

**Finite Element Analyses of Thoracolumbar
Junction ----Investigations of Instantaneous
Axes of Rotation (IARs) and Burst Fracture
Mechanism**

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Summary

In spinal column, the thoracolumbar junction (TLJ) is a transitional region and a frequent site of spinal injuries. Accurate knowledge of the mechanical behaviors of the thoracolumbar spine, in response to external loads, is of importance to clinicians in treating the spinal injuries if the external loads developed in the spine are to be linked in a quantitative fashion to possible modes of injury or abnormal motions of the intervertebral joints. Anatomically realistic finite element (FE) models of the thoracolumbar T11-T12 and T12-L1 functional spinal units (FSUs) were constructed and validated to investigate biomechanical responses of the TLJ under external loads.

Instantaneous axis of rotation (IAR) is one of the kinematics characteristics of a FSU in a plane under load. Reliable identification of IARs can be of diagnostic value in predicting the behaviors of the FSU in response to different injurious vectors. Despite the apparent and putative importance of IARs of FSUs, little is known about the IARs of the thoracolumbar FSUs. The present study established the locations and loci of the IARs at vertebral levels T11-T12 and T12-L1 in sagittal, frontal, and transverse planes using the FE method. The IAR locations and loci were found to vary and the loci of the IARs were observed to move along different tracks with rotation in three anatomical planes. The predicted IARs provided further understanding to the kinematics and biomechanics response of the TLJ, which is important for the diagnosis of disc degeneration and implant study.

Burst fractures are mainly localized in the thoracolumbar region, however, their causes are not clear. Thus, it is definitely beneficial to identify mechanisms that result in the observed fracture patterns. Accordingly, the burst fracture mechanism and the influence of disc degeneration on it were explored by internal changes such as load transmission and stress distribution in the FE analysis of T12-L1 FSU under axial impact. The paths of load transmission and stress distributions in the thoracolumbar vertebral bodies were found to be dependent on the conditions of the intervertebral discs. Thus, the induced fracture modes were also different. The risk of compressive burst fracture in the vertebral body adjacent to degenerated disc is lower compared with the risk in those adjacent to normal disc. This provided new insight into the aid in the management of the treatment of burst fractures and explained that the burst fracture mostly occur in young people.

Chapter 1 INTRODUCTION

1.1 Background

The human spine (also known as the spinal column, the vertebral column or the backbone) is a complex structure which is vulnerable to injury, abuse and the natural ageing process. The unique geometrical shape and articulation of the spine provide the human being much needed flexibility for motion and stability to maintain an upright posture.

The thoracolumbar junction (TLJ) is a transitional region where the normally kyphotic thoracic region shifts to the normally lordotic lumbar region; the coronally oriented facet joints of the thoracic region transform to the sagittally oriented facet joints of the lumbar; and the relatively immobile thoracic region changes to the relatively mobile lumbar region. To accommodate the changes in mobility, the transitional vertebra of the TLJ is uniquely structured with thoracic-type superior and lumbar-type inferior articular processes. The superior facets of such vertebra are coronally oriented to permit rotation (Panjabi et al., 1976a, 1976b; Willems et al., 1996) articulation with the upper level of thoracic spine, while the inferior facets are sagittally oriented to restrict rotation (Ahmed et al., 1990; Lavaste e al., 1992; Markolf, 1972; Panjabi et al., 1994; Shirazi-Adl et al., 1986; Yamamoto et al., 1989) articulation of the lower level of lumbar spine.

Anatomically and physiologically, the twelfth thoracic vertebra - T12, is a transitional vertebra of the TLJ in both man and most quadrupeds. In some subjects,

the transitional vertebra may be at T11, the eleventh thoracic vertebra. Hence, the transitional changes in anatomy and function from the thoracic spine to the lumbar spine occur either at the T11-T12 or the T12-L1 intervertebral joints (Davis, 1955, 1961; Gertzbein, 1992; Giles and Singer, 2000), where L1 is the first lumbar vertebra. However, in most anatomical texts, the TLJ is typically depicted by showing an abrupt change in the orientation of the articular joints of T12 (Chua, 1999; Panjabi et al., 1993; Giles and Singer, 2000; Singer et al., 1989).

The different anatomical characteristics at the functional spinal units (FSUs) of T11-T12 and T12-L1 provide an opportunity to study the associations between pathoanatomical changes in these two levels. In this thesis, T11-T12 and T12-L1 segments, which possess the transitional vertebra T12, are studied using the Finite Element Analysis Method to determine whether the mechanical properties reflect these anatomical changes between the two FSUs and if the TLJ has certain biomechanical features that are unique in the spine.

As the motion of a FSU in a plane can be expressed in terms of the location of instantaneous axis of rotation (IAR), locus of centrodes (centers of rotations) and rotation magnitude under load, the rotation path of a FSU is indicative of the health and pathological disease of a spinal segment (Amevo et al., 1992; Gertzbein et al., 1985, 1986; Sharma et al., 1995). Any changes in the locus of the IARs of the segment under motion are the consequences of disc degeneration, and changes in physical properties of the ligamentous and bony structures (Gertzbein et al., 1985, 1986). Therefore, reliable identification of IARs can be of diagnostic value in predicting the behavior of a FSU in response to different injurious vectors. Despite

the obvious and recognized importance of IARs of intervertebral joints, little is known about the IARs locations of thoracolumbar intervertebral joint region, especially the T12-L1 level. Therefore, the present study endeavored to document the IARs at levels T11-T12 and T12-L1 in the thoracolumbar region.

It is also well known that the thoracolumbar region is a common site of spinal injuries (Bensch et al., 2004; Yoganandan, 1989). The reasons for the vulnerability of this region are manifold. One possible reason is that though the rib cage tends to increase the stability of the thoracic spine (White and Panjabi, 1990), it protects neither the TLJ, nor the lumbar spine. Another reason may be the abrupt change in the superior and inferior facet orientations of T12 as aforementioned.

Injuries in the region of the TLJ contributes to 30-60% of all spinal injuries (Kifune et al., 1997), of which burst fractures, mainly caused by instant, impact axial loading in traffic and falling accidents (Bensch et al., 2004; Yoganandan, 1989), account for approximately 15% (Denis, 1983; Kifune et al., 1997). Moreover, the frequency of neurological deficit in patients with burst fractures was reported to be as high as 50-60% (Abe et al., 1997; Cho et al., 2003; Denis, 1983; Kifune et al., 1997; McEvoy and Bradford, 1985; Panjabi et al., 1995b; Trafton and Boyd, 1984; Willen et al., 1985). Due to the severity of the injuries and the consequent costs associated with the care of paraplegic patients, the prevention and management of thoracolumbar burst fractures have been topics of great interest in the spine research community (Cho et al., 2003; Denis, 1983; Gertzbein, 1994; Hitchon et al., 1999; James et al., 1994; Magerl et al., 1994; Heth et al., 2001; Hitchon et al., 2000a, 2000b, 2002), attracting many clinical and laboratory investigations with respect to

spinal injuries in terms of stability and treatment after trauma (Abe et al., 1997; Fredrickson et al., 1992; Kifune et al., 1995, 1997; Langrana et al., 2002; Leferink et al., 2002, 2003; Lin et al., 1993; Oner et al., 1998; Panjabi et al., 1995a, 1995b, 1998; Tran et al., 1995; Willen et al., 1984, 1985; Zou et al., 1993).

Most of these experimental studies focused on the fracture patterns during trauma, post-trauma kinematics characteristics and treatment of the thoracolumbar region. However, the instant internal changes of the fractured vertebra bodies during the dynamic process could not be observed experimentally. Since analytical method, such as the FE method which not only has the potential to replicate the external responses, but also quantify and qualify the intrinsic parameters (such as stress, strain, strain energy, etc) (Fagan et al., 2002; Gilbertson et al., 1995; Goel and Gilbertson, 1995; Yoganandan et al., 1996) of objects with complex geometry, multiple material compositions and complicated loading conditions, the present study also analyzed the mechanical mechanism of burst fracture of TLJ under impact axial loading to understand spine fracture etiology that may provide information to aid management of these fractures, and even to reduce their incidence by identification of critical loading conditions.

1.2 Objectives

The objectives of this study were:

- To develop and validate the three-dimensional FE models of the thoracolumbar junctional motion segments T11-T12 and T12-L1 using actual human cadaveric specimen for biomechanical study.

- To determine and track the locations and loci of IARs of the T11-T12 and T12-L1 motion segments under all the static physiological loading modes: flexion, extension, lateral bending and axial rotation in the three anatomical planes, using FE method.
- To investigate the burst fracture mechanism under axial impact load by analyzing stress distribution during the dynamic process using FE model of T12-L1.

1.3 Scope

This research focused on the study of IARs and burst fracture mechanism of the TLJ using the FE method.

1.4 Organization

This thesis is divided into eight chapters and a brief description of each chapter is as follows:

Chapter 1 begins by introducing the background, objectives and scope of this study. Chapter 2 includes an introduction of the anatomical terms related to anatomical positions and planes, and presents a brief overview of the relevant anatomy of the thoracolumbar junction.

Chapter 3 is an overview of the biomechanical studies of the TLJ and reveals the present gap in research which motivated the interest for the current study.

Chapter 4 concentrates on describing the procedures used to develop the three-dimensional anatomically realistic FE models of the thoracolumbar motion segments T11-T12 and T12-L1 based on the digitized geometrical data of the embalmed vertebrae T11, T12 and L1. This chapter also presents the validation study of the FE models of the T11-T12 and T12-L1 FSUs, whereby the predicted results by the FE models were compared against those in published experimental studies under flexion and extension, left and right lateral bending, and left and right axial rotation loading configurations.

Chapter 5 is devoted to investigate the locations and loci of the IARs of T11-T12 and T12-L1 under all physiological loading modes within the validated range. The normal locations and loci of the intact FE models T11-T12 and T12-L1 in the three anatomical planes were documented. Chapter 6 presents the burst fracture mechanism by internal changes such as load transmission and stress distribution in the FE analysis of T12-L1 FSU under axial impact. The influence of disc degeneration on burst fracture mechanism is also discussed to explain why the burst fracture is mostly observed in young people.

Chapter 7 concludes the findings in this study and arrives at the practical importance of this research. Finally, Chapter 8 discusses the limitations of this study and recommendations for further research.

Appendix A lists the publications related to this research including original journal papers and conference papers.

It is hoped that the work presented in this thesis will be judged to have fulfilled these objectives clearly and concisely whilst demonstrating that the intelligent use of the structural model (Finite Element Method) can be usefully employed in the biomechanical analysis of biological structures.

Chapter 2 ANATOMY OF HUMAN THORACOLUMBAR JUNCTION (TLJ)

This chapter presents anatomical terms pertaining to anatomical position and provides an overview of the anatomy of the TLJ, with particular attention focus on the functional anatomy of the intervertebral disc, ligaments and facet joints with an abrupt change in the orientation of the articular joints at T12.

2.1 Definition of the Anatomical Position and Planes

Various parts of the body can move in many different directions. In the vertebral column, most of the movements take place (i) at the intervertebral disc and (ii) at the joints or articulations formed between the articular processes (superior and inferior articular processes) of the over- and underlying vertebral arches of the articulating vertebrae providing the flexibility of human movement. Description of human body movements can be difficult and since this study involves the kinematics analysis of spinal motion segments under loads, the anatomical terms used are briefly described below as a foundation for a clear understanding of anatomical positions and planes.

Figure 1 shows the commonly adopted anatomical position and planes with which reference is made to describe the various parts of the body. The anatomical planes and terms are defined with the assumption that the person is standing erect, with the upper limbs by the sides and the face and palms of the hands directed forward, and it is called the anatomical position (Seeley et al., 1991).

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As shown in Figure 1, the midsagittal plane is a vertical plane passing through the midline of the body dividing it into right and left halves. Any vertical planes parallel to the midsagittal plane are termed sagittal planes.

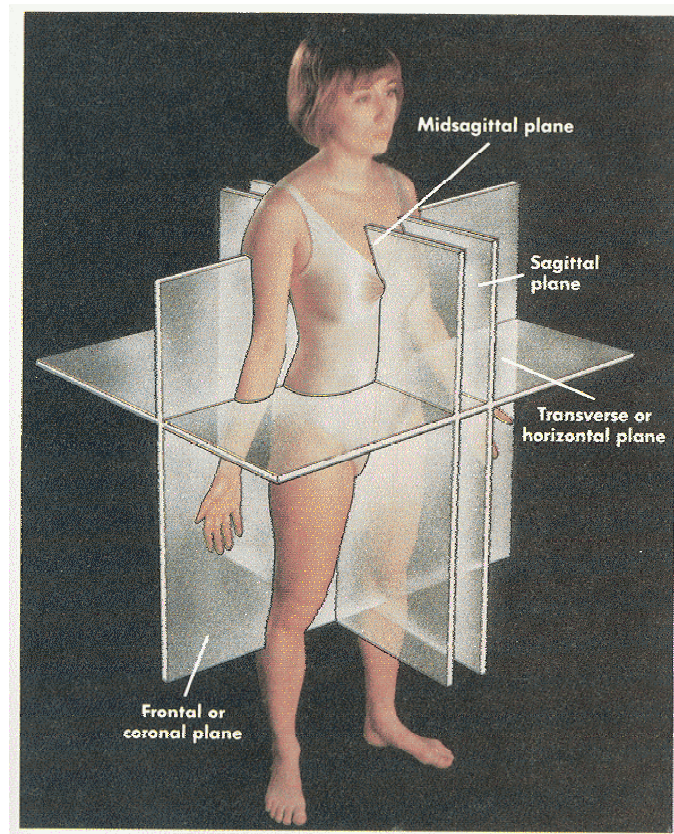


Figure 1 Anatomical planes (Adopted from Seeley et al., 1991)

Frontal or coronal planes are vertical planes perpendicular to the midsagittal or sagittal planes dividing the body into anterior and posterior sections. Transverse or horizontal planes are any planes passing through the body at right angles to both the sagittal and coronal planes dividing the body into superior and inferior portions.

The terms anterior and posterior (or ventral and dorsal) are used to indicate the front and back of the body respectively. The relationship of two structures is said

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to be anterior or posterior to the other insofar as it is closer to the anterior or posterior body surface. The terms superior and inferior are used to denote levels relatively high or low with reference to the upper and lower ends of the body.

2.2 Anatomical Features of the Thoracolumbar Junction (TJL)

In human beings, the vertebral column is divided into five common classifications according to site and characteristics as shown in Figure 2. There are (i) seven cervical vertebrae in the neck region, (ii) twelve thoracic vertebrae in the thoracic cavity, (iii) five lumbar vertebrae in the lower back, (iv) five sacral vertebrae (fused) in the sacrum, and (v) four coccygeal vertebrae (fused) in the coccyx. Various situations impose various tasks upon the vertebrae, which in consequence, exhibit structural modifications to meet them.

The TLJ is a transitional region of the spine, it is well known as a site of anatomical variations and clinical significance (Giles and Singer, 2000, Singer et al., 1989). The TJL junctional segments link together regions of thoracic and lumbar spines, each of different mobility limitations. The transitional changes in anatomy and function from the thoracic spine to lumbar spine occur at the T11-T12 and T12-L1 intervertebral joints (Oxland et al., 1992). Figure 2 shows the lateral view of the vertebral column and depicts the thoracolumbar vertebrae T11, T12 and L1 that appear in the entire vertebral column.

Anatomically, the thoracolumbar transitional zone highlights an increase in the size of the vertebral bodies and intervertebral discs (Berry et al., 1987; Panjabi et

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al., 1991, 1992; Tan et al., 2002, 2004), associated with a larger vertebral canal to accommodate the lumbar enlargement of the spinal cord and conus (Louis, 1983). The anatomical relationships of the bony vertebrae (T11, T12 and L1), intervertebral disc, facet joints and ligaments are emphasized in this section for better understanding of the associate structural elements involved in biomechanical responses of the thoracolumbar FSUs.

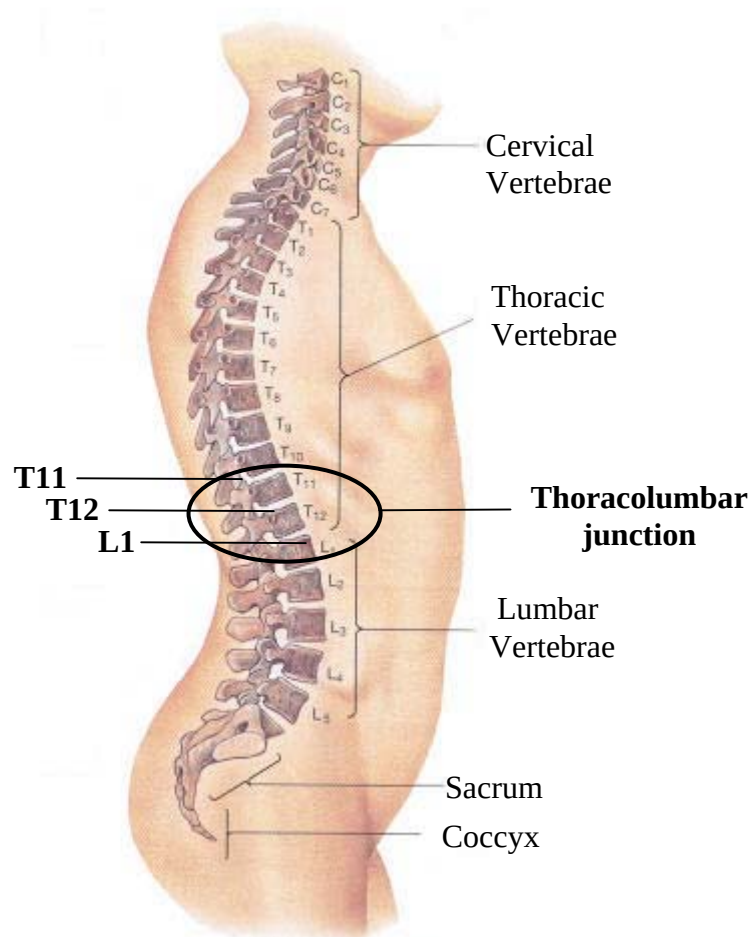
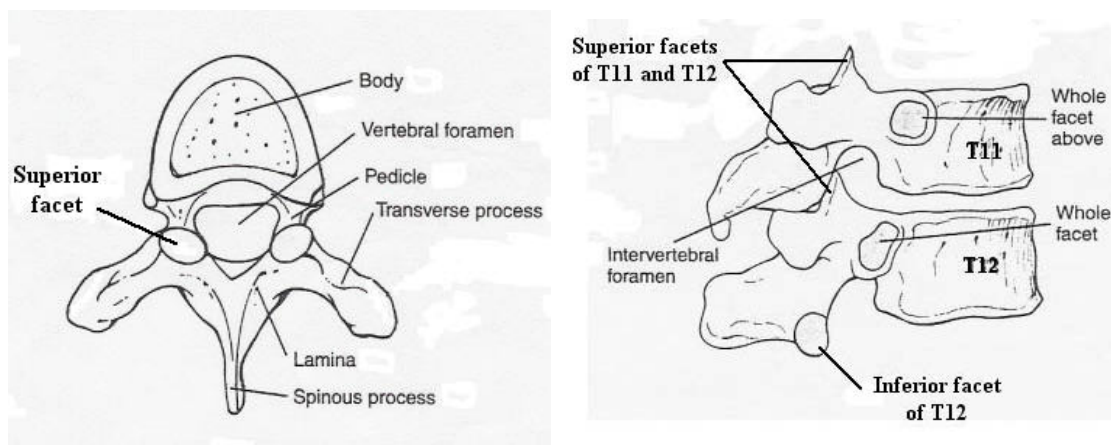


Figure 2 The vertebral column and location of the thoracolumbar junction appears in the entire vertebral column. (Modified from Flynn and Greenman, 1996)

2.2.1 Vertebrae

The vertebrae T11, T12 and L1 of the TLJ are irregular bones consisting of various parts (Figures 3 and 4). Although they show regional differences, they all possess a common pattern. Generally, a typical vertebra consists of a rounded body anteriorly i.e. the body, and a vertebral arch posteriorly, which contains the articular, transverse, and spinous processes. The vertebral body is a large block of cancellous bone contained in a thin shell of cortical bone. Viewed from above or below, the last two thoracic vertebrae T11 and T12 have rounded, heart-shaped bodies (Figure 3A); but the upper lumbar vertebra L1 has a curved perimeter that is more or less kidney-shaped (Figure 4) (Benzel and Stillerman, 1999; Bogduk and Twomey, 1997). The vertebral body is the main load bearing structure of the vertebra. The superior and inferior surfaces of the vertebral bodies are relatively flat to enhance their load carrying capacity.



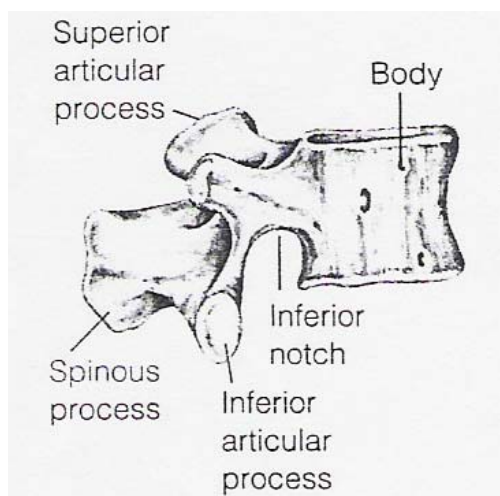
A. Superior view

B. Articulated and right lateral view

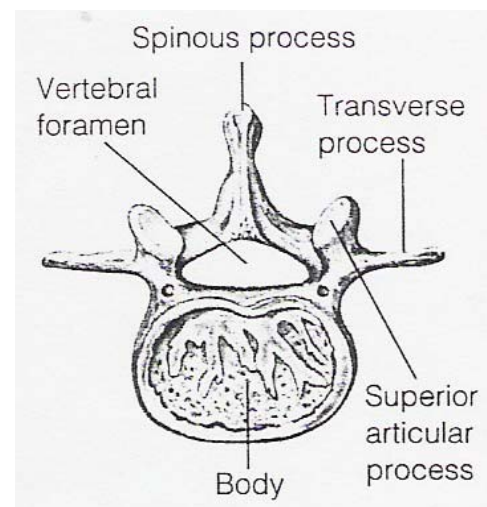
Figure 3 Vertebral characteristics of T11 and T12. (Modified from Eroschenko, 1996)

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The neural arch surrounds the neural elements that pass through the vertebral foramen. The vertebral arch is a composite structure formed by two pedicles and two laminae. Projecting from the back of the vertebral body are two stout pillars of bone; each of these is called a pedicle. The pedicles attach to the upper part of the back of the vertebral body. The last two thoracic vertebrae T11 and T12 usually have one complete facet each, located on the pedicle for articulating with the so-called floating ribs (Figure 3). The pedicles have notches on their superior and inferior borders, forming lateral openings between adjacent vertebrae called intervertebral foramina (Figure 3 and 4) whereby the spinal nerves issuing from the spinal cord pass through. Projection from each pedicle towards the midline is a sheet of bone called the lamina. The laminae are flattened roof plates that complete the arch posteriorly.



A. Right lateral view



B. Top view

Figure 4 Bony anatomy of lumbar vertebra L1 (Modified from Marieb and Mallatt, 1997)

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The superior and inferior articular processes are paired processes that protrude superiorly and inferiorly, respectively, from the pedicle-lamina junctions. The superior processes of one vertebra articulate with the inferior processes of the vertebra immediately superior (Figure 3). Thus, successive vertebrae are joined both at their bodies and at their articular processes. The articulating surfaces of these processes are called facets (Marieb and Mallatt, 1997).

The spinous process is a single posterior projection arising at the junction of the two laminae. A transverse process projects laterally from each side of the vertebral arch. Both spinous process and the transverse processes are attachment sites for muscles that move the vertebral column and for ligaments that stabilize it.

The quantitative information of the vertebra is necessary as spinal biomechanics research becomes more widespread and for precise clinical practice (Figure 5). This information provides a better understanding of the spine, and allows for a more accurate surgical management of spinal problems. The information is also needed for the accurate construction of the mathematical models of the human spine.

The detailed three-dimensional vertebral geometries of T11, T12, and L1 of general adult population have been documented previously (Panjabi et al. (1991, 1992) and Tan et al., (2002, 2004)). The main dimensions of the T11, T12, and L1 are listed in Table 1. Discrepancies in the two sets of data could have arisen due to geographical disparities. These dimensions provide the basis for the selection of representative specimen geometry to build the FE models for the current study described in Section 4.2.1.

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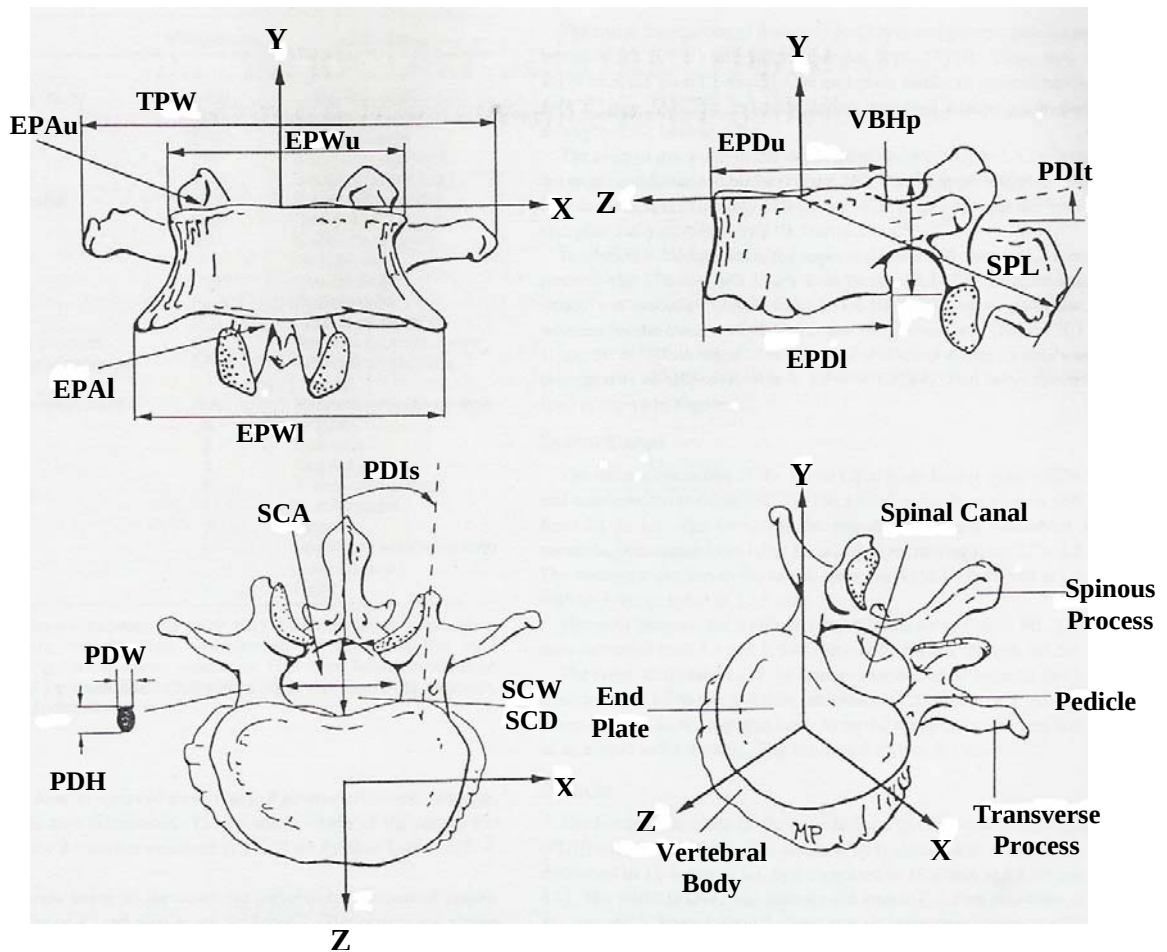


Figure 5 Four views (*front, side, top and isometric*) of a lumbar vertebra. *EPAl* Lower end-plate area, *EPAu* upper end-plate area, *EPDI* lower end-plate depth, *EPDu* upper end-plate depth, *EPWl* lower end-plate width, *EPWu* upper end-plate width, *PDH* pedicle height, *PDI* left pedicle, sagittal inclination, *PDI* left pedicle, transverse inclination, *PDW* pedicle width, *SCA* spinal canal area, *SCD* spinal canal depth, *SCW* spinal canal width, *SPL* spinous process length, *TPW* transverse process width, *VBHp* posterior vertebral body height. (Adopted from Panjabi et al., 1992)

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Table 1 Main Dimensions of vertebrae T11, T12, and L1.

	Vertebral level					
	T11	T12	L1	T11	T12	L1
	Tan et al. (2002, 2004)			Panjabi et al. (1991, 1992)		
	Linear dimension (mm)					
EPWu	31.6±0.3	34.5±0.3	36.3±0.4	34.9±0.97	39.0±0.58	41.2±1.03
EPWl	35.3±0.2	36.4±0.4	39.2±0.5	39.1±0.71	42.1±0.91	43.3±0.78
EPDu	25.4±0.3	26.7±0.3	27.5±0.4	31.9±0.71	32.8±1.21	34.1±1.34
EPDl	26.9±0.2	27.7±0.3	28.5±0.5	31.8±0.78	33.4±0.78	35.3±1.27
VBHp	20.4±0.2	21.5±0.2	22.4±0.4	21.3±0.71	22.7±1.04	23.8±1.03
SCW	15.3±0.2	17.9±0.3	19.4±0.2	19.4±0.95	22.2±1.12	23.7±0.67
PDHl	14.2±0.3	14.2±0.2	13.1±0.4	17.8±0.29	16.8±0.94	15.8±0.74
PDHr	14.3±0.2	14.0±0.3	13.1±0.3	16.9±0.49	16.5±0.71	15.9±0.81
PDWl	7.1±0.2	7.8±0.2	5.6±0.2	10.7±0.84	8.6±0.68	9.2±0.88
PDWr	6.9±0.2	7.3±0.3	5.5±0.3	8.8±0.43	8.8±0.81	8.0±0.95
SPL	48.4±1.1	48.5±1.0	51.5±0.9	45.6±1.15	47.4±1.70	67.7±1.24
TPW	43.0±0.9	41.0±0.5	53.6±1.5	52.2±1.28	46.9±1.84	71.2±1.66
Surface area (mm ²)						
EPAu	706.6±21	795.6±20.8	889.8±25	842.0±41.4	954.0±44.0	1057±60.78
EPAI	803.0±21	877.7±24.7	1009.8±30	945.0±44.3	1024.0±49.8	1117.0±49.00
SCA	148.6±4.1	177.8±5.5	200.0±2.2	220.0±15.7	280.0±16.3	320±18.10
PDAI	95.6±3.4	99.1±2.1	67.8±3.2	98.3±5.93	97.0±12.30	88.5±11.74
PDAr	90.5±3.0	92.0±3.7	68.6±3.9	88.4±7.5	90.9±13.0	86.4±11.42

2.2.2 Facet Joints

The transitional regions of the human spine often have clinical significance due to the prevalence of morphological variation and dysfunction. The change from coronal to sagittal plane orientation of the facet joints at the TLJ is commonly described as occurring abruptly between T11-T12 to T12-L1 (Singer et al., 1989) (Figure 6A).

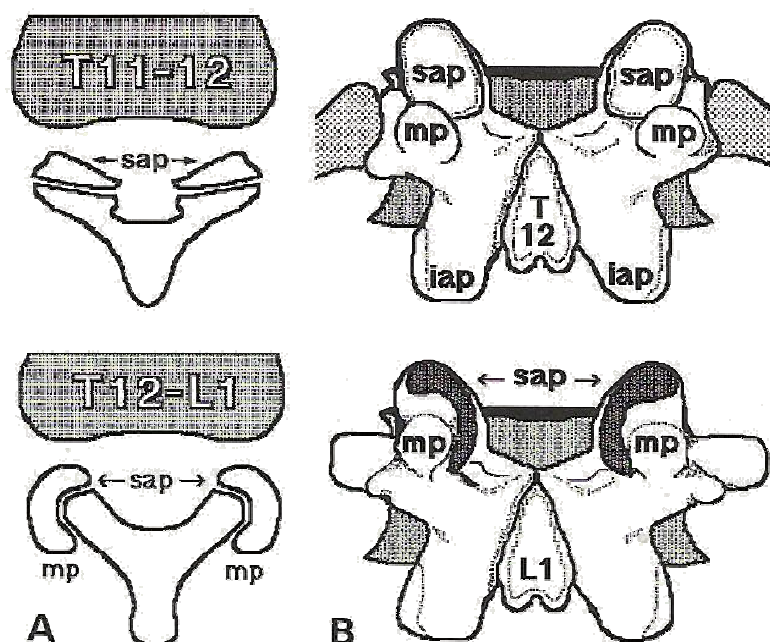


Figure 6 Schematic illustration of the facet joints at T11-T12 and T12-L1. **A.** Schematic illustration of an abrupt transition from T12-T12 to T12-L1, comparing the articular planes of the paired facet joints from a transverse section parallel to and through the superior endplate. The superior articular processes of T12 are directed in the coronal plane and the inferior articular processes are oriented in the sagittal plane. **B.** A mortice joint is demonstrated by the medial projection of the mamillary processes which enclose the inferior articular processes of the vertebra above. *iap* inferior articular process, *sap* superior articular process, *mp* mamillary process. (Adopted from Singer et al., 1989)

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The inferior facets of T11 face anteriorly while the superior facets of the T11 and T12 face posteriorly in the frontal plane. This arrangement facilitates axial rotation of the T11-T12 spinal segment. In contrast, the inferior facets on the mamillary processes (Figure 6) of the T12 and L1 face laterally, while the superior facets of L1 face medially in the sagittal plane. This shape limits the axial rotation of the T12-L1 spinal motion segment (Figure 3 and 4) (Eroschenko, 1996). Davis (1955, 1961) called this the thoracolumbar mortice joint (Figure 6B), formed through the “interlocking” inferior articular processes of the T12 and the enclosing superior articular processes of the L1, noting the “locking” of the joint in axial rotation.

Hence, the TLJ has certain biomechanical features that are unique in the spine. As the superior facet joints of T12 are shaped like those of the thoracic vertebrae, while the inferior ones have the pattern of lumbar facet joints, T12 acts a hinge around the thoracic and lumbar parts of the spine. This means that the harmonious movement of the spine is broken at this site, and explains why this part of the spine is particularly susceptible to traumatic injuries

2.2.3 Intervertebral Disc

The space between two vertebrae is usually occupied by an intervertebral disc (Figure 7) which is a soft tissue. They function as spacers to provide clearance for exiting spinal nerves, as connectors to link adjacent vertebrae together and allow for movement, and also as spinal shock absorbers. The obvious functions of the discs are to provide flexibility to the spine and to facilitate a range of complex movements. In addition, they also transmit mechanical loads through dissipation of energy.

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Intervertebral discs are discrete components of the spine that occupy approximately one-third of its length and separate the vertebral bodies. The three basic components of the disc, shown in Figure 8, are the annulus fibrosus, the nucleus pulposus and the end plates.

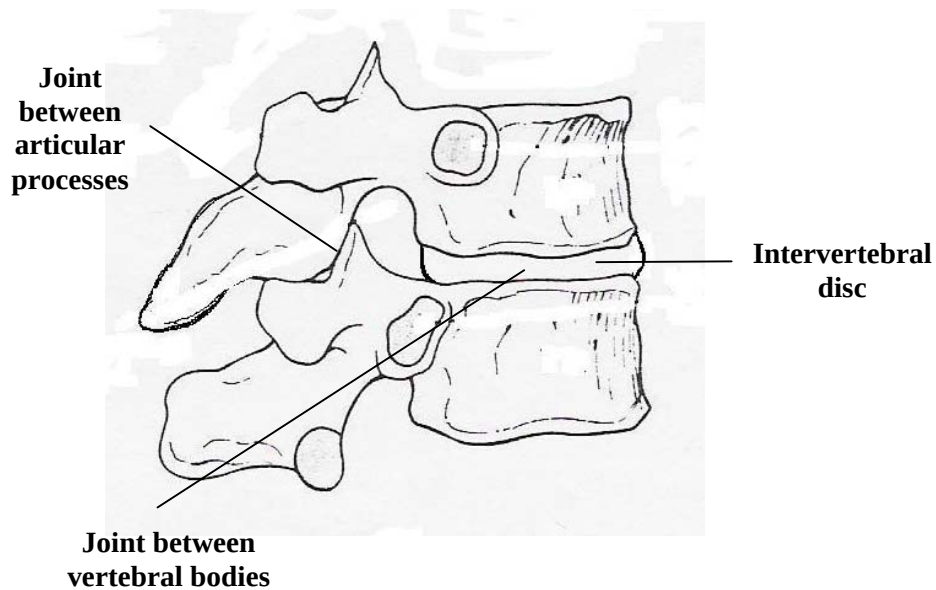


Figure 7 Two adjacent vertebrae joined by joints between articular processes and joint between vertebral bodies. (Modified from Eroschenko, 1996)

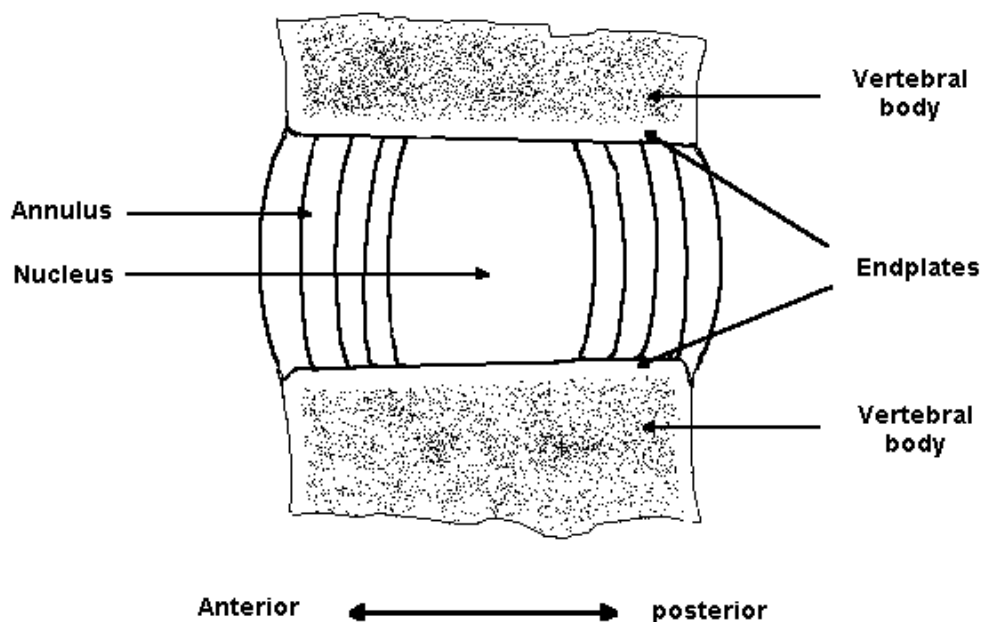


Figure 8 Diagrammatic representation of a sagittal section of a human intervertebral disc showing the attachments of the annular lamellae to the vertebrae and endplates. (Modified from Ghosh, 1991)

With the exception of the sacrum and the first cervical level, all 23 discs in the human spine have a similar structure generally (Figures 8 and 9) (Ghosh, 1991). The annulus fibrosus forms a firm, but flexible, outer layer that surrounds the central incompressible nucleus pulposus (White and Panjabi., 1990).

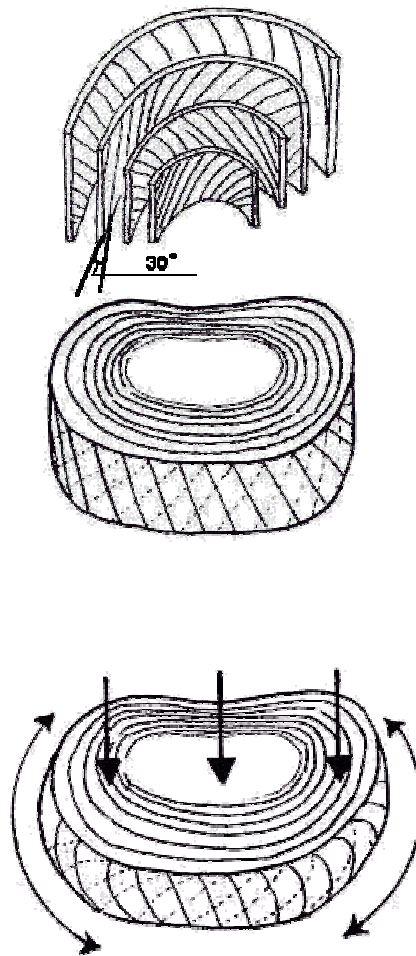


Figure 9 Structure of the intervertebral disc. (Top) The concentric lamellae of the annulus are expanded to show the alternating fiber angle. (Center and bottom) Compression of the disc causes height loss, increased radial bulge, and a change in alternating fiber angle. The nucleus pulposus exhibits a hydrostatic pressure and creates a tangential hoop stress in the annulus. (Adopted from Ghosh, 1991)

2.2.3.1 *Annulus Fibrosus*

The surrounding annulus fibrosus is a composite structure consisting of concentric layers or lamellae of collagen fibers that form the peripheral portion of the intervertebral disc (Figure 9). Alternating fiber layers attach obliquely (Flynn and Greenman, 1996). The confined nucleus pulposus is somewhat posteriorly off center

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(Figure 9) (DePalma and Rothman, 1970, White and Panjabi, 1990). The collagen fibers in each lamella are orientated parallel to one another and form an angle of inclination (about 30°) with the bony endplates (Oliver and Middleditch, 1991) (Figure 9). The lamellae structure and the angle of inclination of the collagen fibers enable the annulus fibrosus to sustain the normal forces of compression, torsion, and flexion that occur during the deformations of the disc.

2.2.3.2 Nucleus Pulposus

The nucleus consists of a highly hydrated gel of proteoglycans containing some collagen. It occupies approximately 40% of the disc's cross-sectional area. The nucleus pulposus has a high water content which decreases with age (DePalma and Rothman, 1970). Being a highly hydrated gel, the nucleus can be deformed under pressure without a reduction in volume (Figure 9). This essential property enables it to both accommodate to movement and to transmit some of the compressive load from one vertebra to the next. The nucleus is usually located between the middle and posterior thirds of the disc (DePalma and Rothman, 1970).

2.2.3.3 Endplates

The endplates separate the discal elements from the cancellous bone of the vertebrae and are composed of hyaline cartilage adjacent to the disc and cortical bone between the cartilage and cancellous bone (Gertzbein., 1992; Moore, 2000) (Figure 8). The endplates of a healthy disc prevent the highly hydrated nucleus from bulging into the adjacent vertebral bone, while simultaneously absorbing hydrostatic pressure that results from mechanical loading of the spine (Broberg, 1983). The endplates

appear to be susceptible to mechanical failure as this is a weak link in the structure of the disc.

2.2.4 Spinal Ligaments

Ligaments are uniaxial structure and they are most effective in carrying loads along the direction in which the fibers run. They readily resist tensile forces but buckle when subjected to compression. When a functional spinal unit is subjected to complex force and torque vectors, the individual ligaments provide tensile resistance to external loads by developing tension (White and Panjabi, 1990).

There are seven well-defined ligaments in the thoracolumbar junctional region (Figure 10).

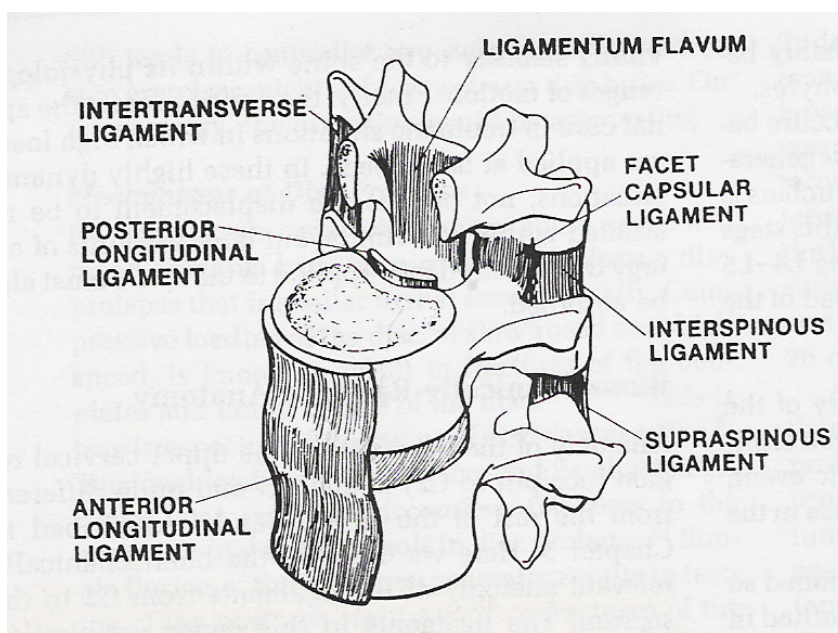


Figure 10 Ligaments supporting the thoracolumbar spine.(Adopted from White and Panjabi, 1990)

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2.2.4.1 Ligaments of the Vertebral Body

The ligaments surrounding the thoracolumbar vertebral bodies include the anterior longitudinal ligament (ALL) and posterior longitudinal ligament (PLL). The anterior longitudinal ligament is a continuous band throughout the axial skeleton. It inserts along the anterior aspect of the vertebral bodies and disc. Its width is less than the width of the discs. It is made of multiple layers of interlacing fibers oriented longitudinally.

The posterior longitudinal ligament also extends continuously throughout the spinal axis. In general, the posterior longitudinal ligament is thicker than the anterior longitudinal ligament (Benzel and Stillerman, 1999). It is required to restrict posterior separation of the vertebral bodies and to limit hyperflexion.

2.2.4.2 Ligaments of the Laminae and Spinous Processes

The ligamentum flavum (LF) connects each overlapping lamina. The ligament originates on the anteroinferior border of the lamina above, and attaches to the superior margin of the lamina below (Benzel and Stillerman, 1999). The ligament is a bilateral structure. It is composed of a large amount of elastic fibers and represents the most pure elastic tissue in the human body. In any case, depending on its origin, the ligamentum flavum possesses significant tension. It is believed that the function of this ligament is to allow limited separation of the laminae of the vertebrae whilst ensuring no encroachment on the nearby nerve roots and spinal cord.

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The interspinous ligament (ISL) joins the spinous processes along their entire length. The supraspinous ligament (SSL) is a multisegmented ligament that connects the posterior edges of the spinous processes along the entire vertebral column (Benzel and Stillerman, 1999). The role of the interspinous and supraspinous ligaments is to limit distraction and movement of the spinous processes during flexion.

2.2.4.3 Ligaments of the Facet Joints

The capsular ligaments (CL) surround the facet joint capsule and are attached just beyond the margins of the adjacent articular processes. The fibers are generally oriented in a direction perpendicular to the plane of the facet joint.

2.2.4.4 Intertransverse Ligament

The intertransverse ligament (ITL) passes between the transverse processes in the thoracic region and is characterized as rounded cords intimately connected with the deep muscles of the back (White and Panjabi, 1990).

2.2.5 Thoracolumbar Curvature

Four curvatures of the vertebral column of an adult can be identified in the lateral view, as shown in Figure 2. Of the physiological curvatures of the vertebral column, the cervical and lumbar spines have lordotic curves (anteriorly convex), but the thoracic spine has a kyphotic curve (anteriorly concave) (Figure 2). The anterior height of thoracic vertebrae is less than their posterior height; this difference accounts primarily for the thoracic kyphosis (Figure 2). The intervertebral discs in

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the cervical region are thicker anteriorly than posteriorly and are entirely responsible for their normal cervical lordosis. The lumbar intervertebral discs tend to be of greater height anteriorly than posteriorly, and thus the lumbar lordosis is due almost entirely to the shape of the disc (DePalma and Rothman, 1970). The TLJ is a transitional region where the curvature is changed from the anteriorly concave of the thoracic curve to the anteriorly convex of the lumbar curve (Figure 2).

In the normal orthograde posture, the location of the line of gravity in relation to the vertebral column appears to be consistently located through the transitional regions (Giles and Singer, 2000). Despite changes in the magnitude of the curves from one region to another, they appear to compensate each other in order to maintain a “balance” in relation to the line of gravity. This reciprocal change in curvature of the thoracic anteriorly concave and lumbar anteriorly convex produces an inflexion point which is located commonly between T11 and L1 (Giles and Singer, 2000). During axial loading, the thoracic spine deforms in kyphosis, the lumbar spine in lordosis resulting in the TLJ experiencing pure compression. Thus, the thoracolumbar spine is predisposed to axial compression injuries while falling from a height and landing on the feet (Bensch et al., 2004).

2.3 Functional Spinal Unit (FSU)

The basic repeating structure of an intervertebral disc, its two neighbouring vertebrae and the connection ligamentous tissues is referred as the functional spinal unit (FSU), or the spinal motion segment (Figure 10). It is the smallest segment of the spine that exhibits biomechanical characteristics similar to those of the entire

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spine (White and Panjabi et al., 1990). The behavior of a FSU is dependent upon, among other things, the physical properties of its components, such as the intervertebral disc, ligaments, and articulating facets

The spine can be considered as a structure composed of multiple FSUs connected in series, its total behavior may be approximated as a composite of the behaviors of the individual motion segment constituting the spine.

Chapter 3 LITERATURE REVIEW

3.1 Introduction

The abrupt changes in anatomy and function of the TLJ from the less mobile thoracic spine to the more mobile lumbar spine (Giles and Singer, 2000; Oxland et al., 1992) and the structural modifications of intervertebral discs and vertebrae to accommodate the change in curvatures from thoracic kyphosis to lumbar lordosis (described in Chapter 2) make the TLJ region susceptible to fractures. Hence, the TLJ is an area of clinical interest and has been the focus of many biomechanical studies (Gregersen and Lucas, 1967; Koeller et al., 1984; Markolf, 1972; Panjabi et al., 1976a, 1976b; Oxland et al., 1992; Singer et al., 1989). It is also a common site of vertebral fractures leading to paraplegias (Abe et al., 1997; Cho et al., 2003; Denis, 1983; Kifune et al., 1997; McEvoy and Bradford, 1985; Panjabi et al., 1995b; Trafton and Boyd, 1984; Willen et al., 1985).

Biomechanics is widely considered to play a major role in the etiology and treatment of spinal disorders or injuries. To study the biomechanics of the human thoracolumbar spine, *in vitro* experiments and computational FE analyses are usually used. Since human cadaveric spines are advantageous in their retention of geometrical, structural, and material properties, the results from *in vitro* cadaver experimental study are most appropriate to explain the biomechanical behavior of the human spine. However, the biological variability, availability difficulty, ethical and religious concerns, and costs associated with numerous physical tests tend to be the limiting factors in conducting *in vitro* study using cadavers. In view of the

difficulties in obtaining fresh cadaveric spine specimens in the present climate of limited research resources, specimens from animals, such as bovine or porcine, and FE models are popularly used in the biomechanical studies of biological structures.

Young porcine and bovine spines are frequently used in *in vitro* studies of spine mechanics (Tran et al., 1995; Wilcox et al., 2002; Yingling et al., 1997) due to the less likelihood of variability. These specimens are not degenerated, compared with the relative much older spines normally available from cadavers. Comparative study of spinal morphology by Cotterill et al. (1986) has shown that spines of 6-8-week-old dairy bovines are similar to human thoracolumbar spines. Furthermore, another study by Cotterill et al. (1987) also showed that reproducible spinal burst fractures could be created in bovine spines.

In addition to *in vitro* studies using human cadaveric and animal specimens, many investigators (Hakim and King, 1979; Kumaresan et al., 1999, 2000; Shirazi-Adl, 1992; Teo and Ng, 2001) have developed various analytical models, such as FE models, to characterize the biomechanical responses of spine under different loadings and conditions. These models are used to predict certain parameters that prove difficult to achieve in *in vitro* studies.

The FE model is the modeling of a continuous system of an infinite number of degrees of freedom, using a representative geometry of that system made up of a finite number of smaller elements and node points. These elements are derived by constructing a set of nodal lines over surfaces and through solid bodies to produce subdivisions, which have simple shapes, of the components. The accuracy of the FE

model and results, in terms of strain, stress and displacements of the structure under loads, depend on the geometrical representation, appropriately attributed material properties and boundary/loading conditions. The material properties, displacements, and other system characteristics are represented by mathematical functions between nodes. Once the FE model has been created and the system characteristics established in the model, a global stiffness matrix is then formed for the whole structure. Given the forces and boundary conditions, the unknown displacements at each of the node points are determined and used to compute the stresses and strains acting on each element (Kardestuncer, 1987).

As the current study is focused on the biomechanics of the TLJ, brief reviews of the biomechanical studies of the thoracolumbar spine using *in vitro* experiments and FE methods conducted by different investigators are presented in the following sections as a foundation to understand some analyses relevant to the study.

3.2 *In Vitro* Experimental Studies of Thoracolumbar Junction

In *in vitro* experimental studies of thoracolumbar spine, static tests were carried out to determine the biomechanical properties of the spinal components, while dynamic tests were performed for fracture-related studies.

3.2.1 Static Tests

The motion of a normal FSU can be established by observing the deformation produced by a given force system. Any deviation from this normal range of motion may then be considered as a sign of degeneration of the segment. Therefore, an

understanding of spinal abnormalities requires knowledge of its normal behavior as a baseline. This behavior includes the relative motion of vertebral bodies. Three dimensional movements and IAR describe the relative motion of a FSU from one position to another.

3.2.1.1 Measurement of Three Dimensional Movements

A few *in vitro* studies under static loading conditions have been performed to study the gross movements of the intact thoracolumbar FSUs (Markolf, 1972; Markolf and Morris, 1974; Oxland et al., 1992; Panjabi et al., 1976a, 1976b; Virgin, 1951). These experimental studies measured the precise movements of the thoracolumbar FSU using conventional material testing machines. The *in vitro* results such as response curves, range of motion, stiffness, flexibility of the cadaveric specimens are usually used for the validation of any mathematical model developed under similar boundary/loading conditions. This allows further analyses using the validated model to be conducted with confidence.

In these previous mechanical testings of cadaveric spinal segments, each FSU was mounted by embedding the superior and inferior vertebrae in polyester resin such that the midtransverse plane of the disc was mounted parallel to the horizontal plane of the fixture (Markolf, 1972; Oxland et al., 1992; Panjabi et al., 1976a, 1976b). The testing apparatus allowed the application of the different modes of loading. Compressive load was uniformly applied by imposing displacement (displacement control) onto the superior surface of the top vertebra (Markolf, 1972). Pure moments were also uniformly applied to top vertebra through a specially

designed circular headpiece attached to this vertebra, and the couple motion developed by opposing pneumatic actuators attached to the circular headpiece with cables (Oxland et al., 1992) generating the pure moment.

Compressive deformation data of the intervertebral disc, preserved in Ringer's solution, under compression was first reported by Virgin (1951). Material was obtained from fifty-one cadavers and one to five discs were tested from each subject. The intervertebral discs were removed at autopsy, a thin slice of bone at each end of every disc was included. A series of experiments were carried out to record the force necessary to compress the object (the load) and the loss of height of the object (deformation) for the load-deformation plots. Virgin concluded that the intervertebral disc holds the property of elasticity to a marked degree. Moreover, the property of elasticity depended largely upon the ability of the intervertebral disc to absorb and lose fluid – it was in fact a viscous elasticity.

In 1972, Markolf designed a series of experiments to measure experimentally the mechanical response of the FSUs from T7-T8 through L3-L4 subjected to external forces and moments and to quantify these load-deformation characteristics by examining stiffness values for tension, compression, flexion, extension, lateral bending, torsion and shear under static tests. In this study, eight or nine vertebrae at the TLJ region were removed intact from the human specimens during the routine autopsies and tested immediately. One T11-T12 and six T12-L1 FSUs were tested under different load vectors, and the findings of the T11-T12 and T12-L1 segments were as follows:

For axial tension and compression, the load-deformation behavior of an intervertebral disc was nonlinear with the stiffness increases with load. The mechanical resistance of the intervertebral joint to lateral bending, flexion, extension, and torsion increases with continued deformation, seen from the loading and unloading curves of T12-L1. The moment-rotation curves for these modes of deformation are nonlinear and the loading and unloading curves were nearly the same.

Markolf and Morris (1974) performed a series of static compression tests (load versus displacement, creep, and load relaxation) on fresh autopsy specimens to determine the contribution of various parts of the intervertebral disc in withstanding compressive forces. In this study, the discs tested were distributed as follows: five from the interspace of T12-L1, seven from the interspace of L1-L2, five from the interspace of L2-L3, six from the interspace of L3-L4, and one from the interspace of L4-L5. The following preparations were tested: intact disc, disc injected with saline, disc punctured through annular wall, disc with nucleus removed, and annulus alone. They measured load versus displacement, deformation versus time with load held constant (creep), and load versus time with displacement with displacement held constant (load-relaxation). They reported same compressive stiffness was shown by all specimens (after initial readjustment of height and bulging in some preparations), indicating that the annulus was much important for the compressive behavior of the disc than has been hitherto believed.

Panjabi et al (1976a, 1976b) used fresh human thoracic specimens to determine three dimensional flexibility and stiffness properties, including the effects

of couple motions. Twelve forces and moments, one at a time, were applied to the top vertebra with the bottom vertebra fixed. Vertebral displacement was accurately measured in 3-dimensional space. Only one T11-T12 specimen from the thoracolumbar junction was tested in this study. They reported that (1) stiffness coefficients were higher in the axial direction than other directions, (2) axial loads resulted in significant horizontal displacements, and (3) motion was generally greatest in the direction of loading and less for the directions of coupled motions.

More recently, Oxland et al. (1992) documented the complete three-dimensional mechanical behavior of the thoracolumbar T11-T12 and T12-L1 intervertebral joints. Pure moments of flexion/extension, bilateral lateral bending and bilateral axial torque up to a maximum of 7.5 Nm were applied to 11 three-vertebrae human cadaveric specimens (T11-L1). The nonlinear load-displacement curves were reported. Average flexion, extension, lateral bending and axial rotation ranges of motion (ROMs) at level T11-T12 were $2.7 \pm 1.3^\circ$, $2.4 \pm 1.3^\circ$, $3.5 \pm 1.1^\circ$ and $1.8 \pm 0.7^\circ$, respectively, at level T11-T12, while the ROMs at T12-L1 were $2.9 \pm 1.4^\circ$, $3.9 \pm 1.4^\circ$, $3.7 \pm 1.1^\circ$, and $1.2 \pm 0.7^\circ$, respectively. The ROMs for extension and axial rotation at level T11-T12 were found to be significantly different from those motions observed at T12-L1. The different geometry in the facet joints explains these observed differences in the mechanical behavior of T11-T12 and T12-L1.

The published results of the load-displacement curves and the ROMs in the above reviewed studies (Markolf, 1972, Oxland et al., 1992) were used for the validation of the FE models of T11-T12 and T12-L1 FSUs used in the current study

(Section 4.3). The compressive load-deformation curves in the studies of Virgin (1951) and Markolf and Morris (1974) were used to compare and justify the load-deformation response of the FE model of T12-L1 in the compression study (Section 6.2.2.1).

3.2.1.2 Study of Instantaneous Axis of Rotation (IAR)

Panjabi et al. (1984) and Broc et al. (1997) reported the approximate IAR locations of various thoracic FSUs of T1-T12 and T4-T12, respectively, in their experimental studies, however, no studies were found investigating the IAR of T12-L1 FSU.

In 1984, Panjabi and his co-authors studied the IARs of thoracic segments in the sagittal plane using fresh cadaveric FSUs covering all levels of the thoracic spine from T1-T2 to T11-T12. In their study, six load types, namely anterior and posterior shear forces, flexion and extension moments, and compression and distraction forces, were applied to produce motions only in the sagittal plane. The resulting motions were measured using dial gauges, and statistical methods were used to analyze data collected at various vertebral levels under each corresponding load magnitude and load type. The authors only reported the load type that was significantly related to the location of IARs. Though there was significant variability in the locations of the IAR related to each load type, the average location of IAR for all thoracic segment was 15-45 mm directly below the geometrical center of the moving vertebra.

Broc et al. (1997) performed nondestructive flexibility testing to assess the biomechanical differences between the normal thoracic spine and the thoracic spine after microdiscectomy and to determine whether microdiscectomy results in spinal instability. Eight motion segments (T4-T5 to T11-T12) from five human cadaveric thoracic spines were studied before and after microdiscectomy. Three-dimensional motion was recorded in response to nondestructive, nonconstraining pure moments. Parameters measured included the IAR, the neutral zone, elastic zone, range of motion, and rotational flexibility.

The neutral zone, elastic zone, and range of motion increased by a small amount but significant (average $P = 0.02$ for range-of-motion increase) in all directions after thoracic microdiscectomy (mean bilateral range of motion increase, 2.1° ; range, 0.5° - 4.2°). Flexibility increased slightly under lateral bending and flexion. The locations of IARs usually did not change, but sometimes shifted slightly away from the discectomy site after microdiscectomy. For all eight motion segments, the approximate locations of IARs were reported to be directly below the geometrical center of the moveable vertebra around the geometrical center of the fixed vertebra in sagittal plane, be situated around the superior endplate and directly above the geometrical center of the lower vertebra of the FSU in frontal plane, be located in the anterior portions in transverse plane, respectively. They concluded that thoracic microdiscectomy had small effects on the immediate mechanics and kinematics of the thoracic spine and did not overtly destabilize the motion segments.

3.2.2 Dynamic Tests

As the loading experienced by the spinal column during an actual traumatic event can rarely be duplicated in motion segments, most *in vitro* biomechanical testing of spinal segments is a common method of simulating traumatic injury (Abe et al., 1997; Fredrickson et al., 1992; Kifune et al., 1995, 1997; Langrana et al., 2002; Leferink et al., 2002, 2003; Lin et al., 1993; Oner et al., 1998; Panjabi et al., 1995a, 1995b, 1998; Tran et al., 1995; Willen et al., 1984, 1985; Zou et al., 1993). The trauma models used in laboratory are only an attempt to duplicate the mode of injury encountered in clinical situations.

In thoracolumbar burst fracture simulations, the loading mechanism of injury is commonly believed to be either pure compression or a combined axial load with flexion (Denis, 1983; Panjabi et al., 1995a, 1995b). Usually, three-vertebrae segments (T11-L1 or T12-L2) of human cadaveric or bovine spine were used for dynamic testing with the top and bottom vertebrae mounted and the middle vertebra horizontally oriented. The top vertebra was then impacted axially by falling mass from certain height to produce injuries in the middle vertebra. The following reviewed some dynamic studies carried out by different researchers to explore methods to reduce fracture dislocation, examine the instability of fractured thoracolumbar spine, and quantify the transient canal encroachment, by replicating fractures in experimentally trauma set-up.

3.2.2.1 Investigation of Reduction of Fracture Dislocation

The most characteristic findings in burst fracture are the comminution of the vertebral body and retropulsed bony fragment into the spinal canal. The comminuted fragments may result in severe neurological injury. Hence, the methods reducing the fracture dislocation are important for the clinician to treat this type of fracture.

Willen et al. (1984) carried out experiments on seven human spinal segments T12-L2 exposed to an instant axial dynamic force to produce a burst or crush fracture. They assess the force by which a burst or crush fracture occurs; the resulting damage to the vertebra including encroachment of the spinal canal, injury to the ligaments and intervertebral discs and how the fracture dislocation may be reduced. The experiments were performed in a test set-up, where the prepared segments were placed on a force transducer (Figure 11). A metal drop weight (MDW) with a mass of 10 kg was allowed to fall freely from a height of approximately 2 m to produce an instant, dynamic, axial loading.

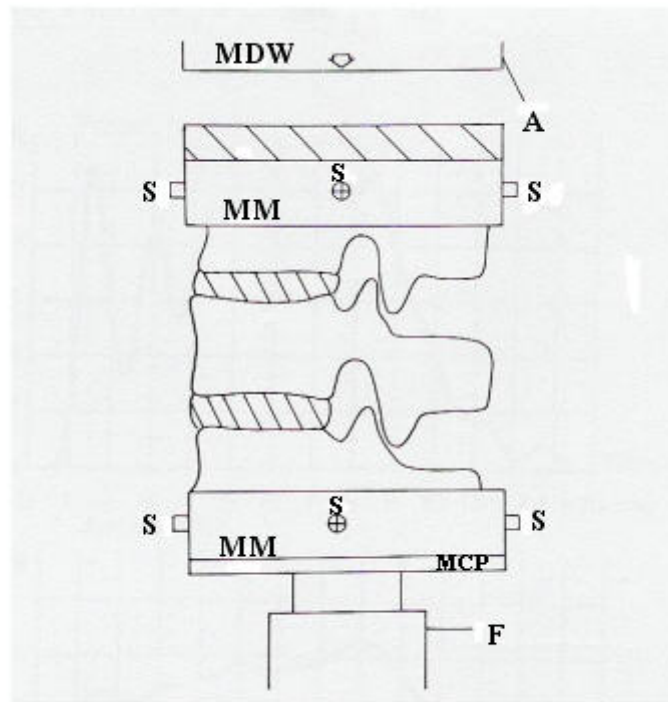


Figure 11 Axial dynamic testing set-up. MDW=metal drop weight; A=accelerometer; PF=plastic foam; MM=metal mould; MCP=metal cylindrical plate; F=force transducer; S=screw fixation. (Adopted from Willen et al., 1984).

After impact, the vertebral preparation then was designed for performing static axial compression of 400 N and distraction of 400 N as well as flexion loading 200 N with a lever of 5 cm and extension loading 200 N with a lever of 5 cm to study the resulting instability of the injured vertebra and how the fracture dislocation may be reduced by these loading conditions.

This study showed that dynamic axial loading caused an initial fracturing of the vertebral body at an average force of 8 kN. Comminution of the vertebral body with severe encroachment of the spinal canal occurred at an average force of about 11 kN. The resulting fractures were similar to fractures observed clinically and

showed a comminuted vertebral body with fractured vertebral endplates, dislocated disc nucleus, bone fragments severely encroaching upon the spinal canal, and facet joint laxity. The flexion-extension range of motion was increased considerably. The authors suggested that the severe thoracolumbar injuries with great encroachment by bone fragments in the spinal canal and fractures of the posterior elements were labeled crush fractures, and this fracture type should be regarded as unstable with a risk of progressive flexion deformity, neurological deterioration and pain. The fracture dislocation could be reduced by an axial distraction but the comminuted vertebral body would extensively narrow the spinal canal under compression. In 1997, Kifune et al. also reported that in the burst fracture specimens, a distraction force was effective in significantly widening the canal, supporting the observations by Willen et al. (1984).

3.2.2.2 Instability Study of Injured Thoracolumbar Segments

A spinal motion segment is considered unstable if it exhibits abnormal large rotational or translational movements under physiological load levels. Information concerning the clinical instability of an injured spine is important when considering a course of treatment for the injury (Panjabi, 2003). Clinically speaking, instability is often associated with hypermobility or hyperflexibility. For this reason, clinical assessment includes the measurement of ROM from functional radiographs, and increase in ROM has been suggested as indicators of spinal instability clinically (Kifune et al., 1995).

In 1994, Panjabi et al. conducted a series of experiments to obtain measures of acute instability of the thoracolumbar fractures by investigating the multidirectional flexibility of the intact specimens pre- and post trauma. The authors produced high-speed axial trauma on thirteen fresh cadaveric thoracolumbar spine specimens (T11-L1) to simulate the burst fracture with disruption of the anterior and posterior walls of the vertebral body (a comminuted vertebral body fracture).

The apparatus for the trauma production consisted of a falling mass impacting the top of the spine specimen. A steel frame, attached rigidly to the concrete floor and ceiling of the laboratory, supported the impounder/linear bearing assembly and a Plexiglas tube, within which was the falling mass (Figure 12). The dropping mass began with 2.3 kg, increased in increments of 2 kg until fractures were observed. The maximum falling mass was 10.3 kg, dropped from a height of 1.6 m resulting in an impact speed of 5.6 m/sec. The fracture pattern produced in the study was similar to that of a clinical burst fracture (Figure 13).

For the flexibility study, the intact and traumatized specimens were subjected to pure moments of flexion/extension, right/left lateral bending and right/left axial torque in three equal steps up to a maximum of 7.5 Nm, and a pair of axial forces of tension and compression up to 300 N to identify load-deformation responses including the neutral zone (NZ) and ROM for the intact and post trauma specimens.

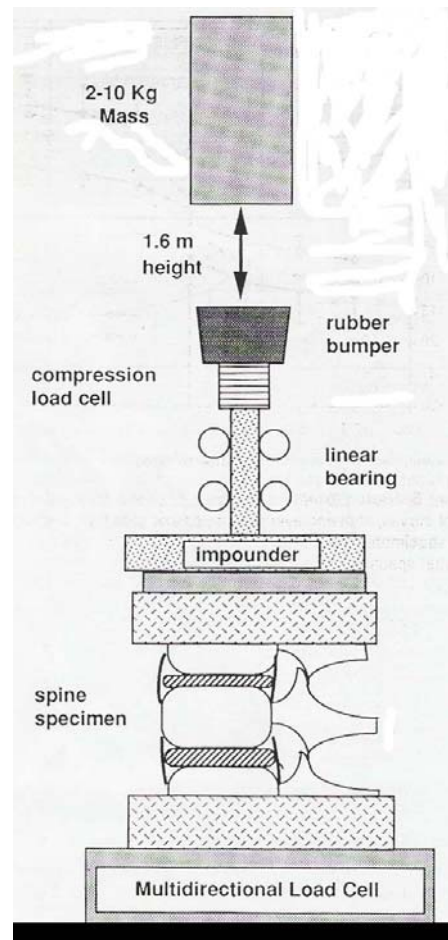


Figure 12 The trauma-producing apparatus.(Adopted from Panjabi et al., 1994).

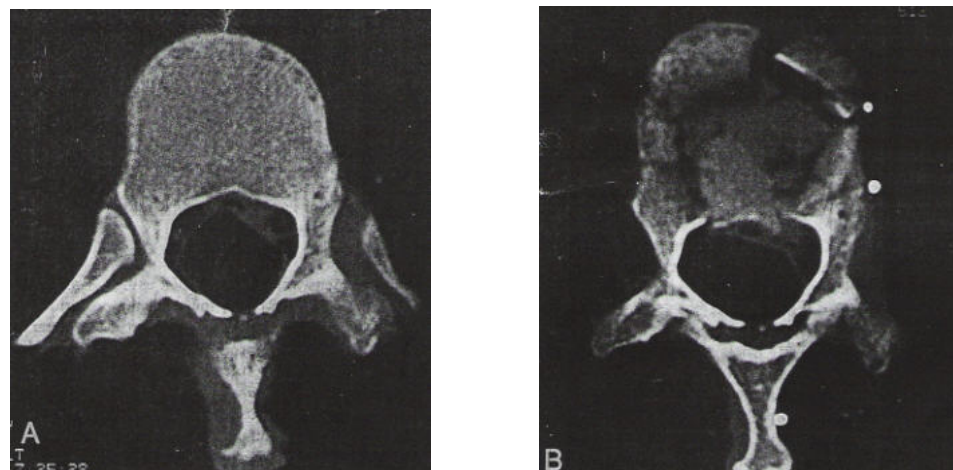


Figure 13 CT scans of a specimen. A: Intact. B: Traumatized, showing a fracture pattern similar to that of a clinical burst fracture. (Adopted from Panjabi et al., 1994)

The authors found that the average ROMs (and NZs) for intact specimens under flexion/extension, axial rotation, and lateral bending of 7.5 Nm as well as compression/tension of 300 N were respectively, 7.9° (1.0°)/6.8° (1.0°), 2.9° (0.2°), 8.0° (1.4°), and 0.6 mm (0.2 mm)/1.6 mm (1.4 mm). The average ROMs of the burst fractures, measured as percentages of the corresponding intact values at 7.5 Nm were 202%, 403%, 266%, and 462% for flexion/extension, axial rotation, lateral bending, and tension/compression, respectively. For the neutral zone motion parameter, the motions of the burst fracture were even greater: 670%, 1650%, 779%, and 650%, respectively. All of the increases were significant. The clinical significance of the study lies in its finding of high multidirectional acute instability of the thoracolumbar burst fracture, especially in axial rotation. Kifune et al. (1995) similarly reported highest instability was seen in the axial rotation for all endplate, wedge, and burst fractures in their experimental studies.

3.2.2.3 Measurement of Dynamic Canal Encroachment

Post-injury CT scans are often used following burst fracture trauma as an indicator for decompressive surgery. Literatures (Panjabi et al., 1995b; Tran et al., 1995; Wilcox et al., 2002) suggest, however, that there is little correlation between the observed fragment position and the level of neurological injury or recovery. This suggests that the spinal cord injuries are related to the location of fracture and the extent that the spinal canal is encroached, and the spinal cord injuries caused during the instantaneous impact. Several experimental studies using different methods had been performed to measure the dynamic events during spinal impact (Panjabi et al., 1995b; Tran et al. 1995; Wilcox et al., 2002).

Panjabi et al. (1995b) produced trauma on 15 fresh cadaveric thoracolumbar spine specimens (T11-L1) to determine the dynamic canal encroachment during thoracolumbar burst fractures. Each specimen was carefully prepared with static canal markers and dynamic canal gauges (Figure 14). To measure the dynamic changes in the spinal canal diameter at several levels simultaneously, the spinal canal transducers were placed in the specimen at the levels of the superior endplate of the middle vertebra (T12), the mid-disc plane of the lower disc (between T12 and L1), and the superior endplate of the lower vertebra (L1) (Figure 14).

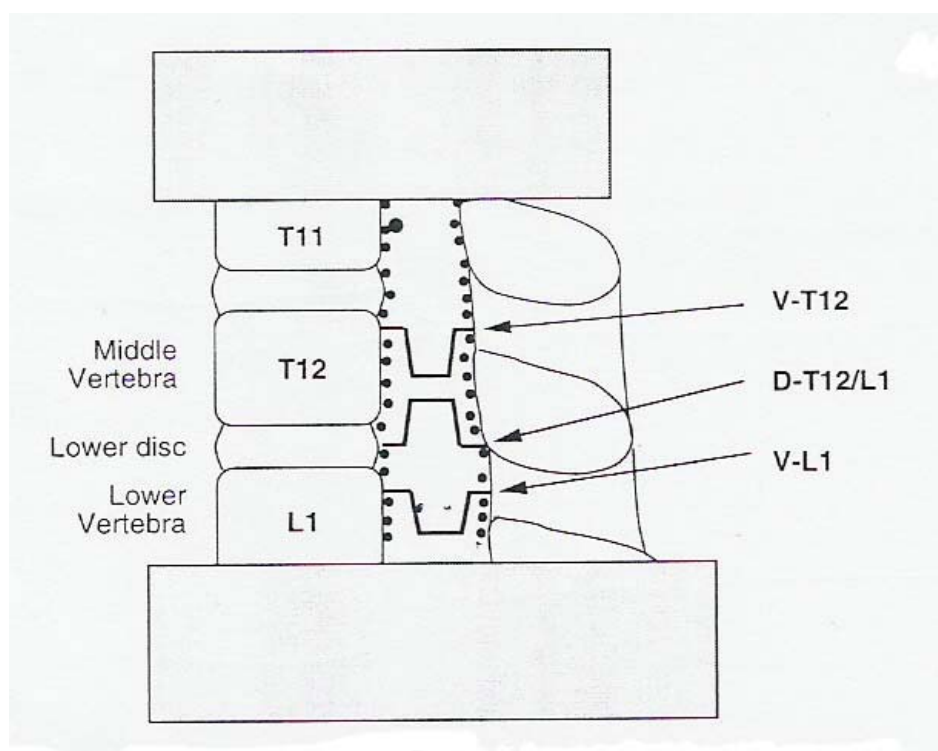


Figure 14 Three-vertebrae T11-L1 specimen with mounts steel balls in the canal. Also shown are three canal gauges to measure dynamic canal diameter at superior endplates of T1 and L1 vertebrae and the disc between T12 and L1. (Adopted from Panjabi et al., 1995b)

A high-speed impact was used to produce a burst fracture, and the canal gauges were monitored during the trauma. The impact trauma apparatus used in this study was shown in Figure 15. A wedge was placed between the impounder and the specimen to force the specimen into flexion so that the mechanism of loading was flexion-compression. The specimens were incrementally impacted, starting with initial 2.3 kg by an increment of 2 kg, in the high-speed trauma apparatus until fracture occurred. The average drop height was 1.36 m, resulting in an average impact velocity of 5.16 m/sec.

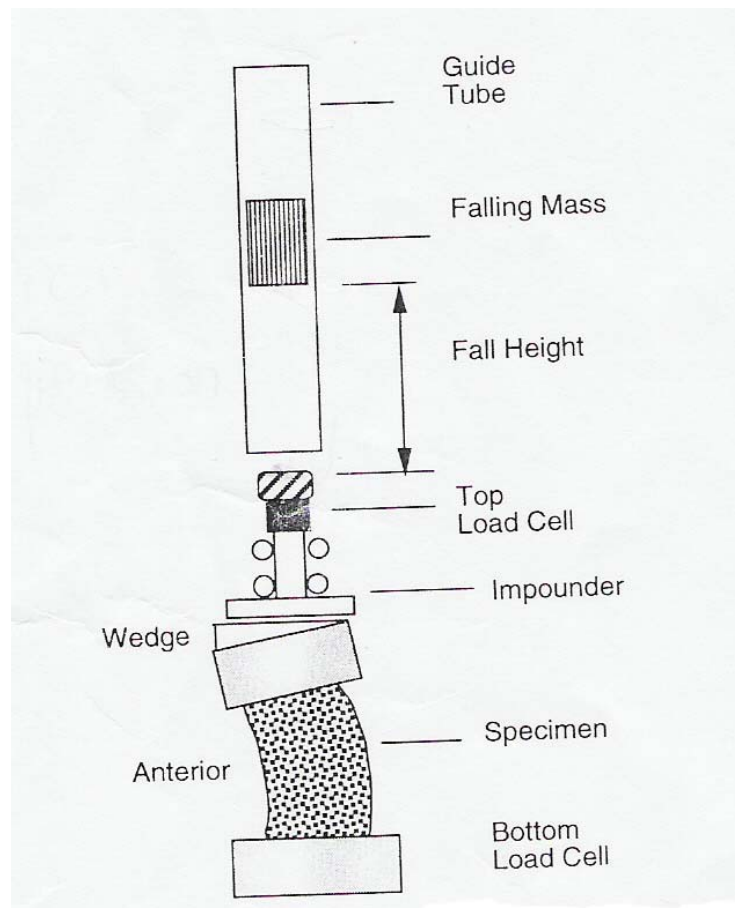


Figure 15 High-speed trauma apparatus to produce the burst fracture. (Adopted from Panjabi et al., 1995b)

During trauma, dynamic canal encroachments were measured by the three specially designed transducers placed in the canal. After the trauma, residual static spinal canal encroachments were measured from the radiographs of the specimens that were prepared with 1.6-mm diameter steel balls lining the canal in the midsagittal plane. The authors found that the average canal diameter was 16.6 ± 1.3 mm and the static canal encroachment was 18.0% of the canal diameter. The corresponding dynamic canal encroachment was 33.3%. Thus, the dynamic canal encroachment was 85% more than the static measurement. Hence, the clinical significance of this study lies in providing awareness to the clinician that the dynamic canal encroachment is significantly greater than the static canal encroachment seen on post-trauma radiographs or CT scans. The finding explained the clinical observation of poor correlation between the canal encroachment measured radiographically and the neurological deficit.

Tran et al. (1995) designed experiments on a total of 12 bovine thoracolumbar motion segments of T9-T11, T11-T13, L1-L3, and L4-L6 (the bovine spine has 13 thoracic segments) to investigate the effect of the loading rate on the occlusion of spinal canal during axial compressive impacts. The specimens were randomly assigned to two groups. After insertion of a transducer capable of measuring transient occlusion of the spinal canal during impact, a low rate of 400 ms (zero to peak load) of axial impact was applied in one group and a high rate load of 20 ms in the other. The direction of impact and total energy delivered by low and high loading rates were the same.

The schematic diagram of the experimental apparatus used in the study of Tran et al. (1995) is shown in Figure 16. The upper and lower vertebral bodies were potted in dental plaster, leaving the middle one free. The low rate axial compressive impact was applied on a group first to determine the mean energy-to-failure, and analysed to obtain the drop weight and drop height for the high loading rate tests. The transient spinal canal occlusion was determined by recording the magnitude of the pressure rise within a flexible plastic tube containing calculating water and placed within the spinal canal of the specimen (Figure 16). After impact, radiographs and CT scans were obtained for all specimens. A representative post-injury CT scans of specimens was shown in Figure 17.

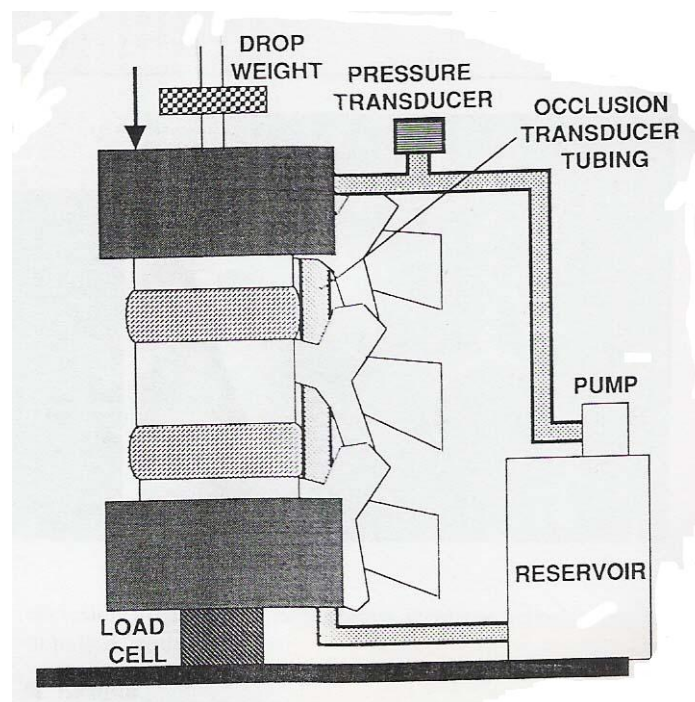


Figure 16 A schematic diagram of the experimental apparatus used in the study of Tran et al. (1995). (Adopted from Tran et al., 1995)

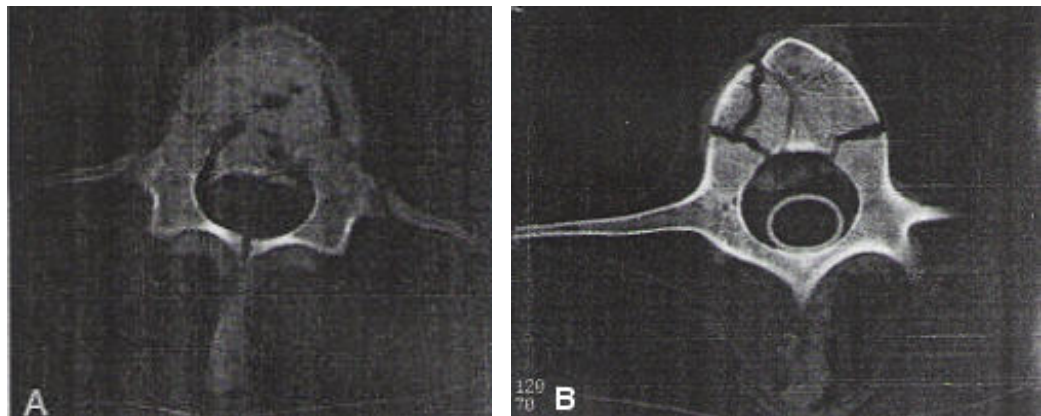


Figure 17 Representative post-injury computed tomographic scans of specimens impacted at (A) a high loading rate and (B) a low loading rate. (Adopted from Tran et al., 1995)

It was found that the post-injury radiographs of the low loading rate (400 ms) group showed compressive fractures with a mean canal occlusion of 6.84%, whereas the high loading rate (20 ms) group had burst fractures with mean canal encroachment of 47.6%. The authors concluded that for randomly assigned vertebral motion segments subjected to axial compressive impact at the same impact energy, those loaded at a higher rate had greater spinal canal occlusion and sustained fractures that likely to involve neurological injury than the low loading rate group.

Wilcox et al. (2002) developed a direct method of measuring spinal canal occlusion during a simulated burst fracture by using a high-speed video technique. The fractures were produced by dropping a mass from a measured height onto nine three-vertebra bovine thoracolumbar spine specimens with T13 in the center in a custom-built rig. The impact energy was in the range of 60 to 140 J. The specimens were constrained to deform only in the impact direction such that pure compression

fractures were generated (Figure 18). The spinal cord was removed prior to testing and the video system set up to film the inside of the spinal canal during the impact. A second camera was used to film the outside of the specimen to observe possible buckling during impact.

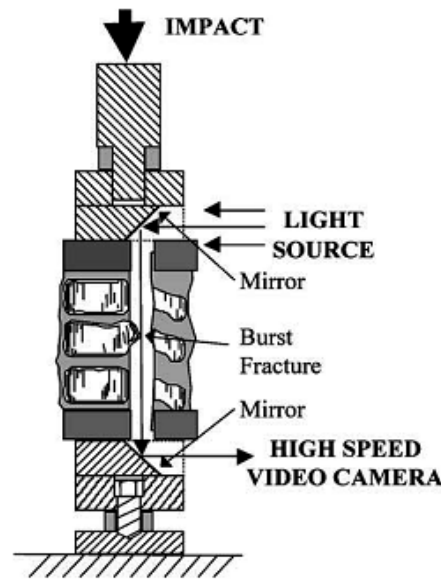


Figure 18 Schematic of experimental set-up used. (Adopted from Wilcox et al., 2002)

The video images were analyzed to determine the change in cross-sectional area of the spinal canal during the event. The images clearly showed a fragment of bone being projected from the vertebral body into the spinal canal and recoiled to the final resting position. The graph of percentage occlusion against time for a typical impact dynamic test showed canal encroachment is significantly larger than the static canal encroachment seen on post-trauma CT scans, and the occlusion obtained by the post-fracture CT scans and the final video images taken at the resting position were in good agreement. This suggested that a burst fracture is a dynamic event, with the

maximum canal occlusion and maximum cord compression occurring at the moment of impact.

3.3 FE Analyses of Thoracolumbar Spine

FE models can provide local and internal mechanical responses that cannot be measured directly in experiments. Hence, FE model simulations are a valuable complement to cadaver experimental models. However, limited FE biomechanics studies of thoracolumbar spine are available in literature.

Gilbertson (1993) constructed three different detailed FE models of the thoracolumbar T11-L1 to simulate the different types of facet orientations in this region. The FE models were built based on the geometrical data obtained from serial CT cross-sections of a cadaver specimen using the commercial FE package ANSYS. These models were assumed sagittally symmetrical (Figure 19), and exercised to investigate the role of facet orientation on thoracolumbar fracture mechanisms. The baseline model (Model T/L) shown in Figure 19 featured an abrupt change in facet orientation, with thoracic-type facet joint articulation between T11 and T12, and lumbar-type facet articulation between T12 and L1. The other two models did not have an abrupt facet orientation transition: Model T/T had thoracic-type facet joint articulations at both levels, while Model L/L had exclusively lumbar-type facet joint articulations.

The models were subjected to complex loading conditions equivalent to 1000 N axial compression plus 7.5 Nm extension bending moment. The predicted

outcomes (e.g., facet contact forces, lamina and vertebral body stress distributions, etc.) were compared among the three models to elucidate the role of the different facet orientations.

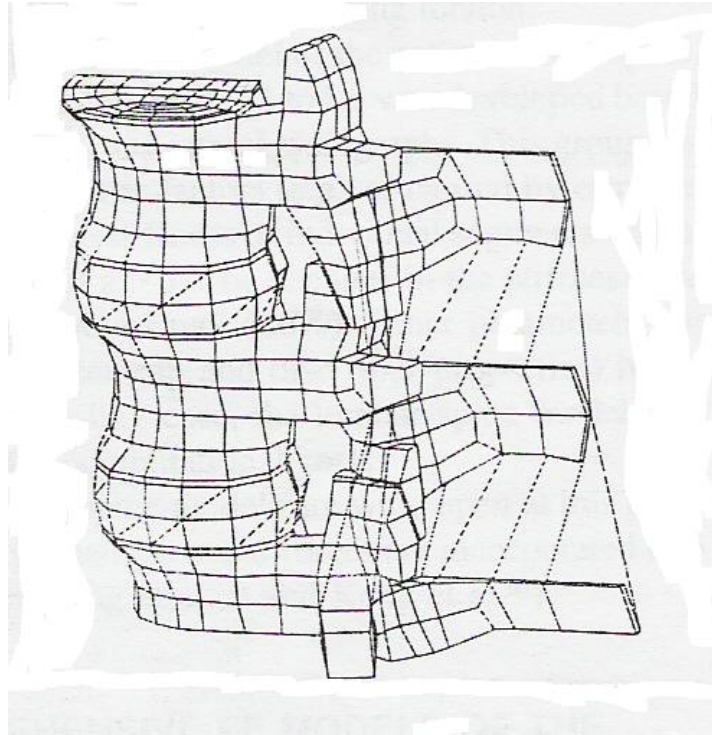


Figure 19 A FE model of thoracolumbar spine segments T11-L1. (Adopted from Gilbertson, 1993)

In all three models, the vertical component or the normal facet contact force was approximately the same (≈ 100 N) regardless of the facet-orientation. However, the normal facet contact forces were approximately 32% higher for the lumbar-type facet articulations than for the thoracic-type facet articulations—this was attributed to the steeper inclination of the lumbar-type facets. These high normal contact forces tended to force the superior lumbar-type facets apart laterally owing to their mainly medial orientation in the transverse plane. This action placed the corresponding

lamina under tension, tending to split the lamina apart in the sagittal plane. Part of this action was also transmitted via the pedicles to the vertebral body, where approximately a 60% increase in von Mises stress in the superior endplate occurred for those vertebrae with lumbar-type superior facets compared to those with thoracic-type superior facets. From this examination of a number of facet orientations, Gilbertson concluded that the abrupt transition itself was not problematic for the loading conditions studied. Rather, the lumbar-type superior facets themselves seemed to have an adverse influence on vertebral body stress. Since the lumbar vertebrae tend to decrease in size cranially, the most vulnerable vertebra will be L1 (or, more generally, the cranial most vertebra with lumbar-type superior facets). The analytical observations from this study are consistent with clinical observations that some burst fractures in the thoracolumbar junction region are accompanied by increased interpediculate distance, vertical laminar fractures, and facet splay. This mechanism may explain the high frequency of certain types of spinal injuries near the thoracolumbar junction.

Gilbertson et al. (1994) developed a simple model (which can span lengthy regions of the spine) of the orthotically stabilized trunk (T1-S1, including the rib cage) combined with a detailed model (which heretofore have been limited to short regions owing to technical limitations) of the T11-L1 motion segments, as shown in Figure 20, to study the biomechanics of orthoses in the treatment of thoracolumbar injuries. The combined model made it possible to apply loads to the trunk and observe the interactions between the orthosis and the detailed intact and injured thoracolumbar segments.

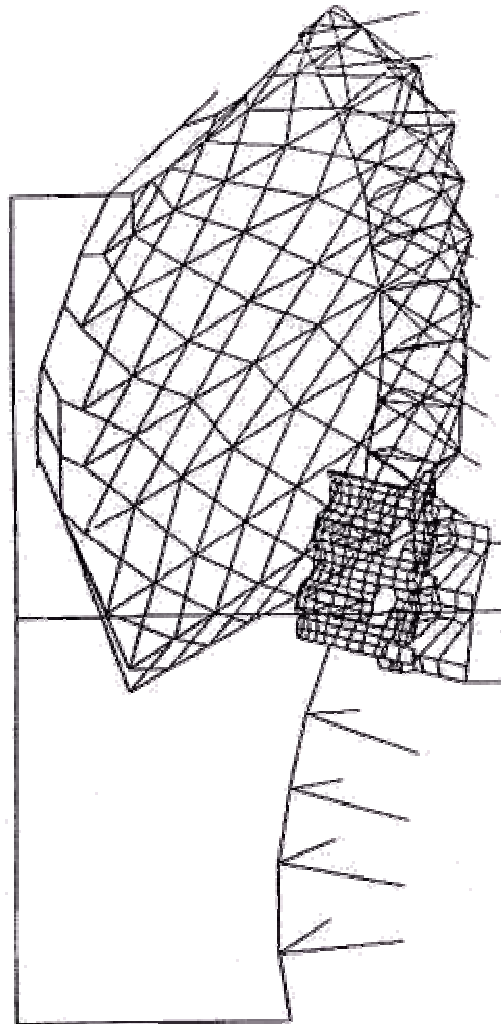


Figure 20 Combined FE model of the orthotically stabilized trunk, featuring detailed T11-L1 motion segments incorporated within a simple beam element representation of the remaining segments of the thoracolumbar spine. (Adopted from Gilbertson et al., 1994)

Under trunk flexion-producing loads, for an injury involving a compromised anterior column, the flexion limitation provided by an initially loose-fitted orthosis reduced stresses in the remaining intact portions of the injured T12 vertebra by about

44% compared to the stresses without an orthosis. The stress relief was not as good (25%) when the orthosis was initially applied with a preload (which produced initial extension of the trunk and individual spinal segments). Clinically, orthosis preload is a variable subjectively controlled by the health care professional at the time of fitting of the orthosis, and it can change with time owing to changes in the orthosis and the patient. Although the effects of gravitational loading and muscles were not considered, this study underscored the importance of establishing (prior to the fitting of an orthosis) whether the primary goal of orthotic treatment in a particular spinal injury case should be done to produce trunk extension or limit trunk flexion.

Liu et al. (2000, 2002) created a three-dimensional FE model of T12-L1 to study the stress distribution of thoracolumbar vertebrae injuries under static pure compression, compression-flexion, and distraction-flexion. The FE model was constructed from the three-dimensional coordinate values of the exterior nodes, measured using a surveying instrument, of an excised and mid-sagittally sectioned T12-L1 vertebrae of a fresh adult cadaver. The final FE model as shown Figure 21 had 337 nodes and 449 elements. The linear elastic material properties were adopted in this study.

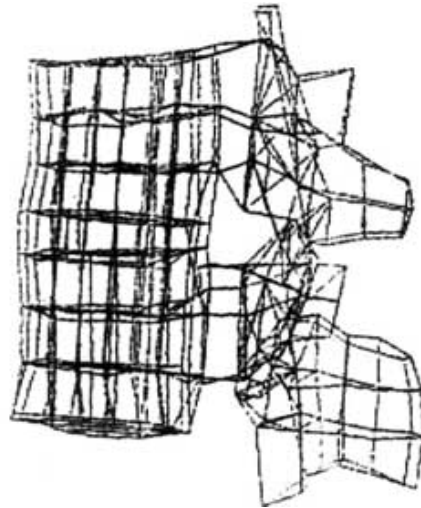


Figure 21 The FE model of T12-L1. (Adopted from Liu et al., 2000)

The intact and degenerated models were used to analyze the stress distribution (Liu et al., 2002). The lower part of the model was fixed, the pure compressive force of 5000 N, compressive force of 5000 N with flexion moment of 60 Nm, and distraction force of 5000 N with flexion moment of 60 Nm, were applied on the superior vertebral body of T12. The stresses of all regions of motion segment were analyzed.

With axial compression and compression-flexion loads, the thoracolumbar motion segment of the normal disc showed that the central part of the upper and lower endplates of the vertebrae and the central part of the trabecular bone adjacent to the endplate were loaded with the most intensive stresses, meanwhile, the posterolateral part of the annulus fibrosus was concentrated with stresses. Degenerative disc showed that the stress distribution of the trabecular bone was relatively averaged, the stresses of the central part adjacent to the endplate were low, while at the same time, the stress of the peripheral part were increased relatively.

With distraction-flexion load, the stresses of the cortex bone, trabecular bone, endplate and annulus fibrosus of the thoracolumbar vertebrae of degenerative discs were low, meanwhile, the stresses of the posterior structure of the vertebral body were relatively increased compared with that of normal discs. The authors concluded that there is a difference in the influence between normal and degenerative discs on the stress distribution of the thoracolumbar vertebrae. Hence, the path of load transfer was changed after disc degeneration.

3.4 Summary

The review of biomechanical studies of thoracolumbar spine shows that both the experimental and FE methods can be used to provide accurate mechanical properties (Markolf, 1972; Panjabi et al., 1976a, 1976b; Oxland et al., 1992), reproduce the spinal injuries (Panjabi et al, 1994, 1995; Tran et al., 1995; Wilcox et al., 2002; Willen et al., 1984), study the instability of the injured spine (Panjabi et al., 1994), investigate the role of individual spinal component in biomechanics (Gilberston, 1993), determine the efficiency of various of instrumentations (Gilberston et al., 1994), and simulate physiological spinal degeneration (Liu et al., 2002). The results of these studies are helpful in understanding the underlying mechanisms of dysfunction and injures, and surgical procedures.

Although a number of biomechanical studies have been conducted to study the kinematics (Markolf, 1972; Panjabi et al., 1976a, 1976b; Oxland et al.,1992) and dynamics (Kifune et al., 1995, 1997; Panjabi et al., 1994, 1995b; Tran et al., 1995; Willen et al., 1984; Wilcox et al., 2002) of the TLJ, there is still a lack of knowledge

on the locations and loci of the IARs of the thoracolumbar junctional segments for the kinematics and the internal transient change of the vertebral bodies during fracture. To date, this internal transient change is impossible to obtain in *in vitro* experimental studies. Furthermore, despite the numerous dynamics studies of the TLJ in literatures, not much has been studied about the mechanism of burst fracture by quantitatively analyzing the internal stress distribution during dynamic process.

Where the *in vitro* studies are inadequate, the FE method can be used to reveal the biomechanical function of the healthy, diseased or damaged spine especially, to obtain the stresses or strains. However, the existing FE models of thoracolumbar spine (Liu et al., 2000, 2002; Gilberston, 1993, 1994) were only used to study the role of the transition of facet orientation, design orthoses for burst fracture, and explain the influence on load transfer of the degenerated disc in thoracolumbar region. These FE models were all only subjected to the static loadings. Heretofore, there has been no attempt to investigate the mechanism of burst fracture under dynamic loadings using the FE method.

This thesis provides the FE analyses of the TLJ to investigate the locations and loci of the IARs under static loading conditions for the FSUs of T11-T12 and T12-L1, and mechanism of burst fracture by analyzing the stress distribution of T12-L1 under dynamic loading conditions.

Chapter 4 MODELING AND VALIDATION OF FE MODELS T11-T12 AND T12-L1

4.1 Introduction

There are several existing FE models of the thoracolumbar junctional segments as described in Section 3.3. However, these models have relatively simplified geometrical representation of facet joints. In human spine, the unique orientation and articulation of facet joints at various vertebral levels determine different motions of the spinal segments under different modes of loadings. Hence, for FE analysis of spine, it is important to characterize the actual geometry of the structure as closely as possible to obtain realistic results applicable to clinical situations.

The FSU is the smallest part of the spine (described in Section 2.3) and exhibits biomechanical characteristics similar to those of the entire spine (White and Panjabi, 1990). Therefore, the primary emphasis of this project was on the development of anatomically realistic FE models of the thoracolumbar junctional FSUs of T11-T12 and T12-L1, based on actual vertebrae T11, T12 and L1, which serve as the fundamental platform for further research applications described in the Chapters 5 and 6.

The predicted responses based on the FE models of the FSUs of the human spine must be representative of the *in vitro* responses. Thus, it is important that the FE models must be firstly validated against experimental data under similar

simulated loading conditions before being used for further investigations. Accordingly, in the following, the details of the modeling of the FSUs of T11-T12 and T12-L1, including the discretization of the associated components, the validation studies, etc are described.

4.2 Finite Element Modeling

To achieve a better geometrical representation of complex vertebrae to be constructed, in particular transitional facet joints of the TLJ, a multi-axis digitizer (FaroArm, Bronze Series, Faro Technologies, Inc) was used to extract the physical geometrical details of the TLJ vertebrae T11, T12, and L1. The geometrical coordinates were then stored in computer for further obtaining of the cross-sections of each spinal components of the vertebra by using an imaging software (Imageware Surfacr 7.1, Imageware, Inc). The coordinate data of cross-sections was imported into a commercial FE software package (ANSYS6.0, ANSYS Inc., Pennsylvania, USA) for processing to develop the FE models. The flow chart for this procedure is shown in Figure 22.

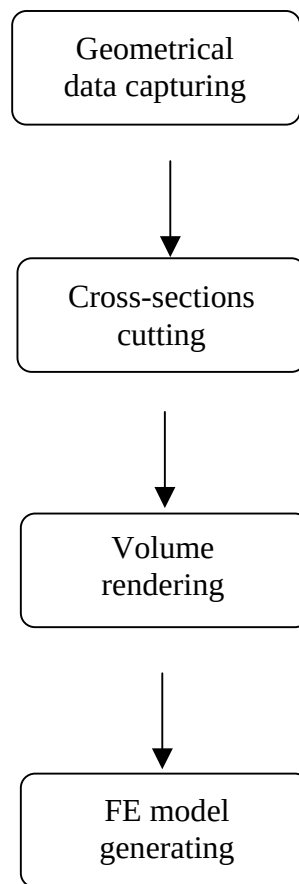


Figure 22 Flowchart showing procedure for construction of FE model

4.2.1 Specimens

In this project, embalmed thoracolumbar vertebrae - T11, T12 and L1, shown in Figure 23, of a 56-year-old male cadaver acquired from the Singapore General Hospital were selected for the three-dimensional coordinates extraction.

Medical records and physical evaluation found the specimen selected to be within normal limits and did not show gross abnormalities physically.

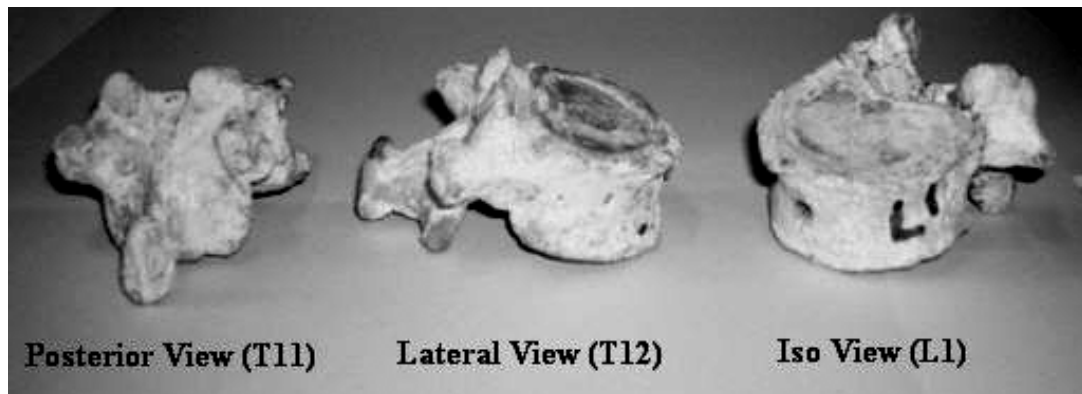


Figure 23 Embalmed vertebrae of T11, T12 and L1 used in the project.

This cadaver belongs to one of the specimens used by Tan et al. (2002, 2004) in their anatomical studies. The age, sex, height, weight and cause of death of the origin of the vertebrae are tabulated in Table 2.

The main dimensions of the T11, T12, and L1 used in the project are listed in Table 3. The geometrical data of selected specimens lie within the mean dimensions of general adults population (Table 1 in Section 2.2.1) reported in the quantitative anatomical studies of the human thoracic and lumbar spines by Panjabi et al. (1991, 1992) and Tan et al. (2002, 2004).

Table 2 Specimen bio-data.

Age	Sex	Race	Body	Height (m)	Cause of	Vertebrae
			Mass (kg)		Death	Extracted
56	M	Chinese	70	1.72	Renal failure	T11, T12, L1

Table 3 Specimen data of embalmed vertebrae T11, T12 and L1.

	Vertebral Level		
	T11	T12	L1
EPWu (mm)	32.1	35.7	37.4
EPWl(mm)	36.0	37.3	42.5
EPDu(mm)	26.5	27.8	29.2
EPDl(mm)	27.8	28.7	29.6
VBHa(mm)	18.9	19.6	21.7
VBHp(mm)	20.8	21.9	22.6
PDHl(mm)	14.8	15.1	14.3
PDHr(mm)	14.5	15.4	14.2
PDWl(mm)	7.5	8.2	6.1
PDWr(mm)	7.2	7.8	5.7
EPAu(mm ²)	751.3	869.7	972.3
EPAI(mm ²)	877.2	957.5	1059.2
SCA(mm ²)	162.1	210.2	266.9

EPWu—Upper endplate width, EPWl—Lower endplate width, EPDu—Upper endplate depth, EPDl—Lower endplate depth, VBHa—Anterior vertebral body height, VBHp—Posterior vertebral body height, PDHl—Left pedicle height, PDHr—Right pedicle height, PDWl—Left pedicle width, PDWr—Right pedicle width, EPAu—Upper endplate area, EPAI—Lower endplate area, SCA—Spinal canal area (refer to Figure 5 in Section 2.2.1 of Chapter 2).

4.2.2 Geometrical Data Capturing

The detailed geometrical coordinates of the cadaveric thoracolumbar vertebrae (T11, T12 and L1) used for FE modeling were obtained using the high definition three-dimensional digitizer (FaroArm, Bronze Series, Faro Technologies, Inc.) (Figure 24).



Figure 24 High definition digitizer (FaroArm) with ball probe used to capture geometrical details of vertebrae.

A threaded rod was inserted through a small hole drilled into the body of each vertebra and secured to allow clamping the vertebra in place for subsequent digitizing using a digitizer connected with a ball probe. Figure 25 shows the set-up for digitizing the vertebra on the flexible arm digitizer.

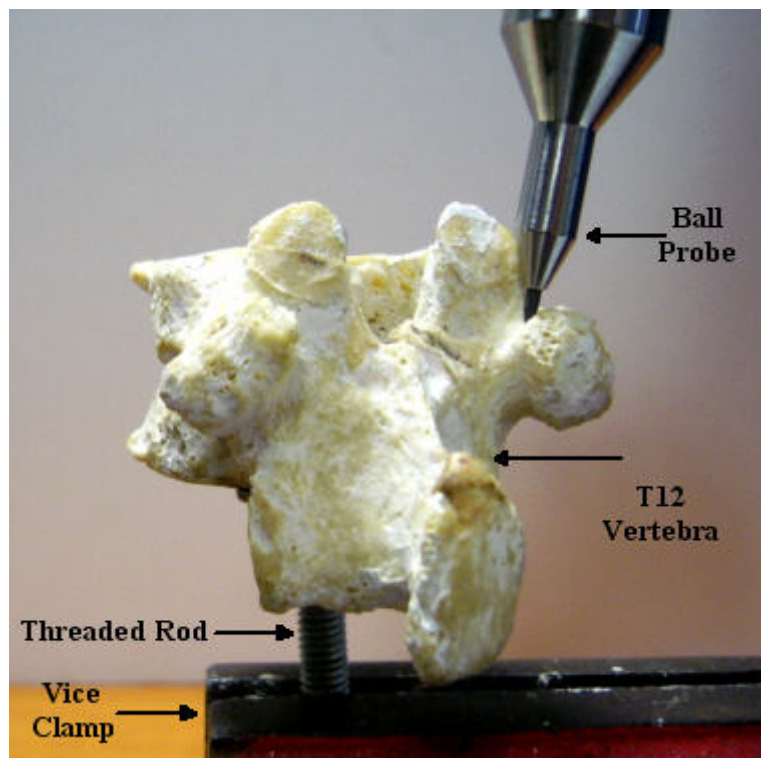


Figure 25 Setup for geometrical data capturing of each vertebra using the digitizer (FaroArm, Bronze Series, Faro Technologies, Inc., Lake Mary, USA). The specimen shown in this Figure is vertebra T12.

Before any digitization, calibration and setting of the reference point were undertaken first. The digitizer was then used to capture the geometrical coordinates on the outer surface area of the whole vertebra at 0.1 mm interval by moving the ball probe over the various surface profile of the vertebra. The automatically registered coordinates data (24152 points) were stored in a computer for further processing. Figure 26 shows the cloud image of the registered data points of thoracic T12 vertebra.

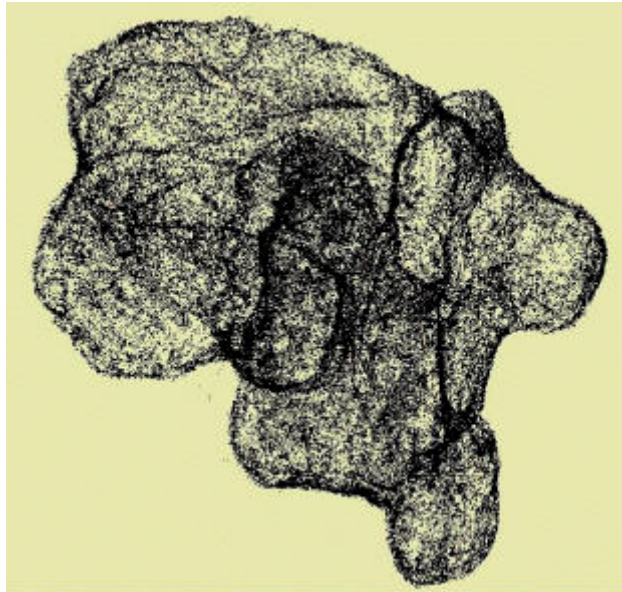


Figure 26 The captured coordinate points of T12 using the digitizer.

4.2.3 Cross-Section Cutting

Using the computer imaging software (Imageware Surfacar 7.1), the registered coordinate data of the cloud image of the vertebra obtained in stage one were processed to obtain the sequential cross section outlines of the vertebra at 1-3 mm intervals, depending on the complexity of the geometry. Since the facet joints play an important role in the motions of the FSU (Markolf, 1972, White and Panjabi, 1990), more sections were cut on the facets of each vertebra for accurate discretization in the FE mesh generation in the next stage. Figure 27 shows the multi-planar sectional views of vertebra T12. The coordinates of these sectional views were saved in data files of universal graphics format (IGES).

Data files (IGES) were then imported into a commercial FE modeling software (ANSYS6.0) for the three-dimensional FE mesh reconstruction.

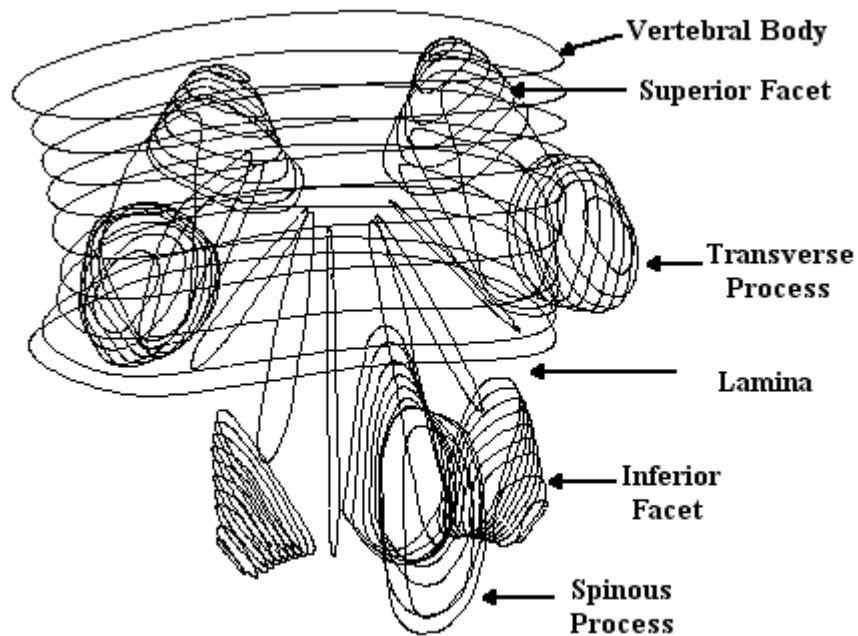


Figure 27 Posterior View of Cross-sections of vertebra T12.

4.2.4 Solid Model Developing

The generation of FE mesh requires the creation of solid volume between the sections. The creation of solid model development included two stages: surface creation and solid volume definition. The solid creation of the bony vertebrae and the intervertebral disc between two vertebrae to form a FSU are briefly described as follows. For the modeling of T11-T12 FSU, the ribs were not considered, as the effect of the ribs at T11-T12 would be minimal, if any, as the ribs are of a floating kind (Oxland et al., 1992).

4.2.4.1 Vertebrae Geometry

In ANSYS 6.0, each closed-loop curve of each cross-section was divided into four segments to meet the minimum requirements of forming an area suitable for volume construction. For creation of areas between sections, the matching curves from each corresponding sections were “skinned” using a built-in NURBS algorithm in ANSYS 6.0 to create a series of surfaces. These surface areas were visually inspected and compared occasionally with actual vertebra for any further modifications to be made to ensure geometric integrity. Repeating the procedure from section to section, the surfaces of the entire model were created as shown in Figure 28. In the next stage, corresponding surfaces of adjacent sections were systematically selected and enclosed to form the solid volume between sections. Repeating the same procedure for all the sections, numerous solid volumes were created and adjoined to form a completed solid model of each vertebra (Figure 28).

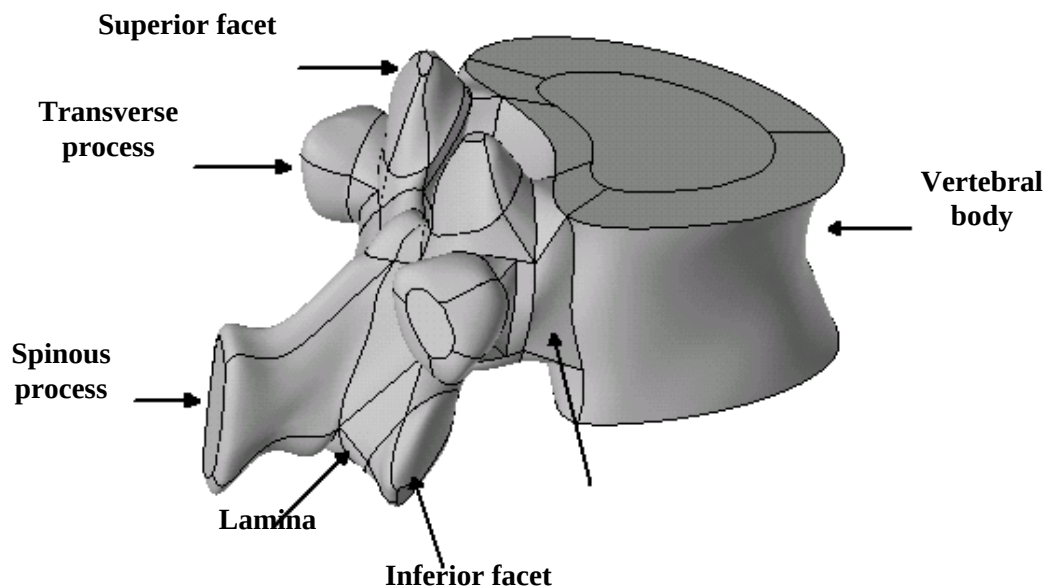


Figure 28 Solid model of the vertebra T12.

4.2.4.2 Intervertebral Disc Geometry

Once the solid models of the vertebrae (T11, T12, and L1) were developed, the next step was to generate the intervertebral discs of T11-T12 and T12-L1 FSUs. In this study, the anatomical data of the intervertebral discs based on published data were used, and the intervertebral disc, modeled as three layers, was stacked between vertebrae ((T11 and T12), (T12 and L1)) to form the FSUs as shown in (Figure 29).

Two layers of solid volumes with thickness of about 0.5 mm (Edwards et al, 2001), representing the superior and inferior endplates of each vertebral body, were modeled. Though the endplates are composed of hyaline cartilage adjacent to the disc and cortical bone of the vertebral body (described in Section 2.2.3.3), however, during adulthood, when the problems of spinal disorders are most likely to occur, the endplates undergo gradually calcification (Bernick and Cailliet, 1982). In addition, Kurowski and Kubo, 1986, and Shirazi-Adl et al., 1986 reported that the cartilaginous tissue does not influence the mechanical properties of a FSU significantly. Accordingly, in this study, the mechanical effects of the cartilaginous endplates were not considered.

A middle layer corresponding to the intervertebral disc consisting of annulus and nucleus was also created. The average values of disc heights were adopted from literature. An anterior height of 6.4 mm and a posterior height of 5.9 mm were adopted for the disc of T11-T12; the anterior height of 7.2 mm and the posterior height of 5.4 mm for the disc of T12-L1 (Brinckmann and Grootenboer, 1991;

Edwards et al., 2001; Giles and Singer, 2000; Kiefer et al., 1997; Malmivaara et al., 1987a; Schultz et al., 1973).

The endplate was assumed to cover the entire contact surface area of the intervertebral disc. The solid model for the entire structure of a motion segment was thus developed after the disc was constructed (Figure 29).

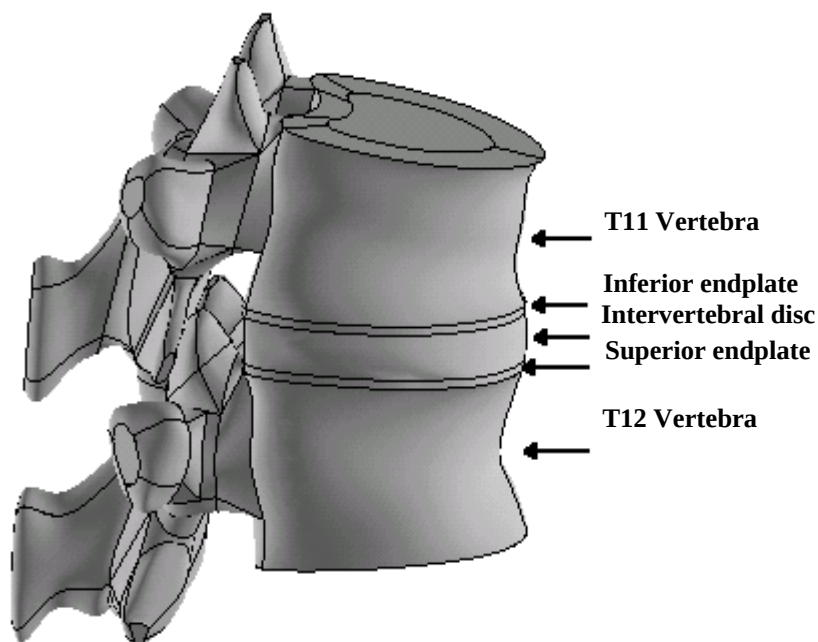


Figure 29 Solid model or volume rendering of a motion segment (T11-T12).

4.2.5 Mesh and Analysis Considerations

After the solid model was defined and developed, the next step was to form element mesh within the solid model. ANSYS allows meshing of solid volume in two modes: free mesh and mapped mesh. Meshing refers to the process of building a model based on the type of elements the user has chosen. When meshing a model the

mesh density may be different at different locations of the model; for example at corners or sharp edges (e.g. facets) a fine mesh (smaller size elements) is necessary, whereas in other areas of the model, a sparse mesh (body and lamina) (larger size elements) can be used. For mapped mesh, the meshing density is manually defined, and the ANSYS FE software will then automatically create the finite elements which conform exactly to the shape of the defined volume. In free mesh, an internal algorithm of ANSYS determines the size and types of the elements.

In order to accurately represent the details of the highly irregular vertebra geometry, different mesh density was assigned for the body and posterior elements, and transition mapped mesh was used to mesh the pedicle. The transition elements change in size from the small elements (posterior elements) to the larger elements (body).

Finally, the FE models of FSUs with geometrical data of bony vertebrae obtained by discretizing, post processed to form the solid model and a mapped mesh technique in the FE generation of T11, T12, and L1 vertebrae, intervertebral disc and endplates, were developed. Figure 30 shows the exploded view of T11-T12 FSU developed.

In this study, eight-noded isoparametric solid elements, instead of four-noded tetrahedron solid elements, were chosen to model the motion segment. The hexagonal elements have superior performance to tetrahedral shaped elements when comparing an equivalent number of degrees of freedom. The use of hexagonal elements also vastly reduces the number of elements and consequently analysis and

post-processing times. In addition, hexagonal elements are more suited for situation where alignment of elements is important to the physics of the problems, such as in the intervertebral disc where the fibers are orientated at $\pm 30^\circ$ to the horizontal plane (Figure 31).

Solid elements were also chosen instead of fluid elements to model the disc nucleus because the fluid element in ANSYS does not support large deflection options. Therefore, when the human thoracolumbar FSU FE model undergoes large deflections, the fluid element, if modeled, will become less accurate due to distortion.

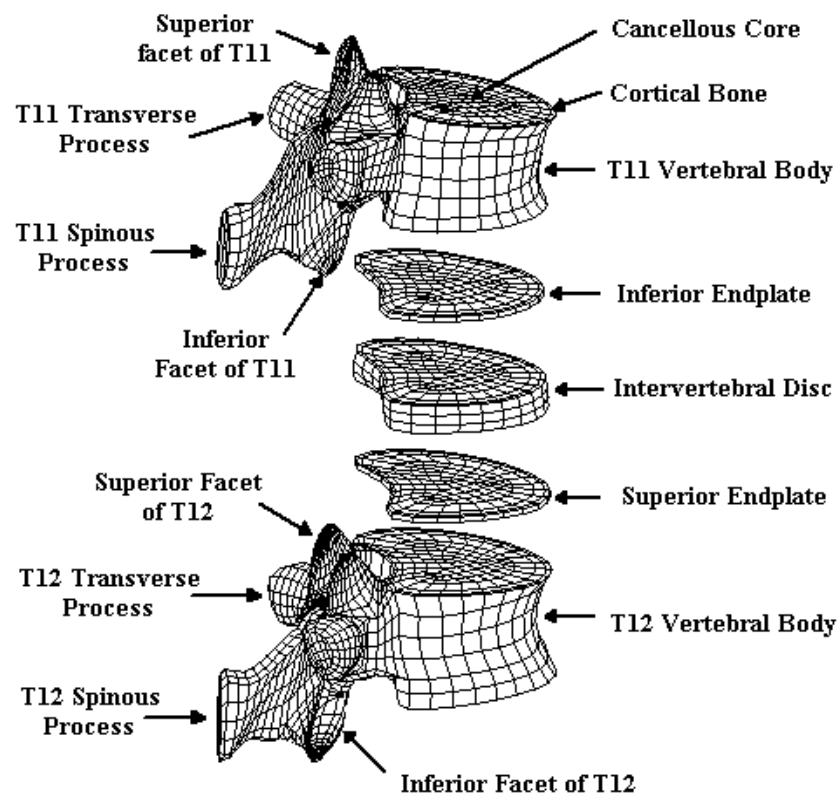


Figure 30 Exploded view of finite element model of T11-T12 motion segment.

4.2.6 The Finite Element Model

After convergence test (mesh sensitivity study), Table 4 summarizes the type and number of elements used to model the various components of the thoracolumbar junctional FSUs T11-T12 and T12-L1. There are a total 12003 elements and 13494 nodes in the FE models.

Table 4 Element types and number used in the FE models of T11-T12 and T12-L1.

Component Name	Element Type	Number of Element
Cortical Bone		532
Cancellous Bone		2508
End plate	8-node brick	640
Bony Posterior Elements		5218
Disc Annulus		588
Disc Nucleus		532
Annulus Fiber		1560
ALL		63
PLL		21
ITL	3D-Cable (Tension only)	12
ISL		18
SSL		6
LF		14
CL		24
Articular Facets	3D surface to surface contact	267

ALL=anterior longitudinal ligament, PLL=posterior longitudinal ligament, ITL=intertransverse ligament, ISL=interspinous ligament, SSL=supraspinous ligament, LF=ligamentum flavum, CL=capsular ligament.

In this project, the vertebral body was assumed to have a cancellous core with the periphery composed of cortical bone (1.0mm thick) (Figure 30). The disc annulus fibrosus was modeled as a matrix of homogenous ground substance reinforced by annulus fibers. It was assumed to consist of 3 consecutive laminar layers (Figure 31). In each layer, the annulus fibers were oriented on average at $\pm 30^\circ$ to the endplate (Shirazi-Adl et al., 1986; White and Panjabi, 1990). The fiber content was assumed to be approximately 19% by volume (Lu et al., 1998; Shirazi-Adl et al., 1986). The fiber diameters or fiber cross-sectional areas, which are averaged in each annulus layer, were calculated based on the fiber volume fraction assumed, the number of fiber elements (simulating the annulus fibers) used, length of the fibers, and the volume of each annulus layer. The annulus fibers were simulated by 3D cable elements that sustained tensile stress only. In the FE model of each FSU, 8 layers of fibers were formed for each disc (Figure 31). The ratio of the cross-section area of nucleus was assumed to be 40% (Depalma and Rothman, 1970) of disc area. The nucleus pulposus of the intervertebral disc was shifted posteriorly by 1.5mm (Benzel and Stillerman, 1999; Depalma and Rothman, 1970).

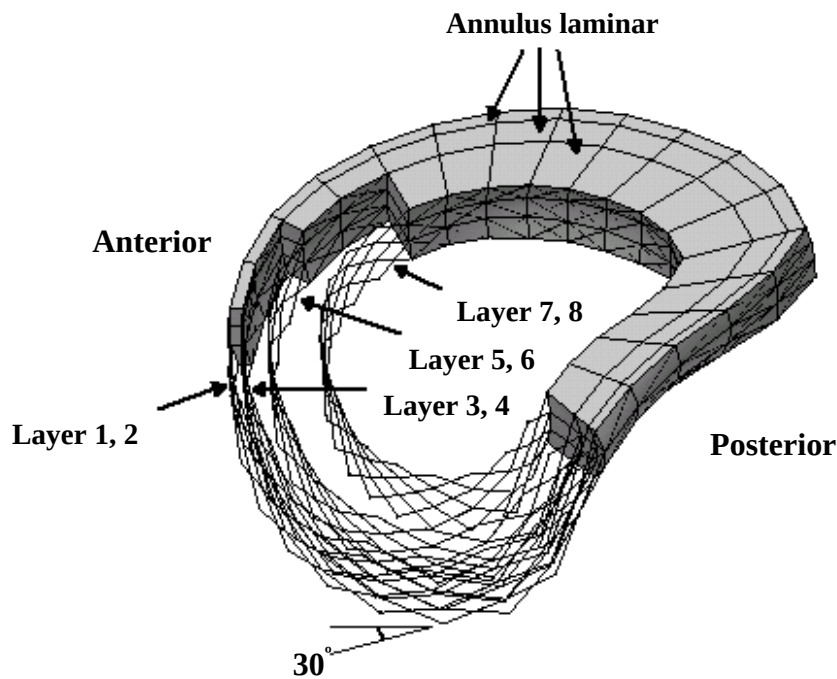


Figure 31 Fibers arrangement in the annulus fibrosus layers of the intervertebral disc (T11-T12) in the finite element analysis.

The friction between the facets is negligible due to synovial fluid in reality (Goel et al., 1988; Zander et al., 2002). Thus, in order to appropriately model the changing areas of contact of facet articulation surfaces with increments in loading, the facet articulations were modeled using sliding surface contact frictionless elements (Goel et al., 1988; Zander et al., 2002).

Seven ligaments were incorporated in the model: the anterior and posterior longitudinal ligaments, ligamentum flavum, the intertransverse ligament, the capsular ligament, the interspinous ligaments, and the supraspinous ligament were modeled by 3D tension-active cable elements (Figure 32). Their insertion points were chosen

based on anatomy books (Benzel and Stillerman, 1999; Blandine, 1993; Giles and Singer, 2000). Figure 33 shows the FE model of T12-L1 FSU with ligaments.

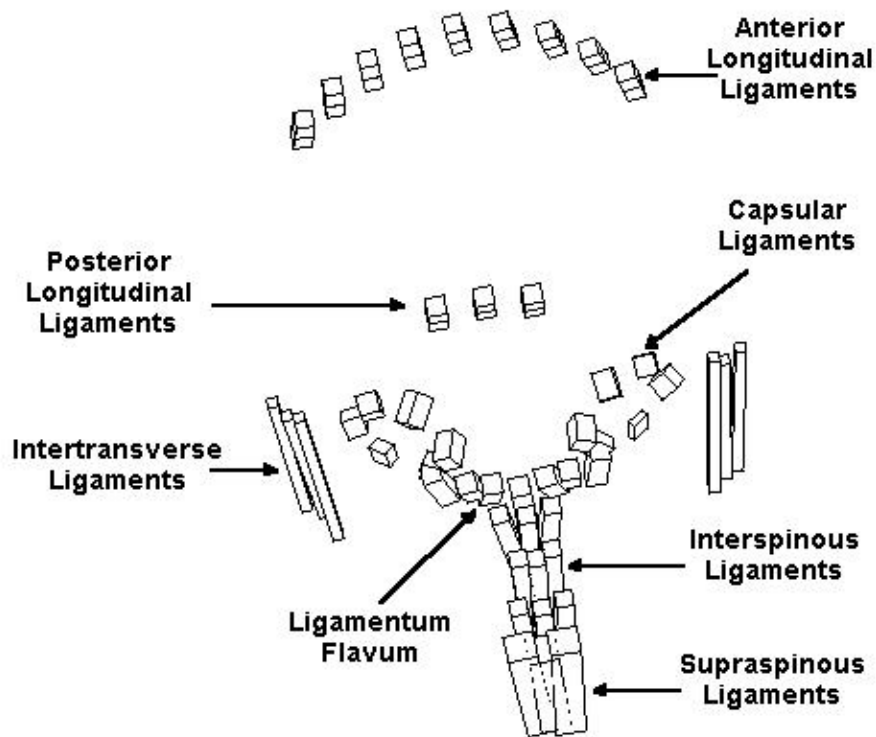


Figure 32 Finite elements (3D-cable element) used to model ligaments.

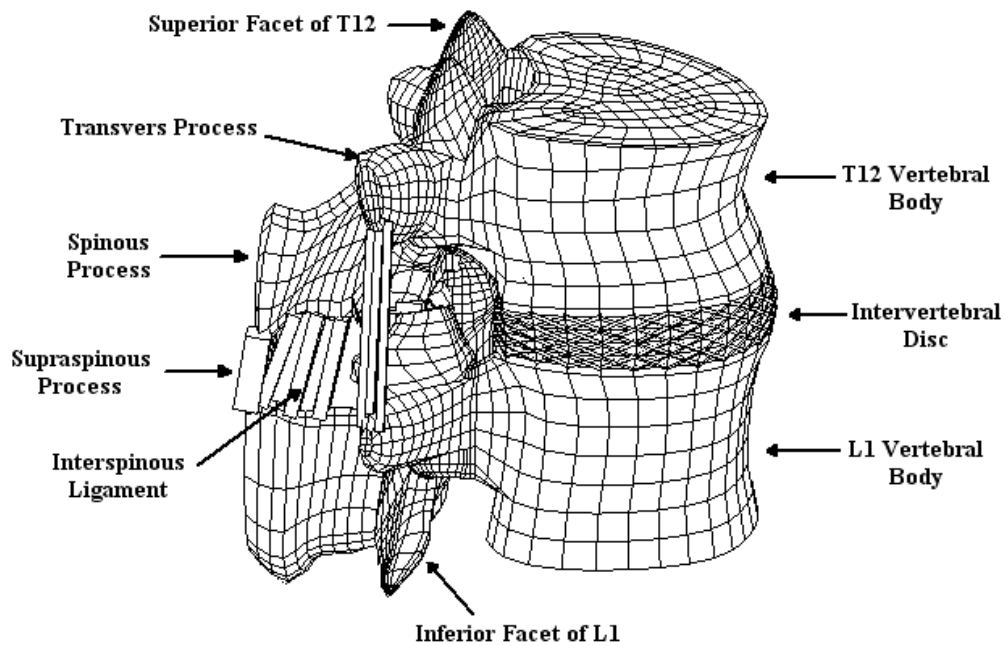


Figure 33 Finite element model of T12-L1.

4.2.7 Discussion

The preceding sections describe the procedures involved in the generation of FE models of T11-T12 and T12-L1 FSUs. The whole procedure involving the digitization process and modeling technique in ANSYS is elaborated herein to illustrate the feasibility of the modeling technique used to generate the anatomical detailed three-dimensional FE models of human thoracolumbar junctional FSUs.

The anatomical realistic three-dimensional FE models of T11-T12 and T12-L1 were developed based on geometrical data, obtained using the digitizer (FaroArm), of the embalmed thoracolumbar vertebrae (T11, T12, and L1). The use of the digitizer to extract crucial geometry features of the vertebra for model generation (Ng et al., 2003; Qiu et al., 2003; Teo et al., 2001, Teo and Ng, 2001)

provided an alternative means of modeling of human spine to the current predominant method of CT scans in literatures (Crawford et al., 2003; Gilbertson, 1993; Kumaresan et al., 1999, 2000; Liebschner et al., 2003; Yoganandan et al., 1996, 1997).

The complex geometry of the bony posterior elements, attached to the vertebral body, complicated the modeling of the thoracolumbar spine and demanded integration of an appropriate digitization process with pre-processing for volume mesh construction. The digitizing procedure used in this project had not only reliably preserve the intricate topography of the original vertebra for FE modeling, but also provided a better realistic approach to determine the influence of spinal components in the gross responses under different load configurations. Digitizing the whole surface of the vertebra by the digitizer (FaroArm), all the important anatomical features of the thoracolumbar spine vertebrae, such as the facet articulation surfaces, lamina, spinous process, transverse process and pedicle, were able to be explicitly defined in this unprecedented model, as the geometrical data of the whole surface of the vertebra can be sectioned along different directions for each individual spinal component in the computer software (Imageware Surfacr) based on the complexity (Figure 27) for mesh generation in ANSYS. In contrast, Gilbertson (1993) and Liu et al. (2000, 2002) assumed planar facet geometry in their studies for articulating facets, which lacked true geometrical form and orientation.

Furthermore, with the solid geometry built in ANSYS and followed by volume mesh therein, the whole process lessened data interoperability problems, ensured geometry integrity and increased suitability for successful simulations. This

was an added advantage as compared to modeling from CT scan images because of a need of a sophisticated edge detection algorithm to extract vertebral cross section outlines and the sequential stacking of the coronal and sagittal images to form 3D solid models (Gilbertson, 1993). Mesh representation from CT scans depended on how efficacious the algorithm is.

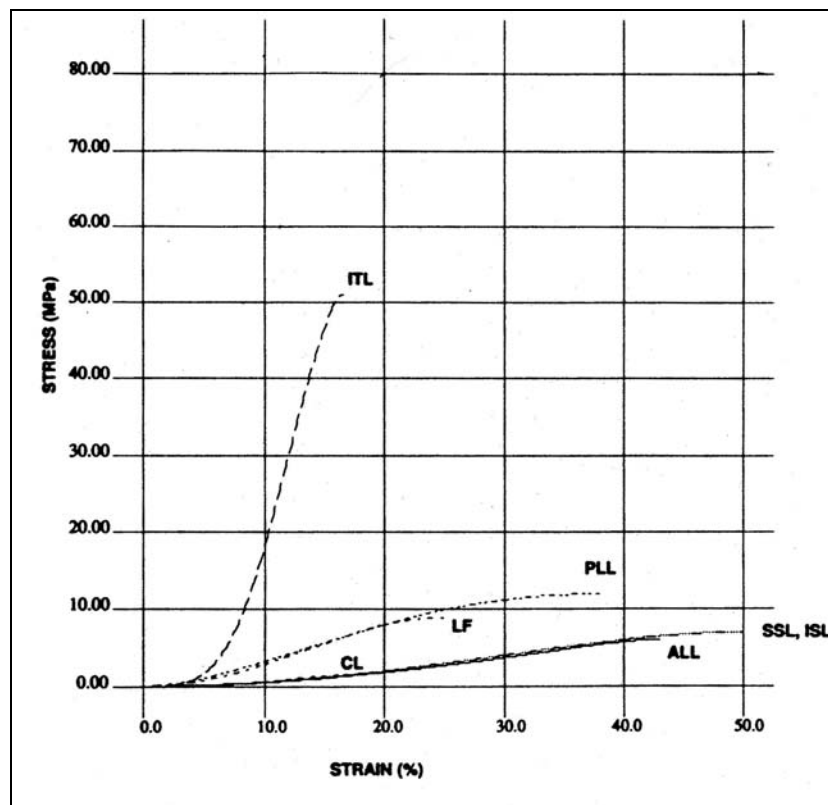
The digitizer was capable of quantifying the human anatomy, thereby achieving the goal of realistic geometry representation. The anatomically realistic FE models of the FSUs T11-T12 and T12-L1 provided a fundamental platform for biomechanical research applications in the following Chapter 5 to identify the locations and loci of IARs, and in Chapter 6 to investigate the mechanism of burst fracture.

4.3 Model Validation

As FE study of any biological component requires several simplifications and assumptions made, in particular to material properties, that would affect the predicted results, the validation process provides a certain level of reliability for the FE model being utilized within the validated range for subsequent analysis. Validation studies form the vital link between the development of the FE model and its final intended usage. Accordingly, the present models were validated under all physiological loading modes - flexion/extension, left/right lateral bending, and left/right axial rotation, by comparing the FE models predicted results with published experimental results obtained by Markolf (1972) and Oxland et al., (1992) in terms of axial rotational displacement responses and ROMs.

4.3.1 Material Properties

Another important feature of the FE analysis is the definition of the material properties of the individual components. All material properties used to model various spinal components of the FE models were assumed to be linear elastic except for the ligaments, which possessed derived nonlinear stress-strain behavior from the experimental study conducted by Chazal et al. (1985), as shown in Figure 34. The cross-sectional areas used for various ligaments were averages of the values reported previously (Chazal et al., 1985; Goel and Weinstein, 1990; Sharma et al., 1995) (Table 5).



ALL=anterior longitudinal ligament, PLL=posterior longitudinal ligament, ITL=intertransverse ligament, ISL=interspinous ligament, SSL=supraspinous ligament, LF=ligamentum flavum, CL=capsular ligament.

Figure 34 Stress-strain curves for ligaments used in the FE models (Adopted from Chazal et al., 1985).

Table 5 Cross-sectional areas of spinal ligaments used in this study..

Ligament	ALL	PLL	ITL	ISL	SSL	LF	CL
Area (mm ²)	35	19	2	29	29	33	21
ALL=anterior longitudinal ligament, PLL=posterior longitudinal ligament, ITL=intertransverse ligament, ISL=interspinous ligament, SSL=supraspinous ligament, LF=ligamentum flavum, CL=capsular ligament.							

Table 6 Material properties used in the study.

Component name	Young's Modulus E (MPa)	Poisson's Ratio ν	Reference
Cortical bone	12000	0.30	Brown et al.(1981)
Cancellous bone	345	0.2	Keaveny et al. (2001)
Posterior elements	3500	0.25	Kumaresan et al. (1999)
Endplate	12000	0.30	Brown et al.(1981)
Disc-annulus	4.2	0.45	Yamada (1970)
Disc-nucleus	1	0.499	Lavaste et al. (1992); Sharma et al. (1995)
Annulus fiber	500	0.3	Wu and Yao (1976); Haut and Little (1972)
Ligaments	Material nonlinearity		Chazal et al. (1985)

A Young's modulus of 1 MPa and a Poisson's ratio of 0.4999 were adopted for the nucleus pulposus to simulate its nearly incompressible behavior (Table 6). The cancellous bone showed anisotropic material property (Silva and Gibson, 1997; Ulrich et al., 1999). However, in this study, it was assumed to be homogenous and isotropic with linear elasticity, this has only a minor influence on the deformation of

the motion segments, since the stiffness of the intervertebral disc is much lower than that of a vertebral body (Zander et al., 2002).

Geometrical non-linearity was incorporated in this study to account for the relatively large deformation that occur in the loaded intervertebral disc by simulating annulus fibers embedded in ground substance and surrounded nucleus. In mechanics, non-linearities are often classified into two types, material and geometrical. The non-linearities exhibited by the FE models of FSUs of T11-T12 and T12-L1 in this study must be of the geometrical (large deformation) and material (non-linearity of ligaments) types.

Many widely scattered values of material properties of each spinal component exist in literatures. The author adopted the following approach for the material property assignment. A sensitivity analysis study was conducted to determine the effect of variations in the material properties on the biomechanical responses under all physiological loading modes. Material properties of each component were varied within the range of values reported in the literatures until an optimum match of the model output and experimental data under all loading cases. Numerous iterative analyses were performed to obtain the specific set of material properties, as shown in Table 6 and Figure 34, that yielded an optimum match to the experimental output. The basic material properties summarized in Table 6 were adjusted values after sensitivity study of the material properties.

4.3.2 Boundary and Loading Conditions

A standard right-hand Cartesian coordinate system established by White and Panjabi (1990) was adopted (Figure 35) in the study. The system was defined as follows: the geometrical center of the upper vertebral body was chosen as the origin, inferior to superior --- positive Y-direction, posterior to anterior --- positive Z direction, and left to right --- positive X-direction. The coordinate system was fixed in space and therefore could record the motion of the upper vertebra.

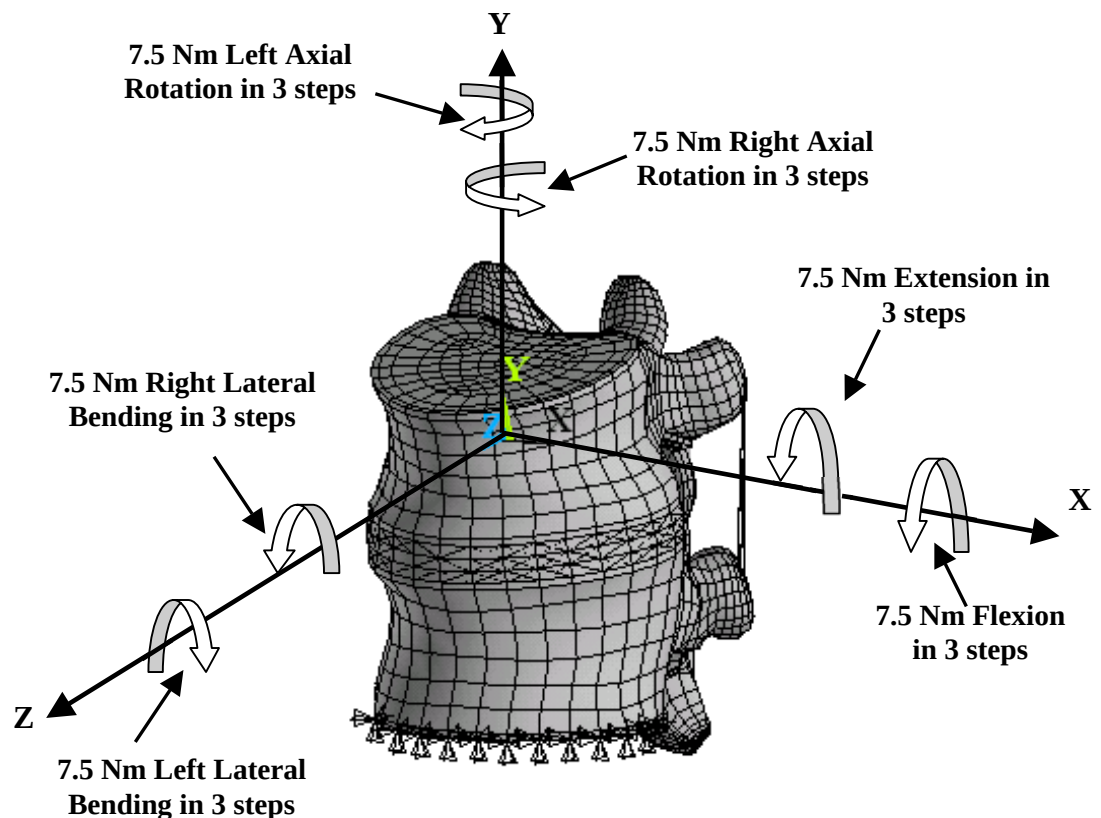


Figure 35 Boundary and loading conditions in a standard three-dimensional coordinate system (T11-T12).

In the validation study, the loading and boundary conditions (depicted in Figure 35) were replicated according to published experimental studies conducted using human cadaveric thoracolumbar spine (Oxland et al., 1992 and Markolf (1972)). In the study by Oxland et al (1992), pure rotational moment of magnitude 7.5 Nm in 3 incremental steps were applied to the moveable vertebra of T11-T12 and T12-L1 segments to create flexion/extension, left/right lateral bending and axial rotation. However, in the study by Markolf (1972), load magnitude of 6 foot-pound (about 8.12 Nm) was applied to moveable vertebra of T12-L1 segment. The load was statically applied to the specimen by suspending weights on weight pans attached to lines running through pulley system to each end of the loading rod. Accordingly, in the FE validation study, the inferior surface and spinous process of the inferior vertebra was fixed in all degrees-of-freedom and pure moments of magnitude only up to 7.5 Nm applied incrementally in 3 steps were applied to the superior surface of the superior (moveable) vertebra to match published experimental study (Oxland et al., 1992) used for validation, the “3 steps” means that the pure moment of 7.5 Nm was applied incrementally in 3 steps:- 2.5 Nm, 5.0 Nm and 7.5 Nm to the FE model.. The predicted FE results were compared against Oxland et al.,(1992) and Markolf (1972) findings appropriately.

The analyses were carried out on SGI Origin 3400 Servers (32 CPUs, MIPS 64 Bit R12000 400MHz/8MB RISC, 44.8 GB/sec System bandwidth, 2 x 18 GB System Disks, IRIX Operating System, On board memory 32GB, TP 9100 Storage, Silicon Graphics Inc.) at the Centre for Advanced Numerical Engineering Simulations (CANES), using a general FE code, ANSYS6.0.

4.3.3 Calculation of Angular Rotation

As the stiffness of the moveable vertebra was larger than that of the intervertebral disc of the FSU, the moveable vertebra was assumed to be rigid and non-deformable. To compute the angular motion of the moveable vertebra with respect to the fixed vertebra, the coordinates of two nodes, before and after loading, along the plane of applied moment were selected on the each of the superior surfaces of superior and inferior vertebrae of each segment (T11-T12, T12-L1) were processed for the angular motion in each corresponding plane of applied moment vector.

Figure 36 shows the schematic calculation of angular rotation degree under flexion in sagittal plane. Points A and B are the two nodes selected on the top surface of the moveable vertebra, points A' and B' show the positions of A and B after flexion, respectively. The angular motion (in degrees) after flexion was calculated in the following manner:

$$\text{Angular motion} = \alpha' - \alpha$$

where, α is the angle between line segment AB and Y-axis,

α' is the angle between line segment A'B' and Y-axis.

The same calculation of angular motion was used under extension, left/right lateral bending, and left/right axial rotation. Sample calculation of each of the six angular rotations of T12-L1 FSU corresponding to pure moment of 7.5 Nm was included in the Appendix B.

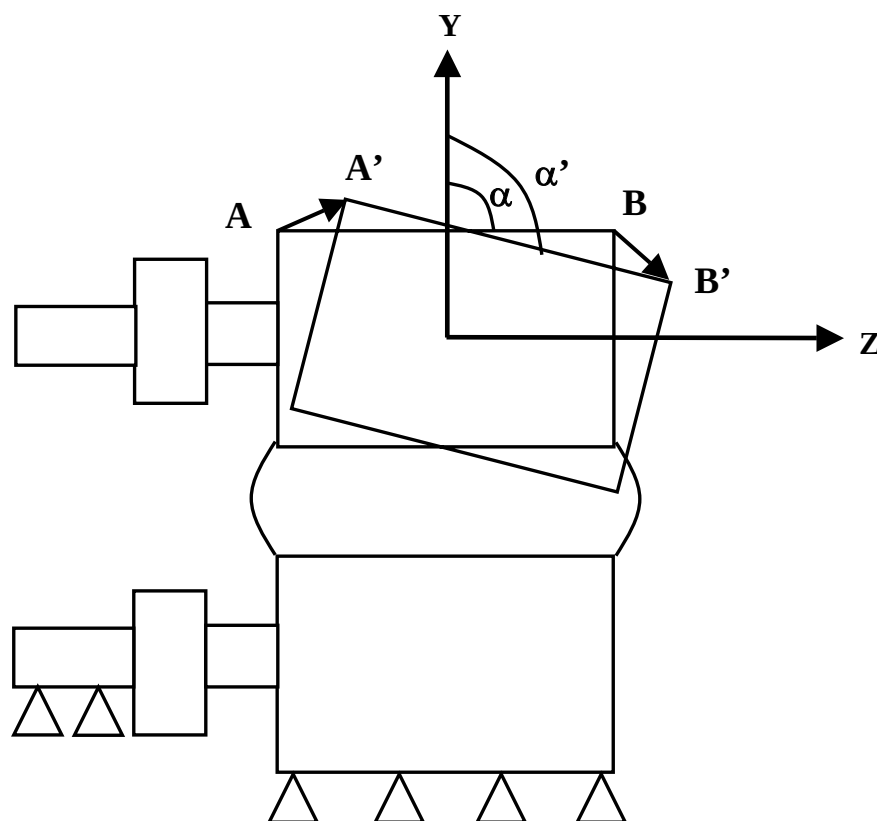
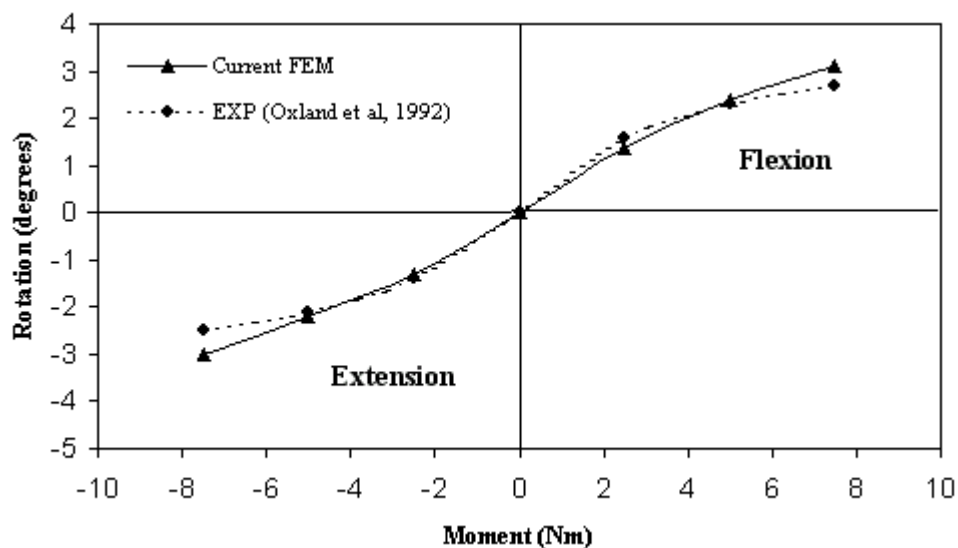


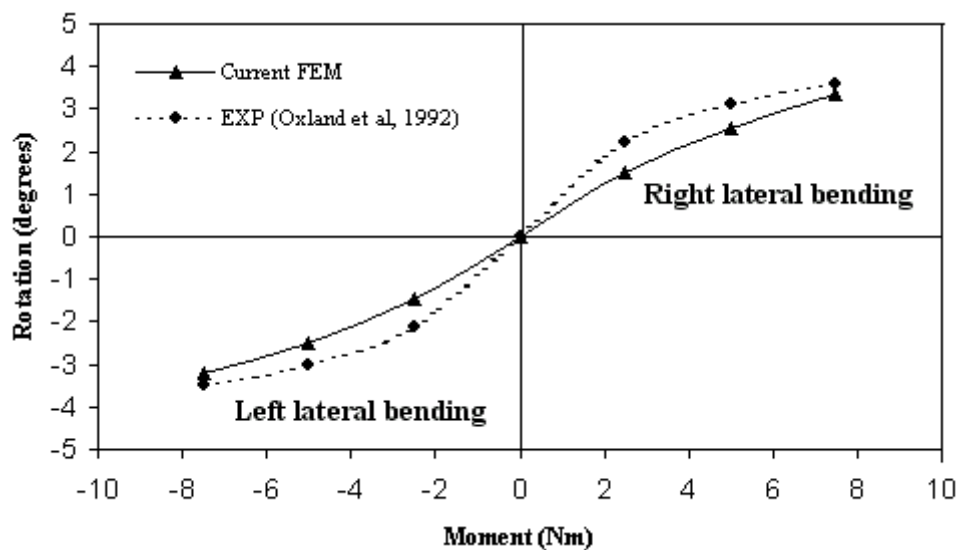
Figure 36 Schematic illustration of the calculation of the angular motion under flexion in sagittal plane (A and B are the two nodes selected on the top surface of the moveable vertebra, A' and B' show the positions of A and B after flexion, respectively).

4.3.4 Results

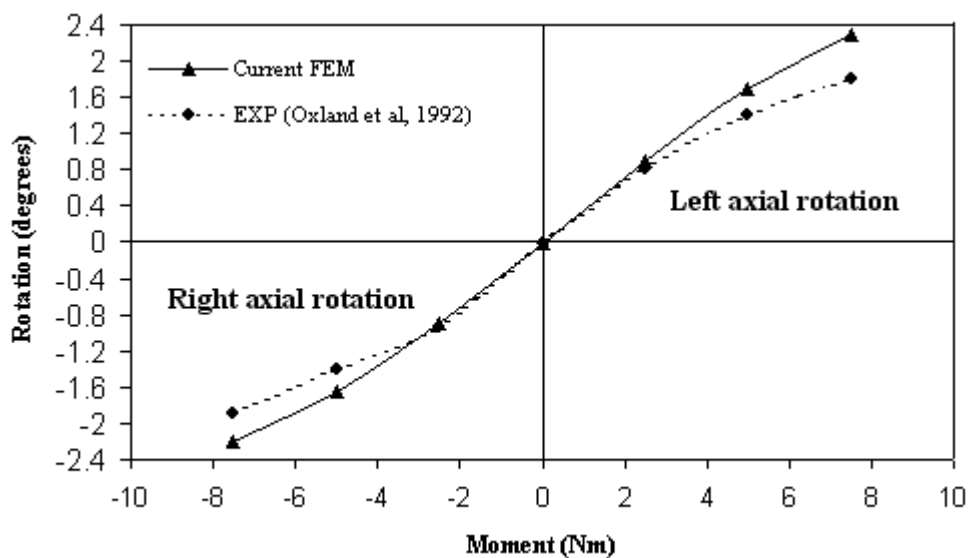
The predicted results from the FE analyses are shown in Appendix C and plotted in Figures 37-39. The load-angular deformation curves predicted by the FE models of T11-T12 and T12-L1 under pure flexion/extension moments, pure left/right lateral bending moments and left/right axial torque were compared with those obtained from published experimental studies (Markolf, 1972; Oxland et al., 1992) (Figures 37 and 38). The predicted ROMs, corresponding to the final load steps under all loading modes were also compared in Figure 39.



A. T11-T12 Flexion and Extension

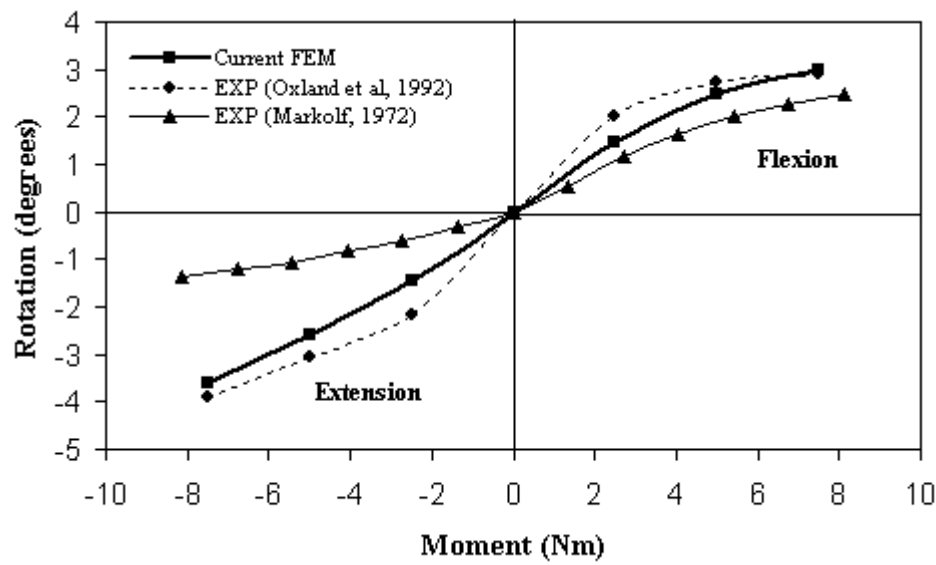


B. T11-T12 Left and Right Lateral Bending

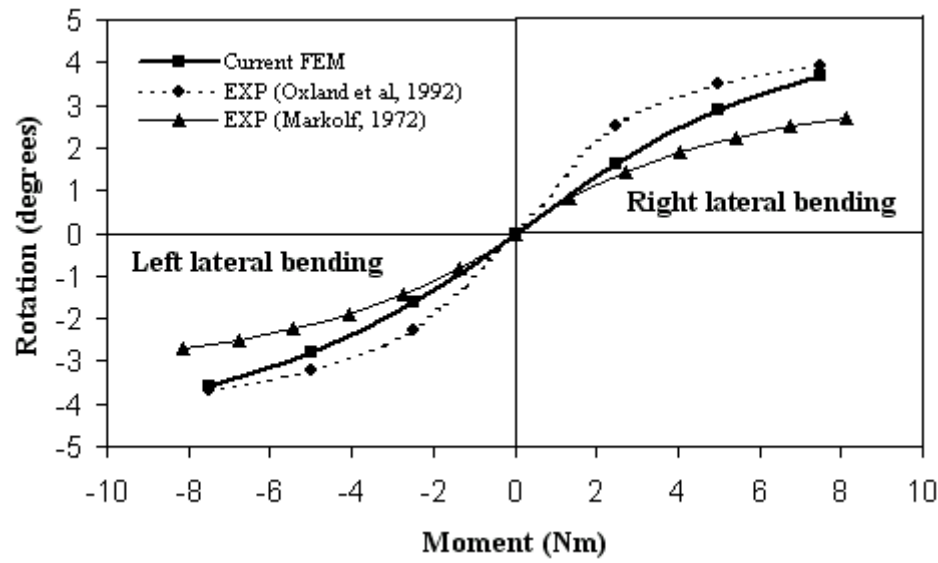


C. T11-T12 Left and Right Axial Rotation

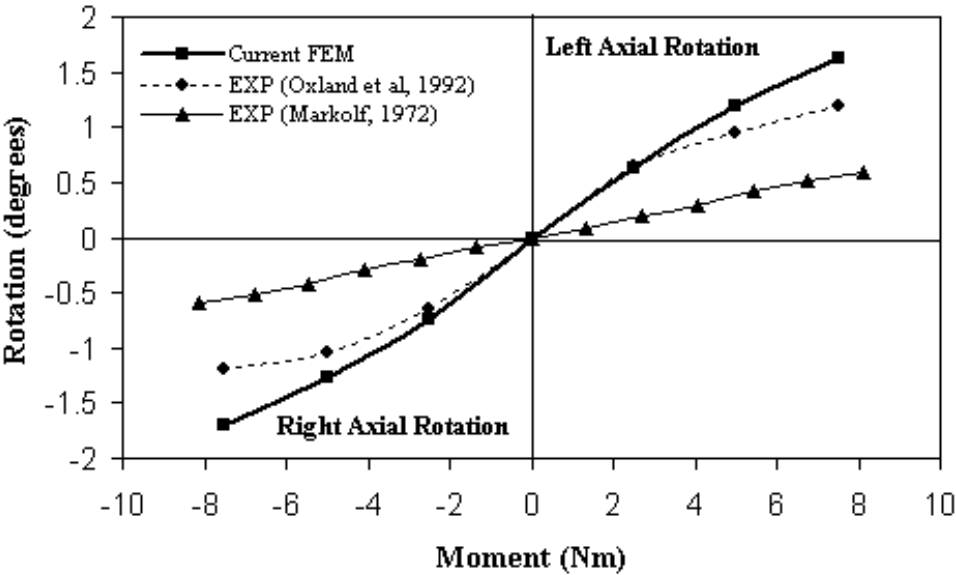
Figure 37 Comparison of load-angular deformation curves predicted by FE model of T11-T12 under flexion/extension moments, left/right lateral bending moments, and left/right axial torque with those obtained by published experimental study (Oxland, et al., 1992).



A. T12-L1 Flexion and Extension

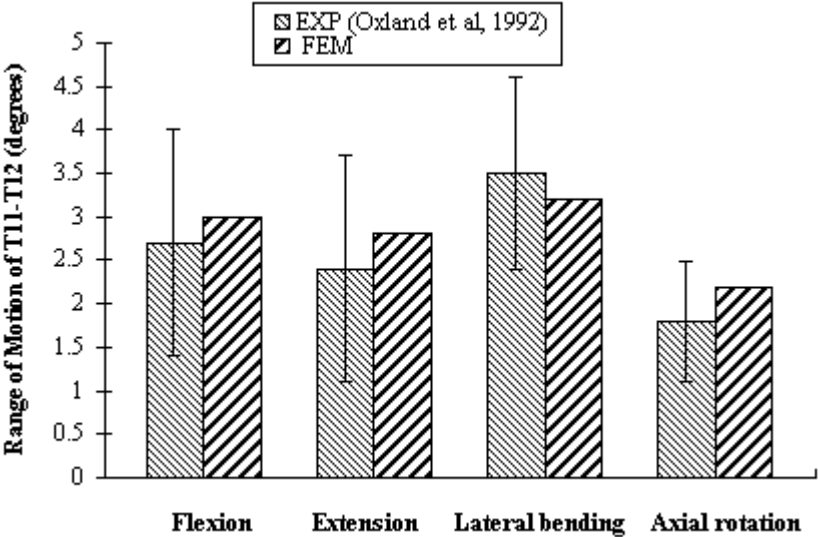


B. T12-L1 Left and Right Lateral Bending

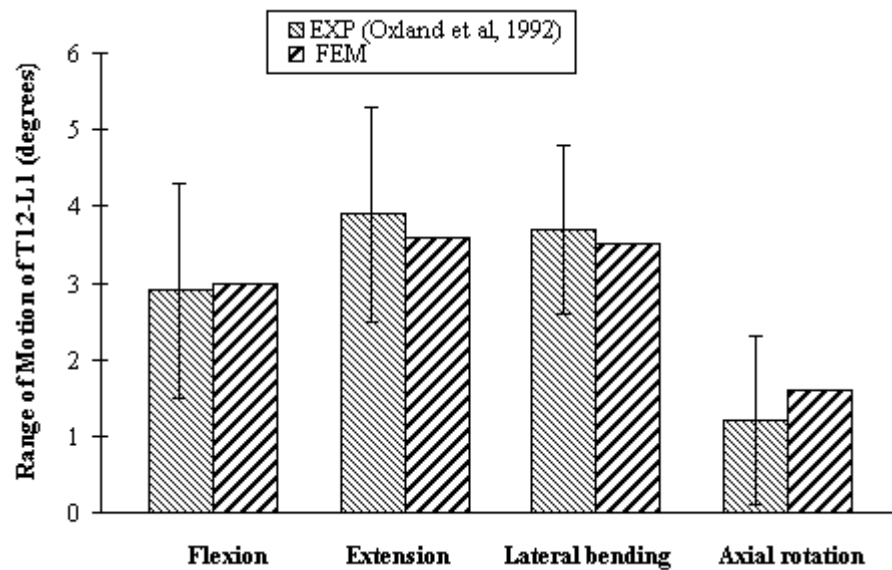


C. T12-L1 Left and Right Axial Rotation

Figure 38 Comparison of load-angular deformation curves predicted by FE model of T12-L1 under flexion/extension moments, left/right lateral bending moments, and left/right axial torque with those obtained by published experimental study (Markolf, 1972; Oxland, et al., 1992).



A. T11-T12 Range of Motion



B. T12-L1 Range of Motion

Figure 39 Comparisons of ranges of motion predicted by FE models of T11-T12 and T12-L1 with those obtained by experimental study (Oxland et al., 1992).

The predicted load-angular deformation curves under various pure moments closely correlated with the biomechanical studies obtained by Markolf (1972) and Oxland et al. (1992). This was evidenced by the similar trends of increasing rotational motion and stiffness of the T11-T12 and T12-L1 segments with applied moment for all the load configurations analyzed (Figures 37 and 38). The load-angular deformation response under various pure moments matched well with the published experimental results. The predicted ROMs of 3°, 2.8°, 3.2°, and 2.2° at T11-T12 fell within the range of the experimental values (Oxland et al., 1992) (Figure 39) of $2.7 \pm 1.3^\circ$, $2.4 \pm 1.3^\circ$, $3.5 \pm 1.1^\circ$ and $1.8 \pm 0.7^\circ$ for flexion, extension, lateral bending and axial rotation respectively. Likewise, the FE model of T12-L1 predicted the same ROMs of 3.0°, 3.6°, 3.5°, 1.7° were also in good agreement with

those published data of $2.9 \pm 1.4^\circ$, $3.9 \pm 1.4^\circ$, $3.7 \pm 1.1^\circ$, and $1.2 \pm 0.7^\circ$ for the above mentioned loading configurations in the same order (Oxland, 1992).

The sagittal rotations of the T11-T12 and T12-L1 under pure flexion and extension are shown in Figure 37A and Figure 38A. The T11-T12 and T12-L1 FE models predicted the nonlinear responses of the motion segments and exhibited similar trends in the angular responses with published results (Markolf, 1972, Oxland et al., 1992). All the predicted and experimental results (Oxland et al., 1992) exhibited that the rotation at T12-L1 was slightly greater than at T11-T12 in flexion, while the rotation at T12-L1 was slightly greater than at T11-T12 in extension.

In lateral bending, Figure 37B and Figure 38B demonstrated that there is not much difference in rotation between T11-T12 and T12-L1, though the motion at T12-L1 was slightly greater than at T11-T12. In axial rotation, the motions at T11-T12 were significantly greater than at T12-L1 as shown in Figure 37C and Figure 38C. The predicted experimental responses in lateral bending and axial rotation were all nonlinear.

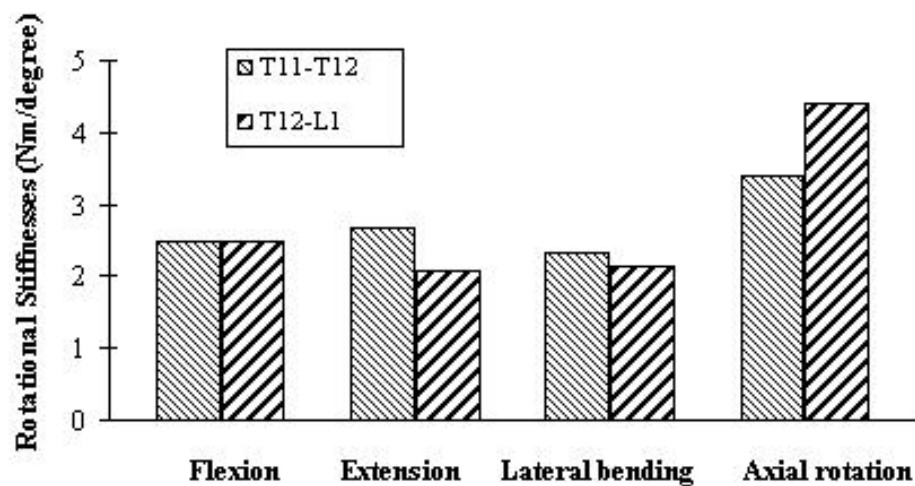


Figure 40 Rotational stiffnesses of T11-T12 and T12-L1 FE models under pure flexion and extension moments of 7.5 Nm, left and right lateral bending moments of 7.5 Nm as well as left and right axial torque of 7.5 Nm.

The overall responses predicted by the T11-T12 and T12-L1 FE models showed differences in stiffness under different load configurations (Figures 37-38). Figure 40 shows the stiffnesses of T11-T12 and T12-L1 FE models corresponding to pure flexion and extension moments of 7.5 Nm, left and right lateral bending moments of 7.5 Nm as well as left and right axial torque of 7.5 Nm. Amongst all loading configurations, the motions at T11-T12 and T12-L1 were the stiffest under axial torque. The lateral bending motions of T11-T12 and T12-L1 were relatively flexible. Under sagittal moments, the motion in extension was greater than in flexion at level T11-T12, while the rotation in flexion was greater than in extension at level T12-L1.

4.3.5 Discussion

The validation studies were conducted for the verification of the FE models of the thoracolumbar junctional FSUs T11-T12 and T12-L1 against published experimental results (Markolf, 1972; Oxland et al., 1992). In the validation study, applied loads were only in the form of three pairs of pure moments to replicate the loading and boundary conditions used in the published experimental studies (Markolf, 1972; Oxland et al., 1992), and therefore the validation study did not address the effect of applied forces on the intervertebral motion.

As there were many assumptions and simplifications in the modeling of the T11-T12 and T12-L1 FSUs, validation is very important before they were used for further studies. The load-angular deformation responses and the ROMs of the FSU under various physiological load configurations correlate closely with those obtained from the experimental studies (Markolf, 1972; Oxland et al., 1992). This showed that the methodology and assumption applied in this study were valid.

The validation study demonstrated that the present models were realistic despite the use of linear elastic, homogeneous and isotropic material properties to define the associated spinal components of the T11-T12 and T12-L1 FSUs except the ligaments.

Only a few studies had documented the mechanical properties of the thoracolumbar junctional region (Markolf, 1972, Oxland et al., 1992). Thus, comparison between the predicted results and published experimental data was

limited. Figure 37C and Figure 38C show that the FE models of T11-T12 and T12-L1 FSUs predicted relatively larger angular rotations than the experimental results. This could be due to the facet articulation defined in these current models. Theories of human joint mechanisms indicated that synovial fluid between the articulating cartilages maintains the fluid-solid interaction, and the cartilages do not come into direct contact during the normal load transmission process (Brand, 1989). The synovial fluid acts as a thin film of lubricant and a load-distributing agent between the two articulating cartilages. Secondly, in actual facet joints, they are enclosed by the capsule and capsular ligaments which limit the axial motions of the spinal segment in particular. However in this study, the articulating facets were modeled as surface moving contact which might not be able to accurately represent the facet articulation mechanism. Thus, the ROMs in axial rotation were most affected; the predicted ROMs by the FE models of T11-T12 and T12-L1 FSUs were larger than those obtained by experimental studies (Markol, 1972; Oxland et al., 1992) (Figure 37C and Figure 38C), since the facet joints with its surrounding capsular ligaments play an important role in axial rotation.

As shown in Figures 37A, 38A and 39, the rotations under extension at level T12-L1 were significantly larger than at level T11-T12. This could be explained by the significantly greater facet resistance to extension caused by thoracic facets in T11-T12 than the more vertically oriented lumbar facets in T12-L1. Under applied axial torque, greater rotation at T11-T12 was expected than at T12-L1 due to the facet geometry; the vertically oriented lumbar facets provided a positive stop to axial rotation.

Figures 37 and 38 show that the load-angular deformation responses predicted by the FE models were in similar trends to published experimental data for flexion, extension, lateral bending and axial rotation (Markolf, 1972; Oxland et al., 1992). The predicted ROMs fell in the range obtained by experimental study (Oxland et al., 1992). At T12-L1, in extension and axial rotation, the predicted ROMs were larger than those in the experimental study of Markolf (1972) (Figures 38A and 38C). There are several possible explanations for such large differences in extension and axial rotation. First, the facet joints defined in the present study were modeled by frictionless contact elements; therefore, no real representation was made for the mechanism of facet joints. Another possible cause may be due to the use of a fixed-axis loading procedure used by Markolf (1972) compared with the study of Oxland et al. (1992). Therefore, Figures 37A and 38C also show that the ROMs obtained by Oxland et al. (1992) are larger than those obtained by Markolf (1972). Differences in the test procedure can produce differing results. The predicted results demonstrated that the FE representation of the FSUs T11-T12 and T12-L1 were accurate enough to predict the biomechanical responses under different loading conditions despite that the experimental results (Markolf, 1972; Oxland et al., 1992) exhibit considerable divergence, which may be due to biological variation among specimens and the use of different experimental techniques.

4.4 Summary

The anatomically realistic FE models of T11-T12 and T12-L1 FSUs were constructed based on the geometrical data obtained from a digitizer. Validation

studies were performed to verify the developed FE models for usage in the next phase of the simulations. The main contents addressed in Chapter 4 are as follows:

- The digitizer was used to capture the three-dimensional geometrical data of the thoracolumbar vertebrae. The digitizing procedure was able to reliably preserve the complicated topography of the original vertebra for the FE modeling.
- The anatomically realistic FE models of T11-T12 and T12-L1 were built with the commercially available software ANSYS, and crucial anatomical features such as facet joints were explicitly represented.
- In the validation study, the close agreement of load-deflection response trends found between the predicted and experimental data provided the necessary confidence to exploit the FE analysis for an understanding of biomechanical responses in clinically relevant situations or injuries study to be described in the following chapters.

The faithful match of FE model responses with experimental mean values could not be possible due to the use of different experimental techniques and the inherent biological variations among specimens. Nevertheless, the agreement found with experimental data provided the necessary confidence for further study. With the realistic geometrical definition of the bony structures, the loci of the IARs characterized in the Chapter 5 and the burst fracture mechanism investigated in Chapter 6 using the models would be good representations of the kinematics and dynamics of thoracolumbar FSUs for the general population.

Chapter 5 INSTANTANEOUS AXIS OF ROTATION (IAR) STUDIES OF T11-T12 AND T12-L1

5.1 Introduction

The movement of a FSU is dependent upon several parameters, namely its complex geometry, facet articulations and material characteristics of the ligamentous tissues, and the applied load vectors. The IAR is one of the in-plane kinematics characteristics of a FSU under load, and knowledge of the exact positions and loci of the IARs of the rotatory motion of the vertebral bodies is needed to understand the normal biomechanical behavior of the spine (Ogston, et al., 1986).

Out of this current research, a related journal article was published (Qiu et al., 2003) which similarly reported the locations and loci of the IARs of thoracic T10-T11 FSU in three anatomical planes. The locations and loci were found to vary with increasing applied loads for each of the pure moments applied to produce flexion/extension, left/right lateral bending, and left/right axial rotation. Under flexion and extension pure moments, the loci of IARs were tracked anterosuperiorly for flexion and posteroinferiorly for extension with rotation between the superior endplate and the geometrical center of the inferior vertebra T11. Under left and right lateral bending pure moments, the loci were detected to diverge lateroinferiorly from the mid-height of the intervertebral disc, then converge medioinferiorly toward the geometrical center of the inferior vertebra T11. For axial rotation, the IARs were located between anterior nucleus and annulus and found to diverge in opposite direction lateroposteriorly with increasing left and right axial torque.

The rotational path of a FSU is indicative of the health of the spinal segment in question. In spinal biomechanics studies, Gertzbein et al. (1984, 1985, and 1986) used *in vitro* and *in vivo* methodologies to study the centrodes of the moveable vertebra in the lumbar spine in the sagittal plane. They reported that a greater scatter of centrodes in flexion/extension was detected for those FSUs with morphologic evidence of disc degeneration.

Therefore reliable identification of IAR could be of value in predicting the behavior of the FSU in response to different injurious vectors as well as in diagnosing disc degeneration (Gertzbein et al., 1984, 1985, and 1986). Very little is known about the locations and loci of the IARs of the thoracolumbar junctional FSUs. Hence, the locations and loci of the IARs of the intact thoracolumbar T11-T12 and T12-L1 FSUs are characterized in this chapter.

5.2 IARs Study of FE Models of T11-T12 and T12-L1

The validated FE models of T11-T12 and T12-L1 FSUs (Section 4.3) were used to identify the locations and loci of IARs in the sagittal, frontal, and transverse planes under physiological loading configurations - flexion/extension, left/right lateral bending, and left/right axial rotation.

5.2.1 Methods

5.2.1.1 Boundary and loading conditions

To investigate the IARs of the intact T11-T12 and T12-L1 FSUs, the same loading and boundary conditions as adopted in the validation study (Figure 35 in

Section 4.3.2) - flexion/extension and left/right lateral bending pure moments of 7.5 Nm each, and left/right axial torque of 7.5 Nm, were applied uniformly to the top surfaces of the movable vertebrae, namely T11 of T11-T12 FSU and T12 of T12-L1 FSU, respectively. The inferior surfaces and spinous processes of the lower vertebrae of the two FSUs were fully fixed. All the six loadings were applied in 10 steps to produce flexion and extension in the sagittal plane, lateral bending in the frontal plane, and axial rotation in the transverse plane in order to characterize the accurate locations and loci of the IARs at each increment load.

5.2.1.2 Determination of locations and loci of IARs

At every single instant, a rigid body in planar motion has a line that either pass through or outside the body that does not move. This line is termed the instantaneous axis of rotation. However, in view of the complex state of deformation present in a FSU subjected to pure planar moments, definition of a precise axis about which the moveable vertebra rotates in each anatomical plane becomes impossible. This complex state of deformation is due to the presence of irregular geometry, diverse materials, and facet interactions. In addition, the IAR is a characteristic of planar motion (White and Panjabi, 1990, Panjabi et al., 1984). Hence, the vertebral rotations under corresponding pure moments in the three anatomical planes were assumed to be planar, without consideration of secondary coupled motions exhibited in the FSU during motions.

As IAR is only for planar motion (White and Panjabi, 1990, Panjabi et al., 1984), therefore, the concept of IAR cannot be applied to three-dimensional motion.

The proper representation of three-dimensional motion is helical axis of motion, which is a unique axis in space that completely defines a three-dimensional motion of rigid body from one position to another position (White and Panjabi, 1990). It is analogous to the IAR for planar motion.

Figure 41 illustrates the method of IAR location. The IAR is defined by an intersection of two perpendicular bisectors of line segments A-A' and B-B' where A, B, and A', B' correspond to the locations of two nodes on the T11 body before and after rotation for flexion load increment in the sagittal plane. This IAR defines the center of rotation under flexion as the motions in all parallel sagittal sections were thought to be same. This defined axis did not undergo flexion and is stationary, and thus can be thought as the IAR of the moveable vertebra T11 in flexion at this particular moment. The location of such an axis is dependent on the relative stiffness of various spinal components.

The same methodology was applied for detecting other locations of centers of rotation for each load increment for flexion/extension in the sagittal plane, lateral bending in the frontal plane, and axial rotation in the horizontal plane. The IARs calculated for each load step of each load type were joined to track the pathway or the loci for flexion, extension, lateral bending and axial rotation.

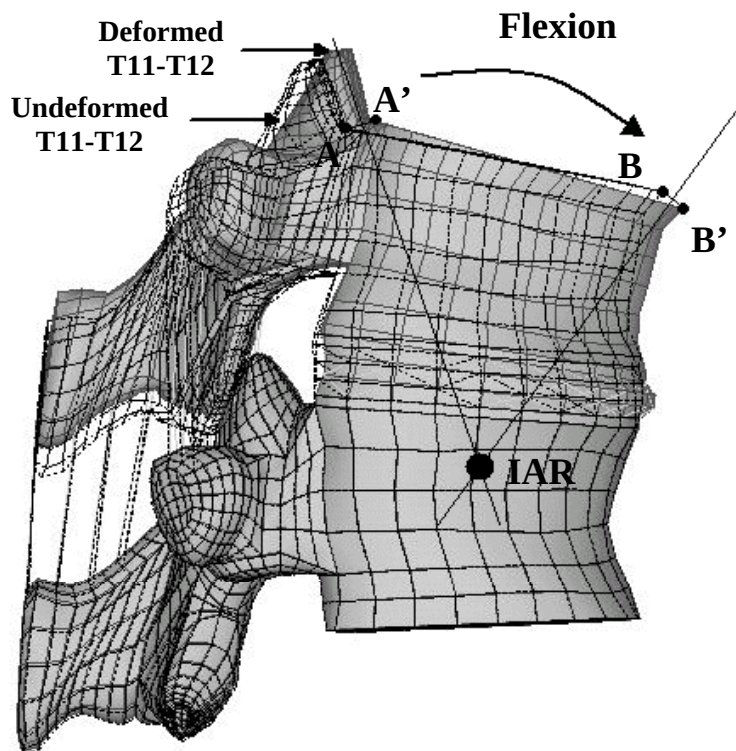


Figure 41 Definition of IAR shown in flexion of T11-T12. This figure shows the concept and the actual method of determining the IAR in uni-planar motion.

5.2.2 Results

The locations and loci of IARs for the intact models are shown in Figures 42-44 for motion segment T11-T12, and Figures 45-47 for T12-L1 FSU. The locations and loci of IARs for the two FSUs were all found to vary with increasing applied moments in the three anatomical planes.

5.2.2.1 The Locations and Loci of IARs for the Model of T11-T12

Under flexion and extension pure moments, the primary motion of T11 was in the sagittal plane and in the same direction as the applied moments. As shown in Figure 42, the model predicted that the centers of rotations in sagittal plane were

located between the region of the geometrical center of the inferior vertebra T12 and the center of the intervertebral disc between vertebrae T11 and T12. At the first initial load of 0.75 Nm, the locations of IARs for flexion and extension were computed to be almost at the same location (j and j' in Figure 42). However, with subsequent increasing moment, the loci of the IARs of rotation were then tracked to diverge in different directions depending on the type of load.

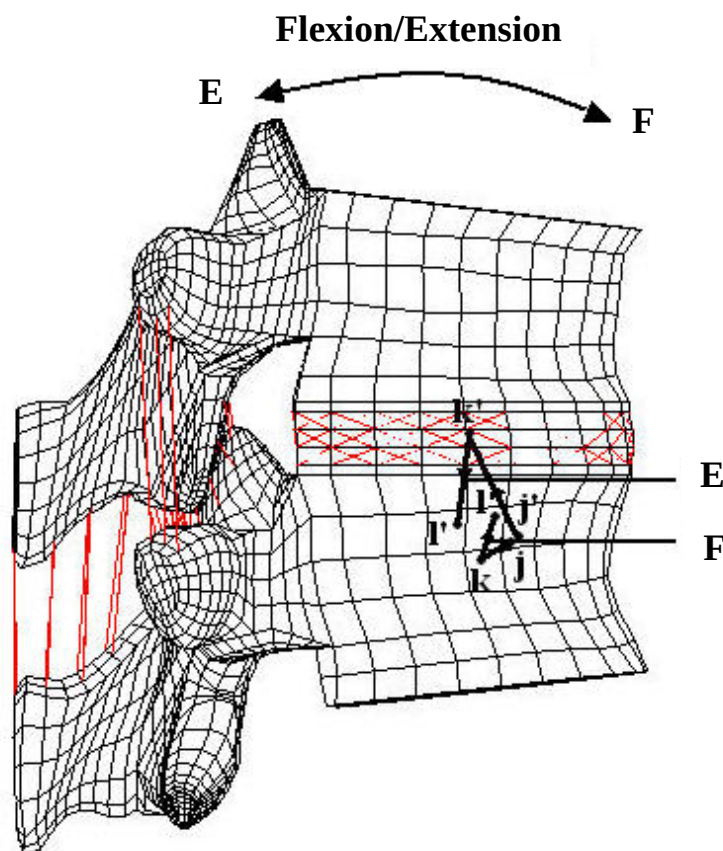


Figure 42 Predicted locations and loci of centers of rotation in sagittal plane with increasing flexion and extension pure moments for T11-T12. The curve j-k-l is the loci of centrodes for flexion, the curve j'-k'-l' for extension; j and j', k and k', l and l' correspond to the flexion/extension pure moments of 0.75 Nm, 1.5 Nm, and 7.5 Nm, respectively. (F for flexion, E for extension).

Under flexion, the loci are shown to move along j-k-l. For moments of 0.75 Nm to 1.5 Nm, points j and k tracked to move inferoposteriorly, then with subsequent increasing load, the loci moved from point k to l in the superioanterior direction in the same rotational direction as the flexion motion of T11. In extension, the T11-T12 model predicted the loci track to move along j'-k'-l'. At initial loads of 0.75 and 1.5 Nm, the loci were tracked to move with larger magnitude in the superioposterior direction (j'-k') towards the superior endplate of T12. At subsequent increase in extension load, the loci moved along k'-l' inferioposteriorly.

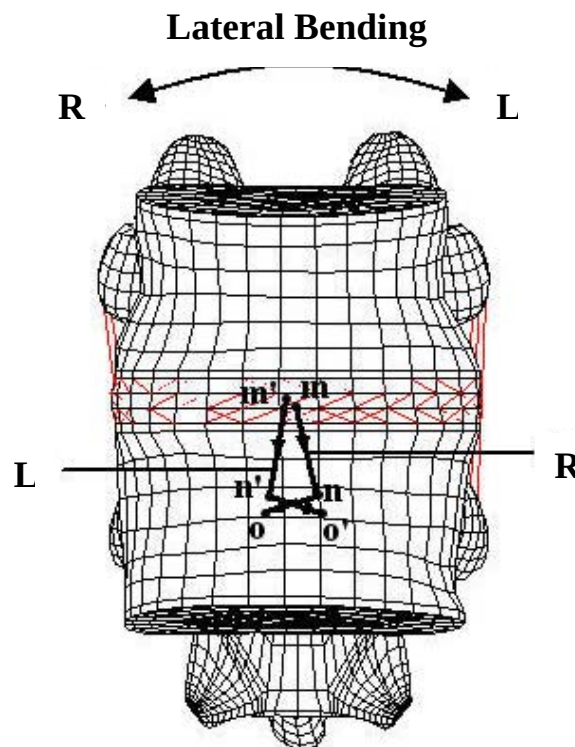


Figure 43 Predicted locations and loci of centers of rotation in frontal plane with increasing left and right lateral bending pure moments for T11-T12. The curves m-n-o and m'-n'-o' are the loci of centrodes for right and left lateral bending, respectively; m and m', n and n', o and o' correspond to the pure right/left moments of 0.75 Nm, 1.5 Nm, and 7.5 Nm, respectively. (L for left lateral bending, R for right lateral bending).

Under lateral bending moment, the model predicted primary motion along the frontal plane. Figure 43 shows that the IARs were all located at the mid-height of the intervertebral disc directly below the geometrical center of the moving vertebra (m , m' in Figure 43) at the first incremented moment of 0.75 Nm of left and right lateral bending. Subsequent increase in moment resulted in the divergence of the loci of the IARs laterally (m - n and m' - n') and then medially (n - o and n' - o') in the opposed directions of the applied moments. The loci of the IARs of T11 under both left and right lateral bending were shifted from between the superior endplate to the geometrical center of the vertebra T12.

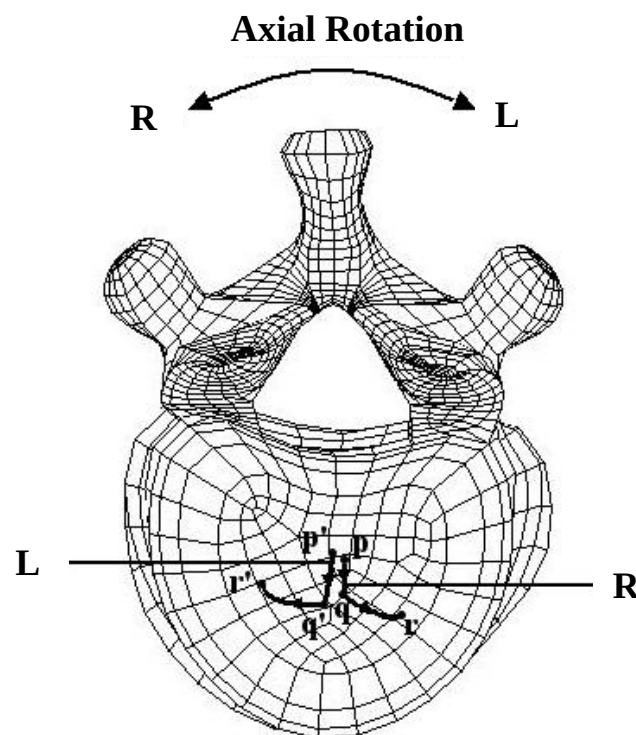


Figure 44 Predicted locations and loci of centers of rotation in transverse plane with increasing left and right axial torque for T11-T12. The curves p - q - r and p' - q' - r' are the loci of centroids for right and left axial rotation, respectively; p and p' , q and q' , r and r' correspond to the axial torque of 0.75 Nm, 1.5 Nm, and 7.5 Nm, respectively. (L for left axial rotation, R for right axial rotation).

Under axial torsion load, the primary motion was in the transverse plane. At small initial right and left torques of 0.75 Nm, the IARs were located in the central portion of the vertebral body of T11 slightly off the mid-sagittal plane (p and p' in Figure 44). With increasing axial torque from 0.75 Nm to 1.5 Nm, the IARs diverted laterally and anteriorly in the transverse plane (p-q and p'-q' in Figure 44). With further increased load of between 1.5 Nm to 7.5 Nm, the model predicted loci to move laterally but posteriorly (q-r and q'-r').

The T11-T12 model predicted that the primary motions of the moveable vertebra (T11) were in the same directions as the applied planer moment in the three anatomical planes. At each corresponding motion in these planes, the loci were located at approximately the same location at initial load, and then the loci diverge in different directions with increasing load.

5.2.2.2 The Locations and Loci of IARs for the Model of T12-L1

The loci of the moveable vertebra T12 with respect to L1 under sagittal, frontal and transverse moments are plotted in Figures 45-47, respectively. The locations and loci of the IARs were also found to vary for every load step under different load types in the three anatomical planes.

The FE model of T12-L1 predicted that the flexion and extension moments on the motion segment of T12-L1 primarily cause sagittal rotations with the T12 vertebra rotating anteriorly and posteriorly with respect to the fixed inferior vertebra of L1, respectively.

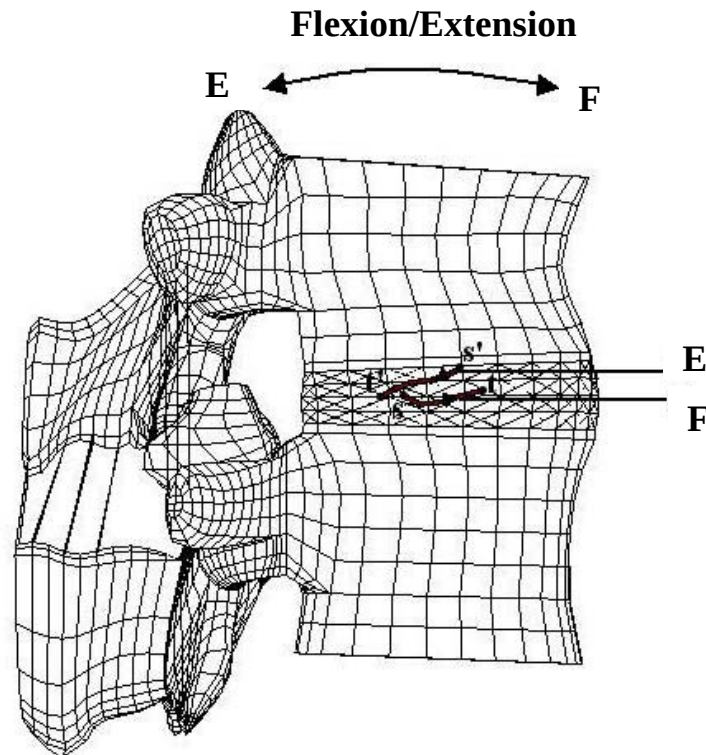


Figure 45 Predicted loci of centers of rotation with increasing flexion and extension pure moments in the sagittal plane for T12-L1. The curve s-t is the loci for flexion, the curve s'-t' for extension. The locations s and s', t and t' are corresponding to the IARs under flexion/extension sagittal moments of 0.75 Nm and 7.5 Nm respectively. (F for flexion, E for extension).

Under pure flexion and extension sagittal moments, the model predicted that the IARs were always located in the region between the center and the posterior region of the intervertebral disc (Figure 45). The loci were identified to lie along curves s-t and s'-t', for flexion and extension respectively, with increasingly applied loads. In flexion, the IARs locations were approximately situated at the mid-height transverse plane of the intervertebral disc between T12 and L1. The loci of the IARs for flexion moved anteriorly from the posterior towards the central region of the intervertebral disc. In extension, the IARs were found to be located above those of the

flexion, and the loci shifted inferioposteriorly from the inferior endplate of the moveable vertebra T12 to the posterior portion at the mid height of the intervertebral disc.

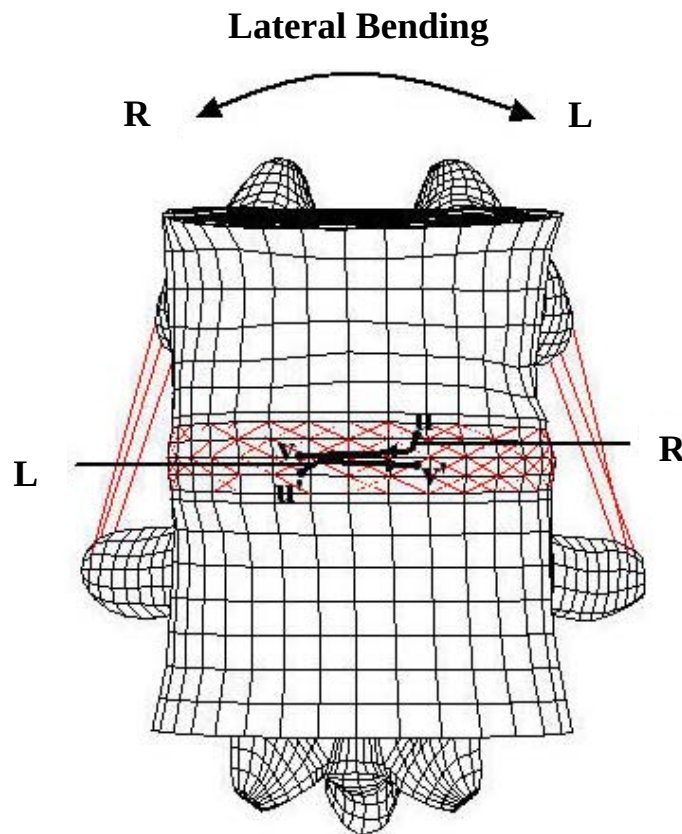


Figure 46 Predicted loci of centers of rotation with increasing lateral bending pure moments in frontal plane for T12-L1. The curves $u-v$ and $u'-v'$ are the loci for right and left lateral bending, respectively. The locations u and u' , v and v' are corresponding to the IARs under right/left lateral bending pure moments of 0.75 Nm and 7.5 Nm respectively. (R for right lateral bending, L for left lateral bending).

In the frontal plane, the pure left and right lateral bending moments caused the T12 vertebral body to undergo corresponding left and right rotations, with respect to the fully constrained vertebra L1. Figure 46 shows that the locations of the IARs under both right and left lateral bending were all situated in the central portion at

mid-height of the intervertebral disc below the geometrical center of the moveable vertebra T12.

In Figure 46, at the first load step of right/left frontal moments of 0.75 Nm, the IARs (u and u') were slightly deviated from the mid-height transverse plane of the intervertebral disc. Subsequently, with increased loadings, the loci of the IARs were situated at approximately the same locations near the mid-height transverse plane of the intervertebral disc under right and left frontal moments. However, the loci are shown to move in opposite directions, along curves $u-v$ and $u'-v'$ for right and left lateral bending, respectively. The loci of the IARs were tracked to shift along almost straight lines, from left to right along curve $u-v$ (Figure 46) under right lateral bending, and from right to left along $u'-v'$ for left lateral bending.

Under axial torque, the vertebra T12 rotated in the transverse plane, with the IARs located in the left and right portions of the vertebral body for corresponding right and left axial rotations. Under the first load step of axial torque, the IAR locations (w and w' in Figure 47) were positioned at the rim and near where the frontal plane sections the middle of the T12 vertebral body. With subsequent increased load, the loci were found to move in opposed directions towards the mid-sagittal plane (Figure 47) and also anteriorly along curve $w-x$ in the left area of the vertebra T12 for the right axial torque, and along curve $w'-x'$ in the right part of the vertebra T12 for left axial rotation.

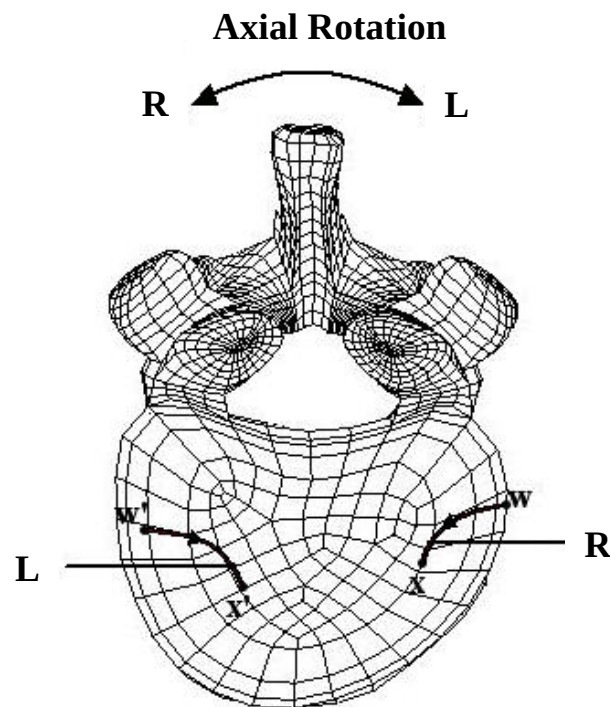


Figure 47 Predicted loci of centers of rotation with increasing axial torque in transverse plane for T12-L1. The curves w - x and w' - x' are the loci for right and left axial rotation, respectively. The locations w and w' , x and x' are corresponding to the IARs under right/left axial torque of 0.75 Nm and 7.5 Nm respectively. (R for right axial rotation, L for left rotation).

5.2.3 Discussion

The characteristics of any planar motion that consists of an arbitrary combination of rotation and translation can always be represented as a pure rotation about a given axis, *i.e.*, the IAR. The rotation magnitude, and the location and locus of the centers of rotation completely define any planar motion. As information regarding the locations and loci of the centers of rotation of the FSUs around the TLJ is not available in previous biomechanical studies, the moment rotations of these

FSUs were considered as having this type of motion in order to identify their locations and loci in this Chapter.

Though complex deformations, as a result of its intricate geometry, various material properties and facet interactions, are present in the FSU when subjected to flexion, extension and lateral bending pure moments, and axial torque, precise *in vivo* and *in vitro* determination of the axes about which the FSU rotates has become possible with the use of MRI technology (Diangelo et al., 2002) and joint measurement system (Farrokhi et al., 2002). Nevertheless, such techniques are still novel and the use of the FE method can complement such findings on IARs under various load configurations.

In this chapter, within the validated range of applied moment for all the loading and boundary conditions (Chapter 4), the FE models of T11-T12 and T12-L1 were analyzed to predict the locations and loci of their IARs. The predicted positions and loci of the IARs in the sagittal, frontal and transverse planes shown in Figures 42-44 for T11-T12 and Figures 45-47 for T12-L1, respectively, were found to vary with increasing pure applied moments. The locations and loci of the IARs, with increasing moment in the three anatomical planes, were found to diverge and converge in different directions. Such characteristics are a result of the influence of the different forms and properties of spinal components in limiting and controlling the rotation motion. The author believes that these findings provide new insight into the kinematics of the FSU motions in the three anatomical planes, as previous biomechanical studies have not explored these kinematic characteristics in detail.

5.2.3.1 IARs of T11-T12 and T12-L1 in Sagittal Plane

In the sagittal plane, the flexion moment primarily caused a rotation of the FSU with the upper vertebra rotating anteriorly with respect to the lower fixed vertebra, resulting in a widening of the gap between the inferior and superior articular facets. The extension moment, on the other hand, caused the moveable vertebra to experience posterior rotation, resulting in downward movements of the spinous process and the inferior articular processes.

Under flexion and extension, the locations and loci of the IARs were located in the area between the superior endplate and the geometrical center of the inferior vertebra T12 for T11-T12 FSU (Figure 42). However, they were situated in the area between the center and the posterior portion of the intervertebral disc for T12-L1 (Figure 45). Thus, the posterior elements such as all the posterior ligaments and facet joints were always posterior to the centers of rotation during flexion and extension. Therefore, also because of their distances from the center of rotation, all ligaments, except for the anterior longitudinal ligaments, were stretched throughout entire flexion duration. As a result, the ligaments experienced large positive strains and played a relatively important role in resisting the flexion moment for T11-T12 and T12-L1 FSUs. With further flexion, the IARs of T11-T12 and T12-L1 all shifted anteriorly and the distances between the posterior ligaments and the centers of rotation increased, (Figures 42 and 45), making the ligaments sustain more tension to resist anterior rotation. At the same time, the separation of the facets increased, resulting in tensile strain in the capsular ligaments, which resisted further widening between the articulation surfaces. These findings deduced from the locations and loci

of the IARs for T11-T12 and T12-L1 FSUs in flexion underscores the crucial role played by the posterior ligaments in resisting flexion moments. This implies that any transection of the posterior ligaments due to surgical procedures would cause clinical instability, resulting in excessive flexion.

In extension, for both the T11-T12 and T12-L1 motion segments, anterior longitudinal ligaments underwent positive strains as they were located anterior to the centers of rotation, and the strains would become larger when the IARs moved posteriorly with increase extension (Figures 42 and 45). However, the ligaments posterior to the centers of rotation experienced inactive strains. The capsular ligaments, however, although situated posterior to the IARs, were stretched due to the opening of facet articulations. Under the application of pure extension moments, the inferior facet of the moveable vertebra moved closer to the superior articulation surface of the fixed vertebra. For the T11-T12 motion segment, the articular processes impinged against one another due to the frontally oriented facet joints and provided more resistance to the extension. For T12-L1, however, there would be less occurrence of facet articulation impingement because of the sagittally aligned facets joints. Hence, the major contributors to spinal stiffness of both T11-T12 and T12-L1 in extension are the facet joints and anterior ligaments.

Finally, the explanation for the differences in the IAR loci under flexion and extension for the T11-T12 and T12-L1 motion segments is the resistances contributed by the different spinal components of the FSUs. The posterior ligaments mainly provide flexion resistance, but the anterior ligaments and facet joints are the dominant factors in resisting extension.

5.2.3.2 IARs of T11-T12 and T12-L1 in Frontal Plane

Under frontal moment for left and right lateral bending, the FE models of T11-T12 and T12-L1 predicted the upper vertebra to primarily rotate to the left and right sides with the spinous process moving toward the convexity of the lateral curvature, with respect to the lower fixed vertebra. At the same time, the facet joints on the same side as the applied moment moved closer to each other, and further apart for the other side.

In the frontal plane, the IARs were found to be in a relatively confined area (Figures 43 and 46). For T11-T12 motion segment (Figure 43), the axes of rotation fell in the region around mid-sagittal plane below the mid-height transverse plane of the intervertebral disc but above the geometrical center of the lower vertebra T12. For the T12-L1 FSU (Figure 46), the IARs were located in the central portion of and around the mid-height transverse plane of the intervertebral disc. Thus, all the locations of the IARs for T11-T12 and T12-L1 were all located around the mid-sagittal plane. Such locations of the IARs indicate that most of the ligaments were not stretched during lateral bending, only the transverse and capsular ligaments on the side with convex curvature sustained positive strains. In addition, though it was seen that the facet joints on the opposite side of the convex curvature became closer, there was only slight impingement of the articular processes during front plane rotation. This demonstrates the minor role played by the posterior spinal elements and underscores the intervertebral disc as the main load bearer during lateral bending.

5.2.3.3 IARs of T11-T12 and T12-L1 in Transverse Plane

In the transvers plane, the applied axial torque motivated the movable vertebra to rotate left and right with the spinous process moving in the same direction as the vertebra rotation. The predicted locations and loci of the IARs fell in the medioanterior region around mid-sagittal plane of the movable vertebra T11, for the T11-T12 (Figure 44), and to the left and right of the mid-sagittal plane near the cortical shell of the upper vertebra T12 around the middle from posterior to anterior for T12-L1 (Figure 47).

It can be deduced, from the IAR locations for T11-T12 and T12-L1 as shown in Figures 44 and 47, respectively, that there will be larger positive strains existing in all the posterior ligaments because of their distance to the centers of rotation during axial rotation. The spinous and supraspinous ligaments will be stretched largest due to their positions located furthest posteriorly from the IARs. This demonstrates that the posterior ligaments greatly contribute to resist axial rotation.

The distinct difference in the locations of the IARs for T11-T12 and T12-L1, especially the IARs were located at the rim of the vertebral body (w and w' in Figure 47) for the right and left axial rotation at the initial loads, may mainly be attributed to the variation in the orientations of the facet joints: T11-T12 has thoracic-type facet articulations which allow more mobility under applied axial torque, but T12-L1 has lumbar-type facet joints which provide positive resistance to axial rotations.

5.2.3.4 Comparison of IARs between T11-T12 and T12-L1

The locations and loci of IARs are shown in Figures 42-44, and Figures 45-47 for levels T11-T12 and T12-L1, respectively, in the sagittal, frontal and transverse planes. The locations and loci of T12-L1 differ greatly from those of T11-T12 in all anatomical planes. It is well known that the anatomic structure of a FSU defines its motion and related biomechanical responses (Giles and Singer, 2000; White and Panjabi, 1990). Hence, some differences in anatomical features of the two FSUs may account for the variation in loci. At level T11-T12, the facet articulation is essentially oriented in the coronal plane; while in the T12-L1 segment, the facet joint surfaces are sagittally aligned (White and Panjabi, 1990). For the intervertebral disc, at T11-T12, the anterior height is slightly larger than the posterior height; whereas, at T12-L1, the anterior height is greater than the posterior height, which results in a lordotic angle. Hence, the different orientations of the facets and the geometry of intervertebral discs may demonstrate the difference in loci at the two levels.

Though there are distinct alterations in the locations and loci of these two segments in the three anatomical planes, some similarities exist. The shift in the direction of the loci under flexion/extension, left/right lateral bending, and left/right axial rotation were in the same direction as the corresponding vertebra motion. This can be observed from the loci j-k-l in Figure 42 and s-t in Figure 45 for the flexion of T11-T12 and T12-L1, respectively; the moveable vertebrae and the loci rotate along the same direction as flexion. The paths j'-k'-l' in Figure 42 and s'-t' in Figure 45 for extension were tracked to shift in the same direction as extension. In the frontal plane, the loci m-n-o in Figure 43 and u-v in Figure 46 of the IARs were moved to the right

during right lateral bending, while the loci m'-n'-o' in Figure 43 and u'-v' in Figure 46 shifted to the left following left lateral bending. In addition, in the transverse plane, during right axial rotation, the loci p-q-r shown in Figure 44 for T11-T12 and w-x shown in Figure 47 for T12-L1 of the IARs shifted along the same direction as the right axial rotation; similarly for the left rotation.

5.2.3.5 Comparison of IARs between T11-T12 and Thoracic Spine FSUs

The T11-T12 FE model predicted locations of IARs shown in Figures 42-44 are similar to those of thoracic spine FSUs reported in the *in vitro* study of Panjabi et al. (1984) in flexion/extension, left/right lateral bending, and left/right axial rotation. Panjabi et al. (1984) used 11 FSUs covering all levels of thoracic spine from T1-T12 and concluded that neither vertebral level nor load magnitude was significant in influencing the location of IARs; except the load type. Similarly, Broc et al. (1997) also reported similar approximate locations of IARs were found in each thoracic level obtained using 8 FSUs from T4-T5 to T11-T12.

The IAR locations predicted by the current T11-T12 FE model under pure flexion and extension moments were similar, but with some variations to those obtained by *in vitro* studies reported by Panjabi et al. (1984) and Broc et al. (1997). The IARs were all found to be directly below the geometrical center of the moveable vertebra. The variations observed are acceptable considering the variation in biological specimens and sizes and the assumptions made in the FE model. In addition, the methodologies used in the determination of the IARs would also contribute to the variations.

Under left/right lateral bending and left/right axial rotation, the predicted IARs locations correlated well with the findings of Broc et al. (1997). The locations of the IARs for left and right lateral bending are located correspondingly at the right and left regions, around the superior endplate and directly above the geometrical center of the lower (fixed) vertebra of the FSU. Under axial rotation, the predicted IARs locations are also in agreement to those reported by Broc et al. (1997); the IARs are located in the left and right anterior portions of the moveable vertebra for right and left axial rotation, respectively.

It is believed that the similar geometrical characteristics in facet articulations (typical frontal oriented thoracic type), intervertebral discs (nearly equal height anteriorly and posteriorly) and shapes of vertebral bodies (anteriorly wedged) may be responsible for the similarities in the locations and loci of the IARs of predicted by the T11-T12 FE model and those of experimental studies (Panjabi et al. (1984) and Broc et al. (1997)).

This comparison between studies also indirectly confirms the reliability of the FE method used in this study to identify the locations and loci of IARs.

5.2.3.6 Comparison of IARs between T12-L1 and Lumbar Segments

The predicted locations and loci of IARs shown in Figures 43-45 in three anatomical planes for T12-L1 are notably different. Under pure flexion and extension sagittal moments, the FE model of T12-L1 predicted that the IARs were located in the area between the center and the posterior region of the intervertebral disc (Figure

45). This site is comparable with those in the lumbar segment, where the IARs are located in the posterior one-half of the disc as reported by Gertzbein et al. (1984, 1985) through experimental studies of normal discs of L4-L5 under flexion and extension. Pearcy and Bogduk. (1988) and Pennal et al. (1972) also reported similar findings using all levels of the lumbar spine in their *in vitro* studies. The similar geometry of the T12-L1 and L1-L5 facet joints and similar lordotic curvature may explain this similarity in loci between thoracolumbar junctional FSU of T12-L1 and lumbar spine.

Since little is known about the locations and loci of the IARs for lumbar segments under lateral bending and axial rotation in literature, comparison between the IARs of T12-L1 and lumbar segments is not possible here.

5.3 Summary

It is recognized that the locations of the IARs under various motions are crucial for the assessment of spinal stability for normal and pathological spine (Amevo et al., 1992; Gertzbein et al., 1985, 1986). Accordingly, this study focused on the use of the FE method to determine the locations and loci of the IARs of the TLJ FSUs under pure moments in the three anatomical planes. The notable findings in the IAR study are as follows:

- The locations and loci of the IARs of the T11-T12 and T12-L1 FSUs were predicted. These locations and loci would provide further understanding into the kinematics and biomechanical responses of the human spine.

- It is believed that the changes in the loci of the IARs of the segment under motion are the consequences of changes in the physical properties of the ligamentous structures and its bony structures. Hence, understanding the loci of the IARs could serve for spinal stability identification.
- The locations and loci of the IARs were found to vary in the three anatomical planes as well as with increasingly applied moments. Therefore, for an accurate prediction of the states of stress and strain, it is essential that the FSU not be forced to rotate about a predetermined axis which remains stationary during the process of the loading. Hence, the predicted locations and loci of the IARs in this study may potentially be used to supplement experimental research in understanding the kinematics and clinical biomechanics of the thoracolumbar spine.

An understanding of spinal abnormalities requires knowledge of normal behavior as a baseline. This behavior includes the relative motion of vertebral bodies. The instantaneous axis of rotation describes the relative motion of a FSU from one position to another in a plane. Hence, understanding the exact location of IAR of the FSU could be of value in the study of its normal function and the prevention, diagnosis, and treatment of pathologic conditions.

Chapter 6 BURST FRACTURE MECHANISM STUDY USING FE MODEL OF T12-L1 FSU

6.1 Introduction

Burst fractures are mainly localized in the thoracolumbar junctional region (Panjabi et al., 1995a, 1995b). The clinically important question of whether the spine is stable or unstable after burst fracture is evaluated by the “column” theories (Panjabi et al., 1995a). Holdsworth (1963, 1970) divided the spine into two columns: anterior-disc and both longitudinal ligaments; and posterior-all posterior elements. This classification has been quite useful in appreciating the complexity of spinal injuries, and Holdsworth insisted that rupture of the posterior column was sufficient to create instability of the spine. Denis (1983) carried out a retrospective review of over 400 thoracolumbar spinal injuries and revised Holdsworth’s two-column theory by introducing a third or middle column composed of middle osteoligamentous complex. The anterior column was redefined to include the anterior vertebral body, anterior annulus fibrosis and anterior longitudinal ligaments. The posterior column remained as before, composed of the entire posterior osseous arch, including the facets, and posterior ligamentous complex. Using this classification, Denis found that disruption of the posterior ligamentous complex was insufficient to produce instability. Denis (1983) and McAfee et al. (1983) pointed out the pathogenesis of burst fractures as failures of the anterior and middle columns under axial compression. Denis divided the anterior and middle columns midway through the vertebral body, whereas Ferguson and Allen (1984) modified the columns by including the anterior two thirds of the vertebral body and annulus fibrosis with the

anterior column, and by placing the posterior third of the vertebral body and annulus fibrosis with middle column. This three division of columns seems to agree better with injuries observed in practice and Panjabi et al. (1995a) verified the three-column theory of Ferguson and Allen (1984) that injury to the middle column significantly alters the mechanical stability of the spine.

Also, many biomechanical studies (Sharma et al., 1995; Shirazi-Adl et al., 1986) have shown that mechanical loads imposed on the human spine during daily living play a significant role in the onset of spinal disorders. Among the various types of loading conditions, impact loading releases energy rapidly over a shorter time period, and burst fracture is one type of such spinal fractures caused by the impact loading.

Over the last few years, great interest has been focused on comminute fractures of the thoracolumbar spine. In particular, fractures caused by axial impact loadings have been discussed in experimental studies (Fredrickson et al., 1992; Kifune et al., 1995, 1997; Panjabi et al., 1995a, 1995b; Langrana et al., 2002; Tran et al., 1995), some of which had been reviewed in Chapter 3. These studies focused on the instability, encroachment of retropulsed fragment into the spinal canal, and treatment methods of the burst fractures produced by trauma set-up in laboratory. However, it is difficult to obtain accurate internal biomechanical information during the dynamic processes using these technically demanding experimental procedures. Therefore, the use of analytical models has been suggested as a way to overcome the limitations of experimental investigations.

Recently, in the study of Lee et al, (2000), a three-dimensional nonlinear poroelastic FE model of a L3-L4 body-disc was used to analyze the biomechanical effects of impact loading on the spinal segment to predict changes in biomechanical parameters such as intradiscal pressure, dynamic stiffness, stress in endplate region, and the shock-absorbing mechanism of the spine under different impact duration/loading rates, and also to investigate the relation between the rate of loading and the fracture potential of the vertebral body. The authors concluded that fractures were likely to occur under shorter duration conditions and depending on the strength of the region, a fracture might be initiated in the endplate region or the posterior wall of the cortical shell. They also concluded that the nucleus pressure was independent of the impact duration and depended only on the magnitude of the impact force.

Despite the fact that it is of immense clinical significance to know the pathomechanism (Shirado et al., 1992) of the spinal fractures and that burst fractures occur during impact rather than during static loading (Tran et al., 1995; Yingling et al., 1997), not much FE studies have been performed to quantitatively and systematically analyze the mechanism of thoracolumbar burst fractures, especially on a predictive basis, under high-speed impact loading conditions.

Shirado et al. (1992) and Willen et al. (1984) observed that burst fractures of the TLJ usually occur in the young but not the very elderly, who usually have degenerated discs and osteoporosis, to produce neurological deficits. The goals of this chapter are to report the clarification of the mechanism of thoracolumbar fracture by the FE model of T12-L1 with normal disc and evaluate the effect of degenerated discs on the mechanism of burst fracture.

6.2 Load Transmission of Normal and Degenerated Discs

In order to predict the mechanism of the burst fractures for spinal segments with normal and degenerated discs under dynamic loading, their load transfer patterns were first investigated. Static compression was performed for the analysis of the load transfer patterns.

6.2.1 Methods

6.2.1.1 *Consideration of Material Properties of Intervetebreal Discs*

A healthy disc contains a soft and highly hydrated central region, the nucleus pulposus, which acts as a hydraulic cushion. The material properties of each component of intervertebral disc were the same as those used in the validation study (Table 6 in Section 4.3.1) for the normal disc. Among them, a Young's modulus of 1 MPa and a Poisson's ratio of 0.4999 were adopted for the nucleus pulposus to simulate its incompressible behavior.

However, with increasing age, the intervertebral disc undergoes striking alteration in volume, shape, structure, and composition that decrease motion and alter the mechanical properties of the spine (Fraser et al., 1993; Iatridis et al., 1997; Thompson et al., 1990).

The degenerated discs have only a little or no hydrostatic region (Shirazi-Adl, 1992; White and Panjabi, 1990). Initial degenerative changes are likely to occur within the intervertebral disc itself, or more specifically, within the nucleus. Hence, in severely degenerated state, some discs exhibit almost no nucleus pressure, which

implies that the annulus and the posterior elements are virtually bearing all applied compressive loads (Iatridis et al., 1997; Thompson et al., 1990).

The effect of degenerative process on the properties of various structures of the intervertebral disc is not documented in the literature. It is, however, known that the disc nucleus loses its gel-like appearance and hydrostatic capabilities as a result of the degenerative process (Ferguson and Steffen, 2003). An increase in the elastic modulus of the nucleus with progressive degeneration has been shown likely the result of an increase in tissue density due to water loss with aging (Ferguson and Steffen, 2003). Therefore, disc degeneration that results from the aging process was modeled by varying the material properties (Kim et al., 1991; Whyne et al., 2003). Since the disc degeneration in T12-L1 is not predominant (Malmivaara, 1987b), and the present study was only concerned with the effects of disc degeneration on the load transmission and stress distribution, the loss of the gel-like nucleus at the disc between the T12 and L1 vertebrae was modeled as follows. The elastic modulus of the degenerated disc nucleus was two times the elastic modulus of the intact nucleus and the Poisson's ratio (Table 6, Section 4.3.1) was adopted to be the same as that of the annulus (Kim et al., 1991; Iatridis et al., 1997; Whyne et al., 2003). The changes in the material properties of all other structures, if any were ignored.

6.2.1.2 Loading and Boundary Conditions

The inferior surface of the L1 vertebral body and the spinous process were fixed in all directions, and an uniform static compressive displacement of 1.6mm (displacement control) corresponding to compressive forces of about 4 kN, sufficient

to simulate moderate manual labour (Shirazi-Adl, 1992; Shirazi-Adl, 1986), was applied in 8 increments to the superior surface of the T12 vertebral body of the thoracolumbar motion segment T12-L1 with normal disc. For simulated T12-L1 FE model with degenerated disc, the same 1.6mm displacement control was applied to simulate about 2 kN of applied load (Figure 48), as the stiffness decreased about 50% due to the loss of incompressibility of the nucleus after disc degeneration (Brown et al., 1957; Shirazi-Adl, 1986).

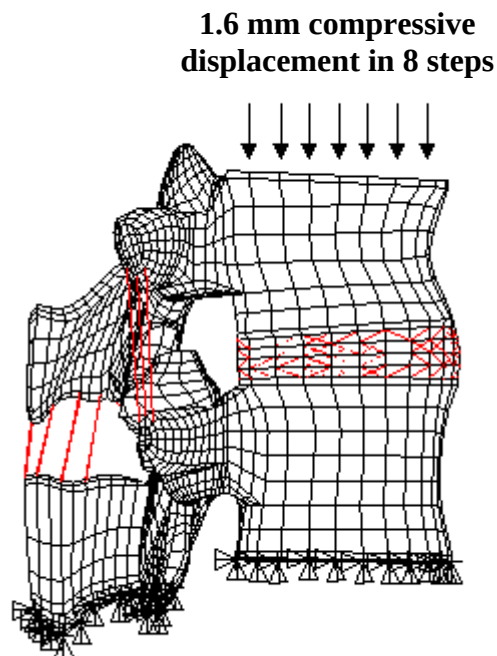


Figure 48 Boundary and loading conditions applied onto the motion segment (T12-L1) under compressive load.

6.2.1.3 Extraction of Reaction Force, and Plotting Of Stress Profile and Disc

Bulge

At each increment of applied displacement, the equivalent axial compressive force was obtained by extracting the total reaction force on the constrained nodes of the L1 bottom surface and spinous process for both the normal and degenerated discs models. For an identical load of about 2 kN, the compressive stress profiles of normal and degenerated discs across the disc at mid-height in the sagittal plane were plotted. The corresponding disc bulges of the healthy and degenerated discs in the sagittal plane were also plotted.

6.2.2 Results

The predicted results for compressive loads from the FE analysis are shown in Figures 49, 50 and 51. Figure 49 shows the comparison of load-displacement response for the FE model of T12-L1 (with normal and degenerated discs) with published experimental results obtained by Markolf and Morris (1974) and Virgin (1951). The compressive stress profiles and disc bulges of normal and degenerated discs across the disc at mid-height in sagittal plane were plotted in Figures 50 and 51.

6.2.2.1 Load-displacement Response

Figure 49 shows the comparison of load-displacement response for the FE model with normal and degenerated disc under compressive load with published experimental results (Markolf and Morris, 1974; Virgin, 1951

The load-displacement response of the intact model was within the range of the published experimental results by Markolf and Morris (1974) and Virgin (1951). This is evidenced by the similar trend of the curve plotted, as shown in Figure 49. The model predicted the stiffening behavior of the disc with increasing compressive loads. For the intact model with normal disc of nearly incompressible nucleus, the stiffening was more significant than the intact model with degenerated disc. The stiffness was reduced by about 47% for degenerated disc.

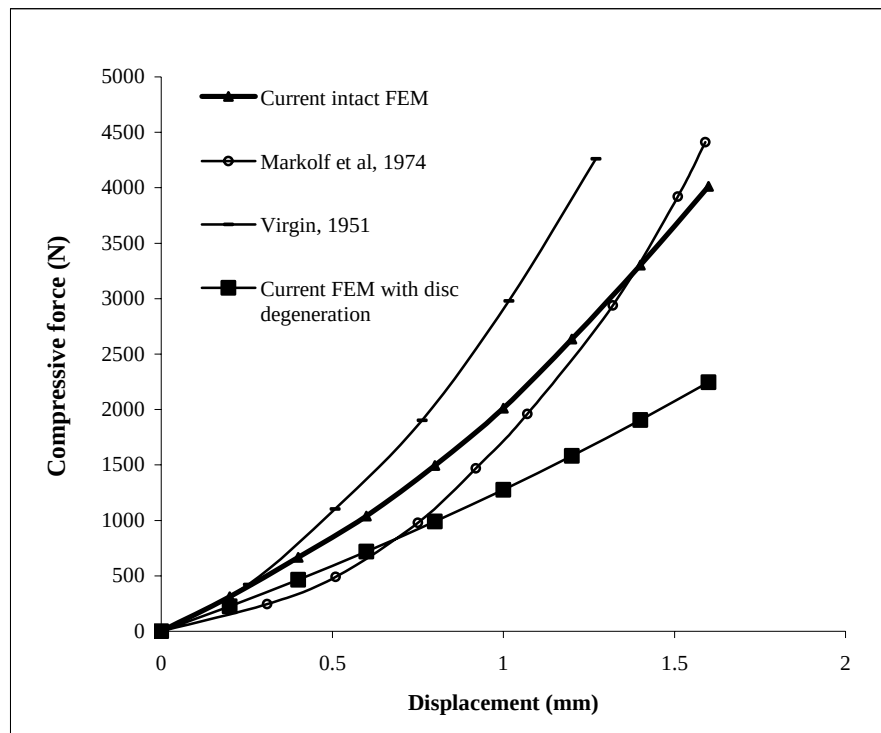


Figure 49 Comparison of load-displacement response for the FE model with normal or degenerated disc under compressive load with published experimental results (Markolf and Morris, 1974; Virgin, 1951).

6.2.2.2 Compressive Stress Profile Across the Disc at Mid-Height in Sagittal Plane

For the normal discs, Figure 50 shows that the compressive stress in the central region of the disc, corresponding to the anatomical nucleus, are almost constant with an average value at 2.4 MPa. For the degenerated disc, the peak stresses rose to high levels in the annulus (Figure 50), to about 2.6 MPa and 2.1 MPa in the posterior and anterior annulus, respectively. Accordingly, the stress increased about 40% and 35% in posterior and anterior annulus, respectively, while the stress in the nucleus decreased about 30%.

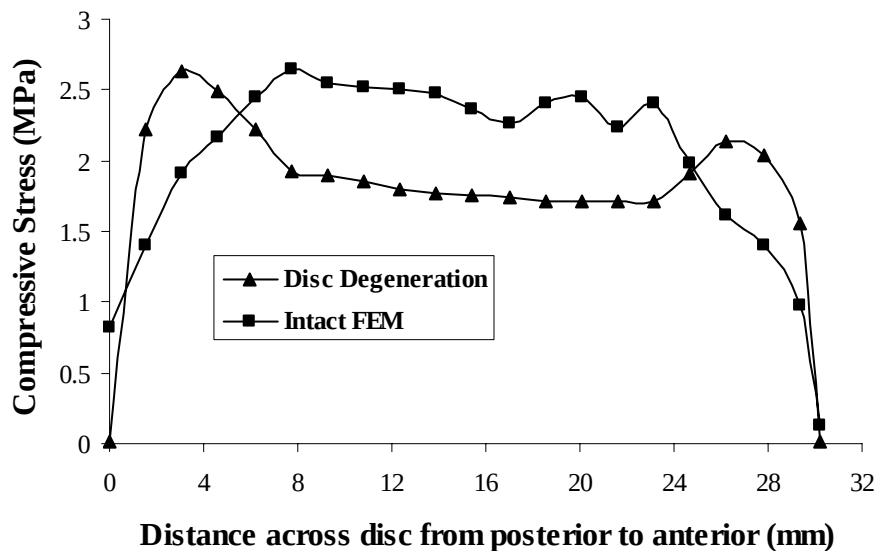
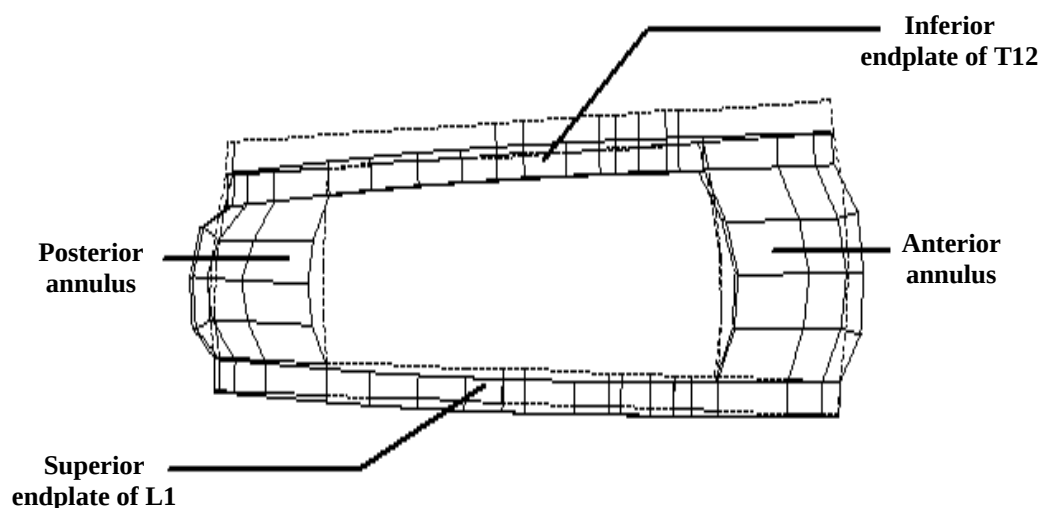


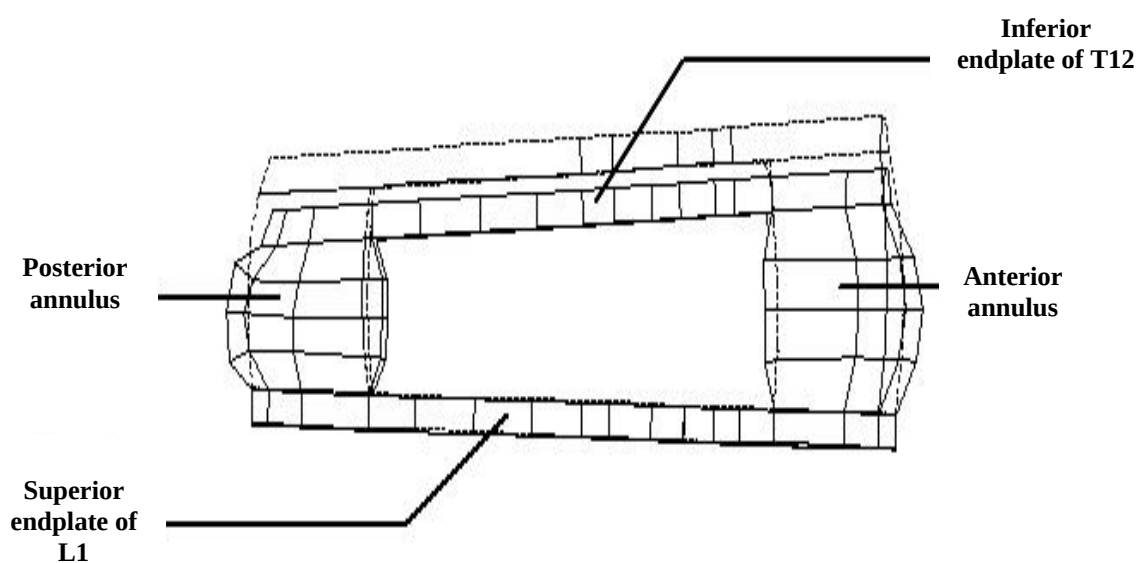
Figure 50 Compressive stress profiles for discs across disc at mid-height in sagittal plane under the identical compressive load of 2 kN.

6.2.2.3 Disc Bulges

Figure 51 shows the disc bulges (the disc deformation in the radial or transverse direction) along the mid-sagittal plane of normal and degenerated discs under the identical compressive load of 2 kN. The disc bulge in the normal disc (Figure 51A) was predicted to be less than that in the degenerated disc (Figure 51B) at both anterior and posterior locations. Along mid-height of the disc, the posterior bulge was larger than anterior bulge for both normal disc (Figure 51A) and degenerated disc (Figure 51B). The posterior disc bulges were 0.8 mm and 1.1mm for normal and degenerated discs at the mid-height, respectively. A compressive stress peak in Figure 50 implies a reversal of the stress gradient in the anterior and posterior annuluses resulting in the outer lamellae being pushed outward and the inner lamellae being pushed inward (Figure 51B). Figure 51 also shows that degenerative changes in the nucleus will result in both the inferior and superior endplate bulges from outwards (Figure 51A) to almost flat (Figure 51B). For identical load of 2 kN, increased peripheral deformations of the annulus were also observed in Figure 51B for the degenerated disc.



A. Disc bulge of normal disc



B. Disc bulge of degenerated disc

Figure 51 Disc bulges of normal and degenerated discs under the identical compressive load of 2 kN. (A) Normal disc (B) Degenerated disc. Dashed line for undeformed edge of disc, solid line for the deformed disc. The nucleus was not included in order to show the bulge of inner annulus layers.

6.2.2.4 Endplate Bulges

Endplate bulge is defined as the axial displacement of the nodes along mid-sagittal plane. Figure 52 illustrates the variation of the maximum axial endplate deformation in the superior direction along mid-sagittal plane of the inferior endplate of the vertebra T12 with axial compressive load for normal disc. Figure 51B shows that the endplate bulge was almost flat for degenerated disc due to loss of pressure in nucleus; thus, the variation of the endplate deformation for degenerated disc was not plotted here. For the normal disc, the endplate bulge is upwards (Figure 51A) to provide the volume for the incompressible nucleus. Figure 52 shows the endplate displacement gradually increasing from 0 to 0.31 mm as the FE model of T12-L1 FSU was compressed by 1.6 mm.

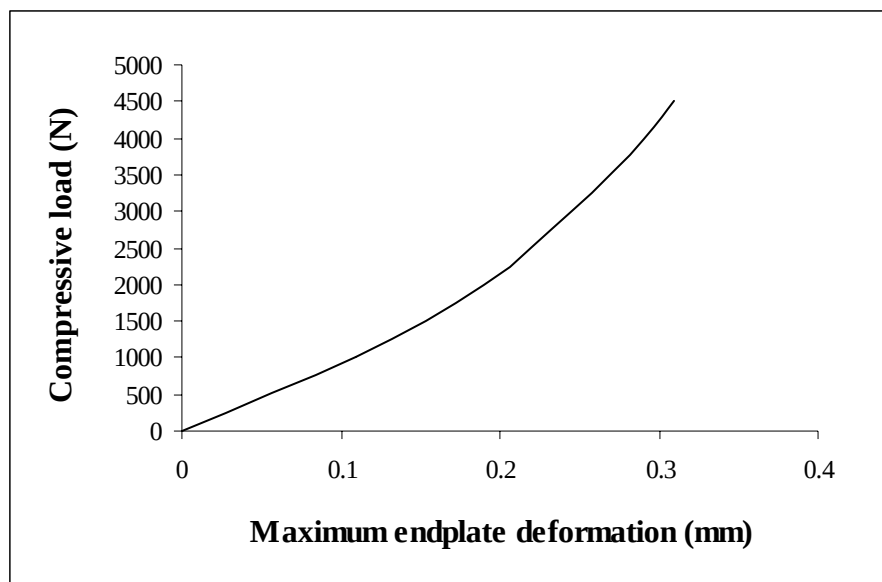


Figure 52 Compressive load plotted against maximum endplate displacement of inferior endplate of T12 for normal disc under static compressive load.

6.2.3 Discussion

The intervertebral disc exhibited a stiffening effect with increased compressive load as shown in Figure 49. The nonlinear behavior of the intervertebral disc stemmed from its composite structure involving the presence of cable elements for annulus fibers, as well as geometrical nonlinearity due to large deformation. The annulus is a composite structure modeled in the current study, involved collagenous fibers embedded in an amorphous ground substance. Initially, the ground substance is the main load-carrying component of the annulus. However, since the ground substance is a relatively soft material, the deformation behavior of the annulus under external loads can be expected to be dependent directly upon the amount of the collagenous fibers present in the annulus (Shirazi-Adl et al., 1986; White and Panjabi, 19990). The collagenous fibers of the annulus, which were subjected to tensile strain, due to the transverse bulge of the disc, became increasingly stiffer. This effect in turn resisted further horizontal deformation and hence, in cases of pressure retaining normal disc, additional axial displacement of the disc. Moreover, the increase in the cross-sectional area of the whole disc, decrease in the slope of the annulus fibers with respect to the horizontal plane, and an increase in the generated nucleus pressure, all contribute to the stiffening behavior of the disc with compressive load (Shirazi-Adl et al., 1986).

Under axial compression, the magnitude of the disc bulge was maximum at the posterior location (Figure 51); this is due to both a thinner annulus at this location and to the cross-sectional shape of the disc. Similar results were in Brown et al. (1957) and Shirazi-Adl et al. (1986) studies. The analysis, when performed

considering a re-entrant posterior shape, predicted a higher posterior bulge (Shirazi-Adl et al., 1986). The posterior disc bulge of 0.8 mm of normal disc appear to be fell in the range obtained experimentally by Brown et al. (1957) and Shah et al. (1978). In contrast to the normal disc, the inner periphery of the annulus of the degenerated disc was computed to bulge inwards (Figure 51). This, of course, is expected in view of the loss of the internal pressure in disc nucleus.

The bulging of the endplate of normal disc increases with the compressive load (Figure 52). This is due to an increase in the intradiscal pressure in the disc nucleus to produce greater upward bulge with the compressive load (Shirazi-Adl et al., 1986). The computed results (Figure 52) appear to be in good agreement with that obtained experimentally by Holmes et al. (1993). Endplate bulge occurs during compression as the vertebral bodies are compressed. Thus, the results described here are consistent to with the view that endplate deformation occurs during compression of the spine (Brinckmann et al., 1983). However, the endplate does not always develop a pronounced bulge during compression (Figure 51B); this is in agreement with the conclusions of Reuber et al. (1982).

Using stress-profilometry, it has been shown (Figure 50) that disc degeneration results in a shift of load from the nucleus to the annulus. The decreased nucleus pressure causes certain “stress concentrations” within the annulus under compression loads. Figure 50 demonstrates that stress profiles have the advantage to examine the load transmission by showing the extent and location of the region as well as the magnitude of the pressure within it, and they also show all three of these

parameters vary between the normal and degenerated disc. Hence, it extends the concepts of internal disc mechanics.

The compressive stress profiles presented here agree in several respects with previous measurements of compressive stress across the disc. They confirmed that compressive stress was more or less uniformly distributed across the disc nucleus in non-degenerated discs (Adams et al., 1996, 2000). The distributions of compressive stress across the posterior-anterior path through the sagittal plane of the normal and degenerated discs (Figure 50) indicated that the compressive load passed mainly through the center of the vertebral body to the nucleus of the disc. With the disc degenerated, more load was transmitted through the cortical wall to the annulus of disc. Similar results were also reported in the experimental studies of lumbar spinal segments by Adams et al. (1996) and McNally & Adams (1992). The FE model predicted that the stress increased about 40% in posterior annulus and decreased about 30% in the nucleus, respectively, which were comparable with an increased percentage of 35% in the annulus and a reduction of 36% in the nucleus, respectively, observed by Adams et al. (1996). Hence, disc degeneration was found to have a profound effect on the mechanism of load transfer through the disc and vertebral body. The author believes that the alteration of the paths of load transmission would cause the alteration of stress distribution in vertebral body.

The degeneration of disc caused a transfer of load from the nucleus to the annulus (Figure 50), resulting in greater deformation of the annulus (Figure 51). The posterior annulus was affected most as it was the lowest and narrowest part of the disc in T12-L1 FSU and was able to deform most. Figure 51 clearly illustrated that

the posterior annulus bulged more than the anterior annulus. At the same time, loss of the nucleus fluid caused an inward bulge at the inner annulus layers (Figure 51). Since the disc bulge is affected by the intradiscal pressures, thus if the pressure is lowered enough, the inner annulus will be forced to bulge inwards by the increased stress in the annulus of the degenerated disc (Figure 51). The inward bulging of the annulus was also reported in the sagittal section through a degenerated disc by Adams et al. (1993), experimentally observed by Seroussi et al. (1989) and in the FE study of lumbar discs by Shirazi-Adl (1992). Hence, the current FE model studied explained why a high intradiscal pressure is essential for the maintenance of the mechanical function of the intervertebral disc under physiological conditions.

6.3 Burst Fracture Mechanism of Normal Disc under Dynamic Loading

The internal changes such as stress distribution in the endplates and intervertebral bodies were investigated to predict the burst fracture mechanism by using the T12-L1 motion segment under dynamic loading condition.

6.3.1 Methods

6.3.1.1 *Consideration of Material Properties of Intervertebral discs used in Dynamic Analyses*

The mechanical responses of FSUs are related to the material properties of each spinal component (Silva and Gibson, 1997), particularly, the intervertebral discs. The biomechanical responses of the intervertebral disc, hence the FSU,

depends greatly on the loading conditions as well (Markolf, 1972; Race et al., 2000; White and Panjabi, 1990).

The intervertebral disc is the most important component for bearing compressive loads experienced by the spine. It exhibits a stiffening effect with increasing compressive load, as shown in Figure 49 predicted by the FE model of T12-L1 and also in experimental studies (Markolf and Morris, 1974; Virgin, 1951). Though it has been known for some time that the intervertebral disc displays viscoelastic and time-dependent properties (Markolf, 1972; Virgin, 1951; White and Panjabi, 1990), the effect of loading rate on its mechanical properties of the disc is not known. Recently, Race et al (2000) used bovine intervertebral discs to quantitatively investigate this effect (Figures 53 and 54). Figure 53 shows the load-displacement responses for a fully hydrated disc obtained in their study under different loading rate, illustrating the stiffening of the intervertebral disc with load and rate of loading. The stress-strain curves derived in the experimental study of Race et al. (2000) were also plotted as shown in Figure 54, it shows that the Young's modulus, defined as the gradient of the curve, increases with the stress and stress rate. At the lower rates of loading, larger variations in the gradients of the curves were observed, while at the higher rates of loading, the gradients of the response curves of the disc was more consistent. Hence, loading rates has great influence in the mechanical response of the disc.

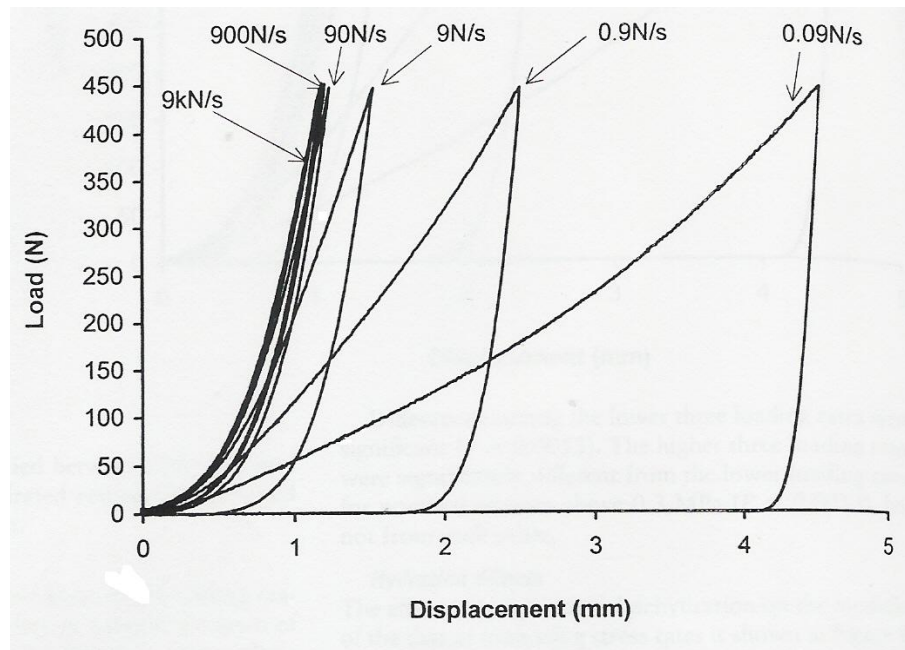


Figure 53 Typical load-displacement curves for a fully hydrated disc covering six different loading rates. The disc had a cross-sectional area of 300 mm^2 and a resting height of 7.3 mm. The stiffness of the disc (given by the gradient of the curves) increased with load and with loading-rate. (Adopted from Race et al., 2000)

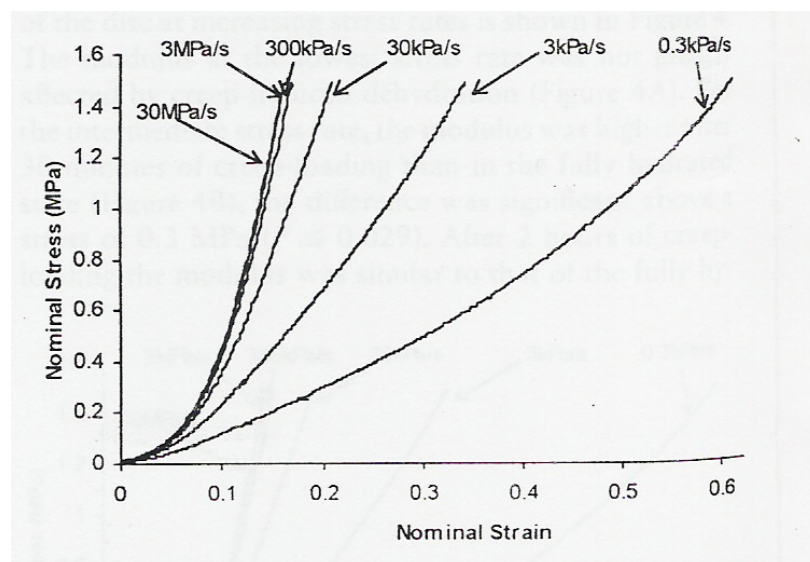


Figure 54 Stress-strain graphs. The gradient of these stress-strain curves provided the modulus of the intervertebral disc. Modulus is a measure of how hard it is to deform a material that is independent of its morphology. (Adopted from Race et al., 2000)

To date, there is no other similar experimental study for human spine available in literature. Though the bovine intervertebral discs (oxtails) used in the study by Race et al. (2000) were smaller than that of humans, comparatively, the magnitude of their mechanical properties may be altered but not the trends. Moreover, it is believed that the ratio of modulus increment should be similar between the bovine and human intervertebral discs; as there are some biomechanical and material similarities between the human and some animal spines such as those of bovine, deer and porcine due to their similar structures (Yu et al., 2002; Kumar et al., 2002; Yingling et al., 1997). The Young's modulus of the bovine intervertebral disc was determined based on the slope of the stress-strain curves in their linear portion corresponding stress 0.3-0.6 MPa (Figure 54). The Young's modulus calculated based on the middle linear portion generally represents the actual properties of the intervertebral disc (Panjabi et al., 1976a) and it is more reproducible and meaningful for axial compression (Markolf, 1972). The slope of the initially curved part of the curve would have little significance for compression, since, in vivo, in the erect position the human spinal column is under continuous compressive preload resulting from the superincumbent weight and muscle tension and the intervertebral disc provides very little resistance under compression at low loads (White and Panjabi, 1990). In addition, the slope of the linear portion at higher load does not represent the general situation in real life because the intervertebral disc seldom undergoes such higher load in real life (Panjabi et al., 1976a). Race et al. (2000) reported that the Young's modulus of the intervertebral disc determined based on the middle linear portion of the stress-strain curve at high loading rate (30MPa/s) were about 6 folds increase over those tested at low loading rate (0.3kPa/s). It is thus assumed that the

modulus of the human annulus and nucleus were also increased 6 times those used in the static analysis (Table 6 in Section 4.3.1) for the healthy disc and applied in the current dynamic analysis of T12-L1 motion segment to predict the burst fracture mechanism.

The material properties of the other spinal components were kept constant as adopted in the static analyses (Table 6 in Section 4.3.1). All materials, except the ligaments, were assumed to be linearly elastic and isotropic including the cancellous bone in the T12-L1 FE model although it is well known the cancellous bone shows anisotropy and does not show elastoplastic properties (Silva and Gibson, 1997; Ulrich et al., 1999). The von Mises stress failure criterion (Alexander, 1981) cannot be used for discussion of the vulnerability to failure of the cancellous bone. However, the anisotropy of vertebral trabecular bone is relatively weak, it is unlikely to have much appreciable effect on the stress distributions (Mizrahi et al., 1993). In addition, Brown and DiGioia (1984) demonstrated that the incorporation of anisotropic properties into two-dimensional models of the proximal femur results in no significant changes in the stress distributions. Hence, the author believed that the use of the isotropic material property for the cancellous bone would not greatly affect the stress distributions.

6.3.1.2 Boundary and Loading Conditions

Figure 55 shows the boundary and loading conditions used in the dynamic analysis. The inferior surface and spinous process of the lower vertebra L1 were fixed in all degrees-of-freedom. A rigid plate, of thickness 3 mm, having the exact

shape as the top surface of the T12 vertebral body was created on the top surface of the upper vertebra T12 to permit uniformly distribute and transfer of the impact compressive loads down to the vertebral body of T12. A ball with a radius of 11 mm and distant 5 mm above the rigid plate was made to impact the rigid plate at a velocity of 5.1 m/s at zero gravity to simulate real-life situation (Panjabi et al., 1995a). This speed corresponded to the ball dropping from a height of 1.4 m obscuring the influence of the gap between the ball and the rigid plate. The center of the ball and the centroid of the rigid plate were aligned in the vertical direction and contact was created between the lowest part of the ball and the top surface of the rigid plate during impact. Accordingly, the ball was constrained to only move along the vertical direction to ensure the direct axial impact in the vertical direction. The mass of the ball was 2.3 kg, which was the minimum required to produce fracture under the speed of 5.1 m/s in the experimental studies by Kifune et al. (1995, 1997) and Panjabi et al. (1995a, 1995b). The corresponding potential energy was about 31 J, and the responses within 30 ms of the entire dynamic analysis were analysed.

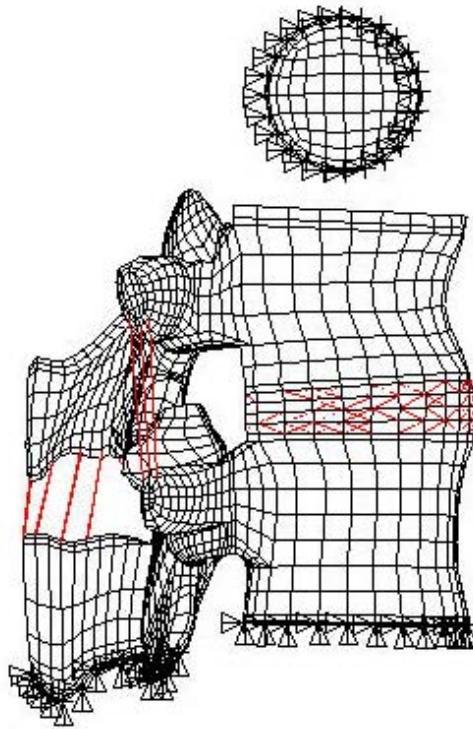


Figure 55 Boundary and loading conditions for the dynamic analysis of T12-L1 motion segment. The inferior surface and spinous process of the lower vertebra L1 are fully fixed, the ball is constrained to move only along the vertical direction. The impact speed of the ball is 5.1 m/s.

6.3.1.3 Extraction of Impact Force, Kinematic Parameters and Stress Contours

The impact force was extracted from the computed contact force in the vertical direction during the contact between the ball and the rigid plate. The displacement, velocity and acceleration of the moveable vertebra T12 during the dynamic process were plotted against time by extracting the vertical values of those parameters of a node (any node) on the superior surface of the T12 (all nodes on the superior surface of the T12 have the same motion), as the analysis was focused on the axial compression.

To relate predictions to vertebral failure risk, the following outcome variables during the impact in this study were extracted: Endplate bulges, von Mises stress - as an effective stress for stress distribution plotting, and the mid-sagittal plane radial displacement of the vertebral body - as an indicator for spinal canal encroachment.

The bulging deformations of inferior endplate of the T12 and superior endplate of L1 in the sagittal plane were plotted along the posterior-anterior direction at the instance when the moveable vertebra T12 reached maximum displacement.

The stress distribution on the endplates, in anterior and posterior cortical bones, the cancellous core and the pedicle base at the same instance were also plotted. As effective stress can be used as a failure criterion for elastoplastic materials, according to the theory of von Mises failure criterion (Alexander, 1981); hence, the von Mises stress was used as the effective stress when plotting the stress distribution. Due to elastoplastic properties of cortical bone (Reilly et al., 1974), the effective stress failure criterion will be used to establish which portions (regions) of the cortical shell are most likely to fail due to mechanical stresses. The vulnerability to failure of the cancellous bone cannot be discussed using the effective stress criterion, because the cancellous bone shows inhomogeneous and anisotropic material characteristics (White and Panjabi, 1990) instead of elastoplastic properties, as though the isotropic elastic material properties were assigned to the cancellous bone in the current study because of the large numbers of elements involved and the related computational difficulties. However, as the cortical bone, not the cancellous bone, is the first osseous structure of the vertebra to fail due to compression (Panjabi et al., 1998; Perrey, 1957), the analysis of cortical shell vulnerability is therefore a

good indicative of failure of the vertebra body structure. Hence, the endplate bulgings and internal stress distributions during the dynamic process could be combined to explain the burst fracture mechanism.

As Panjabi et al., 1995a, 1995b reported that the posterosuperior wall of L1 is the part where most fractures are observed during trauma, experimentally and clinically, the transient changes of the displacement of a node selected from the margin of the posterosuperior wall of L1 of the T12-L1 FSU and the regions of stress concentration during the impact process were plotted as the indicator of the encroachment of bony fragments retropulsed into the spinal canal (Whyne et al., 2001).

6.3.2 Results

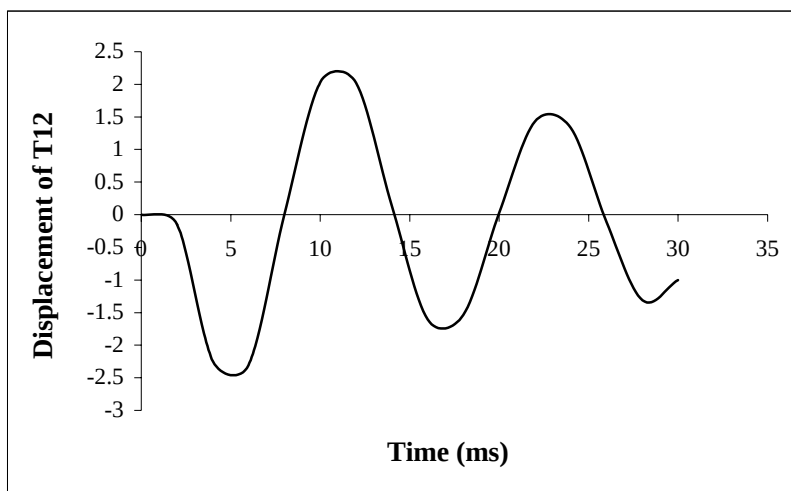
The impact force between the ball and rigid plate was computed to be about 7 kN and the displacement, velocity and acceleration with time of the moveable vertebra T12 during the dynamic process are illustrated in Figure 56. The bulges of the endplates are shown in Figure 57. The stress distributions are plotted in Figures 58-61. The transient change of encroachment is illustrated in Figure 62.

6.3.2.1 Displacement, Velocity and Acceleration of T12

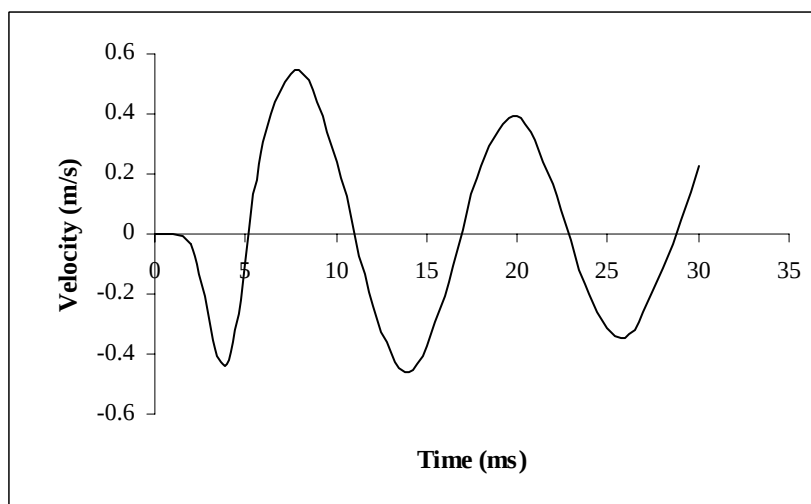
Figures 56A-C show the displacement, velocity and acceleration histories of the moveable vertebra T12 after impact. From these figures, it can be seen that the FSU began to vibrate after subjected to a sudden load. The maximum compressive displacement of about 2.3 mm of T12 was reached at time of 5.5 ms, corresponding

Chapter 6: Burst Fracture Mechanism Study

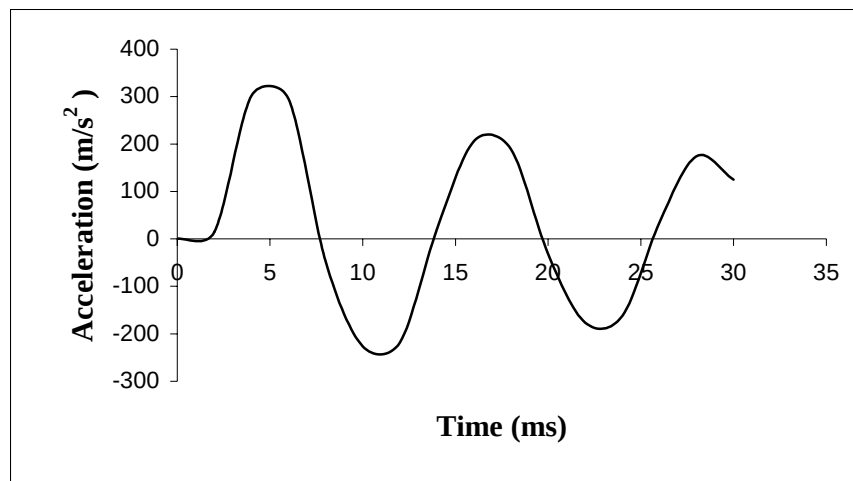
to the maximum displacement of the intervertebral disc (Figure 56A). At the same instance, its velocity reached the minimum value of 0 m/s (Figure 56B) and acceleration reached the maximum value of 300 m/s^2 (Figure 56C).



A. Displacement



B. Velocity



C. Acceleration

Figure 56 The histories of the displacement, velocity, and acceleration of T12 after impact during the whole dynamic process. (A) Displacement; (B) Velocity; (C) Acceleration.

6.3.2.2 Endplates Bulgings

Figure 57 shows the displacements of the inferior endplate of T12 and the superior endplate of the L1. The maximum bulges of both the endplates was similarly about 1.2 mm at the central portion of the endplates adjacent to the nucleus. The inferior endplate of T12 bulged upwards, while the superior endplate of L1 bulged downwards. The upward bulging displacement of the inferior endplate of the T12 was generally larger at the posterior region than at the anterior region.

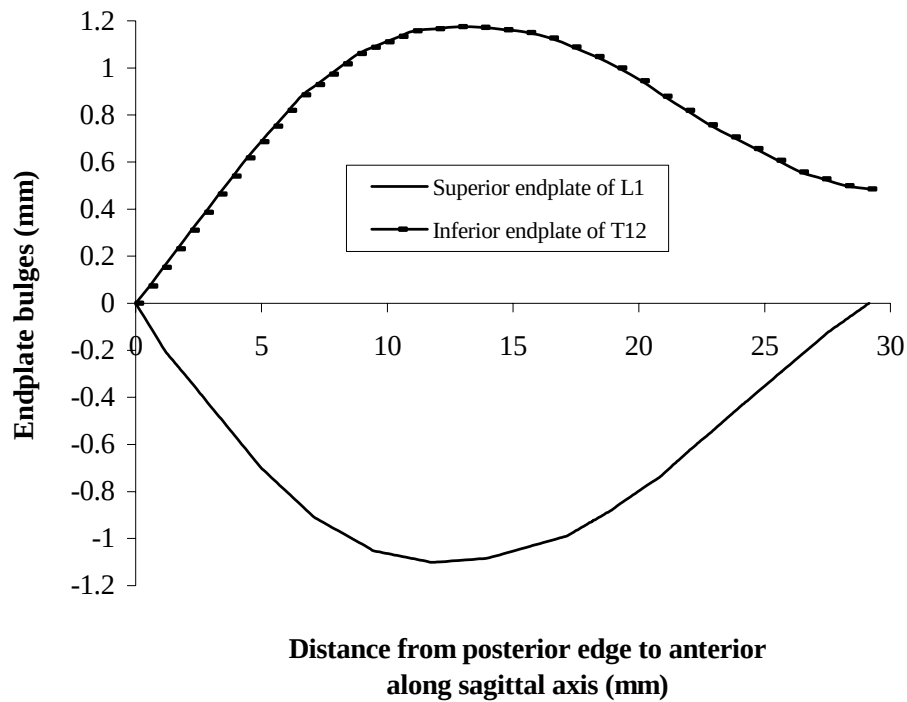


Figure 57 Compressive displacements endplates along mid-sagittal axis at time $t = 5.5$ ms corresponding to the maximum displacement of the intervertebral disc.

6.3.2.3 Stress Distributions

The stress distributions within the endplates, cortical bone, cancellous core and cortical shell near the pedicle base are plotted in Figures 58-61 and analyzed to deduce the burst fracture mechanism.

In Figure 58, it can be seen that the high stress regions of the upper and lower endplates were located around the center adjacent to the nucleus towards the posterior. The stress was found to be as high as 93.7 MPa. Figure 59 shows that, stress reached a maximum of 2.6 MPa at the central parts of the cancellous core adjacent to the endplates.

Stress concentrations within the cortical bone is shown in Figure 60, it shows that stresses were concentrated on the antero-inferior and postero-inferior parts of T12, as well as the middle-superior parts of L1 anterior and posterior. The most intensive stress was 74.8 MPa. The stress concentrated at the bases of the pedicle of the vertebra was 56.4 MPa as shown in Figure 61.

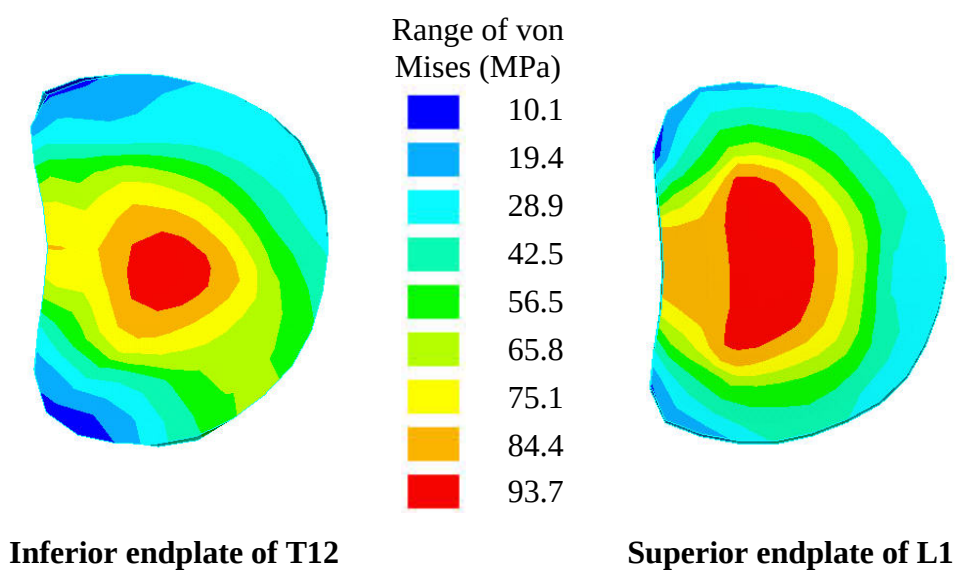


Figure 58 The stress distributions of the endplates at time $t = 5.5$ ms corresponding to the maximum displacement of the vertebra T12. Left: Inferior endplate of the upper vertebra T12. Right: Superior endplate of the lower vertebra L1.

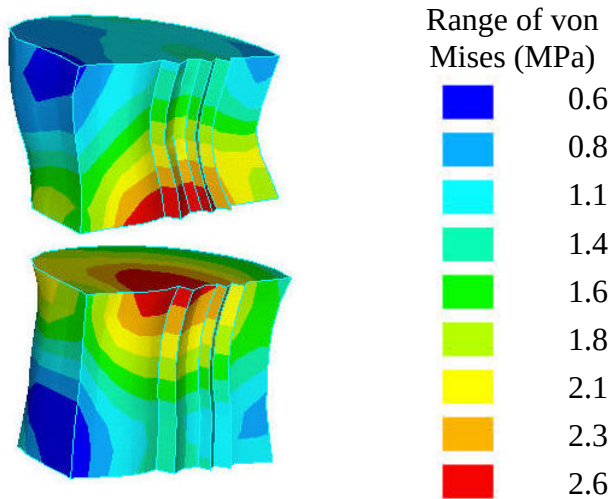


Figure 59 The stress distributions in cancellous core at time $t = 5.5$ ms corresponding to the maximum displacement of the vertebra T12. Half part of cancellous bone sectioned from mid-sagittal plane.

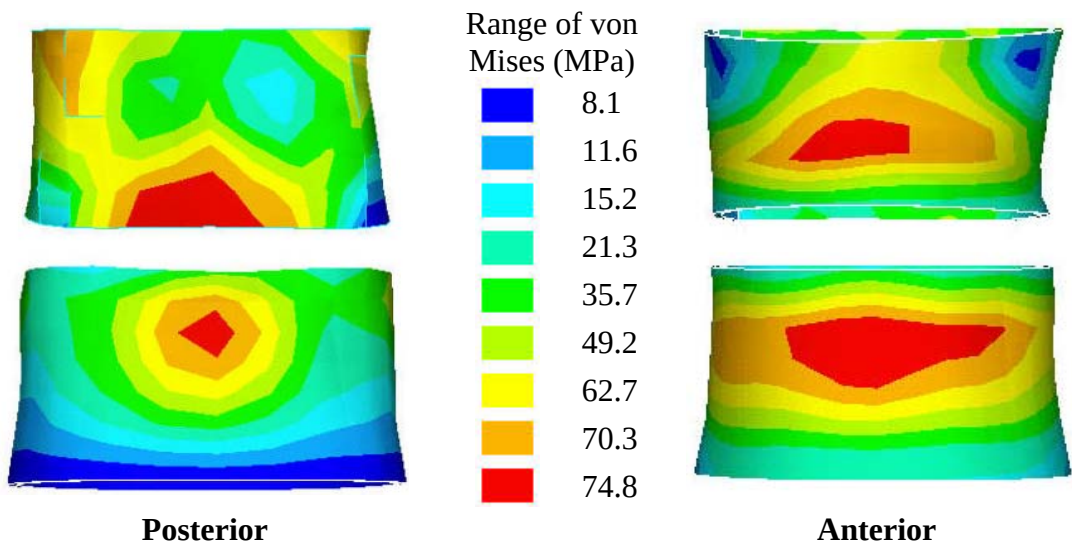


Figure 60 The stress distributions of the cortical bone at time $t = 5.5$ ms corresponding to the maximum displacement of the vertebra T12. Left: Posterior view; Right: Anterior view.

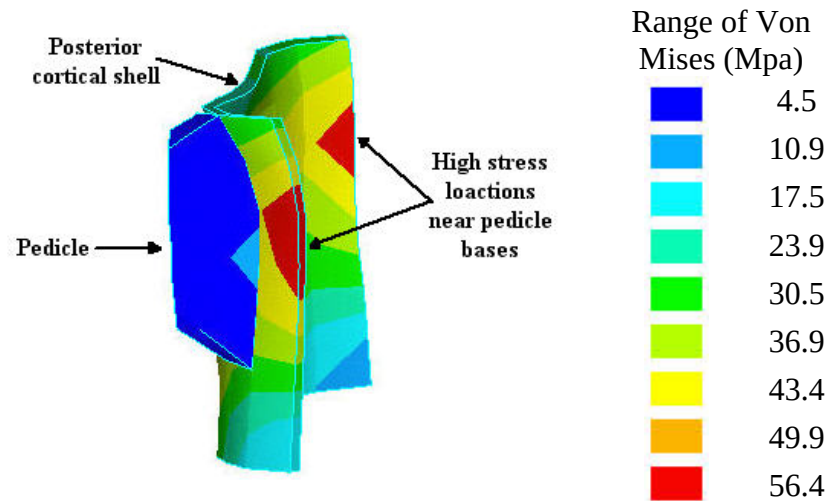


Figure 61 The stress distribution of pedicle bases of L1 at time $t = 5.5$ ms corresponding to the maximum displacement of T12.

6.3.2.4 Fragment Encroachment Representation

Figure 62 shows the history of the radial displacements of the posterior node of L1 selected from the area with high stress. During the impact process, the displacement transiently changed with the vibration of the intervertebral disc (Figure 56), where negative values represent the encroachments of fragments.

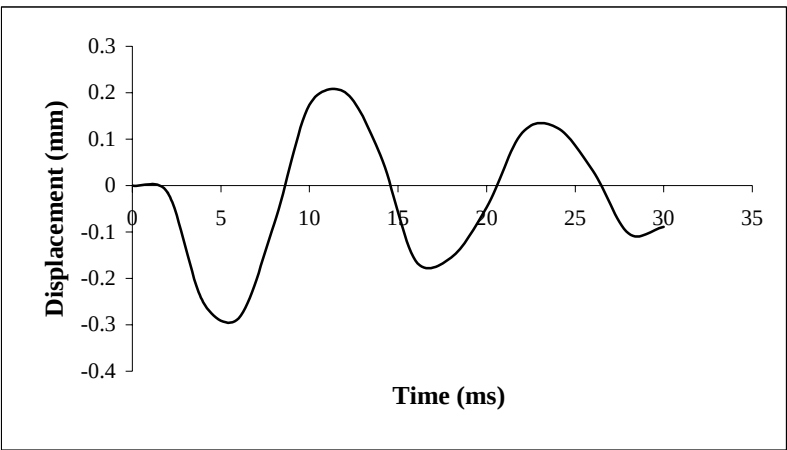


Figure 62 The displacement history of the node selected from the posterosuperior wall of the vertebra L1.

6.3.3 Discussion

Due to the lack of human intervertebral disc properties at high loading rates in literature, the intervertebral disc properties were characterised as those of the bovine intervertebral discs under different loading rates (Race et al, 2000) in the dynamic loading, and the computed internal changes of stress distribution in the FE analysis of T12-L1 under axial impact were investigated to predict the burst fracture mechanism.

In the past decades, spine and spinal cord injuries caused by traffic accidents and falls have increased greatly (Bensch et al., 2004; Yoganandan, 1989). The mechanism, instability and treatment of the injuries have been the research topics in clinical and experimental studies (Denis, 1983; Fredrickson et al., 1992; Kifune et al., 1995; Leferink et al., 2002; Oner et al., 1998, Panjabi et al., 1995a, 1995b; Tran et al., 1995; Zou et al., 1993). With the development and application of biomechanics, the mechanism of various spinal injuries has been further recognized. However, there are still a lot of shortcomings in the current classification of fractures based on the retrospective study on spinal injuries, for example, X-ray and CT provide only statistical results of displacement after spine trauma and do not completely reveal the transient changes of the fractures during the trauma. The mechanism of the burst fracture was never studied in detail during the dynamic process prior to this current study.

6.3.3.1 Burst Fracture Mechanism

For the normal disc model, the distributions of compressive stress across the posterior-anterior diameter of disc (Figure 50) indicated that the compressive load passed mainly through the center of the vertebral body to the nucleus of the disc causing increased pressure in the nucleus. Such increased pressure under axial dynamic load caused the endplates to bulge towards the vertebral body (Figure 57), resulting in the central parts of the endplates adjacent to the nucleus to experience a maximum effective stress of 93.7 MPa (Figure 58) and stress concentration within the central parts of the cancellous bone adjacent to the endplates (Figure 59). This will most likely cause endplate fracture to propagate from the central portion first. In summary, it is hypothesized that as the pressure inside the nucleus increases, the endplates bulges deeper towards the cancellous core and finally cracks during the impact process. This will allow the nucleus material to enter the vertebral body, and thereby pressurizing it more, squeezing the fat and marrow contents of the vertebral body out of the cancellous bone, resulting in vertebral body bursting from the anterior and posterior shell region which has a high stress concentration of 74.8 MPa (Figure 60). Hence, burst fracture can be hypothesized to occur when the rate at which nucleus material enters the vertebral body is greater than the rate at which the fat and marrow contents of the body can be expelled.

In burst fracture, the cause of the fracture of the anterior column can be attributed to the predicted stress concentrations at the antero-inferior part of T12 and the middle-superior parts of L1 anterior (Figure 60) while the retropulsion of the bony fragment into the spinal canal can be explained via the high stress in the

cortical bone in the postero-inferior part of T12, as well as the posterior of L1 (Figure 60).

In burst fractures of spinal segments produced under rapid axial loading, both endplates of the vertebral bodies may fracture resulting in complete loss of axial stability, and the lamina is usually split and accompanied by the widening of the interpedicular distance (Gertzbein, 1992). In this study, as attention was focused only on the clarification of the mechanism of vertebral body bursting, therefore, the stress distributions within the posterior elements were not investigated.

In this study, the mechanism of burst fracture was quantitatively predicted by internal changes such as stress distribution within the FE model of T12-L1 motion segment with a normal disc during the impact process. The fracture patterns deduced from the stress distribution patterns corresponded well with those observed in clinical practice. The deduced mechanism in this study explained why burst fractures are most often seen in young people with healthy discs. The T12-L1 FE model was therefore considered representative of thoracolumbar segment of the human spine. Hence, the author believes that this study provided new insights into the internal changes of the vertebral bodies and disc of the TLJ during the dynamic process, and these informations can be useful in aiding the management of these fractures. In literature, experimental studies have shown that failure caused by impact loading occurs in the endplate or posterior region of the cortical shell, but there is a lack of consensus as to which region fails first. Willen et al. (1984) found that dynamic axial loading cause an initial fracturing of the vertebral body. Conversely, Gertzbein (1992) reported that stresses in the endplate were greater than those within the outer

vertebral cortex, and that fractures could be initiated in the endplate region. However, Lee et al. (2000) found that high stresses could lead to fracture in the endplate and the posterior surface of the cortical wall using FE model, and the initiation of fractures depend on the failure strengths of these regions. In this study, the most intensive stress within the central portion of the endplates was 93.7 MPa, which is larger than the 74.8 MPa in the anterior and posterior cortical bone. This agrees well with the findings of Gertzbein (1992) as mentioned above. Heggeness and Doherty (1997) noted that the cortex of the vertebra canal thinned abruptly near the base of the pedicle in their anatomical study of the thoracolumbar vertebrae, suggesting an explanation for bony fragment retropulsed into the spinal canal during thoracolumbar burst fracture. In the current FE model, the thickness of the cortical cortex was uniform; hence the stress distribution of the posterior cortical shell was not affected by its thickness.

The highly stressed portions in the endplates predicted by the current FE model of T12-L1 motion segment suggested the region where traumatic damage is most likely to originate (Panjabi et al., 1995a, 1995b). In the Liu et al. (2002) FE study of T12-L1 FSU and Shirado et al. (1992) investigation of a two-dimensional FE model of L3-L4, similar high stress concentration of the endplates for the normal disc were also reported. This explains the clinically most observed formation of Schmorl's nodes by the nucleus materials in the central part of the vertebral body adjacent to the nucleus after endplate fracture in young people (Adams et al., 2000).

Hongo et al. (1999) reported the high tensile and compressive strains found at the base of the pedicle of T10, L1 and L4 in their experimental study, and indicated

that the base of the pedicle is the site of fracture initiation. The current FE model of the T12-L1 under impact predicted the high stress also concentrated on the pedicle bases (Figure 61). However, the highest predicted value in pedicle base was 56.4 MPa, less than those of the endplates (93.7 MPa) (Figure 58) and cortical bone (73.4 MPa) (Figure 60). Hence, the current model of T12-L1 with the normal intervertebral disc did not seem to indicate burst fracture initiation from the pedicle base but mainly caused by fracture of the endplate. In 1986, Kurowski and Kubo reported that the failures of the vertebral body in the area near the pedicles were not common. This supported the current result about the pedicle fracture.

The model predicted that stress concentration occurred in the anterior cortex (Figure 60). In 1979, Hakim and King had also reported that the anterior aspect of the vertebra was a region of high magnitudes of strain in their experimental and FE studies of lumbar vertebrae. This is the area where anterior wedge fractures commonly occur (Denis, 1983; Panjabi et al., 1995b).

Stress concentration on the posterior cortical shell of the vertebral body under vertical impact (Figure 60) could cause posterior wall failure, resulting in bony fragments encroaching into the spinal canal. Surgery is often recommended in these cases to decompress the spinal canal. Thus, post-injury computed tomograph (CT) scans are often used as an indicator for decompressive surgery, but there is doubt as to whether the fragment position seen on such scans represents the true extent of the canal occlusion produced during the fracture process. In addition, in burst fractures seen clinically, poor correlation often exists between the neurological deficit and the spinal canal encroachment measured on post-trauma radiographic images. This

implies the spinal cord injuries are caused by instantaneous impact. Thus, the information of transient changes in canal encroachment and stress distribution of fracture location during spinal trauma is very important for diagnosis and treatment of spinal cord injuries for clinicians. In the current FE study, as the real burst fracture could not be simulated; hence, the absolute spinal canal occlusion was not investigated. However, the displacement of the node on the posterior wall with stress concentration indicated that the encroachment of the bony fragments is a transient process during vertical impact, and the final displacement at the end of the simulation in Figure 62 was the least. This is consistent with the findings of Panjabi et al. (1995b) found in their experimental study, in which the authors reported that the dynamic canal encroachment was significantly greater than the static canal encroachment seen on post-trauma radiographs or CT scans. Thus, it can be deduced that in real-life situations, during the vibration (Figure 56) of the intervertebral disc after burst fracture of the vertebral body due to rapid axial loading, the anterior and posterior longitudinal ligaments will stretch due to the collapsed vertebra and force the retropulsed bony fragments back to the vertebral body when the disc stretches further than its normal height (Figure 56A). This explained the poor correlation between the neurological deficit and the spinal canal encroachment measured on post-trauma radiographic images. This also further confirmed that distraction force applied to the injured spine with vertebral fracture via Harrington distraction rods usually used clinically is an effective method to reduce fragment encroachment (Fredrickson et al., 1992; Kifune et al., 1997; Willen et al., 1984).

6.4 Influence of Disc Degeneration on Burst Fracture Mechanism

There appears to be a link between intervertebral loading and disc degeneration of the spine (Adams et al., 2000). The fluid content of the disc is an important determinant of its mechanical response. Disc degeneration and surgical discectomy result in small nucleus pressure and altered loading distribution in compression (Adams et al., 1993).

The investigation of load transmission in normal and degenerated discs (Section 6.2) indicated changes in load transmission, therefore, the stress distributions within the adjoining vertebral bodies could be altered. Hence, this section will discuss the influence of disc degeneration on the burst fracture mechanism through the stress distributions within the vertebral bodies.

6.4.1 Methods

6.4.1.1 *Material Properties*

All the material properties were kept the same as those used in the FE model of T12-L1 with degenerated disc (Section 6.2.1.1) in the static study of load transmission.

6.4.1.2 *Boundary and Loading Conditions*

The same boundary and loading conditions used in the impact study of intact T12-L1 FE model shown in Figure 55 were adopted.

6.4.1.3 Extraction of Stress Contours

The same extraction method of plotting stresses within the endplates and vertebral bodies as described in Section 6.3.1.3 was adopted.

6.4.2 Results

The stress distributions of each of the components in the vertebral bodies with degenerated disc under dynamic loading are shown in Figures 63-65.

Figures 63-65 show that a degenerated disc, with lower intradiscal pressure, introduced stresses to the lateral aspects of the endplate, the cancellous bone and the vertebral body wall. In Figure 63, the high stresses were located around the rim of the endplates, while the central portions of the endplates experienced the lowest stress. Similarly, for the cancellous core, the intense effective stresses occupied the periphery, while the lowest stresses were found in the central parts (Figure 64). In Figure 65, the high stresses were seen in the middle-superior part of the posterior wall and the superior aspect of the anterior shell for L1. The high stresses were also found in the posteroinferior cortex and inferior edge of T12.

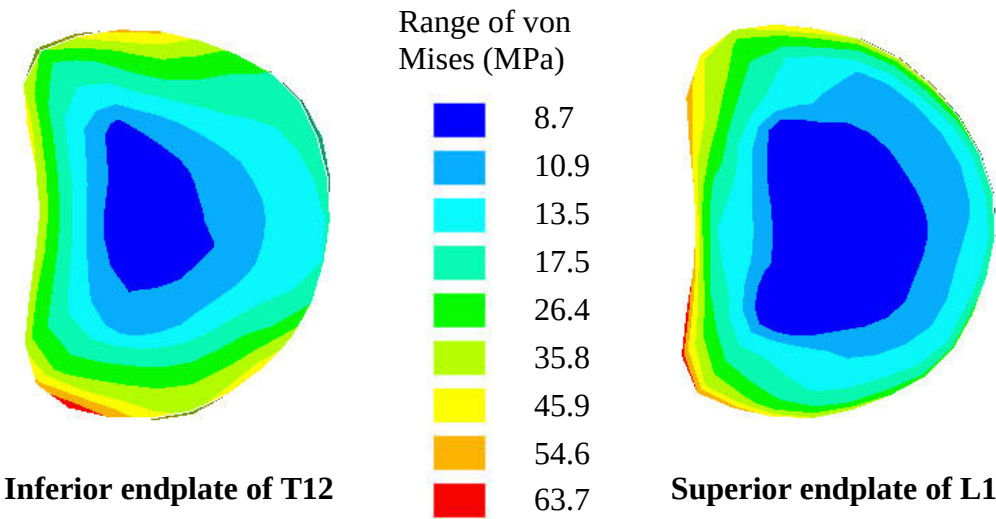


Figure 63 Stress distributions of endplates. Left: Inferior endplate of T12; Right: Superior endplate of L1.

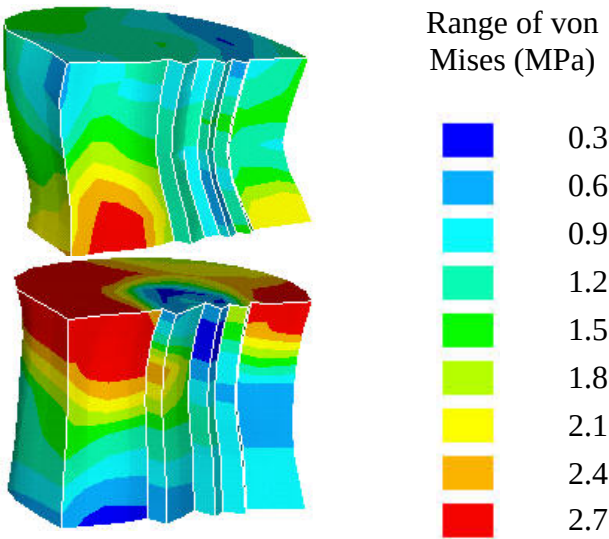


Figure 64 Stress distributions of half parts of cancellous bones cut from mid-sagittal plane. Half part of cancellous bone cut from mid-sagittal plane.

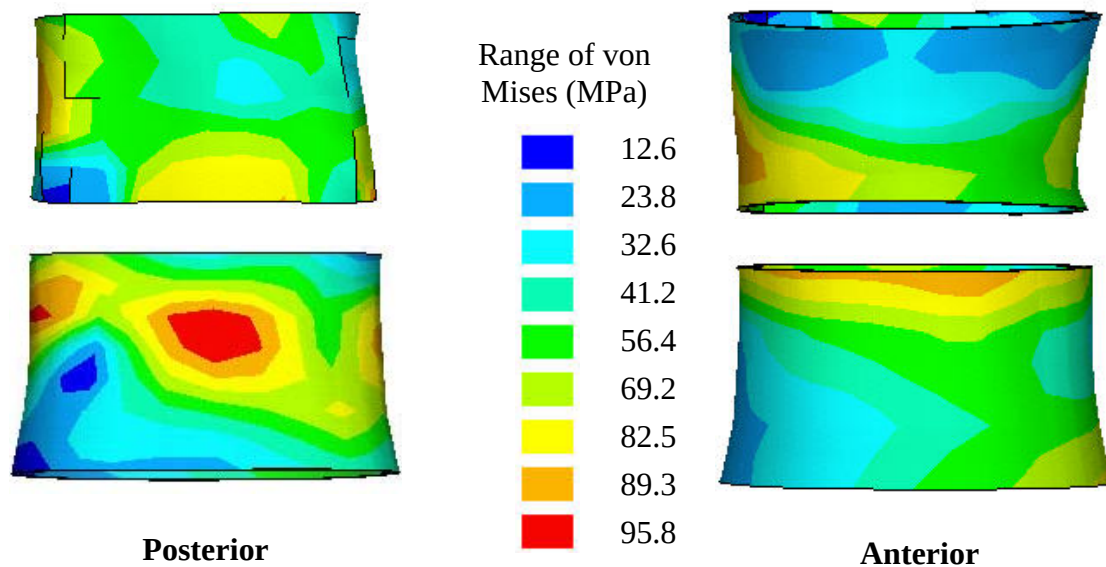


Figure 65 Stress distributions of cortical bones. Left: Posterior wall; Right: Anterior wall. Left: Posterior view; Right: Anterior view.

6.4.3 Discussion

6.4.3.1 Influence of the Degenerated Disc on Burst Fracture Mechanism

With disc degeneration, the stress distributions in each component of the vertebral bodies (Figures 58-61 and 62-65) were altered due to the change in the path of load transmission (Figure 50). Healthy disc with pressure-containing nucleus pulposus introduced loads to the central part of the vertebra, so the central portions of the endplates were highly stressed. Disc degeneration deviated load transmission through the peripheral portions of the vertebral body resulting in high stresses on lateral aspects of the endplates and the vertebral body cortical wall. As a result of these differences in the load transmission and stress distributions, the sites that failed first also change, thus indicating strong dependence of burst fractures on the conditions of the intervertebral disc. Axial impact on motion segment with non-degenerated disc is likely to produce cracks in the central portions of the vertebral

endplates first and then results in the burst fracture of the vertebral body (Section 6.3.3.1). Compression of motion segment with degenerated disc, on the other hand, is likely to collapse the cortical wall.

The present analysis results of thoracolumbar fractures via the stress distributions corresponded well with experimental findings (Brown et al., 1957; Shirado et al., 1992), relating the quality of the disc with the manner of fracture and path of load transmission in the vertebral bodies subjected to axial compression. The influence of disc degeneration on the burst fracture mechanism predicted by the stress distributions were supported by the findings obtained by Shirado et al. (1992) in their experimental study, in which, out of six specimens with degenerated discs, four were totally collapsed, and four had no endplate disruptions. If the vertical force was just in front of the midline of the spine compressing the anterior lip of the affected vertebra, the vertebra collapsed with a wedge fracture most common in elderly patients (Magerl et al., 1994; Panjabi et al., 1995a).

The current FE analysis predicted differences in load transmission and stress distribution for normal and degenerated discs. This may explain the findings reported by Rockoff et al. (1969), that less load is transmitted by way of the central part of vertebra in older subjects than in those under 40 years of age.

Degeneration and dysfunction of the disc decreased the load borne by the vertebral body above the disc. Therefore, the risk for compression burst fracture in this particular vertebral body would be much decreased compared to those adjacent to normal disc, even though osteoporosis has decreased the strength of the vertebral

body. Subjected to the differences in load transmission (Figure 50) and stress distribution (Figures 58-61 and 63-65), modes and locations of damage of the vertebral bodies are also different for healthy and degenerated discs. This would explain burst fracture occurring mostly in young people. From the results of this study, aging changes in the disc are unlikely to make the spine more susceptible to burst fracture in axial impact.

6.4.3.2 Osteophyte Formation

Osteophyte is commonly found in the degenerated motion segment of those used to heavy work (Vernon-Roberts, 1978). Degenerative changes due to increasing age or wear and tear affect the central intervertebral and posterior intervertebral joints (Vernon-Roberts, 1978). In central joints (affected first), greater loads are placed on the annulus fibrosus after disc degeneration (Figure 50). Eventually the annulus is unable to sustain the load; the disc loses height; and this changes the function of all the joints in the affected area. This can lead to osteophyte formation around vertebral bodies (Vernon-Roberts, 1978). In the posterior intervertebral joints the changes are attrition of the articular cartilage and osteophyte formation at the joint margins (Vernon-Roberts, 1978).

In Figures 63 and 65, for a degenerated disc, high stress can be found in the vertebral body rims. The change of stress state in the rims because of disc degeneration corresponds to osteophyte formation. Osteophytes appear in degenerated motion segments and form new bone. This suggests that disc degeneration induces bone remodeling processes in the rims. Change of strains,

obviously related to the change of stresses, seems to be the reason for the bone remodeling. Therefore, the change of stresses in the rims due to disc degeneration may be a reason for osteophyte formation. Similar findings were also observed by Kurowski and Kubo (1986) FE studies of lumbar spine. Shah et al (1976) also reported stress concentration in vertebral body rims by using a photoelastic model of vertebra compressed between two rubber plates. This experiment was setup to correspond to the case of modeling stresses for degenerated discs and the current FE analysis results tallied closely with their results.

6.4.3.3 Disc Degeneration

With aging, the degenerative changes to the intervertebral disc are likely to occur within the nucleus. It is often observed that the nucleus pressure is reduced due to a loss of hydration of the disc material (Panjabi et al., 1988). Degenerated disc shifted load transmission to peripheral portions of the vertebral body and placed high stresses on lateral aspects of the endplates and the vertebral body cortical wall under compressive loads. Thus, the corresponding fracture pattern in this particular vertebral body was much different from those adjacent to normal disc (Section 6.4.3.1).

Although the specific mechanisms by which the degenerative changes to the intervertebral disc occur and progress are not completely clear (Frei et al., 2001), a number of factors may accelerate or contribute to the normal age-related deterioration of discs, including declining nutrition of the central disc regions, loss of viable cells, cell senescence, post-translational modification of matrix proteins,

accumulation of degraded matrix molecules, and fatigue failure of the matrix. The decrease in central disc nutrition caused by increasing volume of avascular tissue with growth, loss of peripheral blood supply, and alterations in the matrix appears to be the most (Buckwalter, 1995).

The endplate between the vertebral body and the intervertebral disc is an important structure for the nutrition of the disc (Holm, 1993). The endplate is also known to deform significantly under compressive loads (Brinckmann et al., 1983). It seems reasonable that this structure may have a connection with the degenerative process of the disc. Hence, the endplate fracture predicted on the endplate central portion due to intense stress concentrated on it in the FSU of T12-L1 with non-degenerated disc (Section 6.3.3.1) would cause the underlying risk of disc degeneration. The mechanism of the internal disc disruption would be probably as follows: The damaged endplate deforms more when under load, allowing more space for the hydrated nucleus pulposus, or allowing more nucleus tissue to pass through it. The nucleus therefore experiences a reduction in pressure, which is similar in amount to the reduction seen in degenerated disc. Thus, the decompressed disc would bulge more and lose height. This process is strengthened by water loss after sustained loading. Less of the applied compressive force is resisted by the decompressed nucleus, so more must be resisted by the surrounding annulus (Figure 50B). High stress gradients in the annulus then force the inner lamellae inward toward the decompressed nucleus and the outer lamellae outward (Figure 51B). The buckling of the lamellae is encouraged by the accompanying loss in disc height. This mechanism of disc degeneration process may be responsible for the degenerative changes to the

intervertebral discs in young people. This indicates that the endplate between the vertebral body and the intervertebral disc is an important structure for the disc degeneration. Frei et al. (2001) had also reported similar observation in which the endplate had a connection with the degenerative process of the disc.

6.5 Summary

The burst fracture mechanism and the influence of disc degeneration on it were explored by internal changes such as load transmission and stress distribution in the FE analysis of T12-L1 under axial impact. The paths of load transmission and stress distribution in the thoracolumbar vertebral body were found to be strongly dependent on the conditions of the intervertebral discs. Thus, the induced fracture modes were also different. The following findings by the current FE analyses were reached:

- Healthy disc with incompressible nucleus pulposus directed load through the central part of the vertebra, so that the central portions of the endplates and the cancellous bone adjacent to nucleus were burdened with high stress.
- Degenerated disc shifted load transmission to peripheral portions of the vertebral body and placed high stresses on lateral aspects of the endplates and the vertebral body cortical wall.
- With non-degenerated disc, cracks would first appear on the endplate central portion due to intense stress concentrated on it. Subsequently, it was

hypothesized that the nucleus material would enter the cancellous core, further pressurizing it to result in bursting of the vertebral body. This provided new insight into the etiology of burst fractures and aid in the management of the treatment of burst fractures. The endplate fracture may cause the underlying risk of the disc degeneration.

- With disc degeneration, compression fractures with collapsed vertebral bodies are most common in the elderly due to the age-related changes in intervertebral disc. Burst fractures seldom occur in the elderly, due to the path of load transmission and stress distribution. The high stresses found in the vertebral rims accounted for the osteophyte formation in the elderly.

Chapter 7 CONCLUSIONS

In human, the thoracolumbar junction (TLJ) is the transitional region where there are changes from the normally kyphotic curvature and relatively less mobile in the thoracic region to the normally lordotic curvature and relatively mobile lumbar region. The transitional vertebra T12 is uniquely structured with thoracic-type superior and lumbar-type inferior articular processes. The unique articulations at the TLJ result in the changes of IARs and its vulnerability to burst fracture. The understanding of the exact location of IAR of a FSU could be of value in the study of its normal function and the prevention, diagnosis, and treatment of pathologic conditions. Also the understanding of the paths of load transmissions and stress distributions in the thoracolumbar vertebral bodies during axial impact loading will help to analyze burst fracture mechanism.

Accordingly, in this study, three-dimensional FE models of T11-T12 and T12-L1 were generated to predict the locations and loci of the IARs at levels T11-T12 and T12-L1 of thoracolumbar spine under flexion/extension, left/right lateral bending, and left/right axial torque in three anatomical planes. The mechanism of the burst fracture and the influence of the disc degeneration on it were also investigated using the T12-L1 FSU under axial impact.

The notable findings of the IAR study were the significant differences of the predicted loci of the normal healthy and degenerated FSUs T11-T12 and T12-L1. The locations and loci of the IARs obtained in this study could provide further understanding into the kinematics and biomechanical responses of the human spine.

The locations of the IARs were found to vary in all three anatomical planes with increasing applied moments. Therefore, for an accurate prediction of the states of stress and strain, it is essential that the FSU not be forced to rotate about a predetermined axis which remains stationary during the process of the loading. Hence, *in vitro* biomechanical studies, in which the axis of rotations was pre-defined by the test machine, are likely to yield erroneous results. It is believed that this experimental design restricts the natural physiological movements of the spine. The kinematics effects of a rotation between two vertebrae are dependent on the location of the center of rotation.

For the dynamic study of T12-L1 motion segment, the burst fracture mechanism and the effect of the degenerated disc on it were revealed by the stress distributions in the vertebral bodies. The paths of load transmissions and stress distributions in the thoracolumbar vertebral bodies were found to be strongly dependent on the conditions of the intervertebral discs and the induced fracture modes were also differed.

Healthy disc with incompressible nucleus pulposus, directed load through the central part of the vertebra, the FE models predicted that the central portions of the endplates and the cancellous bone adjacent to nucleus were burdened with high stress. Thus, the fracture would initiate from the endplate central portion first due to the intense stress concentration. It was hypothesized that the nucleus material would subsequently enter the cancellous core, further pressurizing it to result in the bursting of the vertebral body.

However, degenerated disc deviated load transmissions to the peripheral portions of the vertebral body and placed high stresses on lateral aspects of the endplates and the vertebral body cortical wall. This would cause compression fracture with collapsed vertebral body, which is most common in the elderly due to the age-related changes in intervertebral disc and vertebral body material properties. The burst fracture seldom occurs in the elderly, due to the change in the manner of the load transmission and stress distribution. The high stresses found in the vertebral rims accounted for the osteophyte formation in the elderly.

In conclusion, it is hoped that the results obtained in this project will provide further understanding of the biomechanical response of the human thoracolumbar spine, which is important for the prevention, diagnosis, and treatment of spinal injuries.

Chapter 8 LIMITATIONS AND RECOMMENDATIONS

8.2 Limitations

There were several limitations to this present study. Firstly, trunk muscles, known to be the primary stabilizers in the spine, are not included in the current study. Thus, the predicted locations and loci of the IARs of the TLJ FSUs are only an approximate reflection of the real life situation, and the burst fracture mechanism study was only a qualitative one, though the high stress values were quantitatively analyzed due to lack of the influence of the musculature.

Secondly, the material properties of the normal intervertebral disc under dynamic loading and degenerated disc were derived from the previous studies (Race et al. 2000) due to lack of information in literatures. Hence, the mechanism of burst fracture was predicted by the internal changes of stress distribution based on the assumption of the material properties of the intervertebral disc under dynamic loading. In addition, the disc degeneration simulated in the present study does not refer to the clinical classification of degenerated discs. They correspond to the beginning of the degenerative process when the nucleus pulposus retains liquid-like properties.

Thirdly, the actual burst fracture fragmentation could not be simulated, thus, the real fragment encroachment was not predicted. Likewise, the actual cracking of the endplates and the restricted functionality of the finite elements used for nucleus,

hence, the movement of the nucleus materials entering the vertebral bodies were not simulated.

8.2 Suggestions for Future Works

In order to better understand the kinematics and dynamics of the thoracolumbar spine, the followings are suggested for the improvement of the current study:

- Parametric analyses of the anatomical data of the intervertebral disc and the insertion points of all the ligaments should be considered to examine the sensitivity of changing the value of these parameters on the results, e.g. IAR etc.
- Trunk muscles, known to be the primary stabilizers for the spine, not included in the current analysis, may be incorporated into the FE models, in order to directly apply the predicted results to the real-life situations.
- In order to simulate the real bursting of the vertebral body to produce fragment retropulsed into the spinal canal in the FE model, algorithms may be written to simulate the complex contact relations between discretised elements of vertebral body.
- In order to simulate the fluid movement in between the intervertebral disc and vertebral body after endplate fracture, poroelastic FE model may be constructed. The hypothesis that burst fracture could occur when the rate at

which nucleus material enters the vertebral body is greater than the rate at which the fat and marrow contents of the body can be expelled would be further explored and verified by simulating fluid movement in between the intervertebral disc and vertebral body after endplate fracture in a poroelastic FE model.

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APPENDIX A: PUBLICATIONS

Peer-reviewed journal papers:

Qiu TX, Teo EC, Lee KK, Ng HW, Yang K. Kinematics of the Thoracic T10-T11 Motion Segment: Locus of Instantaneous Axes of Rotation in Flexion and Extension. *Journal of Spinal and Disorder Techniques* 17:140-146, 2004.

Qiu TX, Teo EC, Lee KK, Ng HW, Yang K. Validation of T10-T11 finite element model and determination of instantaneous axes of rotations in three anatomical planes. *Spine* 28:2694-2699, 2003.

Teo EC, Qiu TX, Yang K, Ng HW and KK Lee. Biomechanical Effect of T10-T11 Posterior Elements on the Instantaneous Axes of Rotation in Sagittal Plane. *Journal of Musculoskeletal Research* 7:191-200, 2003.

Qiu TX and Teo EC. Instantaneous Axes of Rotation of the Thoracolumbar T12-L1 Intervertebral Joint. *ASME Journal of Biomechanical Engineering* (Accepted with some revisions).

Qiu TX and Teo EC. Finite Element Modeling of Human Thoracic Spine. *Annals of Biomedical Engineering* (Accepted).

Qiu TX and Teo EC. Investigation of thoracolumbar T12-L1 burst fracture mechanism using finite element method. *Medical Engineering and Physics* (Accepted with some revisions).

Qiu TX and Teo EC. Comparison of kinematics between thoracolumbar T11-T12 and T12-L1 motion segments. *IEEE Transactions in Biomedical Engineering* (Submitted).

Qiu TX and Teo EC. Comparisons of stress distributions of thoracolumbar T12-L1 vertebrae with normal and degenerated discs under dynamics. *Journal of Spinal Disorders & Techniques* (Submitted).

Conferences

Qiu TX and Teo EC. Finite element study of instantaneous axes of rotation of thoracolumbar T12-L1 intervertebral joints. A paper presented at the 3rd Scientific Meeting of the Biomedical Engineering Society (Singapore), 21 May, 2004.

Qiu TX and Teo EC. Finite Element Study on Kinematics of the Thoracic T10-T11 Motion Segment: Assessment of the Locus of Instantaneous Axes of Rotation in the Sagittal Plane. A paper presented at World Congress on Medical Physics and Biomechanical Engineering, Sydney, Australia, 24-29 Aug 2003.

Qiu TX, Lee KK, Ng HW, Yang K, Teo EC. Mathematical modeling of thoracic T10-T11 spinal motion segment: Validation and study of instantaneous axes of rotation in the sagittal plane. A paper presented at International Congress on Biological and Medical Engineering, Singapore, 4-7 Dec 2002.

Lee KK, Qiu TX, Teo EC, Ng HW and Yang K. Mathematical Modeling of Thoracolumbar Spine using ANSYS. A paper presented at 4th ASEAN ANSYS Users Conference, Singapore, 5-6 Nov 2002.

Qiu TX, Teo EC, Lee KK, Ng HW, Yang K. Finite Element Study on Loci of Thoracic Instantaneous Centers of Rotation in Sagittal Plane. A paper presented at 2002 Combined Orthopaedic Meeting, Singapore, 12-16 Oct 2002.

Lee KK, Qiu TX and Teo EC. 3-D finite element modeling of lumbar spine (L2/L3) using digitizer. A paper presented at 3DD-2002, Singapore, 25 Feb 2002.

APPENDIX B: SAMPLE CALCULATIONS OF ANGULAR MOTION OF T12-L1

A.1 The Calculations of Angular Motion of T12-L1 FSU Under Flexion and Extension Pure Moment of 7.5 Nm

Two nodes A (X_a, Y_a, Z_a) and B (X_b, Y_b, Z_b) were selected on the superior surface of T12, after rotation with applied flexion and extension pure moment of 7.5 Nm, the new positions of the two selected nodes became Af (X_{af}, Y_{af}, Z_{af}) and Bf (X_{bf}, Y_{bf}, Z_{bf}) under flexion; Ae(X_{ae}, Y_{ae}, Z_{ae}) and Be (X_{be}, Y_{be}, Z_{be}) under extension, respectively. Under flexion and extension pure moment, the primary motion of the FSU is in the sagittal plane, according to loading and boundary conditions defined in the current study (Figure 35 in Section 4.3.2), the sagittal plane is YZ plane. Hence, only coordinates of Y and Z were used to calculate the angular rotation degrees under flexion and extension.

Before rotation, the coordinates values of Y and Z for nodes A and B were, respectively:

$$\text{A: } Y_a = 11.994 \text{ mm, } Z_a = 6.4847 \text{ mm}$$

$$\text{B: } Y_b = 11.231 \text{ mm } Z_b = 10.2285 \text{ mm}$$

After flexion, A and B moved to Af and Bf, under extension, A and B shifted to Ae and Be, the coordinates values of Y and Z for Af and Bf, Ae and Be are as follows:

Appendix B: Sample Calculations of Angular Degrees

After flexion:

Af: $Y_{af} = 12.1826$ mm, $Z_{af} = 7.7122$ mm

Bf: $Y_{bf} = 11.8311$ mm, $Z_{bf} = 10.3127$ mm

After extension:

Ae: $Y_{ae} = 11.5381$ mm, $Z_{ae} = 5.1120$ mm

Be: $Y_{be} = 11.4251$ mm, $Z_{be} = 10.5720$ mm

A.1.1 The calculation of the angular motion after flexion

The angular motion α_f after flexion was calculated in the following manner:

The angle α between line segment AB and Y axis is:

$$\begin{aligned}\alpha &= \arctan (Z_b - Z_a) / (Y_b - Y_a) \\ &= \arctan (-12.2) \\ &= -85.3204^\circ\end{aligned}$$

The angle α' between line segment AfBf and Y axis is:

$$\begin{aligned}\alpha' &= \arctan (Z_{bf} - Z_{af}) / (Y_{bf} - Y_{af}) \\ &= \arctan (-7.39845) \\ &= -82.30826^\circ\end{aligned}$$

The angular motion α_f after flexion is (refer to Figure 36):

$$\alpha_f = \alpha' - \alpha = -82.30826^\circ - (-85.3204^\circ) = 2.99954^\circ$$

Appendix B: Sample Calculations of Angular Degrees

A.1.2 The calculation of the angular motion after extension

The angular rotation degree α_e after extension was calculated in the following manner:

The angle α between line segment AB and Y axis is:

$$\begin{aligned}\alpha &= \arctan (Z_b - Z_a) / (Y_b - Y_a) \\ &= \arctan (-12.2) \\ &= -85.3204^\circ\end{aligned}$$

The angle α'' between line segment AeBe and Y axis is:

$$\begin{aligned}\alpha'' &= \arctan (Z_{be} - Z_{ae}) / (Y_{be} - Y_{ae}) \\ &= \arctan (-53.01) \\ &= -88.92583^\circ\end{aligned}$$

The angular rotation degree α_e after flexion is:

$$\alpha_e = \alpha'' - \alpha = -88.92583^\circ - (-85.3204^\circ) = -3.60543^\circ$$

A.2 The Calculations of Angular Motion of T12-L1 FSU Under Left and Right Lateral Bending Pure Moment of 7.5 Nm

Two nodes C (X_c, Y_c, Z_c) and D (X_d, Y_d, Z_d) were selected on the superior surface of T12, after rotation with applied left and right lateral bending pure moment of 7.5 Nm, the new positions of the two selected nodes became Cl (X_{cl}, Y_{cl}, Z_{cl}) and Dl (X_{dl}, Y_{dl}, Z_{dl}) under left lateral bending; Cr (X_{cr}, Y_{cr}, Z_{cr}) and Dr (X_{dr}, Y_{dr}, Z_{dr}) under right lateral bending, respectively. Under lateral bending pure moment, the primary motion of the FSU is in the frontal plane, according to loading

Appendix B: Sample Calculations of Angular Degrees

and boundary conditions defined in the current study (Figure 35 in Section 4.3.2), the frontal plane is YX plane. Hence, only coordinates of Y and X were used to calculate the angular rotation degrees under left and right lateral bending.

Before rotation, the coordinates values of Y and X for nodes C and D were, respectively:

$$\text{C: } Y_c = 12.054 \text{ mm, } X_c = -18.685 \text{ mm}$$

$$\text{D: } Y_d = 12.002 \text{ mm, } X_d = 20.485 \text{ mm}$$

After left lateral bending, C and D moved to Cl and Dl, under right lateral bending, C and D shifted to Cr and Dr, the coordinates values of Y and X for Cl and Dl, Cr and Dr are as follows:

After left lateral bending:

$$\text{Cl: } Y_{cl} = 10.9361 \text{ mm, } X_{cl} = -20.1542 \text{ mm}$$

$$\text{Dl: } Y_{dl} = 13.2514 \text{ mm, } X_{dl} = 18.3561 \text{ mm}$$

After right lateral bending:

$$\text{Cr: } Y_{cr} = 12.8497 \text{ mm, } X_{cr} = -17.5094 \text{ mm}$$

$$\text{Dr: } Y_{dr} = 10.3352 \text{ mm, } X_{dr} = 21.6314 \text{ mm}$$

A.2.1 The calculation of the angular motion after left lateral bending

The angular rotation degree β_r after left lateral bending was calculated in the following manner:

Appendix B: Sample Calculations of Angular Degrees

The angle β between line segment CD and Y axis is:

$$\begin{aligned}\beta &= \text{arc tg } (X_d - X_c) / (Y_d - Y_c) \\ &= \text{arc tg } (-753.269) \\ &= -89.9306^\circ\end{aligned}$$

The angle β' between line segment ClDl and Y axis is:

$$\begin{aligned}\beta' &= \text{arc tg } (X_{dl} - X_{cl}) / (Y_{dl} - Y_{cl}) \\ &= \text{arc tg } (-16.63296) \\ &= -93.4342^\circ\end{aligned}$$

The angular rotation degree β_l after left lateral bending is:

$$\beta_l = \beta' - \beta = -93.4342^\circ - (-89.9306^\circ) = -3.5036^\circ$$

A.2.2 The calculation of the angular motion after right lateral bending

The angular rotation degree α_f after right lateral bending was calculated in the following manner:

The angle β between line segment CD and Y axis is:

$$\begin{aligned}\beta &= \text{arc tg } (X_d - X_c) / (Y_d - Y_c) \\ &= \text{arc tg } (-753.269) \\ &= -89.9306^\circ\end{aligned}$$

The angle β'' between line segment CrDr and Y axis is:

$$\begin{aligned}\beta'' &= \text{arc tg } (X_{dr} - X_{cr}) / (Y_{dr} - Y_{cr}) \\ &= \text{arc tg } (-15.566) \\ &= -86.3306^\circ\end{aligned}$$

Appendix B: Sample Calculations of Angular Degrees

The angular rotation degree β_l after right lateral bending is:

$$\beta_r = \beta'' - \beta = -86.3306^\circ - (-89.9306^\circ) = 3.6^\circ$$

A.3 The Calculations of Angular Motion of T12-L1 FSU Under Left and Right Axial Torque of 7.5 Nm

Two nodes E (X_e, Y_e, Z_e) and F (X_f, Y_f, Z_f) were selected on the superior surface of T12, after rotation with applied left and right lateral axial torque of 7.5 Nm, the new positions of the two selected nodes became E_l (X_{el}, Y_{el}, Z_{el}) and F_l (X_{fl}, Y_{fl}, Z_{fl}) under left axial torque; E_r (X_{er}, Y_{er}, Z_{er}) and F_r (X_{fr}, Y_{fr}, Z_{fr}) under right axial torque, respectively. Under axial torque, the primary motion of the FSU is in the transverse plane, according to loading and boundary conditions defined in the current study (Figure 35 in Section 4.3.2), the frontal plane is XZ plane. Hence, only coordinates of X and Z were used to calculate the angular rotation degrees under left and right lateral bending.

Before rotation, the coordinates values of X and Z for nodes E and F were, respectively:

$$E: X_e = 0.087126 \text{ mm}, Z_e = 5.703 \text{ mm}$$

$$F: X_f = -0.35229 \text{ mm}, Z_f = 12.388 \text{ mm}$$

After left axial torque, E and F moved to F_l and E_l , under right axial torque, E and F shifted to E_r and F_r , the coordinates values of X and Z for E_l and F_l , E_r and F_r are as follows:

Appendix B: Sample Calculations of Angular Degrees

After left axial torque:

El: $X_{el} = -0.313264$ mm, $Z_{el} = 5.87594$ mm

Fl: $X_{fl} = -0.94867$ mm, $Z_{fl} = 12.5484$ mm

After right axial torque:

Er: $X_{er} = 0.831946$ mm, $Z_{er} = 6.10371$ mm

Fr: $X_{fr} = 0.59063$ mm, $Z_{fr} = 12.81813$ mm

A.3.1 The calculation of the angular motion after left axial torque

The angular motion α_l after left axial torque was calculated in the following manner:

The angle θ between line segment EF and X axis is:

$$\begin{aligned}\theta &= \arctan(Z_f - Z_{fe} / (X_f - X_e)) \\ &= \arctan(-15.21337) \\ &= -86.23916^\circ\end{aligned}$$

The angle θ' between line segment ElFl and X axis is:

$$\begin{aligned}\theta' &= \arctan(Z_{fl} - Z_{el} / (X_{fl} - X_{el})) \\ &= \arctan(-10.50109) \\ &= -84.56023^\circ\end{aligned}$$

The angular motion θ_l after left axial torque is:

$$\theta_l = \theta' - \theta = -84.56023^\circ - (-86.23916^\circ) = 1.67903^\circ$$

Appendix B: Sample Calculations of Angular Degrees

A.3.2 The calculation of the angular motion after right lateral bending

The angular motion β_r after right axial torque was calculated in the following manner:

The angle θ between line segment EF and X axis is:

$$\begin{aligned}\theta &= \arctan (Z_f - Z_e) / (X_f - X_e) \\ &= \arctan (-15.21337) \\ &= -86.23926^\circ\end{aligned}$$

The angle θ'' between line segment ErFr and X axis is:

$$\begin{aligned}\theta'' &= \arctan (Z_{fr} - Z_{er}) / (X_{fr} - X_{er}) \\ &= \arctan (-27.8093) \\ &= -87.94058^\circ\end{aligned}$$

The angular motion β_r after right axial torque is:

$$\theta_r = \theta'' - \theta = -87.94058^\circ - (-86.23926^\circ) = -1.70132^\circ$$

APPENDIX C: NUMERICAL VALUES OF ANGULAR MOTION OF THE FE MODELS IN THE VALIDATION STUDY

B.1 The Numerical Values of Angular Motion of T11-T12 FSU

Table 7 Angular motion of T11-T12 FSU FE model

Moment (Nm)	Angular Motion (°)					
	Fle	Ext	Left LB	Right LB	Left AR	Right AR
2.5	1.36	-1.3	-1.45	1.52	0.9	-0.9
5.0	2.4	-2.2	-2.5	2.55	1.7	-1.64
7.5	3.0	-2.8	-3.2	3.31	2.3	-2.2

Fle—Flexion, Ext—Extension, LB—Lateral bending, AR—Axial

B.2 The numerical values of angular motion of T12-L1 FSU

Table 8 Angular motion of T12-L1 FSU FE model

Moment (Nm)	Angular Motion (°)					
	Fle	Ext	Left LB	Right LB	Left AR	Right AR
2.5	1.46	-1.45	-1.6	1.6	0.63	-0.75
5.0	2.5	-2.6	-2.8	2.9	1.2	-1.26
7.5	3.0	-3.6	-3.5	3.6	1.67	-1.7

Fle—Flexion, Ext—Extension, LB—Lateral bending, AR—Axial