lumbar motion segments in neutral and flexed torso postures, the following conclusions are drawn:

- 1.) The magnitude of the peak resultant force in 22.5 and 45 degrees torso flexion is approximately double and triple that observed in the neutral posture, respectively.
- 2.) Shear forces remained modest at the lumbosacral joint and were directed anteriorly in neutral and 22.5 degrees of torso flexion and were posterior in direction in 45 degrees of torso flexion.
- 3.) Shear forces are quite significant at the upper lumbar levels in full flexion, and approach limits recommended by some authors in mild flexion. Shear reaction forces were always found to act in a posterior direction at the upper motion segments.

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## **CHAPTER 4**

# **EFFECTS OF TORSO FLEXION ANGLE AND SPINAL LEVEL ON THE FATIGUE FAILURE OF HUMAN LUMBOSACRAL MOTION SEGMENTS**

### **4.1 Introduction**

Recent reviews of the epidemiology literature have concluded that a number of work-related physical factors show consistent and positive associations with the occurrence of low back disorders (Hoogendoorn 1999, National Academy of Science 2001, National Institute for Occupational Safety and Health 1997). Among the occupational factors exhibiting positive associations with low back disorders are jobs involving manual materials handling activities (Frymoyer et al. 1983, Hoogendoorn et al. 1999, Kelsey et al. 1984, Riihimaki 1991), jobs involving frequent bending and twisting (Holmstrom et al. 1992, Marras et al. 1993, Punnett et al. 1991), and jobs where workers bear a heavy physical workload (Burdorf et al. 1990, Heliovaara et al. 1991, Videman et al. 1990). A common theme among the workplace factors linked to low back disorders is that they all involve circumstances where the tissues of the lumbar spine will be placed under high levels of compression, shear, and torsional

forces (Chaffin and Park 1973, Granata and Marras 1995, Nachemson 1965, Schultz and Andersson 1982, White and Panjabi 1978). Furthermore, in jobs involving intensive lifting responsibilities, the lumbar spine may incur high levels of loading hundreds or even thousands of times in a workday (Magora 1972, Marras et al. 1993). Given the association between high spinal loads and low back disorders, it is not surprising that *in vitro* estimates of the ultimate compressive strength of lumbar motion segments and estimates of the loads experienced *in vivo* reveal a relatively small margin of safety (Brinckmann et al. 1988). In fact, given the high incidence of low back disorders in the population, one might question the extent to which a “margin of safety” can be said to exist at all.

If the stress experienced by a material (such as those comprising the lumbar spine) due to a single loading cycle exceeds its ultimate strength, material failure is the inevitable result. However, it is essential to realize that spinal tissues, like all materials, are also vulnerable to fatigue failure (Litsky and Spector 1999). When tissues experience repetitive loading, material failure can and will occur at loads well below the ultimate material strength (Brinckmann et al. 1988, Hansson and Roos 1981). The relevance of fatigue failure to an understanding of the etiology of work-related low back disorders would seem abundantly clear, especially given the repetitive and high magnitude loads that can be experienced by lumbar tissues in occupational settings (Marras et al. 1993).

Though one can state with certitude that fatigue failure occurs in lumbar tissue, only a handful of studies have actually examined the responses to repetitive loading of

human lumbar motion segments. As detailed in Chapter 2, findings from studies examining the issue suggest that failure can occur in lumbar tissues after a fairly modest number of loading cycles when exposed to loads in the physiologic range. For example, Liu et al. (1983) reported fatigue failure in almost half of all specimens after less than 2000 loading cycles. Hansson and Roos (1981) applied loads between 60% and 100% of predicted ultimate compressive strength and found that 16 out of 17 motion segments failed in fewer than 1000 cycles. In Brinckmann's et al.'s (1988) study, greater than half of all motion segments tested failed after only 500 loading cycles at a load predicted to be 40-50% of the predicted ultimate strength of a motion.

A key determinant of the load experienced by the lumbar spine in manual load lifting is the degree to which the torso is flexed (Andersson et al. 1977, Schultz et al. 1982). Torso flexion is quite common in occupational lifting activities (Marras et al. 1993, 1995; Norman et al. 1998, Hoogendoorn et al. 2000), and bending in combination with lifting appears to be a common source of low back pain (Garg and Moore 1992). Biomechanical analyses suggest that spinal loads can be doubled or triple when lifting a given weight, depending on the torso flexion angle (Shultz et al. 1982). Not only is the load magnitude greater when the torso is flexed, but the blend of compression and shear forces acting on the spinal segments will be modified, with an increasing shear component in flexed postures (Macintosh et al. 1993, Potvin et al. 1991). Moreover, the rate at which the load is applied to spinal tissues will be increased when lifting with a flexed torso. All of these factors may play significant roles in the development of fatigue failure in spinal tissues.

Apprised of the information and associations described above, it would seem evident that a better understanding of the effects of torso flexion on fatigue failure of the lumbar spine is needed. To date, no study has attempted to evaluate fatigue failure of lumbar spine motion segments subjected to postures and loads characteristic of occupational lifting tasks performed at various degrees of sagittal torso flexion. Accordingly, the purpose of the present experiment is to examine the fatigue failure of cadaver lumbosacral motion segments to cyclical spinal loads predicted to occur when lifting moderate weights in postures design to carefully simulated various angles of torso flexion.

## **4.2 Method**

### *4.2.1 Cadaveric Material*

Twelve fresh, frozen lumbosacral spines (sacrum through L1) were obtained from the anatomical gift program at Wright State University. Spines were excised within 24 hours after death from subjects having no history of spinal disease or prolonged bed rest prior to death. Surrounding soft tissues were kept intact. The frozen spines were transported to the Biomaterials Laboratory at the Ohio State University, stored in sealed plastic bags, and placed in a freezer dedicated to cadaver specimens.

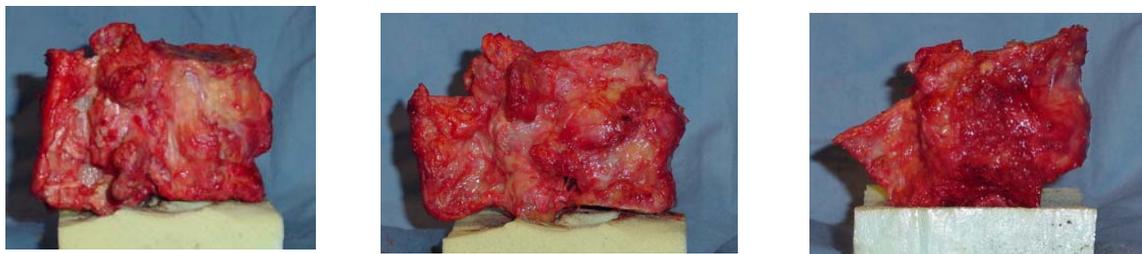
### *4.2.2 Dissection and preparation of motion segments*

Specimens were thawed at room temperature prior to dissection and testing. Intact lumbar spines were dissected using scalpels, rongeurs, forceps and other suitable instruments into three separate motion segments: L1-L2, L3-L4, and L5-S1

(the sacrum was sectioned at a level approximating S1-S2). Motion segments consisted of two complete vertebrae and intervening tissue. Excess musculature, adipose tissue, and fascia were removed from the motion segments and only ligaments spanning multiple levels were sectioned. Tissues dissected out during this process were saved in biohazard disposal bags or appropriate labeled containers for return to the Wright State Anatomical Gift Program. Upon completion of the dissection, motion segments were visually inspected and notations made of any unusual findings. Anterior, posterior, right lateral, left lateral and axial (top view) photographs were taken of each motion segment. Figure 4.1 shows an intact spine, while Figure 4.2 illustrates a specimen sectioned into L1-L2, L3-L4, and L5-S1 motion segments.



**Figure 4.1:** Intact lumbosacral spine specimen prior to dissection.



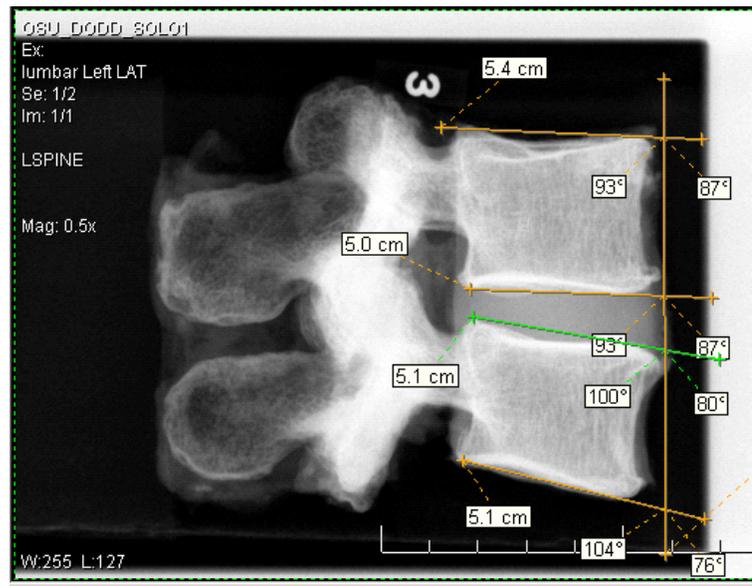
**Figure 4.2:** Motion segments L1-L2, L3-L4, and L5-S1 after spine dissection.

#### 4.2.3 *Pre-test radiological examination*

Anterior-posterior (A-P) and left lateral (LLAT) views of each motion segment were taken using the following settings: 50 kiloVolts, 5 milliAmps, and a film focus distance of 38". These initial radiographs served to detect the presence of existing defects in individual motion segments and were used to make various measurements of intact motion segments via measurement tools provided in eFilm xray viewing software (Merge eFilm, Milwaukee, WI). Lateral films were examined to determine the relative angles of all four endplates of the motion segment, as illustrated in Figure 4.3. The angle of the anterior surface of the sacrum was also measured to assist in positioning the L5-S1 motion segment, as described in greater detail below. Motion segments were held in position for A-P and LLAT views via foam rubber containing cutouts designed to hold the motion segments in the appropriate orientation.

Endplate angles were estimated by determining the location of the four corners of the vertebra, for both superior and inferior surfaces. Lines were drawn to describe the orientation of each endplate in the sagittal plane. Two points on the superior or

inferior surface of the vertebral body in the sagittal plane defined these lines. The points were determined as those that would define the points of contact if a straight edge were laid across the superior or inferior surface of the vertebral body. This definition facilitated appropriate positioning of the motion segments during fixation since inclinometer readings using a straight edge over these same points could be used to position the motion segment in the potting trays. The angles of lines representing the planes of the superior and inferior endplates were determined and the relative angle between the two were calculated using analysis tools provided on the eFilm computer program or, alternatively, on OSURADWEB radiology web site of the Ohio



**Figure 4.3:** Determination of endplate angles from motion segment radiographs.

State University Hospital Radiology Department. These angles were used in the process of establishing the proper orientation of the motion segments when the segments were placed in the specimen trays.

#### *4.2.4 Specimen Mounting Procedure*

Specimens were mounted in two trays containing the mildly exothermic fixative polymethylmethacrylate (PMMA). Each cup was made of a flat base plate onto which was attached two semi-cylindrical steel walls about 4 cm high and approximately 12 cm inside diameter. These walls, which were coated by a thin layer of petroleum jelly before potting of the specimens was performed, were designed to facilitate removal of the specimen still mounted in the fixative. The motion segments were placed in the trays such that the center of the motion segment was in the center of the cylinder. The intent was to align specimens so that the superior endplate of the inferior vertebral body was placed parallel to the edge of the lower tray (i.e., in a horizontal orientation). The horizontal orientation of the superior endplate of the inferior vertebra was determined by determining the sagittal angle at which the inferior endplate should be mounted to obtain a horizontal superior endplate, based on the angles obtained from the radiological measurements described earlier (Figure 4.4).

The motion segment was positioned using a jig which held the motion segment in the proper orientation. This jig was attached to a drill press (positioned in a fume hood) which was used to lower the specimen into the tray filled approximately 2/3 full of PMMA. The spinous process and center of the anterior vertebral body of the motion segment were aligned with a mark drawn on the bottom of the tray which bisected the



**Figure 4.4:** The inferior endplate of the inferior vertebra was measured with a goniometer and positioned such that the superior endplate would be horizontal when potted in the tray, based on endplate angles determined from earlier radiographs.

widest dimension of the tray. This helped to establish the sagittal orientation of the motion segment. A weight was used to hold the drill press in the downward position while the bone cement set. The bone cement was allowed to set until tapping of the cement using a wooden stick indicated the cement had hardened. Inferior vertebrae of all motion segments for a spine were potted in this manner (see Figure 4.5).

#### *4.2.5 Radiographic examination with inferior vertebra potted*

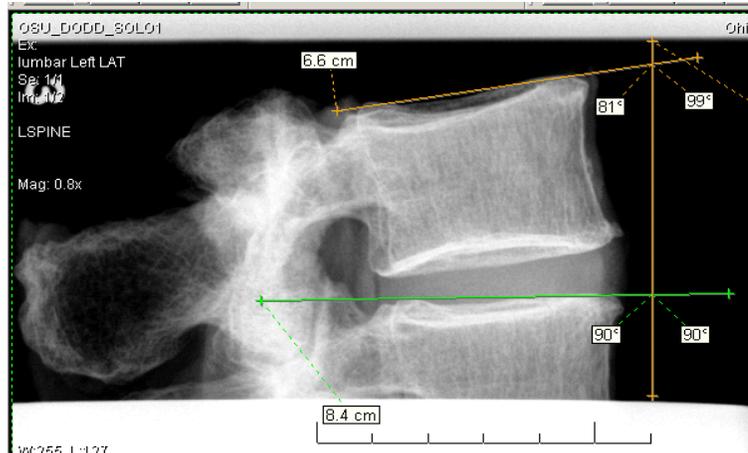
Once all three inferior vertebrae had been potted, specimens were taken back to the radiology clinic. Trays containing the motion segments were placed on the radiology table and a left lateral radiograph was obtained. These radiographs were used to ascertain whether the superior endplates of the inferior specimens were indeed



**Figure 4.5:** Motion segment, held in position by the jig, was lowered into polymethylmethacrylate (PMMA) using a drill press.

in a horizontal orientation (see Figure 4.6), and to determine the Cobb angle of the motion segment (angle between the two superior endplates of the motion segment). If the superior endplate of the inferior vertebra was found not to be in a horizontal orientation, metal shims were used to adjust the specimen so that the superior endplate of the inferior segment was at the desired angle for the cyclic loading test.

Based on the Cobb angles measured in the second radiograph, the superior vertebra of each motion segment was potted such that an angle representative of a neutral posture for each motion segment was achieved when the upper and lower specimen trays were parallel to one another. Neutral posture Cobb angles for the three motion segments studied were operationally defined as: 20 degrees for L5-S1, 8 degrees for L3-L4, and 0 degrees for L1-L2, based on data provided by Chen (2000).



**Figure 4.6:** Crossfire radiograph of L3-L4 specimen potted in inferior tray. Measurements of superior endplate angles taken to ensure inferior segment endplate was horizontal and to establish Cobb angle of the specimen (9 degrees in this case).

As an example of this procedure, Figure 4.6 shows an L3-L4 motion segment with a Cobb angle of 9 degrees. Since the desired Cobb angle in the neutral posture for an L3-L4 motion segment is 8 degrees, this motion segment would be potted “in extension” by one degree, so that when trays were parallel, an 8 degree neutral posture Cobb angle was achieved. Additional flexion of the motion segments, necessary to simulate the 22.5 and 45 degree torso flexion angles, were calculated relative to this neutral motion segment posture. Figure 4.7 illustrates the jig used to pot the superior vertebra of the motion segments.



**Figure 4.7:** Jig used to pot superior vertebra of motion segments.

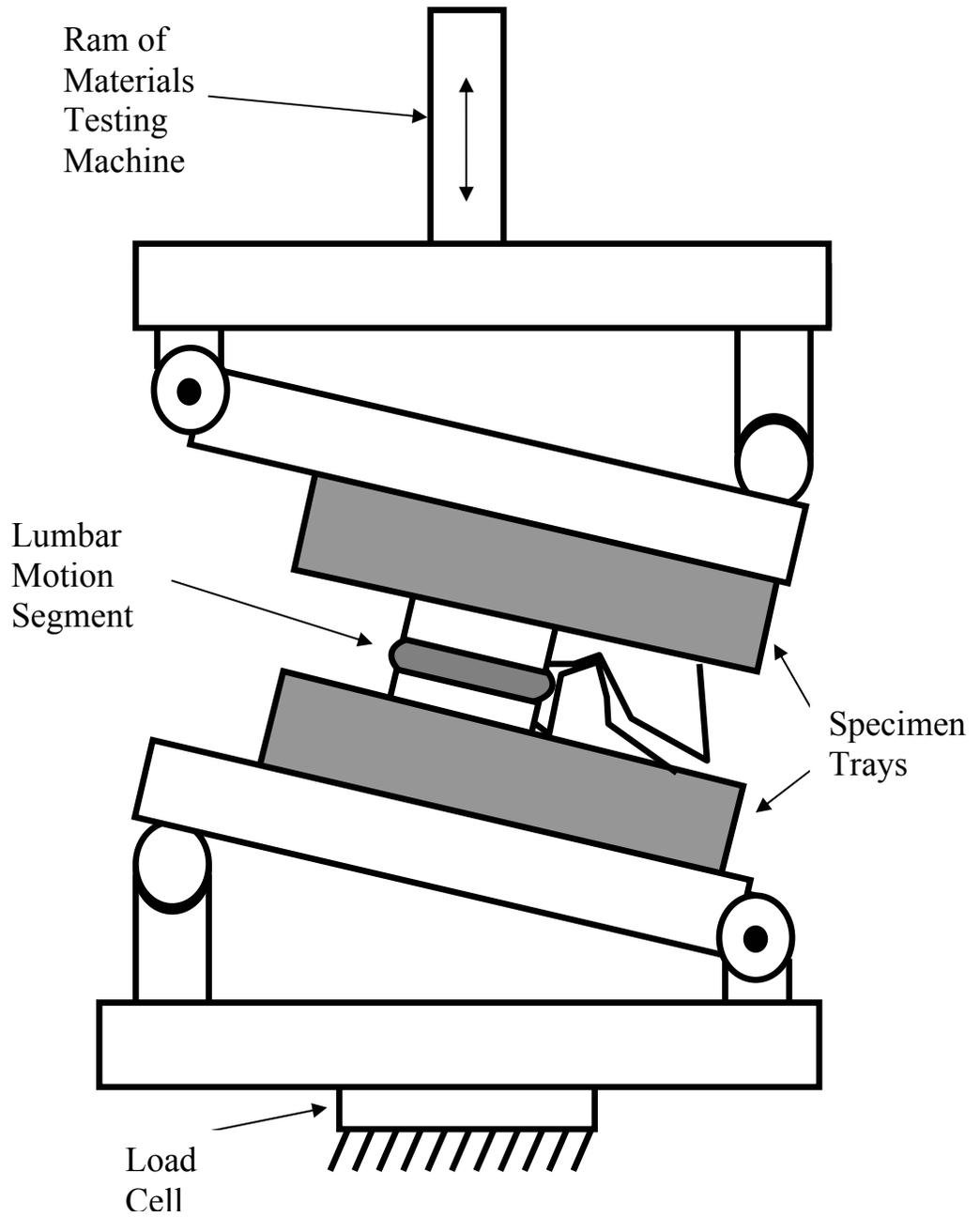
Once both vertebrae of the specimen were potted, the specimen was placed in a test jig designed to mimic the loading experienced during lifting tasks, as determined by a dynamic biomechanical model (Granata and Marras 1995) as detailed in Chapter 3 (see also Table 4.1). A loading system device consisting of angled metal plates (as seen in the schematic in Figure 4.8 below) were used in concert with a servohydraulic test frame (MTS Bionix 858, Eden Prairie, MN) to impose a combination of compression and shear forces on the motion segment for specified levels of flexion. The bottom tray was bolted or clamped to a plate set at the desired resultant loading angle. This ensured that the superior endplate of the inferior motion segment was subjected to the desired resultant force. The angled plate on top was threaded and locked to the ram of the materials-testing machine and the upper tray was bolted or

clamped to this plate. The angle at which the upper plate was set with respect to the lower plate determined the degree of flexion for the segment with respect to the neutral posture. A preload of 500, 750 or 1050 N (dependent on the torso flexion angle) was placed on the specimen for a minimum of 15 minutes to bring the disc hydration level into an appropriate physiologic range for each posture. The creep height loss of the specimen during this period were measured and recorded. The specified cyclic load for each posture was then imposed on the specimen.

Cyclic loads were placed on the specimens at a rate of 0.33 Hz (i.e., one cycle every three seconds) to simulate a repetitive lifting task. Figure 4.9 illustrates the loading cycles for each torso flexion posture. Load deformation curves were collected at 100 Hz using an A/D data acquisition board and were monitored according to the following schedule: Cycles 1-20 were collected, and then cycles 1011-1020, cycles 2011-2020, and so on until the final cycles were completed (10010-10020).

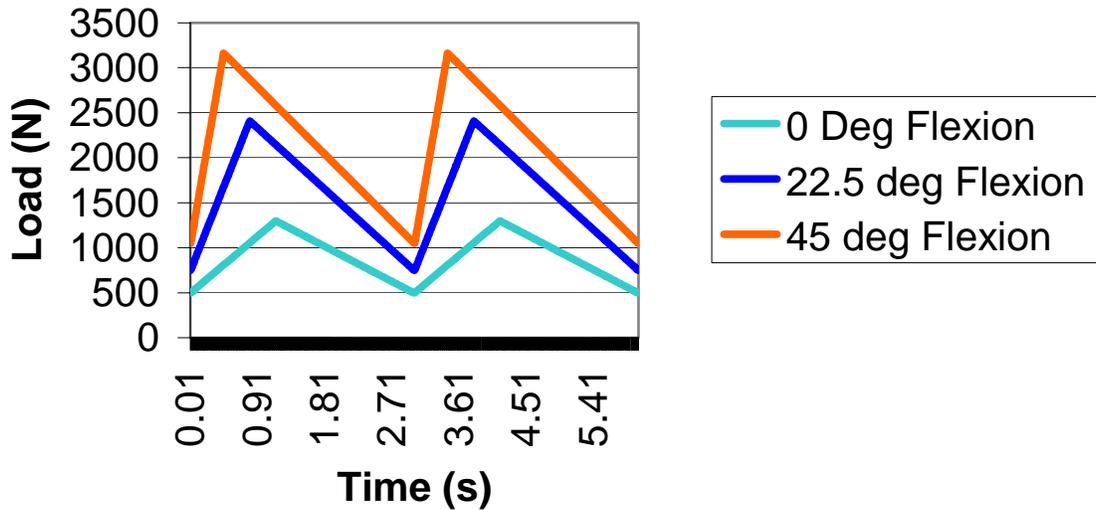
#### *4.2.6 Testing Environment*

An acrylic environmental chamber containing internal heating elements surrounded the testing fixture and specimens and these were set to maintain a temperature of approximately 37 degrees Centigrade to provide an environment similar to body temperature (see Figure 4.10). Testing was performed at body temperature due to the notable impact of ambient temperature on the cycles to failure of cortical bone. Cycles to failure at room temperature may be double that observed at body temperature (Carter and Hayes 1976).



**Figure 4.8:** Schematic of device used to impart specific shear and compressive loads.

## Cyclic Loading Profiles



**Figure 4.9:** Cyclic loading profiles to be used in tests at simulated angles of trunk flexion.

A container of water placed underneath one of the heating elements was introduced into the environmental chamber to maintain a moist testing environment (40-80% RH). These conditions allowed realistic disc creep and ligament extensibility during the materials testing procedure, and prevented the specimen from drying out during the testing.



**Figure 4.10:** Potted motion segment attached to test fixture within the acrylic environmental chamber.

Level	Trunk Posture	Motion Segment flexion from neutral (deg)	Angle of Resultant Force at peak resultant (deg)	Load Rate (N/s)	Min Load (N)	Max Load (N)	Total Cycle Time	Loading Phase (s)	Unloading Phase (s)
L1-L2	0 deg	0	25	700	500	1300	3 sec	1.14	1.86
	22.5 deg	1	26	2100	750	2400	3 sec	0.79	2.21
	45 deg	3	24	4800	1050	3150	3 sec	0.44	2.56
L3-L4	0 deg	0	12	700	500	1300	3 sec	1.14	1.86
	22.5 deg	3	20	2100	750	2400	3 sec	0.79	2.21
	45 deg	6	31	4800	1050	3150	3 sec	0.44	2.56
L5-S1	0 deg	0	-23	700	500	1300	3 sec	1.14	1.86
	22.5 deg	3	-6	2100	750	2400	3 sec	0.79	2.21
	45 deg	6	7	4800	1050	3150	3 sec	0.44	2.56

**Table 4.1:** Summary of loading conditions for motion segments in the nine experimental conditions employed in the experiment. Notes: Motion Segment flexion from Jorgenson (2001). For resultant angle, positive values indicate posterior shear (top vertebrae shearing posteriorly with respect to the inferior vertebrae) – negative values indicate anterior shear. Data from Nachreiner (1992) was used to determine minimum loads experienced in the various postures.

#### 4.2.7 Experimental Design

A randomized block partially confounded factorial design was utilized in this experiment to evaluate the effects of posture and lumbar level on fatigue strength (number of cycles to failure) of lumbar motion segments (Kirk 1995). Each complete lumbar spine was randomly assigned to a block according to the order provided in Table 4.2.

Three motion segments were dissected from each spine (i.e., L1-L2, L3-L4, and L5-S1) and these were tested at three levels of simulated trunk flexion (Table 4.3), using load rates and resultant forces characteristic of those experienced during *in vivo* lifting tasks in these postures (determined using the OSU EMG-assisted dynamic biomechanical model).

Rep 1: 1 <sup>st</sup> spine received assigned to treatment combination $(AB)_2$ [block 4] 2 <sup>nd</sup> spine received assigned to treatment combination $(AB^2)_0$ [block 6] 3 <sup>rd</sup> spine received assigned to treatment combination $(AB)_0$ [block 0] 4 <sup>th</sup> spine received assigned to treatment combination $(AB^2)_2$ [block 10] 5 <sup>th</sup> spine received assigned to treatment combination $(AB^2)_1$ [block 8] 6 <sup>th</sup> spine received assigned to treatment combination $(AB)_1$ [block 2]
Rep 2: 7 <sup>th</sup> spine received assigned to treatment combination $(AB)_2$ [block 5] 8 <sup>th</sup> spine received assigned to treatment combination $(AB)_1$ [block 3] 9 <sup>th</sup> spine received assigned to treatment combination $(AB^2)_1$ [block 9] 10 <sup>th</sup> spine received assigned to treatment combination $(AB^2)_2$ [block 11] 11 <sup>th</sup> spine received assigned to treatment combination $(AB^2)_0$ [block 7] 12 <sup>th</sup> spine received assigned to treatment combination $(AB)_0$ [block 1]

**Table 4.2:** Random assignment of spines to blocks.  $(AB)_x$  = treatment combination in Group 1,  $(AB^2)_x$  = treatment combination in Group 2 (refer to Table 4.3 for specific treatment combinations)

Blocking on spines was done to reduce the large amount of biological variability typically observed when testing cadaver materials. In particular, the influences of age and gender are known to have a large influence on the strength of the lumbar spine motion segments (Jager and Luttmann 1991). The current design allowed control of this variability, thus more precise estimates of the effects of lumbar

	Treatment Comb		ab combination	ab combination	ab combination
Group 1	$(AB)_0$	Block 0	L1-L2 at 0 deg	L3-L4 at 45 deg	L5-S1 at 22.5 deg
		Block 1			
	$(AB)_1$	Block 2	L1-L2 at 22.5 deg	L3-L4 at 0 deg	L5-S1 at 45 deg
		Block 3			
	$(AB)_2$	Block 4	L1-L2 at 45 deg	L3-L4 at 22.5 deg	L5-S1 at 0 deg
		Block 5			
Group 2	$(AB^2)_0$	Block 6	L1-L2 at 0 deg	L3-L4 at 22.5 deg	L5-S1 at 45 deg
		Block 7			
	$(AB^2)_1$	Block 8	L1-L2 at 45 deg	L3-L4 at 0 deg	L5-S1 at 22.5 deg
		Block 9			
	$(AB^2)_2$	Block 10	L1-L2 at 22.5 deg	L3-L4 at 45 deg	L5-S1 at 0 deg
		Block 11			

**Table 4.3:** Experimental Design.

level and torso flexion were possible. The design confounded a portion of the level\*posture interaction; however, intra-block information was available to recover a portion of this interaction. As a result, the interaction of lumbar level and posture remained estimable. Two replications of the design were conducted, using 12 complete spines (36 motion segments). The ANOVA table and expectation of mean squares can be found in Table 4.4. The balance of spines were used to perform pilot tests to ensure experimental procedures were adequate and appropriate, and for use as spares if a spine was excluded due to an error in testing.

Source of Variation	Degrees of freedom (in general)	Degrees of Freedom	F test	Expected Mean Squares (A and B fixed, Blocks random)
Between blocks	$nvw-1$	11		
Groups	$w-1$	1		
AB <sup>2</sup> (between)	$w(u-1)$	2		
Blocks w. AB <sup>2</sup> (bet)	$uw(n-1)$	8		
Within blocks	$nvw(v-1)$	24		
A (lumbar level)	$p-1$	2	[6/9]	$\sigma_e^2 + nqw\sum\alpha_i^2/(p-1)$
B (torso flexion angle)	$q-1$	2	[7/9]	$\sigma_e^2 + npw\sum\beta_j^2/(q-1)$
AB <sup>2</sup> (within)	$(p-1)(q-1)$	4	[8/9]	$\sigma_e^2 + (1/2)nw\sum\sum(\alpha\beta)_{jk}^2 / (p-1)(q-1)$
Residual		16		$\sigma_e^2$
Total	$nuvw-1$	35		

**Table 4.4:** ANOVA table for 2 replications of the design with expectation of mean squares. [Note: n = blocks within groups (3), v = # of ab combinations within each block (3), w = groups (2), u= # blocks within each (AB) combination (2), p= # of levels for treatment A (lumbar level), q= # of levels for treatment B (torso flexion angle)]

#### 4.2.8 *Calculation of Statistical Power*

A detailed analysis of the data provided by Brinckmann et al. (1988) was used to obtain estimates of effect size for the independent variables in this study. These data indicate that the number of cycles to fatigue fracture of the lumbar vertebrae was best predicted by the magnitude of the load applied to motion segments as compared to the maximum predicted ultimate load of the segment (or, the percentage of predicted ultimate strength of the motion segment applied to the segment during cyclic loading). In other words, specimens that were loaded to 80% of the predicted motion segment strength fractured in fewer cycles than segments loaded at 30% of the predicted segment strength.

Regression equations derived from Brinckmann's data after controlling for block variability (variability between spines) indicate that the following relation describes the relationship between cycles to fracture and percentage of ultimate motion segment load applied (percentage given in whole numbers – such as 30%, not 0.30):

$$\text{Cycles to Fracture} = 8377 \text{ cycles} - 123 \text{ cycles} * \text{Percent of Ultimate Load Applied}$$

$$\text{Adjusted } R^2 = 0.63$$

The Residual Mean Square for this regression was 1879863 and the Standard Deviation was 1371. A separate regression on spine variability gave a Residual Mean Square of 3404025 and a Standard Deviation of 1845. This provides an estimate of block variability for the power analysis.

Using the regression equation above, estimates of effect size for posture were calculated in the following manner: The maximum loads (kN) to be applied in each posture (Table 4.1) were compared to an average value for ultimate strength for lumbar vertebrae (5 kN) obtained from a meta-analysis of studies involving destructive tests of lumbar spine motion segments (Jager and Luttmann 1991). In order to compare directly with Brinckmann et al.'s (1988) data, the maximum resultant loads to be applied in this study were multiplied by the cosine of the angle of the resultant force so that the compressive component applied in the current study was comparable to Brinckmann's compression data. This provided percentages of ultimate compressive load for an average specimen in each posture. For the neutral posture condition (simulating 0 deg trunk flexion), the load applied was 24% of the average ultimate strength, for 22.5 degrees trunk flexion the value was 46%, and for 45 degrees trunk flexion, the value was 59%. When these values are plugged into the regression equation from Brinckmann's data, it was estimated that it would take 5425 cycles to fracture an average vertebra in 0 degrees trunk flexion, 2719 cycles at 22.5 degrees trunk flexion, and 1120 cycles at 45 degrees trunk flexion. From these values the average effect size for a 22.5-degree change in trunk flexion was estimated at 2153 loading cycles.

Using this value as an estimate of effect size, and the estimates of block and residual variability shown earlier, power was calculated for 2 replications of the ANOVA design using a power analysis tool developed by Dr. Russell Lenth, a Statistician at the University of Iowa (available at <http://www.stat.uiowa.edu/~rlenth/>)

[Power/index.html](#)). Power was calculated to be 0.8643, indicating that 2 reps of the experimental design would be sufficient to detect differences in cycles to fracture based on changes in posture. It should be noted that many of the data points used in this power analysis consisted of censored data observations (specimens that lasted the entire 5000 cycles of the Brinckmann et al. [1988] experiment). Therefore, the estimates of effect size would be conservative. As a result, the calculated power would also be expected to be a conservative estimate.

#### *4.2.9 Criterion for motion segment failure*

Loads and compressive/shear deformation of motion segments were digitally recorded using analysis software provided by the manufacturer of the materials testing apparatus. Fatigue-induced failure of the motion segments was deemed to have occurred when the specimen deformed greater than 10 mm from its height at the conclusion of the 15 minute pre-test creep loading period. Shear and compressive deformation at failure defined as millimeters of displacement of the upper cup with respect to the lower cup in the sagittal plane. Motion segments lasting the entire 10020 cycles in the fatigue failure test (i.e., censored observations) were subjected to a test of ultimate strength at the same load rate used in the cyclic loading regimen.

#### *4.2.10 Post test analyses*

Upon completion of the mechanical testing, specimens were removed from the loading jigs and taken to the x-ray clinic in Davis Hall to evaluate damage using a

lateral radiograph. These radiographs were read to evaluate the post-test status of the endplates, vertebral bodies, trabeculae, intervertebral discs, and posterior elements.

Visual inspection of failed specimens was performed to evaluate external evidence of failure, particularly facet disruption and/or evidence of fracture to the vertebral bodies. Photographs were taken to document such damage. The fixative-specimen interface was scrutinized to ensure that no failure occurred at this interface. Failed specimens were dissected through the midplane of the intervertebral disc, digitally photographed and a Galante disc degeneration score assigned to each of the discs (Galante 1967).

After the midplane of the discs was examined, the remainder of the disc was removed to expose the inferior endplate of the superior vertebra and the superior endplate of the inferior vertebra. The superior endplate of the inferior vertebrae was examined first followed by the inferior endplate of the superior vertebrae. Endplates were also digitally photographed. Damage to the endplates was classified according to criteria presented in the Results section. Logistic regression analysis was performed to evaluate variables associated with specific patterns of failure for specimens failing via fatigue.

#### *4.2.11 Specimen disposition*

Upon completion of post-test analyses, specimen remains were placed in labeled plastic bags and returned to the freezer in the Specimen Preparation Lab. Specimens were returned to the Wright State Anatomical Gift Program for cremation.

## **4.3 Results**

### **4.3.1 Results of Creep Loading**

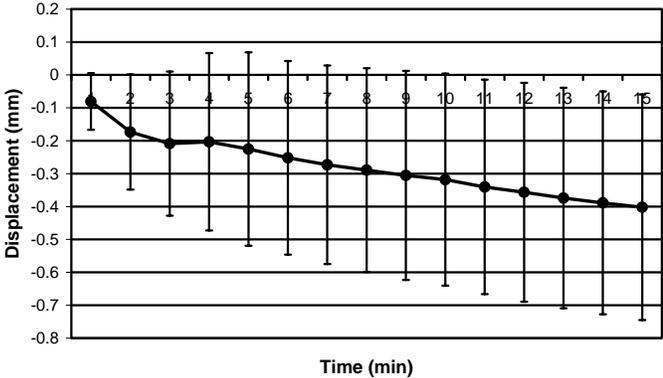
Figure 4.11 illustrates the creep response exhibited by motion segments when preloaded under 0 degrees, 22.5 degree and 45 degree torso flexion conditions (500 N, 750 N, 1050 N preload conditions, respectively). As can be seen from these figures, the creep response was influenced by the magnitude of the preload experienced. One specimen (an L5-S1 motion segment from a 65 year old male) failed on the 1050 N preload condition. This specimen was treated as lasting 0 cycles in the statistical analysis of the fatigue failure experiment.

### **4.3.2 Effects of Posture on Fatigue Failure**

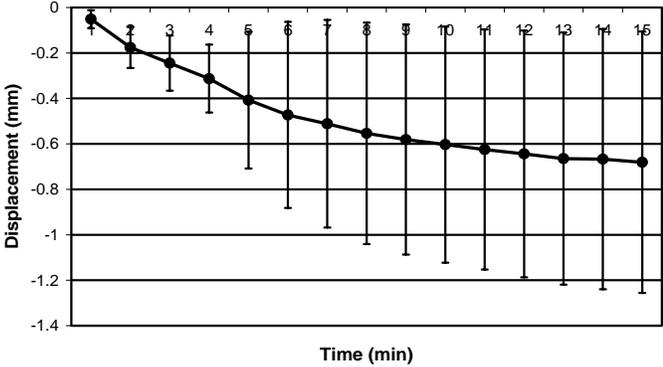
#### *4.3.2.1 Deformation of motion segments over time*

Figure 4.12 provides examples of typical deformation versus time curves for motion segments experiencing cyclic loading simulating lifts in 0, 22.5, and 45 degrees torso flexion. As illustrated in this figure, specimens tended to exhibit a continuous creep response throughout the loading regimen. Extensibility of specimens tended to be quite similar throughout the period of testing. However, as illustrated in Figure 4.13, specimens failed much more quickly in the 45 degrees of torso flexion conditions, and lasted much longer when loaded at 0 degrees torso flexion. The 22.5 degrees torso flexion condition proved highly variable in terms of the response to cyclic loading, with some specimens failing fairly rapidly, and others lasting the entire testing cycle.

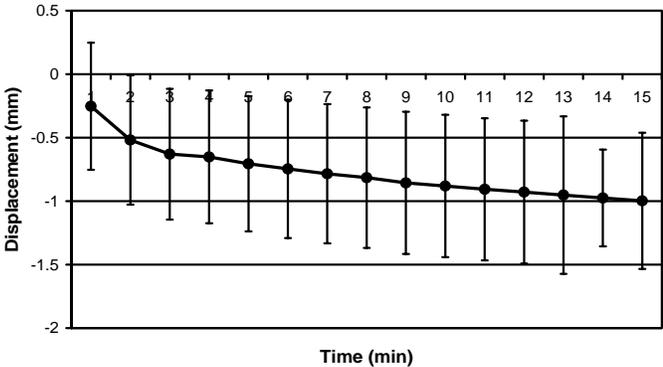
**a) 500 N preload (0 degrees torso flexion)**



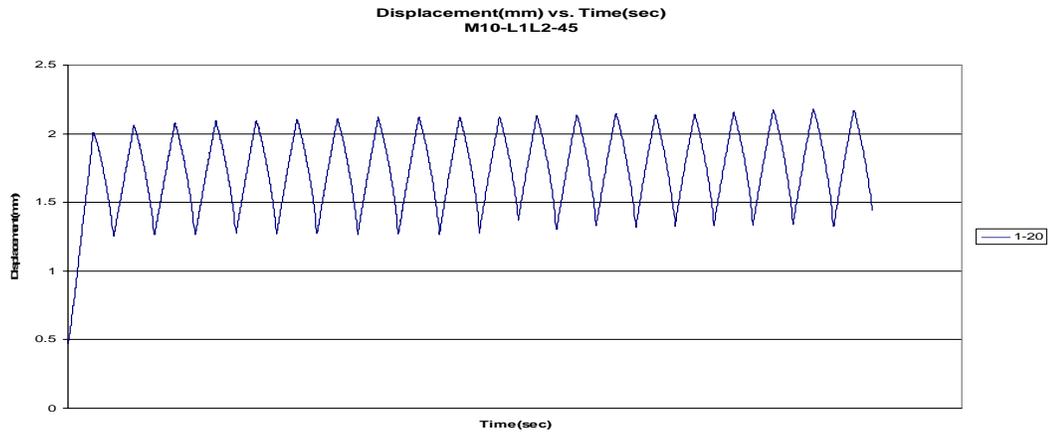
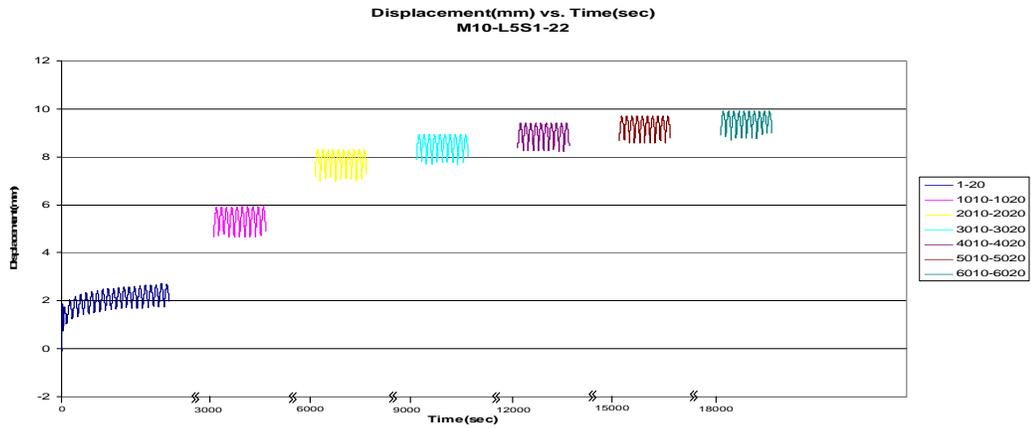
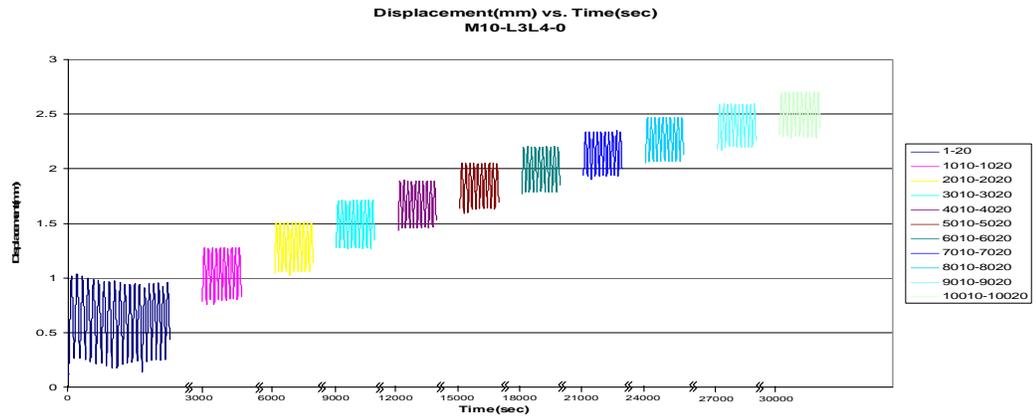
**b) 750 N preload (22.5 degrees torso flexion)**



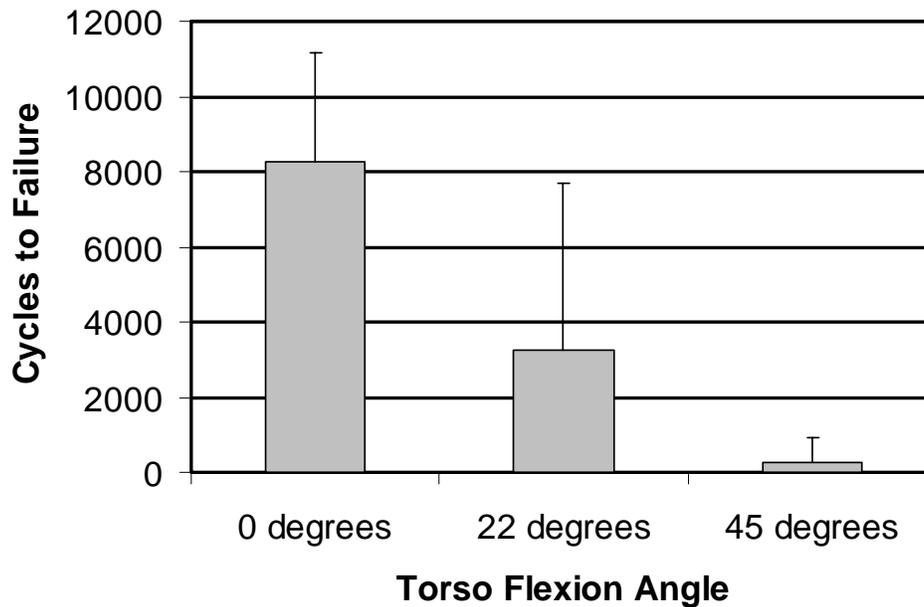
**c) 1050 N preload (45 degrees torso flexion)**



**Figure 4.11:** Average ( $\pm 1$  S.D.) displacement during 15 minute creep loading over all specimens for (a) 0 degree, (b) 22.5 degree, and (c) 45 degree conditions.



**Figure 4.12:** Typical deformation by time curves for: a) 0 degree, b) 22.5 degree, and c) 45 degrees of simulated torso flexion



**Figure 4.13:** Average cycles to failure in different torso flexion angles.

Spinal loading changes attendant to torso flexion had a highly significant influence on the fatigue failure of lumbosacral motion segments ( $p < 0.00001$ ). As can be seen from Figure 4.13, the average number of cycles to failure for segments in 45 degrees of flexion was 263 cycles, while segments in 22.5 and 0 degree torso flexion conditions averaged 3257 and 8253 cycles to failure, respectively. The ratio of maximum to minimum variance between torso flexion conditions exceeded 3.0 (Box 1954), thus a log transformation of cycles to failure was performed and the subsequent ANOVA was found to satisfy the homogeneity of variance assumption.

The ANOVA table for the transformed data is provided in Table 4.5. The F statistic for torso flexion angle in the transformed analysis was ( $F_{2,16} = 36.10, p < 0.00001$ ), indicating the very significant role torso flexion plays in fatigue failure on the lumbar spine in simulated lifting activities. In fact, estimates of the effect size for this factor indicates that it accounted for over 50% of the variance in cycles to failure ( $\theta^2 = 33.714/67.059 = 0.503$ ). Tukey Studentized Range statistics were used to perform pairwise comparisons of the means and it was determined that all three torso flexion angle means were significantly different from one another (Table 4.6).

Source of Variation	SS	df	MS	F	p
Between Blocks	18.08	11			
Groups (G)	0.51	1			
AB' (between)	12.87	2			
Blocks w. AB' (between)	4.69	8			
Within Blocks	48.98	24			
Lumbar Level (A)	2.37	2	1.18	2.54	0.11
Torso Flexion (B)	33.71	2	16.86	36.10	0.00001
AB' (within)	5.43	4	1.36	2.91	0.059
Residual	7.47	16	0.47		
AG	1.28	2			
BG	1.88	2			
AB x BLOCKS	4.31	12			
TOTAL	67.06	35			

**Table 4.5:** Analysis of variance summary table for log transformed data.

	0 degrees	22.5 degrees	45 degrees
0 degrees	-	1.335*	2.364*
22.5 degrees		-	1.029*
45 degrees			-

**Table 4.6:** Results of Tukey Studentized Range tests for pairwise differences between means of various torso flexion angles for the log of cycles to failure. The critical value was 0.722.

Figure 4.14 provides a summary of cycles to failure by lumbar level for segments used in the experiment. While cycles to failure were somewhat less for the L1-L2 segment in comparison to other segments, the difference was not found to be significantly different in the analysis ( $F_{2,16} = 2.54, p = 0.11$ ). It should be noted that loads applied to all segments were equivalent to the resultant vector calculated at L5-S1. Therefore, loads imposed on the upper levels of the spine are probably somewhat in excess of the loads actually experienced by the upper lumbar levels, and this may well be responsible for the trend observed. However, even with this bias, fatigue failure responses did not differ significantly by lumbar level. Table 4.7 provides a summary of experimental conditions and fatigue failure data for the specimens used in this study.

The experimental design also allowed recovery of an estimate of the within blocks portion of the interaction between torso flexion and lumbar level. Neither analysis (transformed nor non-transformed) indicated a significant interaction between lumbar level and torso flexion with respect to fatigue failure of the motion segments ( $p$

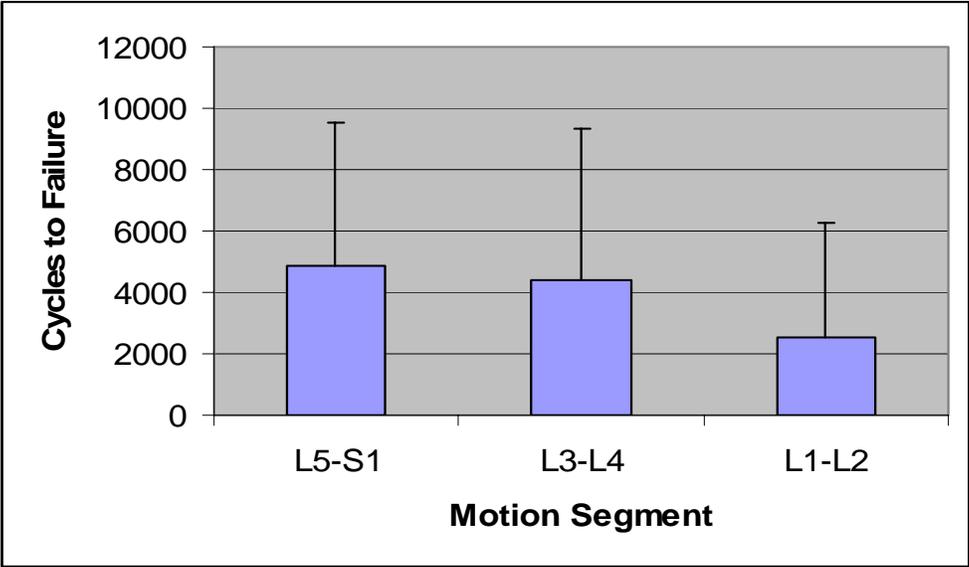
> 0.05). However, the transformed analysis resulted in an interaction probability of 0.059 and it was considered worthwhile to plot the interactions to observe whether the trend towards an interaction disclosed a meaningful or interpretable interaction pattern. Figure 4.15 illustrates the torso flexion\*lumbar level interaction of the log transformed data. Inspection of bone mineral content data (presented in Chapter 5) suggested that the interaction trend may have been driven by the random assignment of some L1-L2 motion segments with low bone mineral content (i.e. weaker specimens) to the 22.5 degree flexion condition, including the two weakest specimens. Since only three specimens were assigned to each block in this design, the weaker specimens assigned to this cell may be responsible for the seemingly anomalous response of L1-L2 specimens illustrated in Figure 4.15.

### **4.3.3 Ultimate Compressive Strength of Specimens with Censored Fatigue Failure Data**

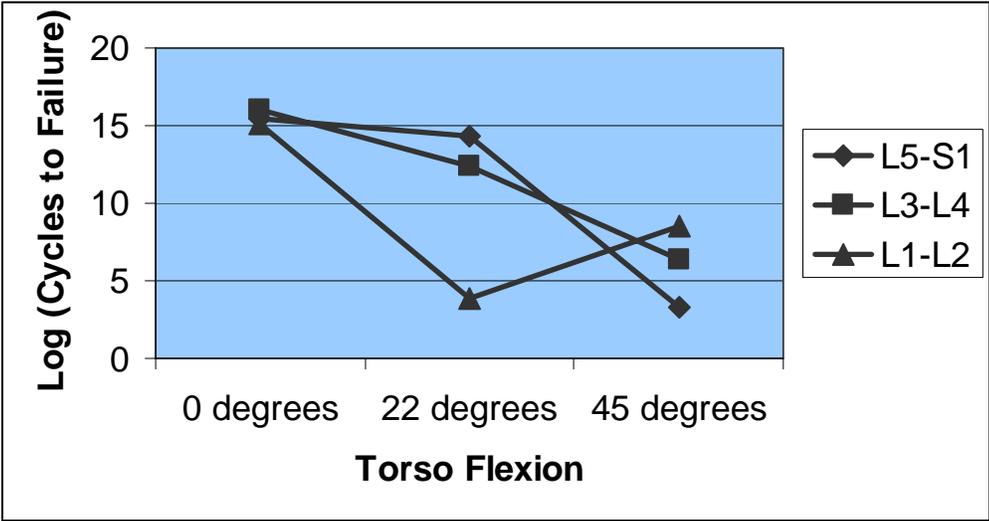
Figure 4.16 illustrates an example of a stress-strain curve from tests of ultimate compressive strength from specimens that lasted the entire testing period. Data from the entire set of tests of ultimate compressive strength of specimens lasting the entire 10020 cycles can be seen in Figure 4.17. Eleven of the 36 motion segments (31 % of specimens) lasted the full test length. Eight of the eleven surviving specimens had been subjected to the 0 degree trunk flexion condition, and the other three survivors experienced the 22.5 degree condition. The average ultimate compressive strength of surviving specimens was 5647.6 Newtons ( $\pm$  1845.7 SD). It should be noted that specimens were tested for ultimate strength at the same load rate experienced in the

Spine	Age	Segment	Torso Flexion (deg)	Creep displacement (mm)	Cycles to Failure	Ultimate Compressive Load (N)
M11	85	L1-L2	22.5	0.39	50	
		L3-L4	45	0.52	92	
		L5-S1	0	0.72	10020	7744
M8	74	L1-L2	45	0.91	2279	
		L3-L4	0	0.11	10020	5800
		L5-S1	22.5	0.32	10020	9600
M1	77	L1-L2	0	0.40	4425	
		L3-L4	22.5	0.54	282	
		L5-S1	45	2.14	90	
F9	85	L1-L2	45	0.45	68	
		L3-L4	22.5	0.76	10020	6200
		L5-S1	0	1.13	8400	
M7	65	L1-L2	22.5	1.09	2	
		L3-L4	0	0.88	10020	2900
		L5-S1	45	*	0	
F6	82	L1-L2	0	0.50	10020	4100
		L3-L4	45	0.70	49	
		L5-S1	22.5	0.53	10020	5200
F17	93	L1-L2	22.5	2.35	5	
		L3-L4	45	2.27	19	
		L5-S1	0	0.43	3406	
M10	80	L1-L2	45	0.46	450	
		L3-L4	0	0.10	10020	4300
		L5-S1	22.5	0.21	6494	
F13	91	L1-L2	0	0.67	2809	
		L3-L4	22.5	0.78	640	
		L5-S1	45	1.48	3	
M16	79	L1-L2	45	1.18	5	
		L3-L4	22.5	0.40	1295	
		L5-S1	0	1.48	10020	6500
M12	84	L1-L2	22.5	0.55	15	
		L3-L4	0	0.46	10020	4900
		L5-S1	45	0.95	73	
M14	73	L1-L2	0	0.10	10020	4900
		L3-L4	45	1.52	26	
		L5-S1	22.5	0.75	300	

**Table 4.7:** Summary of failure modes for all motion segments. Specimens lasting entire 10020 cycles were subjected to test of ultimate compressive strength (\* represents specimen failing on preload).

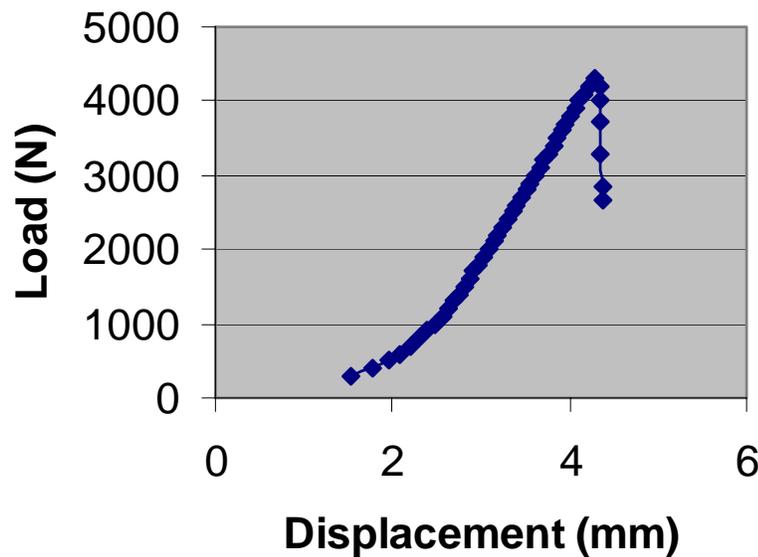


**Figure 4.14:** Average cycles to failure for motion segments at different spinal levels. Error bars represent the standard deviation. Differences between levels were not significant ( $p > 0.05$ ).

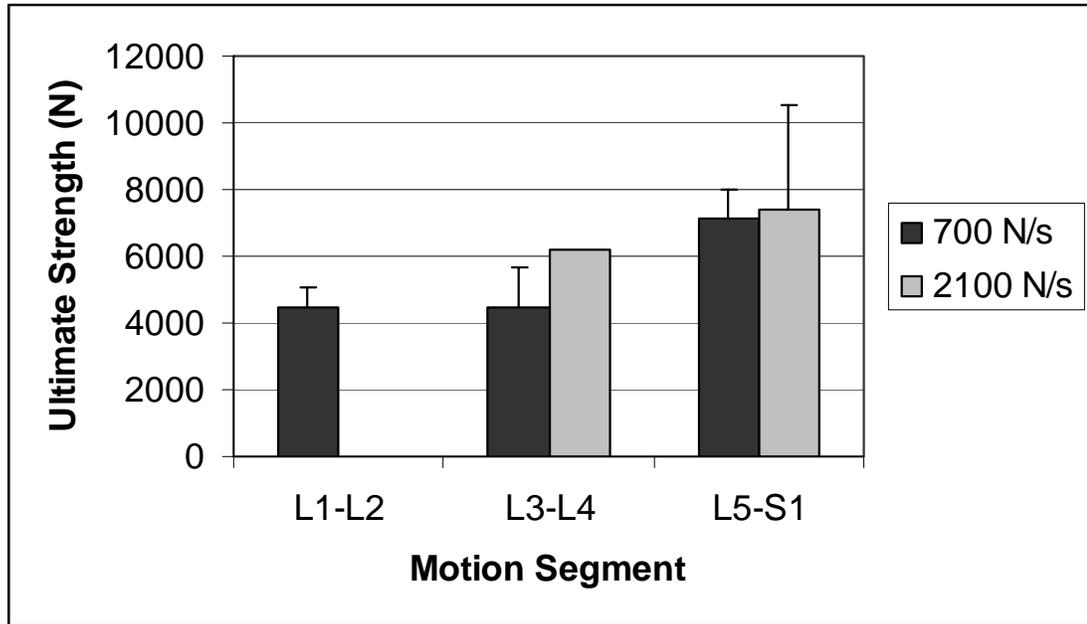


**Figure 4.15:** Interaction trend for Torso Flexion\* Lumbar Level.

experimental portion (700 N/s for the 0 degree condition and 2100 N/s in the 22.5 degree condition). However, regression analyses incorporating independent variables such as load rate, lumbar level, and bone mineral content or density failed to uncover significant associations with ultimate compressive strength ( $p > 0.05$ ). The limited number of specimens tested in this portion of the experiment may have limited the statistical power available to evaluate the influence of these independent variables on ultimate compressive strength of these motion segments.



**Figure 4.16:** Results of ultimate strength test for L3-L4 motion segment of specimen M10 (Load rate = 700 N/s).



**Figure 4.17:** Ultimate strength of motion segments lasting the entire 10020 cycles by spinal level and load rate [Note: number of samples (n) for L1-L2, 700 N/s: 2, L1-L2, 2100 N/s: 0, L3-L4, 700 N/s: 4, L3-L4, 2100: 1, L5-S1, 700 N/s: 2, L5-S1, 2100 N/s: 2].

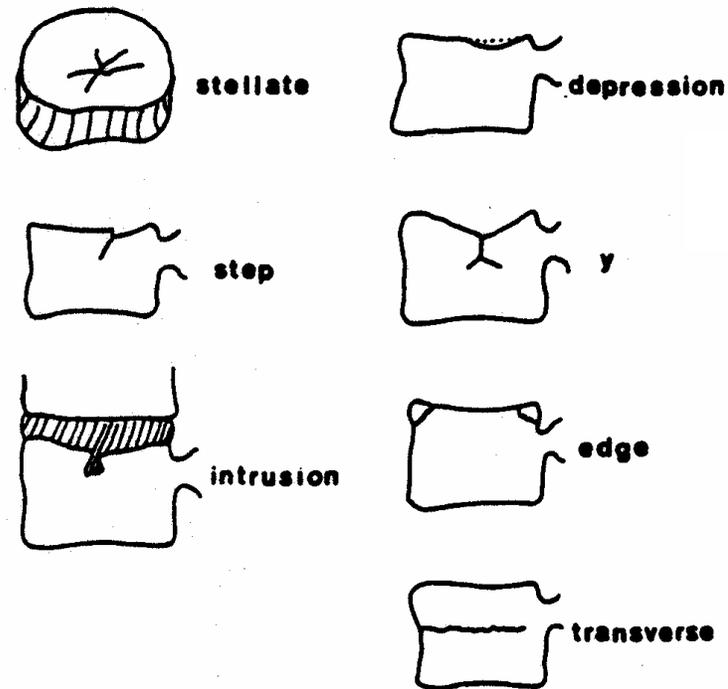
#### 4.3.4 Sites of Failure of Lumbar Motion Segments

Table 4.8 contains a summary of post test analyses of the mechanisms and sites of failure for the lumbosacral motion segments used in this study. As can be seen, failure of the motion segments generally involved endplate fractures, compression/shear damage to the vertebral bodies, and/or disruption of the facet joints. In only two instances was damage to the disc (via annular protrusion) noted. The following sections described typical damage scenarios to these structures.

#### **4.3.4.1 Endplate fractures**

Endplate fractures were frequently observed as a means of failure of the motion segment, often in combination with (and sometimes as part of) other vertebral body fractures. Endplate fractures were more commonly seen in the inferior endplate of the superior vertebral body, but were also a common finding in the superior endplate of the inferior vertebral body. Brinckmann et al (1988) provided a method for classification of endplate fractures (reproduced in Figure 4.18); however, the current author found this classification scheme did not adequately describe the variety of endplate fractures observed in the current study. For example, a common finding in the current data set was an endplate fissure running laterally across the endplate, which was not well-described as either a stellate or a step fracture. Another recurrent finding was an endplate fracture which followed the inside border of the ring apophysis, also not well depicted by the Brinckmann et al. (1988) classification. Other types of damage observed in this study, but not well described in Brinckmann et al.'s (1988) classification scheme, included: (1) a straight endplate fissure running in the A-P direction and (2) endplate "soft spots" discovered via palpation, a possible indication of localized trabecular damage beneath the surface of the endplate.

Since the Brinckmann et al. (1988) scheme did not fully characterize the nature of the endplate fractures observed in this investigation, a modified version of the classification system was developed. This new classification system, illustrated in Figure 4.19, includes several of the Brinckmann classifications, but adds some additional classifications of endplate damage observed in this study.

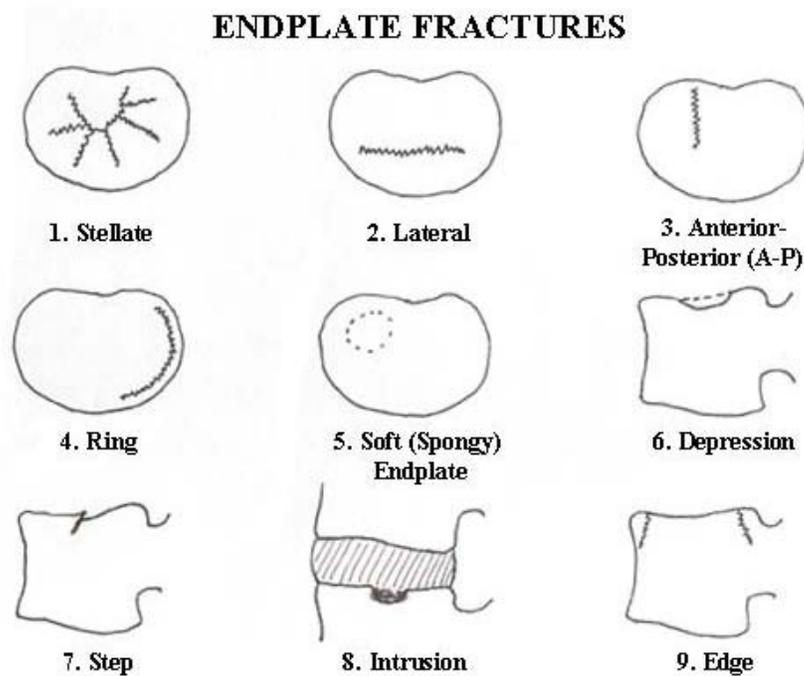


**Figure 4.18:** Classification scheme for endplate fractures by Brinckmann et al. (1988)

Two of the Brinckmann classifications were omitted in this new system. These included the “transverse” fracture, which was considered by the current author to be a vertebral body fracture and not a specific type of endplate fracture, and the “Y” fracture (observed only twice in Brinckmann’s original experiment, and not seen at all in the present study).

As seen in Table 4.8, endplate fractures were common findings in post-test examinations. By far the most frequent findings were endplate depressions (78% of

specimens) and stellate endplate fractures (47% of specimens). Lateral fractures (9 instances), ring fractures (8 instances), and endplates with soft, spongy areas (8 instances) were each present in approximately a quarter of all of the specimens tested. Step fractures (5 instances), intrusions (2), anterior-posterior fractures (2), and edge fractures (2) were seen more rarely. As can be seen in Table 4.8, combinations of different types of endplate damage were often observed. More significant damage was seen in motion segments subjected to tests of ultimate strength than in specimens failing as the result of cyclic loading.



**Figure 4.19:** Classification scheme for endplate damage used in the present study.

#### **4.3.4.2 Vertebral Body Fractures**

A frequent finding in post test examinations was a compression/shear failure of one or both of the vertebral bodies of the motion segment. Figure 4.20 shows an example of the type of damage observed to occur to the vertebral body resulting from compression/shear fractures. Such fractures were more common in superior (present in 53% of motion segment specimens) as compared to inferior vertebral bodies (31% of specimens). Transverse vertebral body fractures were observed in only 3 cases, always in the superior vertebral body of the motion segment. Localized trabecular derangement was observed in radiographs in two inferior vertebral bodies showing no overt external evidence of fracture.

#### **4.3.4.3 Damage to zygapophysial joints**

Examination of the failed specimens also turned up a number of cases where zygapophysial joints exhibited significant laxity or were disrupted entirely. Overall, disruption of the zygapophysial joints was more common in the neutral (0 degree) torso flexion position during fatigue failure tests (5 occurrences) than in flexed postures combined (4 occurrences). On the other hand, zygapophysial joint laxity was a more common finding at 22.5 degrees of torso flexion than at other flexion angles. In fatigue failure tests, facet damage was most common in L1-L2 and L5-S1 motion segments (4 instances each), and least common in the L3-L4 motion segment (1 instance).



**Figure 4.20:** Vertebral compression/shear damage to superior vertebral body. Note deformed shape of superior vertebra.

#### **4.3.4.4 Damage to the intervertebral disc**

Intervertebral discs tended not to incur a great deal of overt damage. However, examination of the sectioned disc on several occasions led to the observation that the internal lamellae of the disc appears to have experienced some distortion, which may have been the result of experimental conditions. Adams and Hutton (1985) reported observing such distortions of lamellar fibers in their study involving cyclic loading of lumbar motion segments. Only 2 discs appear to have experienced a protrusion similar to that occurring with disc prolapse. In both cases, the specimens were subjected to tests of ultimate strength, and an outpouching of annular disc material was observed to occur in the posteriolateral region of the disc. However, this damage did not occur in isolation, but was observed to occur in concert with the failure of other structures, notably damage to the endplates.

#### **4.3.4.5 Logistic Regression Analysis for Modes of Damage in Fatigue Failure**

Table 4.8 presents a summary of results of a series of stepwise logistic regression analysis variables associated with specific failure modes. This analysis used various endplate and vertebral fracture failure modes as dichotomous outcome variables and included Galante grade, motion segment volume, compression forces, shear forces, motion segment flexion, torso flexion, load rate and motion segment weight as co-variates. The analysis was limited to 25 specimens (those failing via fatigue), and the probabilities reported are conditional on the segments failing via these means. The 11 specimens representing censored observations were excluded due to the fact that ultimate strength tests were performed on the specimens and the gross damage inflicted in these tests did not seem comparable to those failing via fatigue. As can be seen, variables such as Galante degeneration grade, amount of shear force imposed, motion segment volume, and degree of motion segment flexion were significantly associated with the occurrence of specific injury patterns. For example, stellate endplate fractures were associated with lower Galante degeneration grades and were more likely in conditions with less imposed shear force. Facet damage was more common in less flexed torso positions, where the facets would assume load-bearing responsibilities.

Failure Mode	Significant variable associations	p-value	Nature of Association
Stellate EP Fracture	Galante grade AP Shear	0.006 0.019	↓ Galante: ↑ Stellate fracture ↑ Post. Shear: ↑ Stellate fracture
Lateral EP fracture	MS Volume	0.003	↑ Volume: ↑ lateral fract.
EP depression	MS Volume AP Shear	0.006 0.015	↓ Volume: ↑ EP depress'n ↓ Post. shear: ↑ EP depress'n
Inferior VB fracture	Galante grade	0.002	↓Galante: ↑ Inf. VB fracture
Facet Damage	Torso Flexion	0.009	↓ Flexion: ↑ Facet damage

**Table 4.8:** Summary of logistic regression results relating variables to specific failure modes (Note: MS= Motion Segment, AP = Anterior-Posterior, EP = Endplate, VB= Vertebral Body).

## 4.4 Discussion

### 4.4.1 *Effects of Torso Flexion on Fatigue Failure of Lumbar Motion Segments*

The spine loading conditions experienced at different angles of torso flexion when simulating loads experienced when lifting a 9 kg box have a dramatic impact on fatigue failure of lumbar motion segments. On the average, simulations of lifting such a box in the neutral posture indicate that the spine can withstand several thousand loading cycles, on average, before the effects of fatigue failure are manifest. However, loads experienced when lifting such a box in a flexed posture indicate a much more rapid onset of fatigue failure – only a couple of hundred cycles to failure, on average.

These results are congruent with evidence from various sources suggesting the dangers associated with torso flexion and lifting. For example, several recent epidemiologic studies have indicated that torso flexion is an important risk factor for low back pain (Riihimaki et al. 1989, Punnett et al. 1991, Holmstrom et al., 1992, Marras et al. 1993). Riihimaki et al.'s (1989) study, for example, found a dose-response relationship between exposure to bent and twisted postures and sciatica (OR 1.5, 95% CI 1.2-1.9). In Punnett et al.'s (1991) retrospective case-referent study, a strong trend was found for increasing length of exposure and risk of back disorders to both mild and severe trunk flexion. Holmstrom et al.'s (1992) study observed that time spent in torso flexed postures were associated with LBP, especially severe disorders (OR 2.6). In the study by Marras et al. (1993), a logistic regression model was developed using five factors which predicts the probability of membership in high versus low risk group membership.

One of the five factors in this model is the maximum sagittal trunk flexion measured during a work task. The greater the degree of maximum sagittal flexion, the greater the probability will be of high low back disorder group membership, with a univariate Odds Ratio of 1.60 (95% CI 1.31 -1.93). Furthermore, interventions designed to reduce trunk flexion (particularly early in the workday) have been shown to be effective in reducing the incidence of low back pain (Snook et al. 2002). Given the results of the current study, it seems reasonable to speculate that the rapid development of fatigue failure observed when lifting in flexed torso positions may have considerable etiologic significance in explaining these associations.

Spine ID	Motion Segment	Torso Angle	Failure Mode	Disc Galante Grade	Inferior Endplate (Sup. VB)	Superior Endplate (Inf. VB)	Superior Vertebral Body	Inferior Vertebral Body	Facet Disruption	Comments
M11	L1-L2	22.5	FF	3	1,6	4	Comp/Shear		L	L. Pedicle of L2 fractured
	L3-L4	45	FF	3		9	Comp/Shear	Comp/Shear		
	L5-S1	0	US		6	1,7		Comp/Shear		Neural arch fracture on L5
M8	L1-L2	45	FF	2	6	6	Comp/Shear			
	L3-L4	0	US	2	6	6	Comp/Shear	Comp/Shear	RL	L4 facet fractured
	L5-S1	22.5	US	2	5	9				
M1	L1-L2	0	FF	3	2		Comp/Shear		L	
	L3-L4	22.5	FF	2	2, 1				L(lax)	
	L5-S1	45	FF	2	2,6		Comp/Shear			
F9	L1-L2	45	FF	3	4,5	5	Transverse			
	L3-L4	22.5	US		7	2	Comp/Shear	Comp/Shear	RL	
	L5-S1	0	FF	3	6	6		Comp/Shear	RL (lax)	
M7	L1-L2	22.5	FF	3	6,5	6	Comp/Shear	LTD		Blood in post. central disc
	L3-L4	0	US	2	6,3	2,4	Transverse			Annular protrusion, L pedicle of L4 fractured
	L5-S1	45	FF	4	6			Comp/Shear	RL	Disc space narrowed
F6	L1-L2	0	US	2	3,4	1	Comp/Shear	Comp/Shear		
	L3-L4	45	FF	2	1			Comp/Shear	RL (lax)	
	L5-S1	22.5	US	3				Comp/Shear		Disc space narrowed
F17	L1-L2	22.5	FF	3	1	6				
	L3-L4	45	FF	2	6	1,6	Comp/Shear		L	
	L5-S1	0	FF	2		6			RL	
M10	L1-L2	45	FF	2	1,4	8	Comp/Shear			
	L3-L4	0	US	2	7	8,4	Transverse		R, L(lax)	Shear fracture of L1 separated EP from VB
	L5-S1	22.5	FF	3		6,4,5		Comp/Shear	RL(lax)	
F13	L1-L2	0	FF	3	6		Comp/Shear		RL	Disc lax in torsional movement
	L3-L4	22.5	FF	2	1,6	6	Comp/Shear			
	L5-S1	45	FF	2	1	1	Comp/Shear			
M16	L1-L2	45	FF	2	1		Comp/Shear			
	L3-L4	22.5	FF	1	1,5		Comp/Shear		RL(lax)	
	L5-S1	0	US	2	1,6	6,4		Comp/Shear	RL	Annular protrusion R posteriolateral aspect
M12	L1-L2	22.5	FF	3	2, 7		Comp/Shear			
	L3-L4	0	US	3	2,7	1,6	Comp/Shear		R,L(lax)	L3 spinous process pushed laterally
	L5-S1	45	FF	3	2,5			LTD		
M14	L1-L2	0	US	3	1,6	2	Comp/Shear	Comp/Shear		
	L3-L4	45	FF	2	1	6,5				
	L5-S1	22.5	FF	3	6,5	6,5				

**Table 4.9:** Summary of failure modes. FF=fatigue failure, US=ultimate strength, LTD=localized trabecular disruption. Refer to Figure 4.19 for endplate failure classifications.

It may be noted that the constant cycle time and varied loads and load rates in simulated torso flexion angles allowed different unloading intervals. For example, while the loads were greater and the rate of loading higher in the flexed torso condition, approximately 40% more time was available each cycle for unloading of the specimen in the flexed posture compared to neutral. However, in this *in vitro* test paradigm, any benefit that might have accrued from the increased recovery time appeared to be totally overwhelmed by the intensity of the loading. Though this is admittedly not a physiologic test condition, it is difficult to imagine any short term physiologic mechanism that would override the *in vitro* finding. This is particularly true given that bone healing repair processes tend to take weeks or months to accomplish (Brinckmann et al. 1988, Hawkins 2001). It is not known the extent to which much longer term recovery cycles may influence responses to cyclic loading of varying magnitudes.

#### 4.4.2 *Effects of Lumbar Level on Fatigue Failure of Lumbar Motion Segments*

In the clinical setting, it is quite clear that the lower lumbar levels experience greater insult and injury than at the upper lumbar levels (Spangfort 1972). However, slightly fewer cycles to failure were observed for the L1-L2 segment. This result is not surprising when one considers that the load imposed on upper motion segments in this experiment was of the same magnitude as the load calculated at L5-S1, and is thus somewhat greater than the motion segment would experience in real life. It is noteworthy; even so, that the trend noted above did not achieve statistical significance,

even though the load imposed was somewhat higher than would be experienced during the experimental task *in vivo*. Unfortunately, there is not a great deal of literature that has predicted the loads at upper lumbar levels, so that a proper adjustment of loads calculated at L5-S1 to the upper lumbar levels. It is generally assumed that the loads at the upper lumbar levels are similar, though somewhat lower, than those occurring at the lumbosacral junction (Daggfeldt 2002).

#### *4.4.3 Ultimate Strength of Motion Segments with Censored Data*

Though a limited number of motion segments (11) were tested for ultimate compressive strength, several comments can be made with respect to the results of these tests. One is that the average strength of the specimens is fairly high compared to average values reported in the literature for the age group represented. This is no doubt the result of the fact that specimens surviving the fatigue loading portion of the experiment represent the strongest specimens of the group tested, thus they represent a biased sample. Expected trends regarding strength and lumbar level (i.e., increasing strength at lower levels) and increased ultimate strength at higher load rates were observed in this sample. However, the limited number of samples and variety of load rates at which they were tested precluded a thorough statistical analysis of this subset of motion segments.

#### *4.4.4 Structures Incurring Failure*

Damage to the motion segments typically involved vertebral endplate fractures, compression/shear fractures of the vertebral bodies, and facet disruption. More often than not, multiple modes of failure were observed.

Endplate fractures or damage were observed in almost all motion segments, and often involved both endplates. This is somewhat in contrast with the fatigue study performed by Brinckmann et al. (1988), where damage was rarely observed in more than one endplate. The difference in findings may be attributable to several factors, one of which may be the increased displacement criterion for failure in the current study (10 mm vs. 5 mm in the Brinckmann study). However, it is quite possible that other factors may have contributed to the differences, including differences in load rate and shear forces present in the current study compared to the pure axial load applied by Brinckmann et al. (1988).

Brinckmann et al. (1988) also reported that in his study, endplate fractures were observed more commonly in the superior endplate of the inferior vertebra of the motion segment than in the inferior endplate of the superior vertebra, even when the specimen was tested upside down. However, this finding was not replicated in the present experiment, where both endplate and compression fractures were more frequent in the superior vertebra of the pair. The reason for this disparity is not entirely clear, but may have to do with differences in the loading regimes, especially the introduction of shear loads in the present study. It is also apparent that the morphology of endplate fractures was considerable different between these studies, as exemplified by the fact that additional fracture types had to be included in the classifications of endplate fractures in the current study. In particular, transverse fissures extending across the endplate were often seen in this experiment, types of fracture also observed by prior investigators (Perey 1957), as well as fractures

following the inner border of the ring apophysis. It is commonly stated that endplate fractures are the result of vertical compressive forces (White and Panjabi 1978); however, the amount of shear force was found to influence the occurrence of certain types of endplate damage, particularly stellate fractures. In addition, occurrence of certain types of endplate fractures appeared to be associated with the size (or volume) of the motion segment, particularly lateral fractures and endplate depression.

Compression and shear fractures of the vertebral body were also common, often in conjunction with fractures of the endplates. Clinically, such fractures occur most frequently in the upper lumbar or lower thoracic regions of the spine (White and Panjabi 1978). Indeed, in the current study the L1 vertebra was by far the most common site of vertebral body comminution in fatigue failure, with 8 occurrences of the 12 specimens tested. L3 and S1 vertebrae exhibited the second highest number of compression fractures with 4 each. In cases of fatigue failure, the upper vertebra appeared to experience an increased rate of failure (generally at or above 50% of specimens tested) compared to inferior vertebrae (approximately 25 % of specimens tested). This may be due to the decreased strength of upper level motion segments relative to the load imposed, and may also reflect differences in the support provided by the facet joints at different levels of the lumbar spine *in vivo* (Bogduk 1997). These fracture patterns are thought to be primarily the result of high compression forces and are often considered to occur subsequent to an initial endplate fracture (White and Panjabi 1978). The pattern of compression and shear deformation seen in Figure 4.20 may have been influenced by the rigid fixation of the upper vertebrae in

PMMA which would induce stresses on the upper vertebral body that would not be present in the physiologic state.

Post-test examinations uncovered a number of facet disruptions and cases where a great deal of laxity was present in the posterior elements. Analysis of this data suggest that facet disruptions occurred in a higher percentage (75%) of motion segments failing in fatigue loading when exposed to the 0 degree torso flexion than in the other torso flexion angles (11% in 22.5 degrees flexion and 16% in 45 degrees of flexion). A very similar pattern was seen in tests of ultimate strength using surviving motion segments. It should be noted that the neutral posture would be a situation where the posterior elements are thought to bear a significant portion of the load, and an increased threat of disruption of the zygapophysial joints has been suggested by some authors when posterior elements bear additional load (White and Panjabi 1978). Data from the present study suggests that the zygapophysial joints may be at risk of fatigue failure even at the relatively lower loads seen in the neutral posture compared to more flexed conditions.

#### *4.4.5 Implications*

Though one cannot draw specific conclusions from in vitro experiments regarding clinical manifestations of endplate fractures in vivo (Brinckmann et al. 1988), there is reason to suspect that results of this investigation of fatigue failure of lumbar motion segments may have implications regarding the development of low-back disorders. Recent reports, for example, have strongly linked disc degeneration to back pain (Luoma et al. 2000). Mechanical loads are thought to be one of several

pathways by which the disc degeneration cascade may be initiated or facilitated (Urban and Roberts 2003). Indeed, endplate failure has been associated with two biomechanical outcomes affecting the disc and which are related to degeneration. One is an irreversible decrease in the disc height, and the other is a permanent increase in the radial disc bulge (Brinckmann and Horst 1985). Mechanical loading in animal models has also been associated with effects on the disc itself, manifest through increased disorganization of annular cells, and an increased percentage of cells undergoing apoptosis, both associated with degeneration of the disc (Lotz et al. 1998). Recent studies have also confirmed a centripetal growth of nociceptors in degenerate lumbar discs, which may provide a morphological basis for discogenic pain (Coppes et al. 1997).

While disc degeneration is an area of intensive investigation regarding its possible role in the development of low back pain, it should be recognized that many other putative sources of pain are available in the spine. Nearly all structures in the spine receive innervation and have the potential to be sites of pain including zygapophysial joints, ligaments and fascia of the spine, the vertebral bodies, and the surrounding musculature (Bogduk 1997). Several motion segments in the current study exhibited damage to the vertebral bodies and zygapophysial joints as well as fractures of the cartilaginous endplates. Avulsion of the zygapophysial joints in the present study was more frequent in the upright posture, where these joints assume a greater load-bearing role (White and Panjabi 1978). In partial flexion, post-test exams findings uncovered more laxity in zygapophysial joints than overt disruptions, and

facet joint damage was rarely found in full flexion. Thus, zygapophysial joint damage appeared to have a dependence on the amount of flexion of the motion segment, with the neutral posture being the position where greatest damage was observed. Use of diagnostic blocks has determined that the zygapophysial joint is a common site of pain among patients with chronic low back pain (Schwarzer et al. 1994, Schwarzer et al. 1995). The prevalence rate among a cohort of injured workers was found to be 15% (95% CI: 10-20%). Prevalence rates in an elderly Australian patient population were approximately 40%.

Vertebral bodies are also innervated and the periosteum is clearly a pain-sensitive structure (Bogduk 1997). However, it is not clear whether fractures of the vertebral body are painful in and of themselves (Gershon-Cohen 1953), or whether the post-traumatic hematoma and edema resulting from such fractures may be a potential cause pain (Bogduk 1997). In the current experiment, fractures were more common in the superior vertebra of the motion segment pair and, of these, more fractures were observed at higher lumbar levels and with increasing lumbar flexion. The finding of increased vertebral fractures at higher levels is concordant with clinical findings, as is the expectation of increased fractures with higher load magnitudes (White and Panjabi 1978). However, it should be noted that the prevalence of pain resulting from fractures of vertebral bodies is not known, but such fractures are relatively rare among patients presenting with back pain under the age of 50 (Deyo and Diehl 1986). The fairly common finding of compression/shear vertebral fractures in the current study

may also be a consequence of the reduced vertebral bone mineral content associated with older specimens.

Overall, the findings of this study appear to have great relevance to injury pathways presumed to be related to the development of low back pain. In particular, the rapid development of fatigue failure due to spinal loads experienced when lifting with a flexed torso suggests an acceleration of the cascade resulting in degenerative disc disease. However, it should be recognized that avoidance of flexion during lifting may not be a panacea for back pain prevention. Specifically, the neutral posture was associated with a greater incidence of damage to zygapophysial joint, a site also associated with chronic low back pain cases. It may well be that no spinal posture is intrinsically safe in terms of preventing low back pain, and that trade-offs will occur in terms of the posture of the spine when lifting and damage to sites of potential pain. However, the rapid development of fatigue failure observed in torso flexion would seem to make this posture particularly important to avoid in the performance of lifting activities.

The findings of this study would also appear to have important implications for the current and future manual lifting guidelines designed to reduce low back disorders. Based on the data presented in this study, current lifting guidelines may not pay sufficient attention to the significant role that fatigue failure may play in the development of low back disorders, and particularly the rapid development of fatigue failure under conditions of full flexion of the lumbar spine. For example, the NIOSH lifting equation (Waters et al. 1991) provides a variable discounting factor for frequent

lifting based upon whether a load originates at above or below waist height (75 cm) in an attempt to address lifting involving trunk flexion. However, it should be recognized that trunk flexion may vary considerably with lifts below waist height. Some lifts from below waist height may involve little or no torso flexion, while others may require full flexion. According to the data collected in this study, the difference in the number of cycles to failure of the spine between the wide variety of torso flexion angles that may exist within the “below waist height” designation may be over 30-fold. Results of this study suggest strongly that torso flexion angles require increased attention in the development of future lifting recommendations. According to the data presented here, full flexion of the lumbar spine in lifting tasks needs to be avoided whenever possible. Partial flexion appears to be a considerable improvement better compared to full flexion, and the lumbosacral spine actually appears quite capable of tolerating repeated loading when lifting moderate weights when the neutral posture is employed.

#### *4.4.6 Limitations*

A complete assessment of the experimental results described above requires the consideration of certain limitations. The first of these deals with the age of the cadaveric cohort studied. Unfortunately, the vendor used to obtain the lumbar spines used in this study was unable to provide specimens under the age of 65. As a result, it becomes difficult to generalize the results of this study to the younger, working age population that is the intended beneficiary of the research. Younger spines would be expected to have increased bone mineral content and density than the older spines

tested in this study, and would be expected to exhibit increased resistance to fatigue failure (Hansson and Roos 1981). Given the effect magnitude demonstrated in this study, one would certainly not expect a reversal of the torso flexion effect. However, effect sizes and the number of cycles to failure may differ between younger and older specimens. Fortunately, efforts are currently underway to address this particular study limitation. Another vendor has agreed to supply lumbar spines aged 18-60 for an additional replication of the experimental design for this study. Data from this additional replication may allow comparison of the responses of younger and older spines to fatigue failure. Furthermore, plans have been made to obtain longer term fatigue data from the younger cohort. Specifically, no arbitrary maximum number (with the possible exception of a very high number) of loading cycles will be imposed with the younger set of spines. Fatigue loading will continue for this group until the failure criterion (10 mm displacement after termination of creep loading) has been achieved.

Another limitation, or set of limitations, deals with the very nature of *in vitro* testing. The lumbar spine is one part of a complex physiologic system and isolation of the spine (or its motion segments) necessarily alters the loading situation compared to the *in vivo* environment. Studies performed on animals have suggested a possible change in mechanical behavior of various tissues after death (Keller et al. 1990); however, the difference attributed to death in this experiment were not much larger than trial to trial variability on the live animal (Adams and Dolan 1995). The process of freezing and thawing of cadaver specimens appears not to have much of an effect

on the mechanical properties of bone (Sedlin and Hirsch 1966) or the intervertebral disc (Nachemson 1960; Smeathers and Joanes 1988). However, the process of dissecting out motion segments necessitates a weakening of ligaments of the spine that have fibers spanning multiple vertebrae (Adams 1995). There are also obvious differences between the manner in which spine segments are loaded with functioning muscles and tendon attachments as opposed to a specimen potted in polymethylmethacrylate and loaded in a frame. Nonetheless, the methods of *in vitro* testing employed here remain one of our best methods to improve our understanding of the mechanical responses of lumbar tissue.

A third limitation of the experiment relates to the estimation procedures used to predict the loads at the upper levels of the lumbar spine. Unfortunately, most biomechanical models provide estimates of loads only at the lumbosacral junction and do not provide estimates at higher levels. Thus, some simplifying assumptions had to be made in the current analysis, including the assumptions that the magnitude of the load vector was equivalent at the lumbosacral junction as at the upper lumbar levels, and treating the spine as a rigid object in the analysis of the compression and shear loads at different spinal levels. As a result of these assumptions, it would be anticipated that the upper lumbar levels experienced a load in this experiment somewhat in excess of that expected to occur when lifting a 9 kg load. It could be considered that the loads experienced at the upper levels were essentially at levels that would be experienced if subjects had lifted a slightly heavier weight.

Fourth, while radiographs were obtained and abnormal external findings were noted during dissection, it is possible that pre-existing damage, particularly to the endplates, may have been present prior to testing but not detected. It is often extremely difficult to detect endplate fractures via regular radiographs or other imaging techniques (White and Panjabi 1978). Thus, it is quite conceivable that some of the damage reported in the post-test evaluations was pre-existing, and not the result of the stresses imposed during testing. Certainly the tests could have also exacerbated pre-existing damage in the specimens, causing failure to occur more quickly than would have occurred to an intact specimen. A couple of the specimens tested were observed to have existing wedging of the vertebral bodies characteristic of comminution injuries, reported to reduce the load bearing capacity by up to 30-40 percent (White and Panjabi 1978). However, given the failure criterion of 10 mm displacement post creep loading, it is likely that a great deal of the damage to the specimens was the direct result of the test procedure.

A fifth limitation can be associated with the failure criterion. The 10 mm displacement criterion was beneficial in a number of ways, likely increasing the variability in cycles to failure of the specimens in the different experimental conditions, and ensuring that macroscopic damage had occurred by the end of the testing period. However, at the same time the amount of displacement used for the criterion often appeared to result in damage to multiple structures of the motion segment. As a result, in many cases it was not possible to isolate a particular structure as having been the initial site of failure. It may be advisable to adjust this criterion in

future studies, especially if identification of initial failure sites becomes a topic of increased interest.

A sixth limitation exists when considering the generalizability of these *in vitro* materials tests to occupational lifting activities. Several differences exist between the loads imposed on the lumbar motion segments and those experienced in occupational tasks. For example, the load experienced *in vivo* is not expected to be a triangular function, will not have a constant load rate, and would not be expected to reflect the precise periodicity of the loading imposed on the motion segments in this experiment. Furthermore, a great deal of individual variability would be expected with respect to the flexion of motion segments in different torso flexion postures. The need for experimental control required establishment of representative values for positioning and loading of motion segments. At this time, there is no way of knowing the degree to which such factors may affect the motion segment responses to loading *in vivo*, and no specific corrections for these factors can be made.

A final limitation associated with this study was touched on earlier in this chapter, but deserves to be recapitulated. Accurate simulation of the changes that occur with torso flexion requires simultaneous manipulation of a number of variables that may impact the resistance of a motion segment to fatigue failure. These include load magnitude, the blend of compression and shear forces experienced, load rate, and flexion angle of the specimen. Because these factors were manipulated simultaneously, it is not possible to ascribe the results to any of these specific factors, only to changes in the entire ensemble. It is hoped that future research can help to

specify the relative importance of these individual factors in the fatigue failure of lumbosacral motion segments.

#### *4.4.7 Future Research Directions*

The dramatic impact of torso flexion angle on the fatigue failure of lumbosacral motion segments demonstrated in this study certainly suggests that additional inquiry along these lines is certainly warranted. An immediate need is replication of this experiment on a younger cohort of spines to ensure that the findings observed in the older cohort are similar to the responses that will be seen in younger spines, and what differences might exist between the two groups. Plans have already been made to execute this research using six additional spines (18 motion segments) between the ages of 18-60 years, so that this need might be addressed.

Additional research will be necessary to develop a complete understanding of the impact of torso flexion on fatigue failure when lifting different loads, the effects of gender and age, and at different levels of torso flexion. Further research will also be required to evaluate the impact of the several factors that change with torso flexion, including the load rate, the load magnitude, the mixture of compressive and shear forces, and the orientation of the motion segments.

Findings of this study also suggest bases for interventions and may also have some important policy implications. For example, the present research suggests strongly that avoidance of flexion may be decisive in the prevention of fatigue failure of tissues of the lumbosacral spine when performing lifting tasks. Intervention studies where

torso flexion is minimized should be developed to ascertain whether low back disorders can be prevented by reducing the threat of rapid fatigue failure in flexed torso postures. Furthermore, this research suggests that lifting guidelines should include recommendations and limits based on the apparent threat of rapid fatigue failure when lifting in forward flexion. Doing so may result in lifting recommendations that might reduce the risk of low back disorders associated with both occupational and nonoccupational lifting tasks.

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## **CHAPTER 5**

### **THE INFLUENCE OF BONE MINERAL CONTENT ON FATIGUE FAILURE OF THE LUMBAR SPINE**

#### **5.1 Introduction**

A vertebral fracture must occur when the strength of a vertebra is exceeded by a transitory load or by cyclic, repetitive loads (Hansson et al. 1987). The strength of a vertebra, and therefore its ability to withstand the stresses imposed on it, is primarily determined by the amount of mineralized tissue of which the vertebra is comprised (Bartley et al. 1966, Hansson and Roos 1981, White and Panjabi 1978). Several studies have found that the relationship between the ultimate compressive strength of a vertebral body and bone mineral content is basically linear (Bartley et al. 1966, Bell et al. 1967, Galante et al. 1970, Hansson and Roos 1981); however, the tendency for a non-linear portion in weaker, demineralized bone has been suggested by some authors (Bell et al. 1967, Hansson and Roos 1981).

There is a general decrease in vertebral strength with age, which is particularly apparent beyond the age of 40 years (Bell et al. 1967, White and Panjabi 1978). This

decrease in strength is associated with an overall decrease in the measured bone mineral density (Ericksen 1976). This decrease in bone mineral density is associated with clear-cut changes in the composition of trabecular bone in the vertebral body (Atkinson 1967). Specifically, there is a significant loss of the horizontally-oriented trabeculae – a loss particularly marked in the central portion of the vertebral body (Atkinson 1967). The disappearance of the horizontal trabeculae deprives the vertical trabeculae of important fortifications, and the load-bearing capability of the center of the vertebral body becomes degraded as a result (Bogduk 1997).

The decreasing strength of the cancellous core of the vertebral body as aging takes place is associated with a shift in the loading paths through which compressive loads are transmitted through the vertebrae (Bogduk 1997). In spines less than 40 years of age, the cancellous bone bears approximately 55 percent of the compressive load, while the cortical shell bears 45 percent of the load. However, the process of trabecular absorption (detailed above) reduces the proportion of the compressive load borne by the cancellous bone to only 35 percent, and increases the demands on the cortical shell, which must assume nearly two-thirds of the compressive load (Rockoff et al. 1969). This change will, in turn, alter the deformation characteristics of the vertebral body, since cortical bone fails at 2% deformation, while trabecular bone can withstand 9.5 % deformation before failing (White and Panjabi 1978). The upshot is that the vertebral body would be expected to experience decreased resistance to deformation and would be more susceptible to compression fracture (Bogduk 1997).

The age-related changes detailed above also impact the vulnerability of the endplate to failure. Specifically, the weakened trabecular network underlying the endplates permits increased deformation of both endplates of affected vertebrae. This increased deformation is thought to render the endplates more vulnerable to fracture when compressive loads are introduced (Bogduk 1997). The finding of increased fracturing in the endplates and vertical trabeculae with increasing age tends to support this notion (Brown et al. 1957, Vernon-Roberts and Pirie 1973).

The morphological characteristics of lumbar vertebrae are also considered to play a role in withstanding the compressive loads experienced by the body. It has not escaped notice that the size of the vertebral bodies of the spine correspond generally to the amount of load each is thought to bear, in accordance with Wolff's law (Wolff 1891). It is also well established that the ultimate compressive strength of lumbar vertebrae becomes increasing greater with more caudal vertebrae, which are also greater in size (Brinckmann et al. 1988, Perry 1957, White and Panjabi 1978). The increase in strength with lower vertebrae does not seem to be associated with differences in the material properties of various lumbar vertebrae (Weaver and Chalmers 1966). This has led some authors to conclude that the difference in strength is most likely due to the size of the vertebrae themselves (White and Panjabi 1978). Thus, it appears that both bone mineral content and the size of the vertebral body may play important roles in response of lumbar motion segments to compression and fatigue failure.

In view of the structural roles played by the size and mineral content of lumbar vertebrae, it was of interest to examine the extent to which these factors play a role in determining fatigue failure of motion segments during simulated lifting activities. Thus, the purpose of this chapter is to present the results of an analysis of the effects of bone mineral content and vertebral body size on fatigue failure in a cadaveric study examining the effects of lifting in various torso flexion postures.

## **5.2 Method**

Details regarding the cadaveric specimens were provided previously in Chapter 4. While 36 segments were tested in the experiment described in Chapter 4, the current analysis is based on analysis of a subset of the entire dataset. This is due to the fact that bone density determinations were available only for the upper lumbar levels, in our case motion segments L1-L2 and L3-L4. Thus, the current analysis was based on 24 of the total of 36 motion segments tested, and excludes the L5-S1 segments.

### **5.2.1 Determination of bone density**

The bone density for vertebral bodies L1-L4 of each spine was measured by means of dual x-ray absorptiometry (DXA) using a Lunar DPX machine (Lunar, Madison, WI). Specimens were removed from the freezer and positioned on a bed of rice (to simulate body tissue) with the anterior aspect of the spine facing up. The bone mineral content (BMC) of the specimen was expressed in grams and divided by the projected area of the bone to derive the bone mineral areal density (BMD) in grams

per square centimeter. Bone mineral content, projected area of the vertebral bone, and the derived bone mineral density values were provided in reports such as those provided in Figure 5.1.

### **5.2.2 Reproducibility study**

DXA scans were performed on one lumbar spine specimen (F5) to establish the reproducibility of bone mineral density scan results both with and without repositioning of the spine on the bed of rice. These tests were performed to ensure reliable measurement of subsequent scans of bone mineral content and density. Five scans were performed without repositioning of the specimen and these were followed by four additional scans for which the specimen was repositioned in between scans. Coefficients of variation were calculated for each set of scans and were compared to assess the both the variability between scans and the possible impact of repositioning the intact spine in the bed of rice.

### **5.2.3 Measurement of Lumbar Morphological Characteristics and Calculations of Volumes**

Measurement of a number of dimensions of the lumbar motion segments used in this study were obtained from lateral and anterior-posterior radiographs taken of each specimen after dissection. Figures 5.2 – 5.5 illustrate the measurements taken for each specimen. A software program designed for the analysis of radiographic data (eFilm, Madison , WI) which possessed tools for measuring linear dimensions and angles from radiographic images was utilized to obtain these measurements. Due to

the fan-shaped beam emitted by the radiographic equipment, it was necessary to apply a scaling adjustment to the measurements obtained from the radiographic image.

Measurements of vertebral body dimensions were taken via calipers and compared to the measurements derived from the radiographic software tools. Based on this analysis, measurements made on the radiographic plate were reduced by 12% to obtain proper sizing of the measurements of the motion segments. Volumes for the lumbar vertebral bodies (i.e., excluding posterior elements) and the intervertebral discs were estimated using the following procedures. The volume of the vertebral bodies was estimated by treating the bodies as two frustrums of elliptical cones. The volumes of both halves of the vertebral bodies were then added together to provide the total volume. The volume of each frustrum was calculated as in the following example.

The formula for the area of an ellipse ( $\pi ab$ ) was used as a basis to develop the volumes of each slice the frustrum. However, for the lumbar vertebrae the dimension of the semiminor axis (sagittal plane) was treated as a constant, since this dimension was determined to exhibit little variability according to the height at which it was measured on the vertebral body. The semimajor axis of the ellipse (the frontal dimension, or AP view) did vary according to the height at which it was measured on the vertebral body.

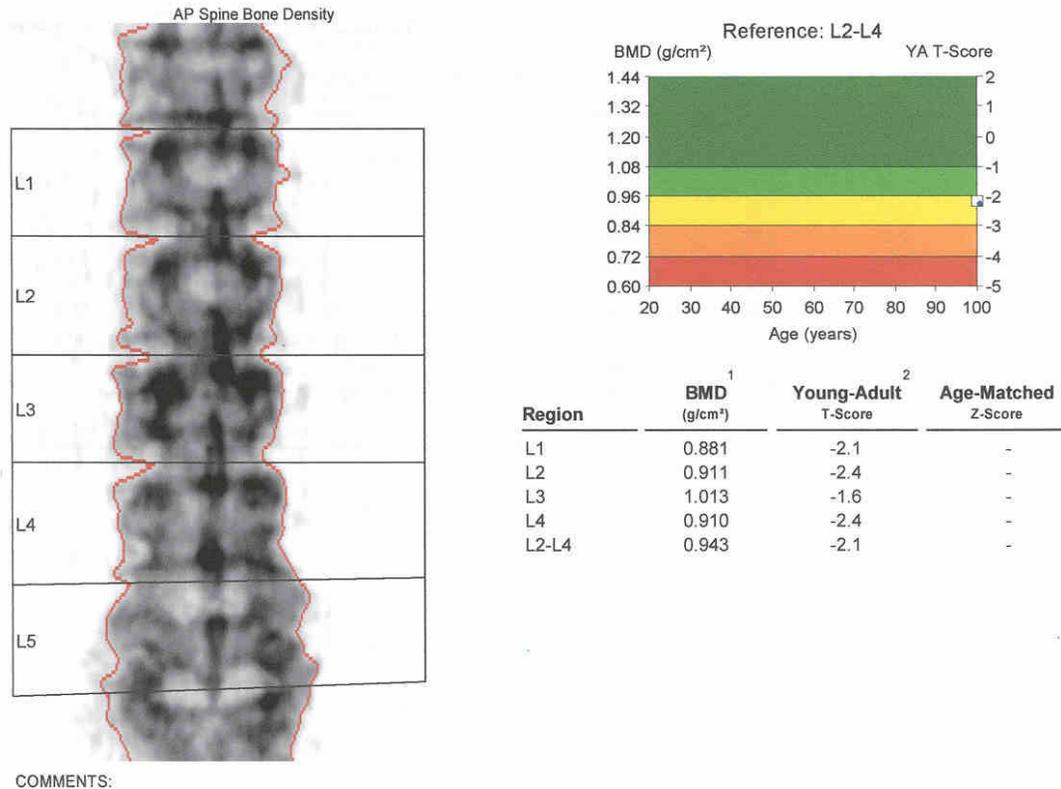
Thus, it was necessary to integrate to obtain the estimated volume as follows:

$$V \approx \sum \pi ab \Delta h$$

Where: a = semimajor axis, b = semiminor axis, and h = height of the frustrum.

Since b is treated as a constant:

$$V = \pi b \int_{h=0}^{h=x} a dh$$



#### ANCILLARY RESULTS [AP Spine]

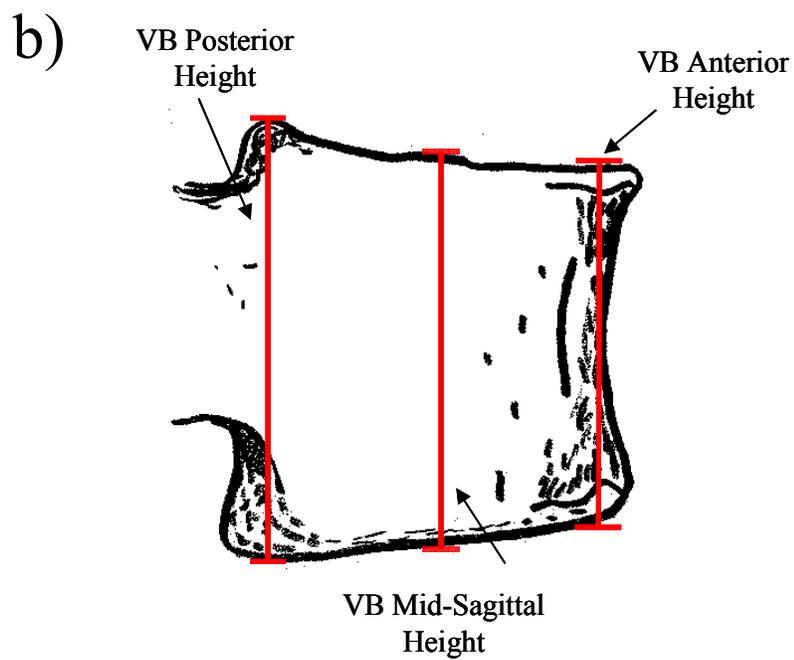
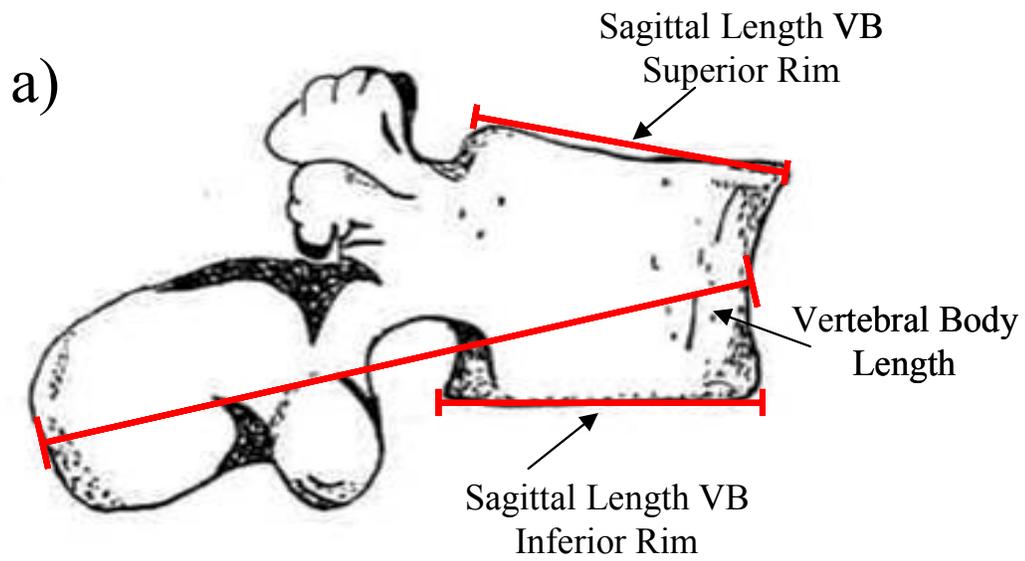
Region	<sup>1</sup> BMD (g/cm <sup>2</sup> )	Young-Adult (%)	<sup>2</sup> T-Score	Age-Matched (%)	Z-Score	<sup>4</sup> Est. BMC (g)	<sup>4</sup> Est. Area (cm <sup>2</sup> )	<sup>4</sup> Est. Width (cm)	<sup>4</sup> Est. Height (cm)
L1	0.881	78	-2.1	-	-	13.7	15.5	4.6	3.36
L2	0.911	76	-2.4	-	-	15.2	16.7	4.5	3.73
L3	1.013	84	-1.6	-	-	16.3	16.0	4.8	3.36
L4	0.910	76	-2.4	-	-	17.0	18.7	5.0	3.73
L5	0.825	-	-	-	-	16.7	20.2	6.0	3.36
L1-L2	0.897	78	-2.1	-	-	28.9	32.2	4.5	7.09
L1-L3	0.936	80	-2.0	-	-	45.1	48.2	4.6	10.45
L1-L4	0.928	79	-2.1	-	-	62.1	66.9	4.7	14.18
L2-L3	0.961	80	-2.0	-	-	31.5	32.7	4.6	7.09
L2-L4	0.943	79	-2.1	-	-	48.5	51.4	4.8	10.82
L3-L4	0.958	80	-2.0	-	-	33.3	34.7	4.9	7.09

**Figure 5.1:** Example of bone density scan report for spine M1.

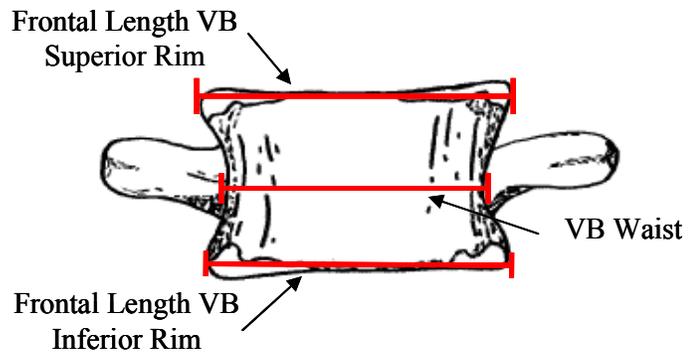
The dimension of  $b$  was calculated using half of the sagittal length of the superior rim of the vertebral body. The dimension of  $a$  ranged from the vertebral waist dimension (measured in the frontal plane) to the measurement of the superior rim dimension in the frontal plane. The value of the height of the frustrum was taken as half of the average value of three height measurements of the vertebral body measured in the sagittal dimension. Thus, the measurement of the vertebral waist width was taken as the frontal width of the vertebral body at the point bisecting the upper and lower halves of the vertebral body. The process described above for estimating the volume of the frustrum of one half of the vertebral body was repeated for the other half and the two volumes were added together to estimate the total volume.

The shape of the sacrum demanded a slightly different treatment in terms of volume estimation of the S1 vertebral body. S1 was also treated as an elliptical cone, but in contrast with the lumbar vertebrae, both elliptical axes varied according to the height of measurement. Furthermore, since the dimensions of both elliptical axes of the sacrum continually increased with increasing height (from the inferior “endplate” surface at S1-S2 to the S1 superior endplate), the volume could be estimated using a single elliptical cone frustrum. The calculations used to estimate the volume of S1 are as follows:

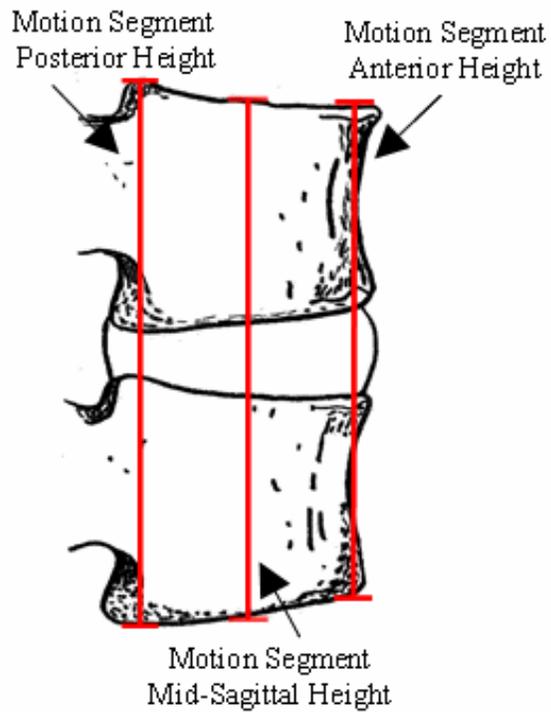
$$V = \pi \int_{h=0}^{h=x} ab \, dh$$



**Figure 5.2:** Measurements of vertebral body a) length and b) height made in the sagittal plane.

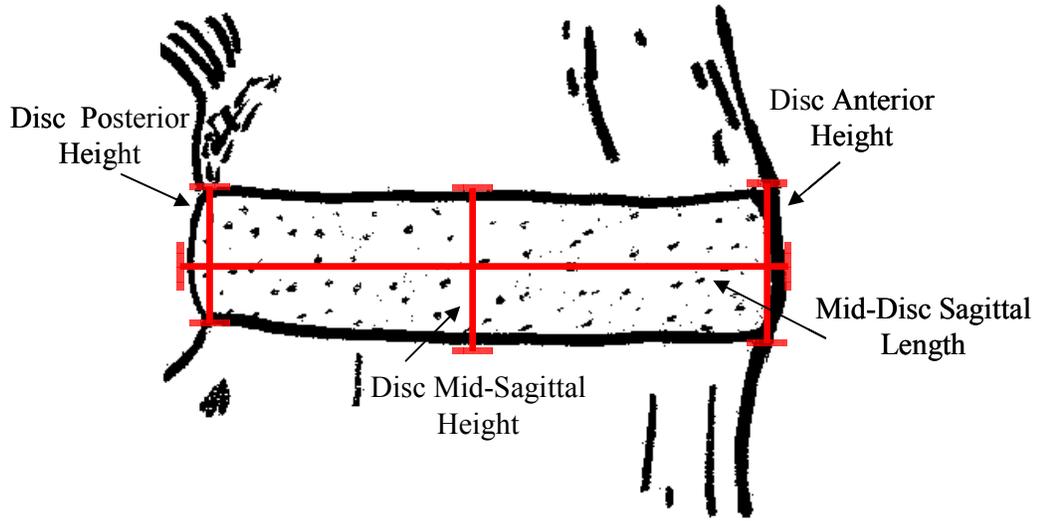


**Figure 5.3:** A-P measurements of lumbar vertebrae.

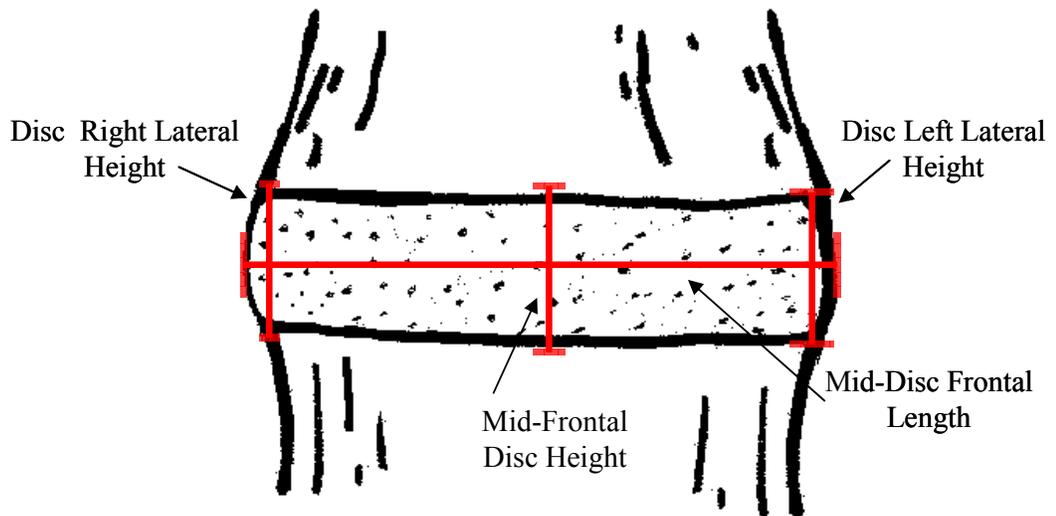


**Figure 5.4:** Sagittal measurements of motion segment heights.

a)



b)



**Figure 5.5:** Intervertebral disc measurements in a) sagittal and b) frontal planes

The volume of the L1-L2, L3-L4 and L5-S1 intervertebral discs were estimated by averaging the dimension of two elliptical cylinders: one describing the outermost sagittal and frontal dimensions of the disc (to the point of the outermost bulges of the disc in each dimension), and one describing a cylinder whose dimension were described by the attachment of the discs at the average dimension of the superior and inferior vertebral rims adjacent to the disc. The height of both cylinders was taken to be the average of the three measurements of disc height taken in the sagittal plane. The semimajor axes of the cylinders were described by frontal plane measurements (of the disc or of the average vertebral rim dimensions), while the semiminor axes were defined using sagittal plane dimensions.

#### 5.2.4. Statistical analysis

A Cox regression analysis (time-to-event regression analysis) was conducted to evaluate the effects of torso flexion angle, lumbar level, motion segment volume and bone mineral content on the number of cycles to fatigue failure. Cox regression analyses were performed to examine all main effects and two-way interactions. The hypothesized model for this analysis can be represented by the following equation:

$$h(t|z) = h_0(t|z)\exp(\beta_1z_1 + \beta_2z_2 + \beta_3z_3 + \beta_4z_4 + \beta_5z_5 + \beta_6z_6 + \beta_7z_7 + \beta_8z_1z_3 + \beta_9z_2z_3 + \beta_{10}z_1z_4 + \beta_{11}z_2z_4 + \beta_{12}z_1z_5 + \beta_{13}z_2z_5 + \beta_{14}z_1z_6 + \beta_{15}z_2z_6 + \beta_{16}z_1z_7 + \beta_{17}z_2z_7 + \beta_{18}z_3z_4 + \beta_{19}z_3z_5 + \beta_{20}z_3z_6 + \beta_{21}z_3z_7 + \beta_{22}z_4z_5 + \beta_{23}z_4z_6 + \beta_{24}z_4z_7 + \beta_{25}z_5z_6 + \beta_{26}z_5z_7 + \beta_{27}z_6z_7)$$

where:

$\beta_k$  = coefficients for variables used in the model,  $z_1$  = 22.5 deg torso flexion,  $z_2$  = 45 deg torso flexion,  $z_3$  = L1-L2 lumbar level,  $z_4$  = bone mineral density (g/cm<sup>2</sup>),  $z_5$

= bone mineral content (g),  $z_6$  = motion segment volume ( $\text{cm}^3$ ), and  $z_7$  = motion segment weight (g). It should be noted that in the current analysis, unlike those in Chapter 4, L5-S1 segments were not included, due to the lack of bone mineral data.

It was also of interest to evaluate whether the survival of motion segments was attributable to specific changes in loading variables associated with various torso flexion angles. Therefore, a univariate survival analysis was performed to evaluate whether compression loads, shear loads, motion segment flexion angle, and/or load rate experienced in the various torso flexion angles might explain the differences in fatigue life observed in the various torso flexion angles. A forward selection procedure was used in the development of all survival models. Variables were selected for inclusion on the basis of the Akaike Information Criterion (AIC). That is, the model whose variable resulted in the lowest AIC was selected at each successive step of the model-building process. The model-building process ceased when the lowest AIC for a step was greater than the lowest AIC obtained in the previous step.

A primary assumption of the time-to-event regression model was that the hazard proportions associated with model variable comparisons did not differ significantly with respect to time during the period of analysis. This assumption was checked for all variables at the univariate stage of the model-building process. If the assumption did not appear tenable, the interaction between the variable and the natural logarithm of time was included in the model whenever that variable was entered into the regression models. A final check of the proportional hazards assumption was performed once the final model was determined.

The Wald statistic was used to calculate the probability that the risk ratio was significantly different from one for covariates with one degree of freedom. Risk ratios for variables with multiple degrees of freedom were tested for significance by subtracting the log likelihoods for the reduced model from the full model, obtaining a chi-square with the appropriate degrees of freedom.

## **5.3 Results**

### **5.3.1 Bone Densitometry Measurements**

#### *5.3.1.1 Reproducibility Study*

Tables 5.1 and 5.2 contain results of the study on the reproducibility of Bone Mineral Density measurements taken on a lumbosacral spine. As can be seen from the table, the error was generally less than two percent whether repositioning was performed or not. This was judged to represent satisfactory reproducibility for the study.

Results of these tests are provided in Tables 5.1 and 5.2. Reproducibility of the DXA scans were found to be excellent both with and without repositioning of the specimen. Coefficients of variation were less than 2.1% in all cases where no repositioning of the specimen was done, and were similar (coefficients of variation less than 2.3%) when the specimen was repositioned between scans. In many instances, the coefficients of variation were less than 1%. Based on these test results, it was concluded that measurements made using DXA were reproducible and were not greatly influenced by repositioning of the lumbar spine specimen in the bed of rice.

### *5.3.1.2 Bone Mineral Density Results*

Table 5.3 and 5.4 present results of the bone density scans for the specimens used in this analysis. Table 5.3 presents the bone mineral content and density data for individual vertebrae from L1 to L4. Table 5.4 presents data estimating the average bone density for motion segments L1-L2 and L3-L4, as well as a combined reading for L2-L4, often used in clinical settings as an estimate of the bone density of the lumbar spine (Szejnfeld et al. 1995). As can be seen from Table 5.3, the average bone mineral content, density, and estimated vertebral body area all increase as one moves from L1 to L4. The variability of these measures remains fairly constant throughout this range. Table 5.4 illustrates that the same trend holds true for combined measurements of motion segments (i.e., the more caudal the position of the motion segment, the higher the bone mineral density and bone mineral content).

Data for the combined bone mineral density and content data for the L2-L4 region can be used to compare the bone mineral density of the specimens used in this experiment to normative data. The T-scores contained in Table 5.4 provide a statistical measure of the bone mineral density values compared to the densities obtained for Young Adults (aged 20-40). The World Health Organization (Kanis et al. 1994) defines T-scores greater than -1.0 to be normal, T-scores between -1.0 and -2.5 are defined as representing osteopenia (mild to moderate bone loss), and T-scores less than -2.5 are considered osteoporotic (significant bone loss). Results of tests on the present specimens indicate that bone mineral densities ranging from near normal values for a young adult (T-score of -0.3) to values indicative of significant osteoporosis (T-score

of -5.0). The average bone mineral density of specimens used in this experiment was in the osteopenic range, indicative of moderate bone loss.

### **5.3.2 Motion segment morphology**

Table 5.6 contains measurements of the intervertebral disc dimensions for L1-L2, L3-L4, and L5-S1. Sagittal measurements of disc height show a pronounced increase in the anterior height of discs with lower levels, a slight increase in posterior disc height at lower levels, but little change of the height in the center of the disc with lower levels. These data suggest that the mid-plane dimensions of disc length stay relatively constant in the sagittal dimension, but show a gradual increase in the frontal dimension with lower lumbar levels. Estimated disc volumes exhibited a linear increase in size with lower motion segment levels.

Measurement of the sagittal heights of the motion segments, contained in Table 5.7 showed that anterior and mid-segment dimensions increased with lower levels, while the posterior heights decreased with lower levels, undoubtedly reflective of the lordosis observed in the lumbar region. The weight of the segments was also found to increase in a linear fashion with lower spinal levels.

SCAN NO.	L1		L2		L3		L4		L2-4						
	BMD	BMC AREA													
1	0.614	8.4	13.6	0.689	9	13.6	0.779	12.5	16.1	0.789	15.5	19.6	0.758	37.3	49.3
2	0.590	8.3	14	0.66	9.5	14.4	0.787	12.6	16	0.807	15.4	19	0.758	37.5	49.4
3	0.615	8.4	13.7	0.673	9.4	14	0.781	12.4	15.9	0.802	15.2	19	0.758	37.1	48.9
4	0.603	8.3	13.8	0.673	9.5	14.1	0.782	12.5	16	0.801	15.4	19.2	0.759	37.5	49.4
5	0.592	8.3	14.1	0.679	9.5	14	0.777	12.5	16.1	0.797	15.2	19.1	0.757	37.2	49.2
AVERAGE	0.603	8.340	13.84	0.675	9.380	14.02	0.781	12.500	16.02	0.799	15.340	19.180	0.758	37.320	49.24
STD DEV	0.011	0.049	0.185	0.009	0.194	0.256	0.003	0.063	0.075	0.006	0.120	0.223	0.001	0.160	0.185
%	1.7	0.6	1.3	1.4	2.1	1.8	0.4	0.5	0.5	0.8	0.8	1.2	0.1	0.4	0.4

**Table 5.1:** Reproducibility of DXA scans for specimen F5 without repositioning of specimen between scans

SCAN NO.	L1		L2		L3		L4		L2-4	
	BMD	BMC AREA	BMD	BMC AREA	BMD	BMC AREA	BMD	BMC AREA	BMD	BMC AREA
1	0.592	8.3 14.1	0.679	9.5 14	0.777	12.5 16.1	0.797	15.2 19.1	0.757	37.2 49.2
2	0.603	8.2 13.5	0.683	9.4 13.8	0.782	12.6 16.1	0.782	15.7 20	0.755	37.7 49.9
3	0.587	8.3 14.1	0.669	9.5 14.2	0.769	12.5 16.2	0.797	15.3 19.2	0.751	37.3 49.7
4	0.602	8.2 13.7	0.68	9.4 13.8	0.773	12.5 16.1	0.79	15.4 19.5	0.754	37.2 49.4
5	0.598	8.3 13.9	0.685	10 14.6	0.782	12.8 16.4	0.801	15.5 19.4	0.761	38.8 50.3
AVERAGE	0.596	8.260 13.86	0.679	9.560 14.08	0.777	12.580 16.18	0.793	15.420 19.440	0.756	37.640 49.700
STD DEV	0.006	0.049 0.233	0.006	0.224 0.299	0.005	0.117 0.117	0.007	0.172 0.314	0.003	0.609 0.385
%	1.0	0.6 1.7	0.8	2.3 2.1	0.7	0.9 0.7	0.8	1.1 1.6	0.4	1.6 0.8

**Table 5.2:** Reproducibility tests of DXA scans for spine F5 where specimen was repositioned between scans.

Spine Label	Gender	Age	L1			L2			L3			L4		
			BMD	BMC	VB Area									
M11	M	85	0.957	11.6	12.1	1.086	17	15.7	1.197	18.7	15.6	1.333	19.9	14.9
M8	M	74	1.117	16.3	14.6	1.125	17.5	15.5	1.125	19.9	17.7	1.067	20.1	18.8
M1	M	77	0.881	13.7	15.5	0.911	15.2	16.7	1.013	16.3	16	0.91	17	18.7
F9	F	85	0.626	7.7	12.3	0.737	10.5	14.3	0.938	12.9	13.8	0.901	15.3	16.9
M7	M	65	0.642	6	9.4	0.627	5.4	8.6	0.645	6.4	9.9	0.648	7.6	11.8
F6	F	82	0.976	11.1	11.3	1.013	12.1	12	1.063	14.1	13.2	1.077	15.8	15.8
F17	F	93	0.58	5.3	9.1	0.722	6.6	9.2	0.682	7.5	11	0.812	8.4	10.3
M10	M	80	0.809	15.3	18.9	0.966	19.4	20.1	1.082	22.9	21.1	1.054	22.7	21.5
F13	F	91	0.716	7.9	11.1	0.784	9.3	11.8	0.894	11.3	12.6	0.91	13.1	14.4
M16	M	79	0.914	12	13.1	0.974	14.3	14.6	1.006	16.3	16.2	1.085	16.5	15.2
M12	M	84	0.871	14.7	16.9	0.869	16.5	18.9	1.029	20.2	19.6	1.032	21.5	20.9
M14	M	73	0.818	11.6	14.1	0.959	13.6	14.2	1.017	15.1	14.8	0.911	16.4	18
	Mean		0.826	11.1	13.2	0.898	13.1	14.3	0.974	15.1	15.1	0.978	16.2	16.4
	SD		0.161	3.7	2.95	0.153	4.4	3.5	0.165	5.0	3.3	0.171	4.72	3.4
	Min	65	0.58	5.3	9.1	0.627	5.4	8.6	0.645	6.4	9.9	0.648	7.6	10.3
	Max	93	1.117	16.3	18.9	1.125	19.4	20.1	1.197	22.9	21.1	1.333	22.7	21.5

**Table 5.3:** Results of bone densitometry studies for individual vertebrae.

Spine Label	Gender	Age	L1-L2			L3-L4			L2-4			T-score
			BMD	BMC	VB Area	BMD	BMC	VB Area	BMD	BMC	VB Area	
M11	M	85	1.029	28.6	27.8	1.266	38.6	30.5	1.203	55.6	46.2	-0.3
M8	M	74	1.123	33.8	30.1	1.096	40	36.5	1.104	57.5	52	-1.1
M1	M	77	0.898	28.9	32.2	0.96	33.3	34.7	0.943	48.5	51.4	-2.1
F9	F	85	0.684	18.2	26.6	0.919	28.2	30.7	0.86	38.7	45	-2.8
M7	M	65	0.633	11.4	18	0.645	14	21.7	0.641	19.4	30.3	-5
F6	F	82	0.996	23.2	23.3	1.031	29.9	29	1.053	42	39.9	-1.2
F17	F	93	0.650	11.9	18.3	0.746	15.9	21.3	0.738	22.5	30.5	-3.8
M10	M	80	0.890	34.7	39	1.07	45.6	42.6	1.035	65	62.7	-1.8
F13	F	91	0.751	17.2	22.9	0.904	24.4	27	0.866	33.7	38.9	-2.8
M16	M	79	0.949	26.3	27.7	1.045	32.8	31.4	1.022	47	46	-1.7
M12	M	84	0.872	31.2	35.8	1.03	41.7	40.5	0.979	58.2	59.4	-2.2
M14	M	73	0.890	25.2	28.3	0.96	31.5	32.8	0.959	45	47	-2.4
	Mean		0.864	24.2	27.5	0.973	26.9	27.1	0.95	44.4	45.8	-2.3
	SD		0.155	8.0	6.4	0.162	14.5	12.9	0.156	14.1	10.0	1.3
	Minimum		0.633	11.4	18	0.645	14	21.3	0.641	19.4	30.3	-5
	Maximum		1.123	34.7	39	1.266	45.6	42.6	1.203	65	62.7	-0.3

**Table 5.4:** Bone mineral density results for various lumbar motion segments. T-scores represent bone mineral density results compared with normal young adults.

		L1	L2	L3	L4	L5	S1
Vertebral Body Length	mean	7.37	8.25	8.43	8.27	7.92	5.82
	SD	1.20	0.75	0.63	0.59	0.76	0.68
	min	4.05	7.04	7.48	7.48	6.86	4.75
	max	8.54	9.06	9.24	9.06	9.06	6.95
Vertebral Body Waist (AP)	mean	3.89	3.96	4.14	4.28	4.84	5.16
	SD	0.47	0.41	0.49	0.38	0.57	0.48
	min	3.17	3.26	3.26	3.61	3.96	4.66
	max	4.49	4.576	4.75	4.75	5.90	6.07
Vertebral Body Anterior Height	mean	2.55	2.74	2.85	2.74	3.09	3.50
	SD	0.27	0.35	0.28	0.37	0.24	0.24
	min	2.20	1.85	2.29	1.76	2.552	3.08
	max	3.08	3.08	3.34	3.08	3.344	3.87
Vertebral Body Mid-Sagittal Height	mean	2.38	2.46	2.55	2.49	2.59	2.95
	SD	0.24	0.35	0.21	0.32	0.28	0.37
	min	2.02	1.58	2.29	1.58	1.94	2.46
	max	2.73	2.82	2.99	2.90	3.08	3.61
Vertebral Body Posterior Height	mean	2.85	2.93	2.88	2.75	2.57	2.76
	SD	0.30	0.34	0.23	0.29	0.19	0.43
	min	2.38	2.11	2.55	2.11	2.29	2.20
	max	3.43	3.43	3.26	3.08	2.90	3.70
Sagittal Length Vertebral Body Superior Rim	mean	3.51	3.64	3.78	3.65	3.68	3.71
	SD	0.42	0.32	0.47	0.24	0.34	0.27
	min	2.90	2.99	3.08	2.99	3.26	3.34
	max	4.31	4.14	4.49	3.96	4.22	4.31
Frontal Length Vertebral Body Superior Rim	mean	4.63	5.03	5.01	5.23	5.38	6.03
	SD	0.51	0.62	0.63	0.63	0.67	0.71
	min	3.70	3.78	3.96	4.40	4.40	5.10
	max	5.54	6.16	6.25	6.51	6.86	7.48
Sagittal Length Vertebral Body Inferior Rim	mean	3.61	3.88	3.79	3.78	3.64	2.00
	SD	0.34	0.39	0.31	0.58	0.32	0.34
	min	3.08	3.17	3.26	2.90	3.08	1.32
	max	4.14	4.45	4.22	5.02	4.22	2.38
Frontal Length Vertebral Body Inferior Rim	mean	5.02	5.02	5.38	5.32	5.73	3.07
	SD	0.66	0.61	0.58	0.76	0.72	0.43
	min	3.96	4.05	4.40	3.96	4.84	2.64
	max	6.07	5.98	6.60	6.78	7.66	3.96
Estimated Volume	mean	32.16	36.77	38.96	37.58	41.65	41.61
	SD	8.35	9.88	9.65	8.87	9.97	14.01
	min	19.90	19.01	24.19	20.15	27.98	25.65
	max	49.58	51.89	56.78	48.28	66.57	65.91

**Table 5.5:** Vertebral body measurements (cm) and estimated volumes (cm<sup>3</sup>)

**INTERVERTEBRAL  
DISCS**

		<b>L1-L2</b>	<b>L3-L4</b>	<b>L5-S1</b>
Disc Anterior Height	mean	0.76	0.87	1.19
	SD	0.22	0.13	0.30
	min	0.35	0.70	0.70
	max	1.06	1.06	1.76
Disc Mid-Sagittal Height	mean	0.86	0.89	0.83
	SD	0.24	0.25	0.31
	min	0.35	0.35	0.44
	max	1.23	1.23	1.32
Disc Posterior Height	mean	0.42	0.44	0.50
	SD	0.14	0.18	0.22
	min	0.26	0.18	0.18
	max	0.70	0.70	0.88
Mid-Disc Sagittal Length	mean	3.92	4.06	4.00
	SD	0.41	0.29	0.30
	min	3.26	3.52	3.52
	max	4.66	4.40	4.58
Left Lateral Disc Height	mean	0.72	0.78	0.43
	SD	0.22	0.24	0.17
	min	0.26	0.35	0.18
	max	1.14	1.14	0.79
Mid-Frontal Disc Height	mean	0.89	0.88	0.77
	SD	0.24	0.22	0.27
	min	0.35	0.44	0.44
	max	1.32	1.14	1.23
Right Lateral Disc Height	mean	0.77	0.70	0.45
	SD	0.25	0.18	0.12
	min	0.35	0.26	0.26
	max	1.23	0.88	0.62
Mid-Frontal Disc Length	mean	5.32	5.71	6.06
	SD	0.66	0.63	0.77
	min	4.14	4.75	5.28
	max	6.34	6.78	7.74
Estimated Volume	mean	10.47	12.61	15.02
	SD	3.36	4.03	4.58
	min	5.53	7.05	8.54
	max	17.41	19.60	24.42

**Table 5.6:** Dimensions and estimated volumes of L1-L2, L3-L4, and L5-S1 intervertebral discs.

**MOTION  
SEGMENTS**

		<b>L1-L2</b>	<b>L3-L4</b>	<b>L5-S1</b>
Weight	mean	216.91	238.58	275.91
	SD	66.16	67.95	56.62
	min	123.50	146.00	218.70
	max	361.00	373.40	401.60
Anterior Height	mean	5.98	6.34	6.94
	SD	0.70	0.68	0.63
	min	4.49	4.93	6.25
	max	6.95	7.39	8.62
Mid-Segment Height	mean	5.64	5.80	5.95
	SD	0.57	0.56	0.52
	min	4.40	4.66	5.46
	max	6.34	6.60	7.39
Posterior Height	mean	6.12	5.96	5.56
	SD	0.64	0.64	0.52
	min	5.19	4.93	4.84
	max	7.13	6.95	6.86

**Table 5.7:** Weight (g) and sagittal height measurements (cm) of motion segments.

### 5.3.3 Survival Analyses

Results of the Cox regression model analysis for the reduced data set are presented in Table 5.8– 5.12. As can be seen from these tables, torso flexion angle and bone mineral content were determined to be significant predictors of the number of cycles to failure. The final model for the analysis of the reduced data set is expressed in the following equation:

$$h(t|z) = h_0(t|z)\exp(2.716*z_1 + 3.380*z_2 - 0.087*z_5)$$

where:

$z_1 = 22$  degrees torso flexion,  $z_2 = 45$  degrees torso flexion,  $z_5 =$  bone mineral content of the total motion segment (grams). However, it should be noted that while bone mineral content was the factor selected in the analysis shown in Table 5.9, motion segment volume and bone mineral density were either significant predictors or very close to achieving this status, even with this relatively small number of samples. From the betas observed in this analysis, all of these variables were predicted to have a protective effect, (i.e., increasing values for these variables would extend fatigue life of lumbar motion segments).

Figure 5.6 presents relative risks based upon this model for various torso flexion angles and motion segment bone mineral content (in grams). The range of bone mineral content in this table (10-50 grams) approximates the bone mineral content observed for motion segments tested in this study (11.5 – 46.7 grams). A reference condition was arbitrarily established to be a motion segment loaded at 0

Variable	Df	Beta	SE	p-value	RR	95% CI	AIC	PH p
<b>Trunk Flexion</b> <b>22.5 deg</b> <b>45.0 deg</b>	2	2.407 2.887	0.832 0.876	0.004 0.004 0.001	11.104 17.937	2.715-56.681 3.224-99.789	<b>79.513</b>	0.053
Lumbar Level	1	-0.586	0.494	0.235	0.556	0.211-1.465	93.169	0.825
BMC	1	-0.040	0.027	0.144	0.961	0.910-1.014	92.416	0.147
BMD	1	-1.656	1.594	0.299	0.191	0.008-4.341	93.554	0.129
MS Volume	1	-0.021	0.012	0.085	0.980	0.957-1.003	91.551	0.229
Weight	1	-0.005	0.004	0.233	0.995	0.987-1.003	93.006	0.430

**Table 5.8:** Univariate Cox regression models for reduced data set

Variable	df	Beta	SE	p-value	RR	95% CI	AIC
Lumbar Level	1	0.986	0.612	0.107	2.680	0.807-8.896	78.777
<b>BMC</b>	1	-0.087	0.038	0.022	0.917	0.851-0.988	<b>76.013</b>
BMD	1	-3.304	1.752	0.059	0.037	0.001-1.138	77.881
MS Volume	1	-0.037	0.016	0.024	0.964	0.934-0.995	76.130
Weight	1	-0.007	0.006	0.204	0.993	0.981-1.004	79.623

**Table 5.9:** Univariate Cox regression models for reduced data set incorporating Torso Flexion

Variable	df	Beta	SE	p-value	RR	95% CI	AIC
Lumbar Level	1	0.823	0.626	0.189	2.277	0.668-7.764	76.297
BMD	1	0.505	3.430	0.883	1.657	0.002-1376.67	78.108
MS Volume	1	-0.021	0.025	0.396	0.979	0.933-1.028	77.376
Weight	1	0.005	0.007	0.512	1.005	0.990-1.020	77.725

**Table 5.10:** Bivariate Cox regression models for reduced data set incorporating Torso Flexion and BMC

Variable	Df	Beta	SE	p-value	RR	95% CI	AIC
Trunk Flexion* Lumbar level ABWG0_1 ABWG0_2 ABWG1_1 ABWG1_2	4	*DNC					
Trunk Flexion* BMC 22 deg*BMC 45 deg*BMC	2	-0.105 0.001	0.105 0.091	0.404 0.999 0.000	0.901 1.001	0.733-1.106 0.836-1.197	78.362
Trunk Flexion* BMD 22 deg*BMD 45 deg*BMD	2	-6.470 0.537	5.149 2.777	0.300 0.209 0.847	0.002 0.847	0.000-37.385 0.007-394.861	77.704
Trunk Flexion* Volume 22 deg*VOL 45 deg*VOL	2	-0.043 -0.002	0.027 0.027	0.259 0.113 0.934	0.958 0.998	0.909-1.010 0.947-1.051	77.348
Trunk Flexion* Weight 22deg*Weight 45deg*Weight	2	0.005 0.005	0.016 0.007	0.798 0.734 0.510	1.005 1.005	0.975-1.037 0.991-1.018	79.702
Lumbar Level* BMC	1	-0.022	0.023	0.342	0.979	0.936-1.023	77.205
Lumbar Level* BMD	1	0.692	0.654	0.290	1.998	0.555-7.191	76.992
Lumbar Level* VOLUME	1	0.008	0.007	0.248	1.008	0.995-1.022	78.755
Lumbar Level* Weight	1	0.003	0.002	0.240	1.003	0.998-1.008	76.736
BMC*BMD	1	0.044	0.104	0.674	1.045	0.852-1.281	77.952
BMC*Volume	1	0.000	0.001	0.823	1.000	0.998-1.001	78.079
BMC*Weight	1	0.000	0.000	0.457	1.000	1.000-1.001	77.612
BMD*Volume	1	-0.042	0.048	0.385	0.959	0.873-1.054	77.344
BMD*Weight	1	0.014	0.012	0.226	1.014	0.991-1.038	76.721
Volume* Weight	1	0.000	0.000	0.873	1.000	1.000-1.000	78.105

**Table 5.11:** Cox regression models examining two-way interactions for reduced data set. Models incorporate Torso Flexion and BMC. \*DNC indicates model did not converge.

Variable	Df	Beta	SE	p-value	RR	95% CI	PH p
<b>Trunk Flexion</b>	2			0.001			
<b>22.5 deg</b>		2.716	0.872	0.002	15.127	2.741-83.487	0.37
<b>45.0 deg</b>		3.380	0.941	0.000	29.371	4.641-185.903	
<b>BMC</b>	1	-0.087	0.038	0.022	0.917	0.851-0.988	

**Table 5.12:** Final Cox regression model

degrees trunk flexion possessing 30 grams of bone mineral content. As can be seen from this table, the relative risk is significantly impacted by both variables. Compared to the reference condition, for example, a decrease by 10 grams in bone mineral content will result in 2.4 times the relative risk. Flexing the torso 22.5 degrees from the neutral posture will result in an 11-fold increase in risk for a given bone mineral content. Examination of this table illustrates that while torso flexion greatly increase risk in general, the relative risk for a severely osteoporotic specimen loaded in the 0 degree condition is greater than a specimen with high bone mineral content loaded in the 45 degree condition.

In the survival analysis examining the relationship of different loading factors to fatigue life, shear loading was found to be the loading variable most highly associated with fatigue life (Beta = -0.002,  $p < 0.002$ ). When beta is exponentiated, the value obtained is 0.998. While it might appear that shear is protective from the

value of this exponentiated coefficient, it must be remembered that posterior shears are treated as negative numbers in the coordinate system used here. Thus, the greater the magnitude of the posterior shear experienced by the motion segment, the fewer the cycles to failure.

Variable	Beta	SE	Wald	p-value	Exp(Beta)	95% CI
Shear	-0.002	0.001	9.531	0.002	0.998	(0.996, 0.999)

**Table 5.13:** Association of shear force to cycles to failure. Neither compression, motion segment flexion or load rate were significantly associated

## 5.4 Discussion

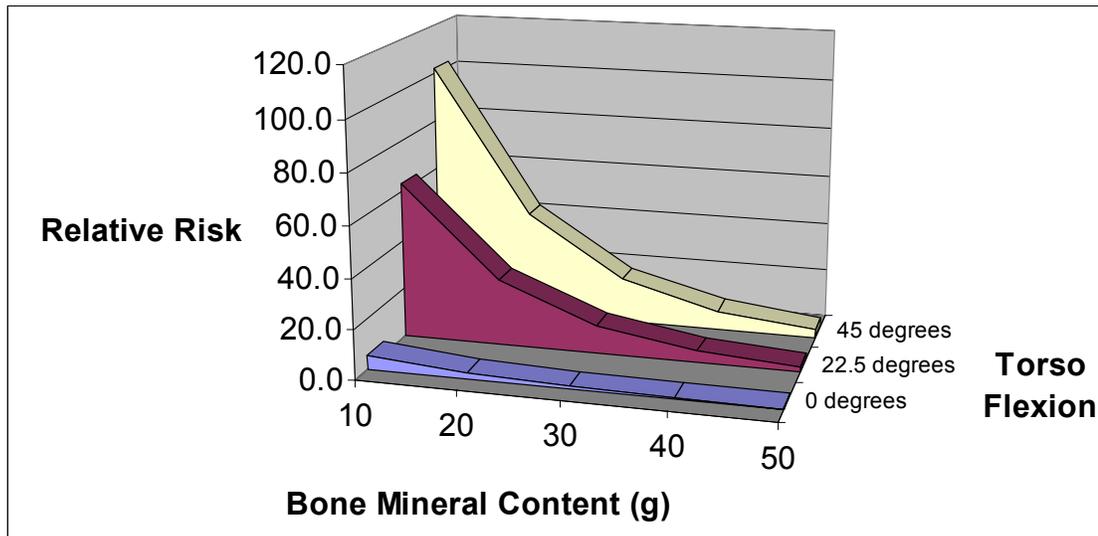
### 5.4.1 Bone mineral content and bone mineral density

Results of the bone mineral content/bone mineral density analyses appear similar to other studies examining older populations of spines (Sjeznfeld et al. 1995). A widerange of bone densities were seen in this study, from near normal for a young adult to values indicative of severe osteoporosis. In comparison with studies examining the bone mineral content of men and women aged 35-40 years of age (Sjeznfeld et al. 1995), the present specimens exhibit a bone mineral content of

approximately 20% less. As would be anticipated, bone mineral densities were lower in the female specimens (averaging  $0.879 \text{ g/cm}^2$ ) compared to the male specimens ( $0.986 \text{ g/cm}^2$ ), probably as a consequence of the bone loss typically observed in post-menopausal women (Bell et al. 1967, Rockoff et al. 1969, Sjeznfeld et al. 1995). It would be expected that reductions in bone content as seen here would result in an increased fracture risk among the specimens tested in this experiment as opposed to specimens that might be obtained from a younger population. Furthermore, the response to load of older specimens experiencing an involution of bone would be expected to be altered as less of the load is borne by the energy-absorbing trabecular bone (Evans 1957), and a greater proportion is resisted by the cortical bone (Rockoff et al. 1969).

#### **5.4.2 Survival Analysis**

Results of the Cox regression model analysis for the reduced data set are presented in Table 5.8– 5.12. As can be seen from these tables, torso flexion angle and bone mineral content were determined to be significant predictors of the numberAs a result of the different loading paths that result as bone content is lost, it will be important to evaluate the impact of higher bone content and density on fatigue failure in younger specimens as research continues along these lines.



**Figure 5.6:** Cox regression model estimates of relative risk by torso flexion angle and bone mineral content, using a segment loaded at 0 degrees flexion with 30 grams of bone mineral content as a referent value.

As mentioned in the introduction, several authors have demonstrated a significant linear relationship between bone mineral content and the ultimate compressive strength of lumbar vertebrae (Bartley et al. 1966, Bell et al. 1967, Galante et al. 1970, Hansson and Roos 1981). However, the relationship between bone mineral content and fatigue failure has been less well characterized. Surprisingly, while Brinkmann et al. (1988) spent considerable effort in evaluating the bone mineral density beneath the endplates of their specimens, they failed to characterize the relationship between bone density and fatigue failure, even though fatigue failure was the primary focus of their experiment. Results of the survival analysis in this chapter

indicate that increased bone mineral content does afford a degree of protection against fatigue failure in lumbar motion segments. This result certainly suggests that aging workers, but particularly females (in whom bone loss is more rapid), would become increasingly susceptible to the effects of fatigue failure as they experience a loss of bone mineral content. Furthermore, this finding reinforces the importance of testing a younger cohort of specimens, as the loss of bone mineral will induce a quicker fatigue failure in older specimens. As a result, the number of cycles to failure in younger specimens would be expected to be somewhat higher than the values obtained in the present study.

The strength of a slender column of circular cross section under a compressive load is given by Euler's formula:

$$F = \frac{\pi^2 ED^2}{16L^2} \times A$$

where F = compressive load

$$\pi = 3.14$$

E = modulus of elasticity of the columnar material

D = diameter of the column

A = cross-sectional area of the column,

and L = free length of the column.

For a column of made of a given material, this formula reduces to:

$$F = C \times \frac{A^2}{L^2}$$

where C = a constant for the column of given material.

Thus, the strength of a column is directly proportional to the square of its cross-sectional area and inversely proportional to the square of its length (White and Panjabi 1978). Therefore, the size of the motion segment would be expected to play a role in both the ultimate strength of a motion segment and in its resistance to cyclic loading. It is not surprising, therefore, that increasing volume of the motion segment was found to have a protective effect in the survival analysis performed on the full data set. However, it is interesting to note that this effect was only present in the analysis where data on bone mineral content was not available. When bone mineral content data was available for inclusion in the survival analysis, it was incorporated in the model, whereas motion segment volume was not. This may suggest that while size of the motion segment is a concern as far as its resistance to cyclic loading, that bone mineral content of a motion segment may be a more critical consideration than its size.

Table 5.20 provides a comparison of selected morphological measures of the motion segments taken in the current study and those taken in previous studies (Berry et al. 1987, Gilad and Nissan 1986, Panjabi et al. 1992). In general, measurements made in the current investigation compare more closely to those of Gilad and Nissan (1986) and Berry et al. (1987), as opposed to Panjabi et al. (1992). The latter study, when compared with the others, appears to have sampled from a smaller (in size) and much more homogeneous sample than any of the other comparison morphology studies. In fact, the sample in the Panjabi et al. (1992) study appears not to accurately reflect the range of variability in morphological measures seen in other studies.

Measurements of anterior and posterior vertebral body heights in the current sample appear similar, if slightly larger, than those in the Gilad and Nissan (1986) and Berry et al. (1987) studies. The former study was a radiographic study involving an all male sample of 157 living subjects, while the latter was a cadaver study involving measurement of 15 male and 15 female spines. One would anticipate that measurements in the current study would be slightly greater than those in the Berry et al. (1987) study due to the greater proportion of male spines, and indeed, this is generally what is observed. In fact, the measurements of vertebral body heights in this study would seem to be more similar in size to those of the Gilad (1986) study employing an all male sample.

Measurements of the endplate widths (superior and inferior) as well as the width of the vertebral waist in this study appear to be quite comparable to the measurements reported by Berry et al. (1987). Again, the measurements reported by Panjabi et al. (1992) appear considerably smaller than those reported by Berry et al. (1987) or those reported in the current investigation. Gilad and Nissan (1986) provide some data on anterior and posterior disc heights in living subjects that can be compared to the cadaver data in the current study. In general, the disc heights observed in the Gilad and Nissan study are somewhat greater than those in the current study. It is not clear whether this is due to an increase in the disc height that might be present in vivo, or whether the difference might reflect increased disc degeneration in the older population used in the current study.

Parameter		Present Study 8 Male/ 4 Female	Gilad (1986) 157 Male	Berry (1987) 15 Male / 15 Female	Panjabi (1992) 8 Male/ 4 Female
Anterior Vertebral Body Height	L1	25.5 (2.7)	25.4 (2.2)	25.0 (2.9)	
	L2	27.3 (3.5)	27.2 (2.0)	27.9 (1.9)	
	L3	28.5 (2.8)	27.9 (2.1)	27.4 (1.7)	
	L4	27.4 (3.7)	27.4 (2.2)	26.7 (1.5)	
	L5	30.9 (2.4)	28.3 (2.1)	28.7 (1.9)	
Posterior Vertebral Height	L1	28.5 (3.0)	27.1 (2.1)	25.8 (2.1)	23.8 (1.0)
	L2	29.3 (3.4)	28.0 (2.1)	25.2 (2.2)	24.3 (1.0)
	L3	28.8 (2.3)	27.9 (2.1)	26.0 (1.6)	23.8 (1.1)
	L4	27.5 (2.9)	27.1 (2.3)	26.4 (1.7)	24.1 (1.1)
	L5	25.7 (1.9)	25.7 (2.5)	23.1 (1.5)	22.9 (1.0)
Superior Endplate Width	L1	46.3 (5.1)		45.2 (4.6)	41.2 (1.0)
	L2	49.2 (7.0)		47.7 (4.7)	42.6 (0.7)
	L3	50.1 (6.3)		49.6 (3.2)	44.1 (0.9)
	L4	51.5 (7.4)		51.2 (5.6)	46.6 (1.2)
	L5	53.8 (6.7)		53.4 (4.4)	47.3 (1.2)
Vertebral Body Waist	L1	38.9 (4.7)		39.5 (3.8)	
	L2	39.6 (4.1)		44.8 (3.1)	
	L3	41.4 (4.9)		42.3 (3.5)	
	L4	42.8 (3.8)		40.8 (3.2)	
	L5	48.4 (5.7)		46.1 (4.5)	
Inferior Endplate Width	L1	50.2 (6.6)		49.1 (3.7)	43.3 (0.8)
	L2	50.2 (6.1)		54.8 (4.8)	45.5 (1.1)
	L3	53.8 (5.8)		53.8 (3.7)	48.0 (1.2)
	L4	53.2 (7.6)		50.9 (4.6)	49.5 (1.4)
	L5	57.3 (7.2)		52.7 (4.3)	49.4 (1.4)
Superior Endplate Depth	L1	35.1 (4.2)	33.5 (2.8)	31.9 (3.7)	34.1 (1.3)
	L2	36.4 (3.2)	34.4 (2.9)	33.3 (3.7)	34.6 (1.1)
	L3	37.8 (4.7)	34.7 (2.7)	33.9 (3.3)	35.2 (1.1)
	L4	36.5 (2.4)	34.3 (2.7)	34.9 (3.4)	35.5 (0.9)
	L5	36.8 (3.4)	34.2 (2.7)	35.1 (2.8)	34.7 (1.2)
Inferior Endplate Depth	L1	36.1 (3.4)	34.1 (2.9)	32.3 (3.5)	35.3 (1.3)
	L2	37.5 (6.4)	34.7 (3.0)	33.4 (3.4)	34.9 (0.7)
	L3	37.9 (3.1)	34.6 (2.8)	34.2 (3.3)	34.8 (1.2)
	L4	37.8 (5.8)	34.9 (2.8)	35.6 (3.1)	33.9 (0.9)
	L5	36.4 (3.2)	33.9 (2.7)	34.5 (3.0)	33.2 (0.9)
Anterior Disc Height	L1-L2	7.6 (2.2)	7.6 (1.4)		
	L3-L4	8.7 (1.3)	10.3(1.8)		
	L5-S1	11.9 (3.0)	14.1 (2.2)		
Posterior Disc Height	L1-L2	4.2 (1.4)	6.5 (1.7)		
	L3-L4	4.4 (1.8)	7.2 (1.8)		
	L5-S1	5.0 (2.1)	7.5 (1.6)		

**Table 5.14:** Comparison of average values of selected vertebral body dimensions from this study and previous studies. Numbers in parentheses represent standard deviations.

A word needs to be said regarding comparisons of measurements of endplate depth which appear to be consistently larger (by 2-4 millimeters) in this study compared to those seen in other studies. In some cases, this difference is due to a discrepancy in the definition of this dimension, which results from differences in measurement technique. The use of sagittal radiographs in the measurement of endplate depth in this study required that this dimension be defined as the anterior part of the vertebral rim to the most posterior projection of the posterior rim (see Figure 5.2). If one uses calipers to measure endplate depth, as done in Berry et al. (1987), one can take measure the measurement in the mid-sagittal dimension of the endplate, where the measurement will be smaller due to the indentation of the kidney-shaped endplate. It would be expected that the endplate depth would be a couple of millimeters greater using the former definition as opposed to the latter, and indeed, such a difference is reflected in Table 5.20. The differences between the Gilad and Nissan (1986) values and the current may data stem from the fact that these authors apparently measured this dimension using selected points on the vertebral body itself, and not the full edge-to-edge dimension of the endplate. Panjabi et al. (1992) apparently use a similar definition of endplate depth as that used in the current study (though this is not absolutely clear from their article); thus, the differences in the size of the endplate depth dimensions may be due to the apparently smaller overall dimensions of their sample, as reflected in other measurements derived from this study.

## 5.5 Conclusions

Based on the results of the analyses reported in this chapter, the following conclusions are drawn:

1. Bone mineral content was found to have a protective influence with respect to fatigue failure of lumbosacral motion segments using Cox regression analysis. Results suggested protective influences for increases in motion segment volume and bone mineral density as well; however, bone mineral content proved a better overall predictor.
2. Given the protective influence of bone mineral content on fatigue failure, and the lower average bone mineral content associated with this older sample of motion segments, younger spines (having increased bone mineral content) would be expected to exhibit a somewhat increased resistance to fatigue failure.
3. Morphology measurements of the current lumbar motion segments indicate that the specimens are similar to data reported by other authors, but tend towards the higher end for many measurements.
4. Bone mineral content of the specimens is similar to other studies examining the content of older lumbar spines, and values represent an approximate 20% decrease in bone mineral content when compared with younger populations.

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## CHAPTER 6

### THE ROLE OF TORSO FLEXION IN THE FAILURE OF LUMBOSACRAL TISSUES

#### 6.1. Introduction

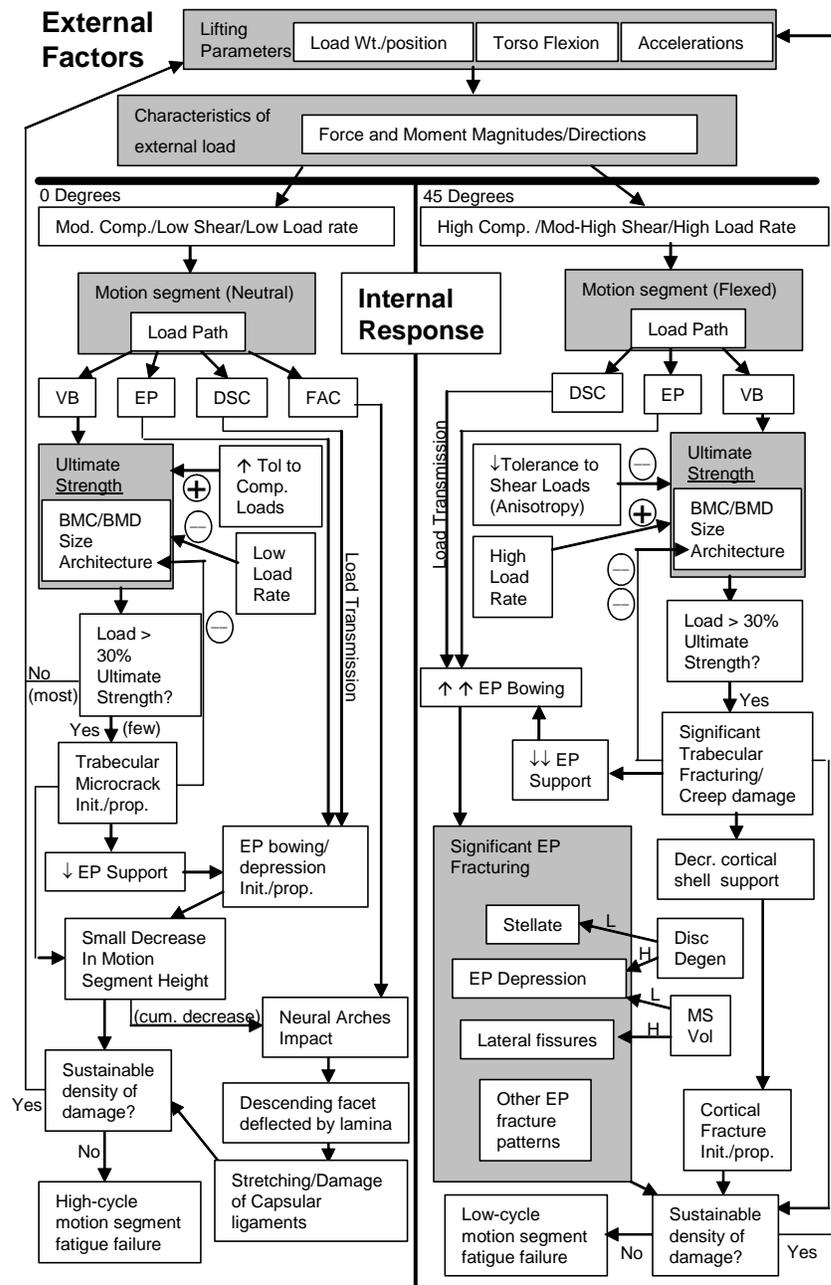
In evolutionary terms, the attainment of a bipedal posture has reaped substantial rewards for *homo sapiens*, particularly in the promotion of manual dexterity and the ability to manipulate objects and tools (Evans 1992). In the context of the present work, it has also fostered the human's ability to move items from place to place with relative ease, especially compared to species that rely on a quadrupedal gait as a means of transportation (Evans 1992). However, it is thought that the frequent lifting and carrying of heavy burdens is a relatively recent phenomenon, and may have been instituted primarily in order to transport back to the home setting food obtained at a considerable distance (Davis 1968).

Despite the benefits derived from the erect posture, there appear to be significant costs, one of which is the high loads experienced by lumbosacral tissues during lifting tasks using bipedal support. These loads are often magnified by the

need to bend the torso forward when reaching to retrieve an object, creating a large moment that must be counteracted by muscle and ligament forces, which may reach several thousand Newtons due to the small moment arm available to these tissues (McGill and Norman 1987). However, as will be discussed in greater detail below, flexion of the torso has a wide-ranging impact that affects not only the external loading experienced by the spine, but also the mechanisms used to counteract the imposed moment and the ability to withstand the strain imposed upon spinal and paraspinal structures. The conceptual model presented in this chapter will be used to describe the numerous effects of torso flexion on loading and stress response of tissues comprising the lumbar spine.

## **6.2 Overview of the Model**

Figure 6.1 illustrates a conceptual model describing pathways to the injury of spinal tissues when exposed to loadings associated with lifting in various angles of torso flexion. As can be seen in this model, external factors associated with a lifting a load in various torso flexion angles will lead to changes in the internal responses of (and failure patterns observed in) tissues of the lumbosacral spine. The model highlights theorized pathways to fatigue failure of spinal tissues at the two extremes of torso flexion simulated in this experiment: the neutral posture (0 degrees torso flexion) and the fully flexed posture (45 degrees torso flexion). As will be seen in the discussion that follows, the differences in the internal responses between these two postures are quite different, and will lead to remarkably different outcomes as a result. The balance of this chapter will be devoted to an examination of this model and a discussion of the available evidence regarding the proposed failure pathways.



**Figure 6.1:** A conceptual model of the effects of lifting in torso flexion on damage to spinal structures (Note: VB=vertebral body, EP=endplate, DSC=intervertebral disc, FAC=facet joint, BMC=bone mineral content, BMD=bone mineral density, MS Vol.=motion segment volume,  $\ominus$  = decrease,  $\oplus$  = increase).

### **6.3 External Factors**

The degree of loading ultimately experienced by the lumbosacral spine in a lifting task is a direct result of several external factors. As seen in the box at the top of Figure 6.1, several parameters associated with the lifting task (such as the load weight and its position, the degree of torso flexion, and the accelerations applied to the torso and load) are critical determinants of the magnitude of the external load placed on the spine. The following paragraphs consider the influence of these three factors on the forces and moments applied to the spine in a lifting task.

Both the weight of the load lifted and its position relative to the spine are important influences in the determination of the external load experienced by the spine in sagittal plane lifting tasks. In fact, the product of these two factors (the “load moment”) appears to represent one of the best measures of low back disorder risk for workers (Marras et al. 1999). In a neutral posture, where the trunk center of mass is closely aligned with the lumbar spine (Chaffin et al. 1991), the load moment will be the primary determinant of the forward bending moment imposed on the spine. However, increasing torso flexion in a bipedal posture will always increase the horizontal distance of the center of mass of the torso to the lumbosacral joint of interest and will add to the forward bending moment experienced by the lumbar spine. It should be recognized; however, that torso flexion may increase, decrease, or have no impact on the horizontal distance to the load to the lumbar spine. For example, torso flexion may decrease the horizontal distance of the external load center of mass compared to a squat lifting technique, especially where the load cannot be lifted

between the knees (Chaffin and Andersson 1991). The masses of the upper body and the external load are always subject to accelerative forces, which at a minimum will be restricted to gravitational acceleration. However, torso flexion is commonly associated with lifting tasks where objects are lifted from the ground or some other low level where both the masses of the upper body and load must be accelerated to pick up and move an object to some higher level. Dynamic lifting has been shown to significantly increase the forces acting on the lumbar spine according to several authors (Dolan et al. 1994, Granata and Marras 1995).

The discussion above should make clear that torso flexion can have a significant impact on several variables that help determine the external loading imposed upon the lumbosacral spine in a lifting task. Specifically, torso flexion may affect the acceleration of the two masses involved in the lifting task and the horizontal distance to the centers of mass of the object and upper body. These factors are fundamental in establishing the magnitude and direction of the forces and moments acting on the lumbar spine, which in turn set the stage for the internal loading response and ultimately the stress experienced by spinal structures. The overall effect of torso flexion on spine loading in lifting tasks is often quite large, with a doubling or tripling of the load on the lumbar spine as torso flexion is performed (Nachemson 1992, Schultz et al. 1982).

## **6.4 Internal responses to lifting in neutral posture (0 degrees torso flexion)**

### **6.4.1 Magnitude of the loads**

The external loading factors associated with lifting a moderate weight in a neutral torso posture lead to internal responses that are estimated to place spinal tissues under a moderate compressive load and low shear forces at a relatively low load rate. Measurements of intradiscal pressure reported by Schultz et al. (1982) indicated that subjects standing and holding an 8 kg weight close to the chest (a load similar to the 9 kg weight modeled in the current experiment) experienced an average intradiscal pressure of 550 kPa. This may be compared to intradiscal measurements obtained when holding the same load in a 30 degree torso flexion with arms partially outstretched, resulting in a 1620 kPa pressure measurement. These and other direct measurements of pressures in the spine demonstrate that load handling in the upright posture places a relatively moderate burden on the lumbar spine, especially when compared to flexed torso postures (Nachemson 1992).

Biomechanical models of the lumbar spine have estimated the compressive and shear loads on the lumbar spine in neutral postures, and results consistently predict moderate compressive load and low shear loads when lifting in upright postures (Bogduk et al. 1992, Marras et al. 2001, Schultz et al. 1982). Bogduk et al. (1992), using a detailed examination of the fascicles of the lumbar extensor muscles, found that in the upright postures, shear forces developed by the lumbar musculature are directed in an anterior direction at the lower levels, with a reversal to the posterior direction at the upper levels, a finding mirrored by estimates of shear forces at various

lumbar levels when lifting a moderate load in the upright posture in the present lifting analysis.

### **6.4.2 Load Path**

As illustrated on the left-hand side of Figure 6.1, the moderate load experienced when lifting a 9 kg weight in the neutral torso flexion posture is distributed across the interbody joint (vertebral bodies and discs), which takes on the majority of the load, and the zygapophysial joints, which bear the minority (Bogduk 1997, White and Panjabi 1978). While there is some controversy on this point (Adams and Hutton 1983), some authors have indicated that the posterior elements can bear approximately 30-40% in normal motion segments (Lorenz et al. 1983, Ashton-Miller et al. 1983). Other research has suggested a slightly lower proportion on load borne by the facet joints, with minor variations between levels of the lumbar spine, with higher levels bearing a lower proportion of the load through the facets (11% on average), while lower level may transmit approximately 20% in this fashion (Adams and Hutton 1980). Load bearing via the zygapophysial joints, however, requires one of two specific loading scenarios. The first of these involves posterior rotation of the motion segment, which will cause the articular facets to make contact, allowing a load to be transmitted through the facets (Dunlop et al. 1984). The other scenario, more relevant to the current discussion, involves prolonged or repeated compression of the motion segment, which will cause the inferior articular processes to contact the superior surface of the lamina of the inferior vertebra (El-Bohy et al. 1989), allowing forces to transmit along this route. If the motion segment experiences severe or

repeated compression, the superior facet may be deflected backwards by the lamina, causing stretching and/or damage to the joint capsule (Adams et al. 2000). While in normal circumstances the load transferred via the facets appears to be fairly low, under conditions of severe disc degeneration it has been shown that the posterior elements may bear as much as 70% of the load burden (Adams and Hutton 1983).

### **6.4.3 Structures under load in the neutral posture**

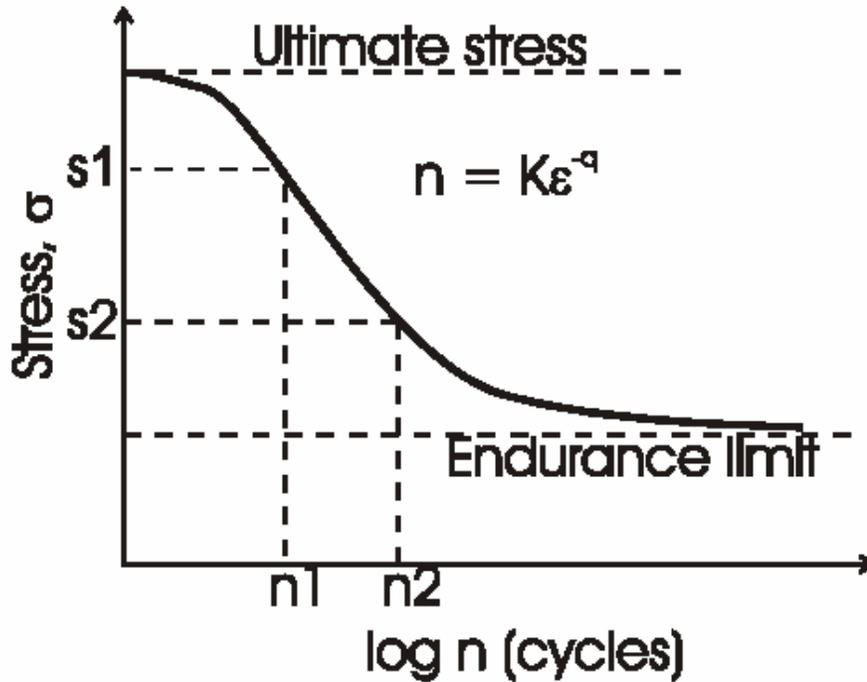
#### **6.4.3.1 The Vertebral Body and Endplate**

It is apparent that when axial compressive loads are applied to normal motion segments, the trabeculae of the vertebral body (VB in Figure 6.1) located under the central region of the disc will fracture and then the endplates (EP) themselves will be the first structures to fail (Brinckmann and Horst 1985, Brown et al. 1957, Perey 1957). The health of the endplate is highly dependent on the strength of the underlying trabecular arcades (Bogduk 1997). When a load is borne by the inter-body joint, the cartilage endplate will begin to bow away from the disc (Adams and Dolan 1995, Hickey and Hukins 1980). This is due to the fact that the cortical bone of the vertebral body strongly supports the periphery of the intervertebral disc; however, the central portion is supported by the somewhat weaker trabecular bone (Bogduk 1997).

The structural integrity and ultimate strength of the vertebral body is largely dependent on three major factors, as indicated in Figure 6.1. These factors include the bone mineral content or density, the size of the bone, and the integrity of the bone architecture (Hayes 1986). Data from the current study reaffirmed the significance of the former two factors, as increases in both factors were found to be protective in

terms of the number of cycles to failure of the motion segment. However, as indicated in Figure 6.1, the strength of the bone is also highly dependent on how closely the load vector is orientated to the longitudinal axis of the vertebral column. Bone is an anisotropic material, meaning it has different mechanical properties dependent on the direction from which loads are applied (Litsky and Spector 1999). Specifically, bone is much stronger in compression than tension, and is weakest in shear (Martin et al. 1998). In the neutral posture condition under consideration the bone will be aligned in a fashion that is quite close to its optimal load bearing capacity. The ultimate strength of the motion segment is also influenced by the load rate. The load rate in the neutral posture is modest and would result in a lower ultimate strength than faster load rates (Hawkins 2001). Nonetheless, the relatively modest compressive load and the favorable alignment of the vertebrae with respect to the imposed load appear to provide a situation in which the strength of the vast majority of motion segments is sufficient to withstand many thousands of loading cycles.

Figure 6.2 illustrates the relationship between injury threshold based upon the magnitude of the load and the number of cycle applications (Martin et al. 1998). As can be seen from this figure, when undergoing loads close to the ultimate stress, few cycles can be tolerated without injury. Decreasing loads allow more cycles to be withstood, and at some level of stress the load applications can be tolerated indefinitely (usually about 30% of the ultimate strength of a specimen, no matter what manner of loading is used). In the current experiment, the loads experienced in the neutral posture appeared to be lower than the endurance limit for the majority of motion segments. Eight of twelve of the specimens assigned to this loading condition



**Figure 6.2:** Fatigue failure relationship with injury threshold dependent on magnitude of the load and the number of loading cycles (Martin et al. 1998)

lasted the maximum of 10,020 cycles. Thus, even in this elderly and osteopenic cohort of lumbosacral spines, two-thirds of the motion segments easily tolerated thousands of cycles of loading. Of the four specimens failing by fatigue failure in the neutral posture, three were females that have been found to exhibit an average compressive strength of 4 kN (Jager and Luttmann 1991). The 1300 N load experienced in the neutral postures very close to the endurance limit value (30% of ultimate strength), and it would not be surprising to find that some of the elderly

segments would be somewhat weaker than the value stated above, leaving them subject to fatigue failure at high cycles of repetitive loading.

While most specimens tolerated loading in the neutral posture quite well, a few experienced high-cycle fatigue failure under this loading condition. As illustrated in Figure 6.1, it is thought that this high-cycle fatigue failure is the result of the accumulation of small cracks in the trabeculae that propagated slowly with repeated loading. When considering the relative size of the cracks manifest in the trabeculae, it is important to realize that the size of the fracture is related to the amount of energy stored by the bone in the loading cycle (Beardsley et al. 2002, Martin et al. 1998). If the energy absorbed by bone in a loading cycle exceeds its energy absorbing capacity, fracture will result and stored energy will be released in the process. The load rate plays an important role in the determination of the energy absorbed by bone. At a relatively low loading rate, energy can dissipate through formation of a single small crack. At high loading rates, the greater energy stored cannot be dissipated rapidly enough through a single crack and extensive fracturing and comminution of the bone tissue will result (Beardsley et al. 2002).

Trabecular bone is composed of a network of vertical struts and horizontal crossbeams (Bogduk 1997). It appears as though fracture in trabeculae loaded via compression is initiated by a shear fracture in the transverse crossbeams (Yeni et al. 2003). Such a fracture will originate in an area where localized stresses exceed the local strength (Hayes 1986). Fracture occurrence, as specified in Figure 6.1, will adversely impact the architectural integrity of the bone and will act to degrade the ultimate strength of the bone in future loading cycles. Furthermore, the loss of support

experienced when a transverse crossbeam is severed by fracture would be expected to result in increased stress concentrations in and around the region where the failure occurred. This creates a scenario where repeated applications of a load will cause overload of remaining structures in the region, which having a decreased capacity to absorb the imposed energy, will result in additional fracture propagation in the region (Martin et al. 1998). As fractures in this and other local regions accumulate, a gradual decrease in the height of the vertebral body will be experienced (Brinckmann et al. 1988), as represented in Figure 6.1.

The accumulation of fractures, usually concentrated in the central region of the vertebral body (underneath the disc nucleus), will decrease the ability of the trabecular network to support the bowing of the endplate which occurs in response to a compressive load (Figure 6.1). The bowing of the endplate is a result of the transmission of loads via the disc to the endplates (Bogduk 1997). When the spine experiences a compressive load, the tendency is for the nucleus of the disc to want to expand radially, an expansion resisted by the anulus fibrosus. The load is thereby transmitted to the endplates by the disc (Figure 6.1), which will bow away from the nucleus, proportionate to the load applied (Adams and Dolan 1995, Hickey and Hukins 1980, Roaf 1960). Degradation of the underlying trabeculae will permit further bowing of the endplate to occur (Brinckmann et al. 1983). In the current experiment, it should be noted that the predominant damage mode observed in the endplates in fatigue failure in the neutral torso posture was an increased depression of the endplate (4 occurrences) – only one endplate experienced fracture under these conditions. It is also worth noting that no intervertebral discs were found to exhibit

failure during the fatigue-testing portion of the experiment. For this reason, the role of the discs are represented in Figure 6.1 as a vehicle for load transmission from one vertebral body to the next, and discs otherwise do not figure in the discussion of failure mechanisms.

Motion segments experiencing trabecular fractures under moderate compressive loading in the neutral trunk posture will experience small cumulative decreases in height (Brinckmann et al. 1988). Due to the gradual increments in the development of trabecular damage, motion segments subject to the loads associated with lifting in a neutral posture would be expected to sustain numerous loading cycles. The current experiment confirms that this is the case. As noted previously, two-thirds of motion segments loaded under these conditions never reached the failure criterion, even after 10020 loading cycles. Of the four specimens that did fail, the average number of cycles it took to achieve the failure criterion was 4760 cycles, and the least number of cycles to failure for any specimen exposed to the neutral position condition was 2809 cycles. Thus, it is apparent that while failure of motion segments is possible when exposed to moderate compression/low shear conditions (at least in an older cohort of specimens), in general, the spine seems quite capable of tolerating such loads for extended periods of cyclic loading.

#### **6.4.3.2 Zygapophysial joints**

It is apparent from the current data set that the zygapophysial (facet) joints (FAC in Figure 6.1) were at increased risk of damage in the neutral posture compared to more flexed motion segments. It is apparent these joints were involved, or with decreasing motion segment height became involved, in the load-bearing process in the

neutral posture. Previous research has indicated that under a constant and severe compressive load, the motion segment can be compressed enough that the descending facets from the superior vertebra make contact with the lamina below (El-Bohy et al. 1989). This process may be exaggerated if the motion segment is compressed and tilted backwards (Adams and Hutton 1980, Dunlop et al. 1984, El-Bohy et al. 1989, Yang and King 1984). If sufficiently compressed, the lamina will deflect the inferior articular process backwards, causing stretching and damage of the capsular ligaments (Adams et al. 2000), and eventually avulsion of the joint (Bogduk 1997). All of the motion segments experiencing fatigue failure in the neutral posture in the present investigation experienced some degree of zygapophysial joint damage. Three of the four specimens exhibited complete disruption of the joint, while the other exhibited increase joint laxity. A previous study of fatigue failure of lumbar motion segments, where all motion segments were tested in a neutral posture, did not report damage to the facet joints (Brinckmann et al. 1988). It is possible that this may have been due to the fact that the focus of this investigation was to examine endplate fractures, and damage to the zygapophysial joints was not considered. However, the displacement failure criterion in the previous study was less than the current study and it may well be that the damage to the zygapophysial joints was either more apparent with the increased displacement criterion used presently, or perhaps did not occur until displacement was beyond the criterion used by Brinckmann et al. (1988).

Data from this study suggests that most motion segments (even elderly) were able to withstand the modest compressive and shear loads associated with lifting 9 kg in a neutral posture. However, in the minority of segments where failure occurred, it

appears that all load-bearing structures (with the exception of the intervertebral disc) were at risk of damage. The development of damage was very incremental in nature, and it required a high number of loading cycles to achieve the failure criterion for loads associate with lifting in a neutral posture.

## **6.5 Failure of motion segments when lifting in flexion**

Torso flexion has important ramifications regarding the magnitude and characteristics of the load experienced by the lumbosacral spine and the manners in which those loads are distributed, as represented on the right side column on Figure 6.1. Even when lifting the same load, measurements of intradiscal pressure measurements indicate that the load on the lumbar spine is double or triple greater than when the same load is handled in the neutral posture (Nachemson 1992, Schultz et al. 1982). Biomechanical models also demonstrate that the greater trunk moments experienced with lifting in torso flexion result in higher compression and shear forces on the spine (Drury et al. 1989, Potvin et al. 1991). Furthermore, lifting in a flexed position will result in the spinal tissues being loaded at a much higher load rate. Tissues of the spine are all sensitive to the rate of loading, but especially bone (Hawkins 2001), which is one of the tissues at most risk of injury. Thus, if one examines Figure 6.1, the simple matter of changing the angle of torso flexion when lifting a load will have a substantial impact on the characteristics of the load applied to the lumbosacral spine. The result of the increased external load on the lumbar spine in

flexion is to expose the lumbar spine to significant internal compression and shear forces (Bogduk 1997, Drury et al. 1989, Potvin et al. 1991).

### **6.5.1 Load path**

The flexion of individual motion segments in the process of spinal flexion is also believed to affect the path that loads follow along the spinal column (Bogduk 1997, White and Panjabi 1978). Compared to the neutral posture, where the zygapophysial joints bear a portion of the spinal load (Adams and Hutton 1983), when the motion segment undergoes a flexion movement, the facets disengage and reduce or eliminate this load-bearing pathway. As a result, not only is the overall load magnitude transmitted through the spinal column much greater when lifting in a flexed posture, the increased load becomes increasingly concentrated in the portion of the column consisting of the vertebral bodies and discs as the spine flexes (White and Panjabi 1978), as indicated in Figure 6.1.

### **6.5.2 Ultimate Strength of Motion Segments in flexion**

While certain attributes contributing to the ultimate strength of motion segments are equivalent in neutral and flexed posture (i.e., the internal motion segment characteristics of bone mineral content/density, the size, and the architecture of the bone), the ultimate strength of the motions segment will additionally be modified by the change in the direction of loading relative to the motion segment and the increased load rate experienced in the flexed posture, as compared to neutral. As mentioned

previously, bone is an anisotropic material and is only about one-third as strong in resisting shear forces as compared to a longitudinally-imposed compressive force. Thus, not only are the larger compressive forces an issues in terms of fatigue failure, the increased shear forces imposed on the flexed motion segment in torso flexion will reduce the ultimate load that the motion segment can withstand.

The ultimate strength of the vertebral bone will also be modified by much higher load rate experienced when lifting in a flexed posture (Hawkins 2001). Higher load rates will increase the ultimate strength of the motion segment (Yingling et al. 1998); however, the increase in ultimate strength due to this factor (approximately 20-40%) may be more than offset by the reduced strength associated with increased shear exposure (approximately 65%), as discussed above (Martin et al. 1998). In addition, the higher load rates experienced in torso flexion will decrease the amount of strain experienced before the bone fails (Martin et al. 1998).

### **6.5.3 Imposed loads and the fatigue endurance limit**

In a paper mentioned previously, Jager and Luttmann (1991) indicate that the compressive strength of lumbar motion segments indicated that the average compressive strength of male lumbar motion segments averages approximately 6 kN while females average approximately 4 kN. Thus, the compressive loads experienced by the lumbar segments in the flexed posture (averaging around 3 kN) would already be expected to be well above the threshold for the development of fatigue failure. However, the exposure to high levels of shear force, especially at the upper levels of the spine also appears to be at a high proportion of ultimate stress values for shear,

estimate to be between 1800-2800 N (McGill 2003). Thus, it would be expected given these loading conditions that fatigue failure would occur among all but the very strongest specimens, those capable of withstanding 10 kN or greater. Given the age of the spines used in this study, the scenario just mentioned would not be anticipated. In fact, in Figure 6.1, no option is provided for escaping the imposed stress undamaged, which is congruent with the experimental results obtained in this experiment.

#### **6.5.4 Development of fatigue failure in the fully flexed spine**

The extent to which the loads experienced in torso flexion exceed the endurance limit for fatigue failure and the increased energy absorbed by the bone as a result of the higher load rate experienced in this condition indicate that the fracturing and crack propagation in the trabeculae will be much more extensive than those observed at 0 degrees torso flexion (Martin et al. 1998). Furthermore, creep damage would be expected to be quite significant given the degree of loading experienced under these conditions (Tsai et al. 1998). The result will be a rapid acceleration of damage processes in the lumbosacral spine.

Both the increased compressive load and the increased shear forces imposed on the trabecular network develop conditions under which the struts will experience significant shear (due to compressive forces) and perpendicular fractures (from shear loads) leading to a rapid breakdown in the internal architecture of the bone, as indicated in Figure 6.1 (Martin et al. 1998). The significant derangement of the trabecular structure from these more widespread fractures would result in a more

substantial reduction in the ultimate strength of the motion segment per loading cycle than that observed in the neutral posture, and will act to hasten fatigue failure. In addition to the fracturing that occurs with the higher spinal loads, an increase in damage of the vertebral bone resulting from creep will also be manifest (Martin et al. 198), leading to additional weakening of the affected vertebra.

The more rapid deterioration of the trabecular network under the loading conditions experienced in flexed lifting would be expected to quickly and substantially undermine the support of the endplate, especially when compared to the localized microcracks experienced in a neutral posture. In the current experiment, the latter condition resulted in overt endplate fracture on only one occasion, with endplate depression being the predominant finding. However, the increased derangement of the endplate supporting structure in the flexion posture was associated with the development of a variety of endplate fracture patterns. It is hypothesized that the reduced structural support will lead to an increased bowing of the endplate to the extent that the strain experienced by cartilage exceeds its strain capacity, leading to development of fractures.

Analysis of endplate fractures using the categorization scheme employed in the current experiment suggests that endplate fracture patterns are dependent on certain attributes of the motion segment, as indicated in Figure 6.1. For example, the occurrence of stellate endplate fractures in fatigue testing was associated both with increased shear forces and a lower disc degeneration grade. It may be that the increased shear forces lead to a derangement of the trabeculae underlying the endplate, and sets up a condition where a less degenerated disc exerts a centralized pressure on

the endplate leading to the characteristic endplate fracture pattern of fissures radiating from the central region of the endplate. Finite element analysis studies have suggested that degenerated discs result in a distribution of loading that is more peripheral – with more load concentrated near the cortical shell of the vertebra (Homminga et al. 2001). It seems reasonable to speculate that less degenerated discs, with a more centralized load distribution may exert the necessary pressure in the central region of the endplate to result in this fracture pattern.

The depression endplate damage classification was also associated with certain loading and motion segment attributes. Specifically, this pattern of damage was associated with a lower degree of shear force and a decrease in the estimated volume of the motion segment. The lower shear forces would be expected to be associated with decreased damage of the underlying trabeculae and thus a higher degree of support for the endplate. This may lead to a situation where the endplate experiences a certain degree of deformation, but is still sufficiently supported to resist endplate fracture. The reason that smaller motion segments may be associated with endplate depressions may stem from the fact that larger endplates have a larger central region supported by weaker trabecular bone, which may allow greater deflection and fracturing, as opposed to smaller specimens that have a smaller proportion of the endplate supported by the weaker trabecular bone.

Lateral fissures in the endplate were associated with increased volume of the motion segment. This fracture pattern is indicative of a lateral stress concentration across the face of the endplate. One may speculate that the reason such fractures may be evidenced in larger specimens is that the proportion of load along the lateral aspect

of the endplate that is supported by the cortical bone is reduced in larger specimens. The greater reduction in cortical support in the lateral axis may lead to a structural weakness in the lateral dimension of the endplate, resulting in a fracture tending to run along the endplate's lateral axis.

Other fracture patterns observed in the experiment did not show specific relations with load of motion segment characteristics. It may be that the relatively limited number of fracture occurrences for many of these patterns inhibited the identification of factors that may be influential to their development.

The rapid accumulation of creep damage and trabecular fractures associated with loads experienced in the flexed posture not only will reduce the ability of the trabeculae to support the endplate, the extent of damage may be sufficient to expose the cortical shell to significant damage as well (Bogduk 1997), as indicated in Figure 6.1. Deprived of the full measure of reinforcement provided by the vertebral spongiosa, the cortical shell will rapidly lose its ability to sustain a significant compressive or shear load (Bogduk 1997), and will experience fracture initiation in areas of localized stress (Hayes 1986). These fractures will continue to propagate as future loading cycles are experienced.

It should be noted that the zygapophysial joints were not found to sustain damage in the fully flexed posture. It appears that flexion of the motion segment associated with this posture prevented the inferior articular facet of the superior vertebra from making contact with the lamina of the inferior vertebra. As a result, even significant compressive loading of the motion segments did not result in damage to the posterior joints.

The relative safety of the zygapophysial joints, however, appears to represent a rather insignificant triumph when one considers the rapidly accumulating damage in the load bearing inter-body joints when lifting in full flexion. The damage to the trabecular bone, the endplates and the cortical bone under these conditions resulted in damage that could be sustained for a relatively low number of loading cycles. In fact, under these loading conditions only a few specimens lasted over a hundred loading cycles. Brinckmann et al. (1988) found that at loads of about 50-60% of predicted ultimate compressive strength, approximately 40% of specimens failed before 100 cycles had elapsed. Specimens in the current study, although tested at under different conditions, showed a similarly short fatigue life. It seems clear that damage associated with exposure to the significant loads experienced in lifting in torso flexion can only be sustained for a limited number of loading cycles, as illustrated in the right-hand column of Figure 6.1.

## **6.6 Discussion**

The model presented in Figure 6.1 describes the internal processes and mechanisms of failure expected to occur at the two extremes of spinal loading examined in the current experiment. This was done for the sake of simplicity and for ease of presentation. However, it should be recognized that between these two extremes there exists a continuum of torso flexion postures and associated loading conditions that will be experienced by the lumbar motion segments. Results obtained in tests at the midpoint of this continuum (22.5 degrees torso flexion), but not included

in Figure 6.1, suggest that the midpoint does show many elements representative of a transition between the two extremes. Clearly the number of cycles to failure in the partially flexed posture (averaging 3257 cycles) was intermediate between the 0 degree condition (8253 cycles) and the 45 degree torso flexion condition (263 cycles). In fact, the high variability observed in cycles to failure at the intermediate posture was the result of several specimens having sufficient strength to last the entire test period (similar to the 0 degree torso flexion condition), while others lasted only a limited number of cycles (similar to the 45 degree condition).

Failure modes also appeared to illustrate the transitional status of the mid-flexion condition. While in the 0 degree torso flexion condition, many motion segments that failed due to fatigue loading exhibited complete disruption of the zygapophysial joints, specimens in partial flexion tended to exhibit increased laxity (but not disruption) in these joints, while facet damage was a rare finding at all in full flexion. In fact, logistic regression analysis identified flexion angle as the key determinant in facet damage. Endplate damage patterns also seem to be reflected in this transition. In the neutral posture, endplates became depressed with fatigue loading, but only rarely showed an overt fracture. Fatigue failure in partial flexion also showed evidence of depressed endplates, but also exhibited endplate fractures in many cases. Motion segments assigned to the fully flexed condition almost always exhibited fracturing in the endplates.

It is quite remarkable how well these results correspond with the findings of a recent high-quality epidemiological study by Punnett et al. (1991). This case-control study examined relationships between back pain and occupational exposures in

workers in an auto assembly plant. One of the key exposures assessed in this study was time spent in non-neutral trunk postures. Results of this study disclosed a dose-response relationship with respect to the degree of torso flexion and the odds of experiencing low back pain. Specifically, Punnett et al. (1991) reported that mild (20 – 45°) trunk flexion resulted in an odds ratio for low back pain of 4.9, while severe (>45°) trunk flexion resulted in an odds ratio for low back pain of 5.7. The effect of torso flexion on fatigue life reported in the current investigation seem to mirror the increased risk of low back pain observed in the conduct of occupational tasks, and may present a rationale with which to explain the results of Punnett et al. (1991). That is, it is quite conceivable that work activities in non-neutral trunk postures among autoworkers resulted in fatigue failure of spinal structures, resulting in symptoms of low back pain. It is also interesting to note that this study showed a much greater association for torso flexion than for lifting (OR = 2.16) with respect to low back pain. Indeed, results of the current study suggest that lifting, in and of itself, is not necessarily that hazardous if one maintains a neutral trunk posture. Instead, results of this study indicate that the threat of injury to the low back becomes increasingly prominent as the torso continues to flex forward, a message concordant with that of Punnett et al. (1991).

Previous studies examining the fatigue failure of lumbosacral motion segments have clearly shown that repeated sub-maximal loadings can result in rapid damage (Brinckmann et al. 1988, Hansson et al. 1987), and that the rapidity appears to be associated with predictions of the percentage of ultimate stress that the motion segment experiences (Brinckmann et al. 1988, Hansson et al. 1987). However, it is

important to recognize that results of previous studies may not be readily applicable to the manner in which lumbosacral motion segments are loaded in various angles of torso flexion. Two critical considerations in this regard are the combination of compressive and shear loading experienced by the motion segment and the rate at which the tissues are loaded.

Bone is an anisotropic material, and is considerably weaker in shear loading than in compression (Hawkins 2001). In fact, the shear stress that can be tolerated by bone is less than 40% of the compressive stress that can be tolerated, on average. Shear damage appears to be concentrated in the cancellous bone; thus, such damage may be particularly relevant to the failure of vertebral endplates (Martin et al. 1998). Since greater shear forces are experienced by the lumbar spine in torso flexion, any understanding of the fatigue life of motion segments when lifting in torso flexion cannot be based on tests examining pure axial loading (e.g., Brinckmann et al. 1988, Hansson et al. 1987, Liu et al. 1983). Others have failed to specify the nature of compressive and shear loads applied to motion segments, or the rationale used to determine these loads (Adams et al. 1983b, 1985). As a result, previous fatigue life research is difficult to relate to the loads actually experienced when lifting in specific non-neutral trunk postures.

## **6.7 Summary**

*In vitro* studies are performed so that the chain of events leading to mechanical failure of the lumbar spine can be better understood (Brinckmann and Pope 1990).

Such experiments allow one to evaluate the effects on spinal structures due to varying loading modes and magnitudes and evaluating the components or structures that are susceptible to injury. Structures receiving the majority of the focus of this research include the vertebral bodies, cartilage endplates, zygapophysial joints, and the intervertebral disc. Damage to other structures, such as the ligaments is much more difficult to detect and quantify in fatigue failure studies (Brinckmann and Pope 1990). The analysis presented here is based on studies involving an elderly cohort of lumbar spines. As a result, further research will be required to evaluate whether the development and sites of fatigue failure are similar in older versus working age individuals. Younger spines would be expected to possess higher amounts of bone mineral content, identified as a protective factor in this experiment. It should be realized that even relatively small increases in bone mineral content produce significant gains in strength. As a result it would be anticipated that fatigue life would be increased in a younger sample of spines; however, loading paths associated with the various postures would be expected to be similar in both groups, thus the sites and mechanisms of failure would be anticipated to be similar.

The results of the current study suggest that torso flexion has significant effects on both the fatigue life of lumbosacral motion segments and on the nature of the damage that will occur when motion segments are exposed to spinal loads predicted to occur in various torso flexion angles. Lifting a moderate load in an upright torso posture resulted in a load that was lower than the endurance limit for fatigue failure for most of the motion segments tested. Weaker segments did exhibit failure, which in all cases required over 2800 cycles to achieve. Failure modes in the few segments

exhibiting failure in the neutral posture involved damage to the zygapophysial joints and depression (not usually involving fracture) of the endplates. The flexed posture resulted in rapid failure of almost all motion segments, with fatigue life averaging 263 cycles. The changed loading path resulted in damage being concentrated in the interbody joints. Different types of endplate damage were associated with the degree of shear loading present and on certain attributes of the motion segment (including disc degeneration grade and motion segment volume).

The model presented in this chapter illustrates the effects of torso flexion on fatigue life of lumbosacral motion segments. As reflected in the conceptual model, torso flexion has a very large effect due to posture in terms of fatigue life and sites of failure. Failure modes observed in this experiment bear similarities to histologic evidence of trabecular fractures in cadaveric specimens (Vernon-Roberts and Pirie 1973). Furthermore, results of fatigue life in various torso flexion postures bear a correspondence to epidemiologic studies of increasing low back disorder risk in postures involving increasingly flexed torso postures (Punnett et al. 1991). Based on results of the current study, the model presented would seem to deserve further evaluation and study.

*In vitro* tests, such as those performed in the current study, have inescapable contextual concerns. It is not known for example, to what extent minor damage observed in this investigation might actually be physiologically beneficial in terms of spurring bone remodeling (Brinckmann and Pope 1990). However, in spite of the limitations associated with this ambiguity, evidence clearly indicates that *in vitro* testing is a tool that has a great deal of utility in developing an enhanced understanding

of several aspects relating to the theorized mechanical pathways leading to the development of low back disorders (Adams 1995).

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## **CHAPTER 7**

### **CONCLUSIONS AND FUTURE RESEARCH**

#### **7.1 Conclusions**

The primary goal of this research was to determine the influence of torso flexion and lumbar level on fatigue failure of motion segments of the human lumbosacral spine. Secondary goals were to evaluate the role of bone mineral content and morphological characteristics of the specimens with respect to resisting the repetitive loads often experienced by the lumbar spine and sacrum. The major conclusions of this research are as follows:

1. A review of the literature involving mechanical testing of cadaver motion segment specimens uncovered several issues that inhibited the ability to relate the results of existing studies to loads experienced in occupational lifting tasks involving repetitive loading in various positions of torso flexion. These included:
  - a. Few studies have employed a fatigue failure model, despite the repetitive loading commonly experienced in occupational lifting.

- b. Motion segment orientations in existing studies have not reflected the positions observed in various trunk flexion angles *in vivo*.
  - c. Loads imposed on motion segments have not been based on physiologically realistic dynamic biomechanical models.
  - d. Differences in load rates experienced with lifting weights in different trunk flexion angles have not been accounted for in previous studies.
2. A study was performed in which spinal loads associated with lifting a 9 kg weight in three torso flexion angles (0, 22.5, and 45 degrees) were cyclically imposed on human lumbosacral motion segments. Results of this study indicated:
- a. The spinal loads experienced at different torso flexion angles had a dramatic impact on the number of cycles to failure of the motion segments. The average number of cycles to failure in 0 degrees trunk flexion was 8253, specimens at 22.5 degrees flexion averaged 3257 cycles to failure, and specimens at 45 degrees torso flexion lasted an average of only 263 cycles.
  - b. Regression models examining the relationship of compression, shear forces, load rate, and motion segment flexion angle indicated that load rate and shear forces were the best predictors of the number of cycles to failure.

- c. The level of the motion segments did not significantly influence the number of cycles to failure.
  - d. Tests of ultimate strength were performed on motion segments surviving the full 10,000 cycles of the fatigue tests. Similar to findings in prior studies, motion segments at lower lumbosacral levels were found to have higher ultimate strength than higher levels. Specimens tested at higher load rates tended to exhibit higher ultimate strength, a typical finding for viscoelastic materials.
  - d. Endplate fractures, vertebral body fractures and damage to the zygapophysial joints were the most common failure modes observed in fatigue testing. Intervertebral discs were not overtly damaged in fatigue testing; however, annular protrusions were observed in two tests involving ultimate strength.
3. The influence of motion segment morphology and bone mineral content on fatigue failure were examined using survival analysis techniques. These analyses demonstrated the following:
- a. Increasing size (volume) of the motion segment and increasing bone mineral content were associated with protective influences with respect to fatigue failure in separate survival analyses using different data sets.

- b. Bone mineral content of the current specimens represents typical values for older individuals and is approximately 20% less than that of a young adult population. Given the protective influence of increasing bone mineral content, it would be expected that a younger cohort of spines would be somewhat more resistant to the effects of cyclic loading.
- c. Morphology of the vertebrae and discs of the present sample are similar to those reported in other studies, though they tend towards the high end of reported mean values for several measurements.

## **7.2 Future Research**

1. A clear and immediate need is to replicate this experiment with a younger cohort of cadaver spines. There are reasons to believe that the response of younger spines to cyclic loading would differ from the responses of an older cohort. One obvious difference is the loss of bone mineral content in older specimens. Since analyses from this study indicate a protective influence for higher bone mineral content values, it would be expected that younger specimens would be more resistant to fatigue failure. However, the nature of this response needs to be quantified in specimens more typical of the age range of manual material handlers, who are the intended beneficiaries of this research. Given the dramatic impact of torso flexion on fatigue failure, it is

anticipated that large differences in cycles to failure would also be observed in younger specimens; however, it is important to quantify the response in order to develop appropriate lifting recommendations for workers involved with manual lifting tasks.

2. This study examined fatigue failure of motion segments associated with lifting a relatively moderate load (9 kg) in different torso flexion postures. However, there is a wide range of loads (both lighter and heavier) that are repetitively handled by workers in occupational tasks every day. As a result, it is important to understand how varying load weights would influence the fatigue failure response of lumbar motion segments. Furthermore, it would be useful to quantify the effects of different cycle times for comparison with the 3 second cycle used in the current study.
3. A maximum of 10000 cycles was imposed on motion segments in the present experiment. Since many of the motion segments lasted the entire number of cycles imposed, the differences in the number of cycles to failure between the experimental conditions are undoubtedly larger than presented here. Since repair mechanisms for bone are quite long in duration (on the order of weeks to months as opposed to days), it would be useful to observe fatigue responses of lumbosacral motion segments for greater numbers of cycles, especially in the 0 and 22.5 degrees of torso flexion conditions.

4. Compression and shear loads at the upper levels of the lumbar spine were assumed to be of an equivalent magnitude and direction of those experienced at the lumbosacral junction, even though the magnitude of loading would be expected to be somewhat lower at higher spinal levels. Furthermore, it is not clear that the direction of the resultant vector would necessarily be pointed in the same direction at upper as compared to lower levels. Unfortunately, these simplifying assumptions were necessary due to a dearth of information about the nature of loading at upper lumbar levels in various torso flexion angles. Additional research is needed to improve model estimates of the loads on the upper levels of the lumbar spine.
5. The purpose of this study was to determine the overall effects of torso flexion on fatigue failure of lumbar motion segments in a physiologically realistic manner. However, torso flexion affects many variables, including load magnitude, mix of compression and shear forces experienced, the rate of load application, and orientation of the vertebral column (motion segments). Since all of these factors had to be simultaneously changed to achieve physiologic fidelity, it is not clear from the current experiment the extent to which each of these individual variables influenced the overall results. Future research should be performed to examine the effects of each of these variables individually, their relative contributions and possible interactions,

so that the numerous effects associated with torso flexion can be more fully understood.

6. The dramatic effect of torso flexion angle on fatigue failure of the lumbosacral spine suggests that intervention studies based upon these research results be developed. It is quite possible that lifting recommendations based on a fatigue failure criterion based on torso flexion angles observed during lifting tasks may reduce the likelihood of experiencing low back disorders resulting from fatigue failure of tissues of the lumbosacral spine.

### **7.3 Summary**

The data from this dissertation may prove quite useful in developing improved methods of reducing low back disability from repetitive occupational lifting tasks. Armed with an improved understanding of the fatigue tolerance of spinal tissues in different torso flexion postures, and having the ability to estimate the dynamic loads experienced by workers *in vivo*, it may be possible to estimate the fatigue life, within reasonable confidence limits, of spinal tissues of individual workers given the loads lifted, torso flexion angle, and/or compression and shear loads experienced during repetitive lifting. Such estimates may be made more precise if data regarding the size and bone mineral content of the workers lumbar spine can be obtained, which is certainly feasible given our current technology. Our improved understanding of the risk of spinal tissue damage in different lifting postures may also lead to important

modifications of lifting recommendations, job design, job rotation, and interventions to reduce low back disorder risk.

It is also clear from this research that bone mineral content is critically important in determining the fatigue life of spinal tissues. Given the fact that obtaining bone scans is straightforward and not prohibitively expensive, one might conceive that monitoring this important variable (on a consensual basis) for manual material handlers could provide substantial financial and health benefits for employers and employees, respectively. In fact, obtaining data on lumbar spine bone mineral content (and relating the result to estimated spine loads) might provide a much more effective rationale for worker selection than measures of job strength demands and muscular capabilities. The latter measures have demonstrated only modest correlations to low back disorders; however, the former have been shown to be good predictors of vertebral strength, both in ultimate strength and (in this study) fatigue failure tests.

While additional research will be necessary to validate the findings of this study, the lessons learned from the current study seem unambiguous with respect to the vital role played by torso flexion in increasing the risk of endplate and vertebral fractures when lifting. These fractures appear to be important in instigating degenerative changes in the disc, and studies examining the development of annular fissures in degenerated discs suggest that migration of nucleus material to the outer portions of the disc may be a potent source of low back pain. Given these facts, it is concluded that repetitive lifting in flexed torso postures plays a crucial role in determining the risk of spinal tissue fatigue damage, and may be a significant contributor to the development of low back disorders.

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### Chapter 1

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