Investigation of spinal dynamic stabilization device using finite element method

Zhou, Yuanli

2008

Zhou, Y. (2008). Investigation of spinal dynamic stabilization device using finite element method. Master's thesis, Nanyang Technological University, Singapore.

<https://hdl.handle.net/10356/5596>

<https://doi.org/10.32657/10356/5596>

Nanyang Technological University

Downloaded on 26 Apr 2023 17:54:57 SGT

**NANYANG
TECHNOLOGICAL
UNIVERSITY**

**INVESTIGATION OF SPINAL DYNAMIC
STABILIZATION DEVICE USING FINITE
ELEMENT METHOD**

ZHOU YUANLI

**SCHOOL OF MECHANICAL & AEROSPACE
ENGINEERING**

2008

INVESTIGATION OF SPINAL DYNAMIC STABILIZATION DEVICE USING FINITE ELEMENT METHOD

ZHOU YUANLI

2008

Investigation of Spinal Dynamic Stabilization Device Using Finite Element Method

Zhou Yuanli

School of Mechanical & Aerospace Engineering

A thesis submitted to the Nanyang Technological University
in fulfilment of the requirement for the degree of
Master of Engineering

2008

Acknowledgements

The author would like to take this opportunity to express her sincere gratitude and appreciation towards the followings:

- The Project Supervisor, A/P Teo Ee Chon, for his patient guidance and consistent attention in the progress of this project.
- Dr. Qiu Tianxia and Dr. Zhang Qinghang, for their invaluable help in teaching the author ANSYS software and selfless sharing of their knowledge and experience in the biomechanical field, and also for their earnest advice for the author to overcome the difficulties encountered during the course of the research.
- My husband, Mr. Gu Hanyang, and my parents, for their continuous support, encouragement and selfless love.

Table of Contents

Acknowledgements	i
Table of Contents	ii
Summary	v
List of Figures	vi
List of Tables	x
Chapter 1: Introduction	1
1.1 Background	1
1.2 Objectives	4
1.3 Scope	4
1.4 Organization	5
Chapter 2: Literature Review	7
2.1 Biomechanical studies on lumbar spine	7
2.1.1 <i>In vitro</i> experimental study	7
2.1.1.1 Typical experimental set-up procedure	8
2.1.1.2 Biomechanical behavior of the intact motion segment	9
2.1.1.3 Biomechanical behavior of the pathological motion segment	15
2.1.2 <i>In vivo</i> experimental studies	17
2.1.3 Biomechanical studies using FE method	18
2.2 Material Properties of disc-body unit	24
2.3 Changes in Degenerative Disc Disease (DDD)	27
2.3.1 Morphological Changes in DDD	27
2.3.2 Biochemical Changes in DDD	28
2.3.3 Effect of degenerative changes on disc function and pathology	30
2.4 Surgical treatments for Disc Degeneration	31
2.4.1 Fusion	32
2.4.2 Dynamic Stabilization Devices in lumbar spine	35
2.5 Discussion on the need for the current project	47
Chapter 3: Materials and Methods	49

3.1 Introduction of FE Method	49
3.2 FE Application in Biomechanics	50
3.3 Intact FE Model	51
3.3.1 Development and validation of L4-L5 model.....	51
3.3.1.1 Specimen.....	51
3.3.1.2 Procedure for the FE model development of lumbar vertebrae..	53
3.3.1.3 Capturing of 3D geometrical data using the digitizer	54
3.3.1.4 Intact Solid Model Developing.....	56
3.3.1.5 FE Model Generating.....	58
3.3.1.6 Material Properties.....	63
3.3.2 Development of L3-S1 model.....	65
3.3.3 FE model of nucleotomized spine with implant	66
3.4 Boundary and Loading Conditions	70
3.5 Data Analysis	71
3.5.1 Rotational motion across motion segment.....	71
3.5.2 Facet forces	72
3.5.3 Von Mises stresses.....	73
Chapter 4: Results and Discussions	74
4.1 Validation of L4-L5 model	74
4.2 Validation of L3-S1 model	77
4.3 Results.....	79
4.3.1 ROM for all the models	79
4.3.2 Facet forces for all the models	85
4.3.3 Stresses in the annulus fibrosus and nucleus pulposus for all the models	91
4.4 Discussion.....	104
Chapter 5: Conclusion and Future Work	113
5.1 Conclusion	113
5.2 Limitations of this study	114
5.3 Recommendation for Future work	115
Reference	116

Appendix A: Human Lumbar Spine	128
A.1 Anatomical Terms and Definitions	128
A.1.1 Anatomical position	128
A.1.2 Directions and planes of the body	129
A.2 Vertebral Column	130
A.2.1 General Characteristics	130
A.2.2 Divisions and Curvatures	131
A.3 Lumbar Vertebrae	132
A.4 Intervertebral Disc	135
A.4.1 Nucleus Pulposus	136
A.4.2 Annulus Fibrosus	136
A.4.3 Vertebral End-plates	136
A.5 Spinal Ligaments	137
A.5.1 Intrasegmental system	138
A.5.2 Intersegmental system	139
A.6 Functional Spinal Unit (FSU)	139
Appendix B: Convergence Test of L4-L5 FE model	141
Appendix C : Publications	143

Summary

In the present study, an anatomically accurate three-dimensional finite element model of L3-S1 lumbar spine was developed and validated. This model was used to evaluate the biomechanical effect of a new dynamic stabilization device (FlexPLUS) in comparison with rigid rod system. The intact model was nucleotomized and implanted with the FlexPLUS system and the rigid rod system bilaterally at L4-L5 level, respectively. Various biomechanically relevant parameters like range of motion, facet force and disc stress at different levels under pure moments were evaluated in the three main motion planes.

The results showed that the FlexPLUS system significantly reduced the range of motion, facet force and disc stress at the instrumented level with a corresponding increase in the adjacent segments in all the loading modalities. The FlexPLUS system was more flexible than the rigid rod system in various loading configurations but not flexible enough to preserve motion and restore the natural load sharing ability of the operated segment.

List of Figures

Figure 2-1: Displacement-force curves for different L4-L5 segment.....	11
Figure 2-2: Mean displacement-force curve for lumbar disc levels	12
Figure 2-3: The ligamentous L1-S1 segment with LEDs	14
Figure 2-4: The first model of an intervertebral disc and adjacent vertebrae developed by Belytschko et al. (1974).....	20
Figure 2-5: The non-linear FE model of the disc-body unit.	20
Figure 2-6: A simplified circular symmetric model of a disc developed by ..	21
Figure 2-7: An exploded view of FE model of lumbar L2-L3 motion segment	21
Figure 2-8: FE model of Dynesys: fixation in the motion segment.....	23
Figure 2-9: Moment versus rotational angle: impact of Dynesys.....	24
Figure 2-10: A typical stress-strain curve of collagen fibers.....	27
Figure 2-11: (A) The non-rigid fixation “Wallis” implant; (B) Schematic view of the Wallis implant in place.	36
Figure 2-12: A sagittal, transverse, posterior, and orthogonal view of the X- STOP implant.....	38
Figure 2-13: Graf artificial ligament instrumentation.....	40
Figure 2-14: Schematic diagram of FASS system.	43
Figure 2-15: (A) Components of the dynesys system; (B) Schematic view of dynesys system in place.	44
Figure 3-1: Four orthogonal views (front, side, top and isometric) of a lumbar vertebra.	53
Figure 3-2: Flowchart for FE model generation of L4 and L5	54
Figure 3-3: Fixation of the specimen in the vice clamp.....	55
Figure 3-4: Captured coordinate point data for L4 from digitizer	55
Figure 3-5: Volume rendering of L4-L5 motion segment	58
Figure 3-6: Collagen fiber arrangement in the annulus fibrosus laminar	60
Figure 3-7: FE model of ligaments	61

Figure 3-8: FE model of L4-L5 motion segment.....	61
Figure 3-9: Stress-strain curve for the lumbar spine ligaments	63
Figure 3-10: FE model of L3-S1 lumbar spine.....	65
Figure 3-11: Detail view for the link between rod and vertebrae (left side)...	66
Figure 3-12: The structure of FlexPLUS	67
Figure 3-13: The ISO view of a FlexPLUS structure	68
Figure 3-14: Exploded FE model of FlexPLUS	68
Figure 3-15: Comparison of FE result with experimental result for FlexPLUS under compression	69
Figure 3-16: Incorporation of different devices in nucleotomized L3-S1 lumbar spine model. (a) FlexPlus; (b) Rigid Rod	70
Figure 3-17: Boundary and loading conditions	71
Figure 3-18: Schematic showing the locations and numbers of the nodes and the equations used to determine the rotational angle of motion segments in the FE model.....	72
Figure 4-1: Displacement-force response curve under axial compression.	75
Figure 4-2: Comparison of maximum FE rotation for rotatory loads with experimental results	76
Figure 4-3: Comparison of experimental results and current FE predictions in response to 9Nm pure moment. (a) under flexion and extension; (b) under lateral bending; (c) under torsion.....	78
Figure 4-4: Comparison of range of motion at all levels of the lumbar spine for all models in 9Nm pure flexion.....	80
Figure 4-5: Comparison of range of motion at all levels of the lumbar spine for all models in 9Nm pure extension.....	81
Figure 4-6: Comparison of range of motion at all levels of the lumbar spine for all models in 9Nm pure right lateral bending.....	82
Figure 4-7: Comparison of range of motion at all levels of the lumbar spine for all models in 9Nm pure left lateral bending.....	83
Figure 4-8: Comparison of range of motion at all levels of the lumbar spine for all models in 9Nm pure right torsion	84

Figure 4-9: Comparison of range of motion at all levels of the lumbar spine for all models in 9Nm pure left torsion.....	85
Figure 4-10: Total facet forces (N) in response to 9Nm pure moment in extension for all the models	86
Figure 4-11: Total facet forces (N) in response to 9Nm pure moment in right lateral bending for all the models.....	88
Figure 4-12: Total facet forces (N) in response to 9Nm pure moment in left lateral bending for all the models.....	88
Figure 4-13: Total facet forces (N) in response to 9Nm pure moment in right torsion for all the models	90
Figure 4-14: Total facet forces (N) in response to 9Nm pure moment in left torsion for all the models	90
Figure 4-15a: von Mises stress plots in the annulus fibrosus at all levels for all the modes in 9Nm pure moment in flexion	93
Figure 4-15b: von Mises stress plots in the nucleus pulposus at the adjacent levels for all the modes in 9Nm pure moment in flexion	93
Figure 4-16a: von Mises stress plots in the annulus fibrosus at all levels for all the modes in 9Nm pure moment in extension	94
Figure 4-16b: von Mises stress plots in the nucleus pulposus at the adjacent levels for all the modes in 9Nm pure moment in extension	94
Figure 4-17a: von Mises stress plots in the annulus fibrosus at all levels for all the modes in 9Nm pure moment in right lateral bending	97
Figure 4-17b: von Mises stress plots in the nucleus pulposus at the adjacent levels for all the modes in 9Nm pure moment in right lateral bending ..	97
Figure 4-18a: von Mises stress plots in the annulus fibrosus at all levels for all the modes in 9Nm pure moment in left lateral bending.....	98
Figure 4-18b: von Mises stress plots in the nucleus pulposus at the adjacent levels for all the modes in 9Nm pure moment in left lateral bending.....	98
Figure 4-19a: von Mises stress plots in the annulus fibrosus at all levels for all the modes in 9Nm pure moment in right torsion	101

Figure 4-19b: von Mises stress plots in the nucleus pulposus at the adjacent levels for all the modes in 9Nm pure moment in right torsion	101
Figure 4-20a: von Mises stress plots in the annulus fibrosus at all levels for all the modes in 9Nm pure moment in left torsion	102
Figure 4-20b: von Mises stress plots in the nucleus pulposus at the adjacent levels for all the modes in 9Nm pure moment in left torsion	102
Figure A-1: Anatomical position	128
Figure A-2: Directions and planes of the body	130
Figure A-3: The vertebral column	132
Figure A-4: The structure of a typical lumbar vertebra	133
Figure A-5: (A) An intervertebral disc consists of a nucleus pulposus surrounded by concentric laminates of the annulus. (B) The fibers are oriented at about $\pm 30^\circ$ with respect to the horizontal plane.....	135
Figure A-6: Ligaments of the spine.	137
Figure A-7: A motion segment from lumbar spine, showing two adjacent vertebrae and the intervertebral disc between them.....	140

List of Tables

Table 2-1: Some mechanical tests on L4-L5 segment	11
Table 2-2: Comparison between various in vitro studies.....	13
Table 3-1: Specimen Data (in mm). See Figure 3-1 for the interpretation of parameters	52
Table 3-2: Types and number of elements used in FE Model of L4-L5.....	62
Table 3-3: Material properties adopted in this project.....	64
Table 3-4: Cross-sectional areas of lumbar spine ligaments used in this study	64
Table 3-5: Material properties used for different stabilization systems	70
Table 4-1: The maximum rotations (degrees) and percentage changes compared to the intact case for the nucleotomized model and instrumented models in 9Nm pure moment in flexion and extension. (Negative % change indicates a reduction in the motion)	81
Table 4-2: The maximum rotations (degrees) and percentage changes compared to the intact case for the nucleotomized model and instrumented models in 9Nm pure moment in lateral bending. (Negative % change indicates a reduction in the motion)	83
Table 4-3: The maximum rotations (degrees) and percentage changes compared to the intact case for the nucleotomized model and instrumented models in 9Nm pure moment in torsion. (Negative % change indicates a reduction in the motion)	85
Table 4-4: Total facet forces and the percentage changes compared to the intact case for the nucleotomized and instrumented models in 9Nm pure moment in extension. (Negative % change indicated a reduction in the force.)	87
Table 4-5: Total facet forces and the percentage changes compared to the intact case for the nucleotomized and instrumented models in 9Nm pure	

moment in lateral bending. (Negative % change indicated a reduction in the force.)	89
Table 4-6: Total facet forces and the percentage changes compared to the intact case for the nucleotomized and instrumented models in 9Nm pure moment in torsion. (Negative % change indicated a reduction in the force.)	91
Table 4-7: Maximum von Mises stresses in the annulus fibrosus (MPa) and percentage changes at all levels with respect to the intact case in 9Nm pure moment in flexion and extension for the nucleotomized and stabilized models. (Negative % indicated a decrease in the stress)	95
Table 4-8: Maximum von Mises stresses in the nucleus pulposus (MPa) and percentage changes at all levels with respect to the intact case in 9Nm pure moment in flexion and extension for the nucleotomized and stabilized models. (Negative % indicated a decrease in the stress)	95
Table 4-9: Maximum von Mises stresses in the annulus fibrosus (MPa) and percentage changes at all levels with respect to the intact case in 9Nm pure moment in lateral bending for the nucleotomized and stabilized models. (Negative % indicated a decrease in the stress)	99
Table 4-10: Maximum von Mises stresses in the nucleus pulposus (MPa) and percentage changes at all levels with respect to the intact case in 9Nm pure moment in lateral bending for the nucleotomized and stabilized models. (Negative % indicated a decrease in the stress)	99
Table 4-11: Maximum von Mises stresses in the annulus fibrosus (MPa) and percentage changes at all levels with respect to the intact case in 9Nm pure moment in torsion for the nucleotomized and stabilized models. (Negative % indicated a decrease in the stress)	103
Table 4-12: Maximum von Mises stresses in the nucleus pulposus (MPa) and percentage changes at the adjacent levels with respect to the intact case in 9Nm pure moment in torsion for the nucleotomized and stabilized models. (Negative % indicated a decrease in the stress)	103
Table A-1: Commonly used directional terms	129

Table B.1: The effect of mesh density on the FE predictions.....	142
--	-----

Chapter 1: Introduction

1.1 Background

The human spine (also known as vertebral column, spinal column or backbone) is one of the most fascinating and elegant mechanical structures in the animal kingdom. It consists of 33 rigid vertebrae with soft tissues (such as intervertebral discs, ligaments and muscles) connecting and surrounding them, interactions between which provide flexibility and stability of the whole spine. One of the functions exerted by the spinal column is to protect the vital spinal cord and nerves. The lumbar spine is located in the lower part of the spinal column and is the region linking the thoracic and sacral regions with a curve convex forward. This region is the physical center of the body to carry load transmitted from the top of the head down to the low spine and to support bones and soft tissues during movement. Therefore, it is most susceptible to external mechanical injury and degenerative disc diseases, which will threaten the cord and nerves and finally induce low back pain.

Low back pain is very common, costing millions of dollars in lost work, as well as millions in medical and insurance every year. In western industrialized societies, back pain constitutes a major public health problem, and affects a large number of people; the point prevalence rates in a number of studies ranged from 12% to 35% (Maniadakis et al. 2000), with around 10% of sufferers becoming chronically disabled. It also places an enormous economic burden on society. According to the results of 'cost-of-illness' studies of back pain reported by van-Tulder et al. (1995)

and Maniadakis et al. (2000), its total cost, including direct medical costs, insurance, lost production and disability benefits, is estimated at 1.7% of the gross national product in the Netherlands and \$12 billion per annum in the UK. Back pain takes a backseat only to headache as the most common medical complaint, and is second only to the common cold as a reason for medical leave (White 1990). Although the exact source of low back pain is difficult to identify and often remains unknown, it is generally admitted that degenerative changes of the intervertebral disc are responsible for most cases of low back, and the classic concept is that loss of integrity of the disc is followed by the development of low back pain (Robert 2004).

Intervertebral disc degeneration is considered one of the main causes for low back pain, and is an extremely common but complex phenomenon, with up to 90% of all persons older than 60 years of age having at least one degenerate intervertebral disc (Boden et al. 1990). The first unequivocal findings of degeneration in the lumbar discs are observed in the age group 11-16 years (Boos et al. 2002). About 20% of people in their teens have discs with mild signs of degeneration; degeneration increases steeply with age, particularly in males, so that around 10% of 50-year-old disc and 60% of 70-year-old discs are severely degenerated (Miller et al. 1988). The process of disc degeneration is an aberrant, cell-mediated response to progressive structural failure. A degenerated disc is one with structural failure combined with accelerated or advanced signs of ageing. Early degenerative changes should refer to accelerated age-related changes in a structurally intact disc.

Degenerative disc disease should refer to a degenerated disc, which is also painful (Adams et al. 2006).

Lumbar fusion is a frequently performed operative procedure for the treatment of low back pain due to disc degeneration. Although successful fusion rate has been achieved with recently improved fusion techniques, the corresponding clinical outcome is not satisfactory (Jackson et al. 1985; Wetzel et al. 1994; Boos et al. 1997) and a variety of complications associated with fusion has been reported, especially its adverse effect on adjacent segmental levels (Wiltse et al. 1994; Kumar et al. 2001a; Kumar et al. 2001b; Guigui et al. 2002). Therefore, the new concept of dynamic stabilization becomes more and more popular since it was introduced in 1980s. Based on this concept, some dynamic stabilization devices have been designed and put into clinical application.

Experimental test (*in vitro* or *in vivo*) is considered to be the only real method to examine healthy or pathological disc behavior, but it has inherent limitations such as high cost, difficulty in obtaining specimen, and large variability from specimen to specimen in human cadavers. The process of preparing and conducting an experimental study is also very time and labor consuming. However, the finite element (FE) study can overcome these difficulties and deficits associated with experimental tests. With the advent of high-speed and large-memory computers and the availability of advanced FE software, the FE method has become a powerful tool in the spinal research and is even considered as a substitute for *in*

vivo and *in vitro* study by providing some information which is difficult if possible to obtain from experiments.

Therefore, the present study is to develop a geometrically accurate three-dimensional (3D) nonlinear FE model of L3-S1 lumbar spine, which will be adopted to investigate biomechanical effects of a newly designed dynamic stabilization device (FlexPLUS) on the instrumented and adjacent segments.

1.2 Objectives

The objective of this study is to evaluate the biomechanical effects of the FlexPLUS system under various physiological loading modes using the developed FE model.

1.3 Scope

This study was focused on:

- The development and validation of an anatomically accurate 3D nonlinear FE model of L3-S1 lumbar spine based on the geometrical data extracted from the human cadaver specimen using a high-definition multi-axis digitizer.
- The investigation of the biomechanical effects of the new dynamic stabilization system (FlexPLUS) on the instrumented and the adjacent segments in terms of range of motion (ROM), facet force and disc stress, and compare them with that for the rigid rod system.

1.4 Organization

This report consists of five chapters and a brief description of each chapter follows as:

Chapter 1 starts with the introduction of the background, objectives and scope of this study. Chapter 2 includes an overview of biomechanical studies performed so far, especially those using FE method. Disc degenerated diseases (DDD) and surgical treatments for DDD including fusion and currently used dynamic stabilization systems in lumbar spine are also reviewed.

Chapter 3 outlines the materials and methods adopted in this project. The methodology and procedure to develop the intact and the instrumented FE models of L3-S1 lumbar spine are described. This chapter also presents the validation study of the intact FE model by comparing the FE predictions with the experimental results reported in literature.

Chapter 4 provides the results obtained from this study and the discussion. The effects of the FlexPLUS on the range of motion, the facet force and the disc stress distribution are documented and compared with that of the rigid rod system. Chapter 5 summarizes the findings of this study and recommendations for future works.

Appendix A describes the anatomy of the human spine to provide a basic understanding of the fundamentals. Appendix B provides the convergence test

results of the L4-L5 model. Appendix C lists the publications including original journal papers and conference papers.

Chapter 2: Literature Review

2.1 Biomechanical studies on lumbar spine

The human lumbar spine is composed of five individual vertebrae, connected by intervertebral discs, facet joints and other soft tissues, all of which play a certain role in the response of the whole spine to different external loading conditions. With an attempt to obtain a full picture of the contribution each functional structure of the lumbar spine, the effects of degenerative lumbar disease on lumbar biomechanical properties, and load-bearing characteristics of these components, numerous studies have been conducted by numerous researchers, which fall into three types: *in vitro*, *in vivo*, and numerical study.

2.1.1 *In vitro* experimental study

Many *in vitro* studies have been carried out to measure the mechanical response of a motion segment or a multi-segmental lumbar spine with either healthy or degenerated intervertebral discs to external forces and moments (Virgin 1951; Brown et al. 1957; Markolf 1972; Markolf et al. 1974; Schultz et al. 1979; Panjabi et al. 1984; Yamamoto et al. 1989; Goel et al. 2005). These *in vitro* results (i.e., displacement-force response curves, range of motion, etc) on the cadaveric specimens are usually used for validating any mathematical model developed under similar boundary/loading conditions, thereby further analyses using the validated model could be viewed with confidence.

2.1.1.1 Typical experimental set-up procedure

Although different testing machines were employed by individual researchers in experimental studies, the experimental set-ups were similar. The testing apparatus allowed the application of the different loading modalities, including axial compression and tensile, flexion, extension, lateral bending and axial torsion. The loading applied could be either displacement-controlled or force-controlled, and the maximum magnitude was always achieved in several increments. The superior and inferior surfaces of the specimen were embedded in epoxy resins or capped by quick-setting plasters and then mounted into the upper and lower portions of the test fixture. The inferior fixture, into which the inferior vertebra of the motion segment had been embedded, was clamped to the base of the machine. The resulting translations and rotations were recorded using transducers.

However, the specimen preparations for compression test and various rotatory tests were slightly different. For compression test on mono-segmental specimens, each test specimen consisted of a vertebra-disc-vertebra combination with the posterior bone elements removed and the exposed superior and inferior endplates cut off. The mid-plane of the intervertebral disc was usually aligned to be parallel to the loading surface of the testing system. The reading from measuring apparatus was the overall axial displacement of the vertebra-disc-vertebra unit. There was no information available for pure compression test on multi-segmental specimens.

For rotatory tests on either mono- or multi-segmental specimens, both the posterior bone elements of vertebrae and all ligamentous components were preserved. All

rotations were measured based on the assumption that the vertebrae were perfectly rigid, therefore no local rotation occurred. Stereometry and electro-mechanical methods are two general types of various measurement techniques. In the stereometry, two positions of a body in space, at the beginning and at the end of a step of motion, are recorded. The positions of the body are defined by the position of three non-collinear points of the body. The positions of these points are determined in a global coordinate system with the help of two or more views utilizing some portion of the electromagnetic spectrum. The images of the points are recorded on film and digitized. Using geometric procedures, the 3D coordinates of the three points of the body in two positions are determined. Principles of kinematic are then utilized to compute the rigid body motion. This method is the basis of the motion measurement techniques, which will be also used in current project. In electro-mechanical method, various motion transducers have been utilized to measure 3D motion.

2.1.1.2 Biomechanical behavior of the intact motion segment

Virgin (1951) investigated the physical properties of the intervertebral disc under compression. The specimen used included an intervertebral disc and a thin slice of bone at each end. It was compressed to the elastic limit by successive increments of 50lb. The experimental result showed that the intervertebral disc held the property of elasticity to a very marked degree and was capable of maintaining very great loads without disintegration.

In order to obtain quantitative data on the physical properties of the intervertebral disc, Brown et al. (1957) performed a series of tests on the lumbar intervertebral discs including axial compression tests and bending tests. For the compression tests, test specimens of L4-L5 motion segment were prepared from three cadavera. Each test specimen was composed of a vertebra-disc-vertebra combination with the posterior bone elements removed. An axial compressive load was applied in increments of fifty pounds or 100 pounds to the superior surface. At each increment, the load was maintained for two minutes and the axial deformation of each vertebra-disc-vertebra unit was measured by means of Ames dials. Figure 2-1 shows the plot the axial force against axial displacement of each disc, and Table 2-1 tabulates the ultimate compressive load capacity and the stiffness under load of each disc tested. It can be seen that all specimens behaved in a similar manner. However, it was obvious that there were considerable quantitative differences in both the initial and in the major slopes of the displacement-force curves of the various discs although they showed qualitative similarity in behavior. Since the procedure of preparation and test for all specimens were the same, the quantitative variations in the results should be due to the inherent difference between the specimens such as age, gender and disc degeneration grade.

Figure 2-1: Displacement-force curves for different L4-L5 segment (adopted from Brown et al., 1957)

Cadaver	Ultimate Axial Compressive Load (N)	Axial Stiffness of Disc (N/mm)	
		Initial Slope	Major Slope
A	4440	525	2695
B	5550	1444	3500
C	5772	82	2100

Table 2-1: Some mechanical tests on L4-L5 segment (adopted from Brown et al., 1957)

In 1972, Markolf (1972) also measured the mechanical response of a single lumbar intervertebral joint to an axial compressive force. In their experiment, twelve vertebra-disc-vertebra lumbar specimens from different levels were prepared. The specimens were removed of the posterior structures and fixed on the test device.

The upper and lower surfaces of the specimens were placed to be parallel to the loading heads. A maximum load up to 2200N was applied to the superior surface through a crosshead. The mean displacement-force curve with standard deviation was given in the paper, as shown in Figure 2-2. As observed by Brown and his colleagues (1957), the compression curve also showed stiffening and a wide range of deviation.

Figure 2-2: Mean displacement-force curve for lumbar disc levels

Other researchers (Schultz et al. 1979; Panjabi et al. 1984; Yamamoto et al. 1989; Goel et al. 2005) conducted in vitro experiments using a whole lumbar spine or certain lumbar motion segment to evaluate the kinematical response under flexion, extension, lateral bending and axial torsion. The results varied over wide ranges between different researchers, as tabulated in Table 2-2. The large variation may

be caused by factors, such as the different magnitude of load used by individual researchers, different specimen preparations, and different grades of disc degeneration. Although the range of motion showed large deflection, a similar stiffness trend was obtained by all researchers. The motion segment is most flexible in flexion while least flexible in axial torsion.

		Schultz et al., 1979	Panjabi et al., 1984	Yamamoto et al. 1989	Goel et al. 2005
Specimen		L1-L2, L2-L3, L3-L4 or L4-L5	L3-L4 or L4-L5	L1-S1	L1-S1
Applied load		400N+10.6Nm	150N+7.5Nm	10Nm	9Nm
ROM at L4-L5 (degree)	Flexion	5.51(1.00)*	4.3(1.8)	8.9(0.7)	5.02(2.8)
	Extension	2.99(1.02) *	1.6(0.4)	5.8(0.4)	2.95(1.18)
	Right lateral bending	5.64(1.22) *	3.2(1.6)	5.9(0.5)	3.25(1.52)
	Left lateral bending	4.9(0.79) *	3.6(1.3)	5.5(0.5)	3.55(2.54)
	Right torsion	1.5(0.67) *	0.2(0.4)	2.7(0.5)	1.29(0.48)
	Left torsion	/	0.4(0.6)	1.7(0.3)	2.24(0.88)

Table 2-2: Comparison between various in vitro studies. ()=Standard deviation; *=mean for different levels

In the study conducted by Panjabi et al. in 1984, three L4-L5 segments without musculature were tested under different loading conditions. During the experiment, the lower vertebra of a motion segment was fixed rigidly to the test table and the

midplane of the disc was adjusted to be horizontal. A moment of 7.5Nm with an axial preload of 150N was applied to the center of the upper vertebra in an incremental manner in the three main planes. The 3D motion of the upper vertebra at its center was recorded by six electrical displacement transducers rigidly attached to the upper vertebral body.

In the experimental study performed by Geol et al. in 2005, L1-S1 spine specimens with ligaments were used. Specimens were potted in a rigid base secured to the sacrum and a loading frame likewise was secured to the L1 vertebral body, as shown in Figure 2-3. A set of three light-emitting diodes (LEDs) was attached to each vertebral body to determine the load-displacement behavior of the specimen. The spatial location of the LED-markers was tracked by Opto-Trak motion measuring system. A maximum of 9Nm pure moment in all the six degrees of freedom was applied to the specimen.

Figure 2-3: The ligamentous L1-S1 segment with LEDs (adopted from Goel et al. 2005)

2.1.1.3 Biomechanical behavior of the pathological motion segment

The intervertebral disc is the most important load bearing and transmitting structure in the lumbar spine and therefore any change occurring in the disc will inevitably affect other structures and maybe initiate disc degeneration. Reuber et al. (1982) explored the behavior of lumbar disc in response to different external loading modalities on fourteen lumbar mono-segmental units from different levels. The chief finding of their study included that bulging occurred under all modes of applied loading with the largest bulges produced in right lateral bending, degenerated discs bulged more than non-degenerated discs under the same load, and endplate bulge was relatively small compared to disc bulge. Brinckmann et al. (1991) investigated the effect of disc degeneration on lumbar spine mechanics on human lumbar specimens from 20 to 40 years of age. The specimens without posterior elements were tested in the intact state and after discectomy, respectively. They found that the disc height and the radial disc bulge changed approximately in proportion to the amount of disc tissue excised. On average, the disc height was reduced by a rate of 0.8mm/g and the radial disc bulge increased by a rate of 0.2mm/g when the axial load increased from 1000N to 2000N. Removal of 3g of central disc tissue lowered the intradiscal pressure to approximately 40% of the initial value (about 1.3MPa on average) under an axial load of 1000N. From the experimental results, the authors concluded that a high intradiscal pressure was a prerequisite for the mechanical function of the disc under physiological conditions. In the other word, the decrease of the intradiscal pressure associated with disc degeneration would lead to a decrease of disc height and an increase of the radial

disc bulges, which would in turn overload the posterior elements and compromise the nerve tissue, and eventually cause pain.

Some other authors experimentally investigated the role of lumbar facet joints play in the stability of the lumbar spine. Ahmed et al. (1990) and Noren et al. (1991) examined the effect of lumbar facet joints morphology on physical response of lumbar motion segments and on disc degeneration. The results in their study indicated that the facet joint asymmetry did not affect its role in resisting axial rotation, but it was a risk factor in the development of disc degeneration. Abumi et al. (1990) and Haer et al. (1994) evaluated the role of the lumbar facet joints in spinal stability by performing facetectomy in mono- or multi-segmental cadaver. Abumi et al. tested 12 motion segments from L2-L3 to L4-L5 subjected to a maximum load of 8Nm in the three main planes after graded facetectomy. They reported that the relative increase of ROM was 76.42% in flexion, 42.5% in extension, 110% in right torsion, 135% in left torsion, and 40% in lateral bending after bilateral total facetectomy. However, based on the experimental study on T12-S2 spine with the facet joints of L3 destroyed, Haer et al. (1994) found the relative loss of stiffness after bilateral facets was about 20% in compression and extension. Their results indicated the facet joint did play a role in stabilizing the lumbar spine, especially in torsion. Although motion patterns may be slightly affected in compression and extension after facetectomy, the load must be transferred to the disc to maintain spinal stability and the degeneration of the associated and adjacent discs may be accelerated.

The experimental studies conducted by Lorenz et al. (1983) and Ahmed et al. (1990) showed that there exhibited a significant variation of facet joint geometry and facet loading between segmental levels in the lumbar spine. At L2-L3 level, the median facet angle was found to 29° and the median facet inclination was 5°, whereas for the L4-L5 level, the mean facet angle and facet inclination were 51° and 11° respectively, both of which were significantly larger than that for L2-L3 level. The facets at L2-L3 generally took more load than those at L4-L5, especially at low axial load. These findings indicated that biomechanical studies of the various segmental levels of the lumbar spine should probably be performed separately.

2.1.2 *In vivo* experimental studies

Most *in vivo* studies were focused on the measurement of intradiscal pressure since its change under different stages was a typical indication for disc degeneration. Sato et al. (1999) measured the intradiscal pressure in 8 healthy volunteers and 28 patients with ongoing low back pain at L4-L5 level with a specially constructed pressure sensor. The 28 patients included 10 women and 18 men with a mean age of 45 years (19-74 years), a mean body weight of 68kg (45-88kg), and a mean body height of 165cm (155-182cm). They found that there was a statically significant difference between grade of degeneration 1 and 2, which meant there was a difference between the degenerated and non-degenerated discs in the pressure-transmission in the nucleus pulposus. A reduction in the intradiscal pressure with increasing degeneration was also observed. Haughton et al. (2002) examined the ability of intervertebral disc in resisting axial rotation by studying

three groups of subjects with discs subjected to different grades of degeneration using a specially built table. Their preliminary study showed that the average axial rotation between vertebrae that had degenerated discs was 3.2° , which exceeded the average 0.8° of between vertebrae that had normal discs.

2.1.3 Biomechanical studies using FE method

FE analysis has been a widely accepted tool used in many research activities since the method was first developed in the 1950s in the aircraft industry. In spine research, this method is also becoming an indispensable complement to in vivo and in vitro studies by exerting contributions in four ways: “(a) to provide an assessment of the spine in health; (b) to provide an assessment of the spine as altered by disease, degeneration, ageing, trauma or surgery; (c) to provide an assessment of the spine with spinal instrumentation and (d) to assist in the design and development of that spinal instrumentation” (Fagan et al. 2002).

Fagan et al. (2002) reviewed the application of FE analysis in spine research. Belytschko et al. (1974) was the first to present details of a FE analysis of an intervertebral disc and adjacent vertebrae. In their model, the disc was modeled as axial symmetry with linear orthotropic material properties, as shown in Figure 2-4. This model was subsequently extended by Kulak et al. (1976), assuming that the annulus had non-linear orthotropic properties. The nonlinear motion segment model of the L2-L3 disc body unit developed by Shirazi-Adl et al. (1984) was used as the basic component in a number of 3D models of the lumbar spine. In their model, the geometry of the FE model was based on direct measurement from the

enlarged images of specimen, and for the first time the annulus fibrosus portion of the intervertebral disc was composed of the ground substance with collagenous fibers embedded in it to simulate the geometrical nonlinearity of the disc, and the motion segment was assumed to be symmetrical about its sagittal plane (Figure 2-5). The fibers of the annulus were modeled using axial elements in eight layers in a crisscross pattern inclined with an average angle of 29° with respect to the horizontal plane of the disc. Based on the geometric data from computer tomography (CT), Rao et al. (1991) used a similarly constructed model to that of Shirazi-Adl but assumed a simplified circular and symmetric geometry and varied the properties of the constituent materials to examine their effects on the disc's mechanical behavior (Figure 2-6). Instead of using the geometrical data from *in vitro* measurement or CT scan images, Teo et al. (1994) first adopted the actual geometrical data obtained from a digitizer to truly describe the vertebra used in FE model, and then Lee et al. (2004) extended the same method, but used a multi-axis digitizer to develop a FE model of L2-L3 spinal motion segment (Figure 2-7).

Figure 2-4: The first model of an intervertebral disc and adjacent vertebrae developed by Belytschko et al. (1974). (Adopted from Belytschko et al., 1974)

Figure 2-5: The non-linear FE model of the disc-body unit. (Shirazi-Adl et al. 1984)

Figure 2-6: A simplified circular symmetric model of a disc developed by (Rao et al. 1991)

Figure 2-7: An exploded view of FE model of lumbar L2-L3 motion segment (adopted from (Lee et al. 2004))

A number of researchers have developed a FE model of lumbar motion segments to study its load-bearing pattern and stress distribution. Shirazi-Adl et al. (1986) used the model developed in 1984 to examine the biomechanical behavior of the motion segment under axial torque loading and demonstrated that torsion was primarily resisted by the contact of the facet joints and the annulus as well, but not the ligaments. Similar conclusion was obtained by Ueno et al. (1987) using a model similar to that of Shirazi-Adl et al. They found that 10 to 40% of the torque loading was carried by the facets depending on the facet gap and the load level. Goel et al. (1995) created a 3D FE model of L3-L4 motion segment based on the geometrical data obtained from CT scans of a cadaveric specimen to examine interlaminar shear stresses and lamina separation under axial compression. They concluded that circumferential tears were most likely to occur in the posterolateral regions of the disc. Also based on the CT data of lumbar spine, a FE model of L2-L3 motion segment was developed by Lu et al. (1996) to investigate the influence of disc height changes on the mechanical behavior of the intervertebral disc. They found that disc height changes significantly affected the axial displacement, the posterolateral disc bulge, and the tensile stress in the peripheral annulus fibers but not the intradiscal pressure and the longitudinal stress distribution at the endplate-vertebra interface. Goto et al. (2002) established a 3D FE model of L4-L5 motion segment using CT images and assessed von Mises stress on the vertebral endplates, the annulus fibrosus and the facet joints in normal and degenerative disc. They found that there was increased von Mises stress on the endplate in the anterior and center portions and on the annulus fibrous in the posterior portion. Von Mises

stress on the facet joints was increased to 2.5 times that in the normal disc but no changes were observed on the endplate or annulus fibrosus in the degenerative disc.

Few models with dynamic stabilization devices have been reported in literature. To the best of the author's knowledge, the only one was developed by Eberlein et al. (2002) to assess the biomechanical behavior of the implanted motion segment (L2-L3 level) with Dynesys under physiological loading conditions. The FE model built by them is shown in Figure 2-8. Their results showed that the stiffness of the treated motion segment was considerably increased under flexion and extension, especially under torsion, which was so high that non-physiological biomechanical response was observed (Figure 2-9). In addition, they found that general functionality of Dynesys was independent on the applied preload forces.

Figure 2-8: FE model of Dynesys: fixation in the motion segment (adopted from Eberlein et al. 2002)

Figure 2-9: Moment versus rotational angle: impact of Dynesys (adopted from Eberlein et al., 2002)

2.2 Material Properties of disc-body unit

It is very difficult to obtain material properties of biological materials, such as the elastic modulus, shear modulus and Poisson's ratio, since they are generally inhomogeneous, anisotropic, viscoelastic and nonlinear. Experimentally determined material properties show significant variations between different studies because the procedure and specimen preparation vary from one to others.

The vertebral body is composed mainly of porous cancellous bone in the center and denser cortical bone at the periphery. Tensile tests showed that the response to be almost linear but highly time-dependent. It was also stronger in the physiological loading direction than in directions perpendicular to it (Carter et al. 1977; Carter et al. 1978). The material properties of cortical bone have been tested by various

researchers (Reilly et al. 1974; 1975; Yoon et al. 1976; Ashman et al. 1984; Choi et al. 1990), and the experimental results showed considerable variation, ranging from 5.44GPa to 18.8GPa. The material properties of trabecular bone were closely correlated with its physical characteristics such as porosity, apparent density, aging and anatomical location (Hansson et al. 1987). A huge variation in modulus was reported by various researchers, with a range from 10MPa to 428MPa. A typical stress-strain response curve of cancellous bone was characterized by a linear region followed by yield at an extended plastic region maintaining constant stress due to collapse within the cellular framework (Goldstein 1987). The cancellous bone undergoes large compressive deformation before it fails (Panjabi et al. 1990). Although the elastic properties of cancellous bone can be generally characterized as a heterogeneous and anisotropic material, it was assumed to be homogeneous and isotropic in most FE models developed to date. This is reasonable because the annulus of the intervertebral disc is the softest substance in the disc-body unit, and therefore cancellous bone material properties must have a small impact on the overall disc behavior. The experimental tests and μ -FE analyses conducted by Kabel et al. (1999) also demonstrated that anisotropic tissue properties have negligible, if any, impact on apparent elastic properties.

The intervertebral disc consists of three distinct parts: the nucleus pulposus, the annulus fibrosus, and the cartilaginous end-plates. There are no experimental data available for the mechanical properties of vertebral body end-plates. Referring to cartilage from various other origins, which has been subjected to experimental tests,

it has been confirmed that cartilage is quasi-linear and inhomogeneous in nature (Kempson et al. 1971). However, its non-linear material behavior has not yet been accounted for by the state-of-art FE models involving cartilage because the cartilage disappears with time and is replaced by bone. FE models still treat the cartilage as an isotropic homogeneous material (Ueno et al. 1987; Galbraith et al. 1989) or even neglect the effect of the cartilage. The nucleus pulposus of a non-degenerated disc is now accepted generally to act as an inviscid, incompressible fluid. In FE models, a bulk modulus of 1666.7MPa (by defining $E=1.0\text{MPa}$ and $\nu=0.4999$), which is close to the value for water (2150MPa), has been used for the nucleus pulposus due to its high water content. The annulus fibrosus is a composite material mainly made up of an isotropic ground substance reinforced by helically wound collagen fibers. Numeric studies (Hirsch et al. 1968; Haut et al. 1972; Sanjeevi et al. 1982) have been carried out to explain the mechanical behavior of the collagen fibers, and it has been established that a typical stress-strain curve (Figure 2-10) will have three regions: the toe in which the fibers straighten; the transition region in which the fibers reach their full unstretched length and are about to elongate; the linear region in which the strain is proportional to the stress. It is generally believed that all the fibers normally act within the linear region. However, there are no data available for the mechanical properties of the ground substance itself. In FE models (Shirazi-Adl et al. 1984; Ueno et al. 1987), the initial portions of the stress-strain curves for the annulus and collagen fibers have been used to extract the properties of the ground substance with the assumption that the

collagen fibers, because of their coiled nature, appear initially extensible relative to the ground substance.

Figure 2-10: A typical stress-strain curve of collagen fibers (adopted from Haut et al., 1972)

2.3 Changes in Degenerative Disc Disease (DDD)

Disc degeneration is a natural procedure of ageing and characterized by changes in the morphology and biochemistry of the disc.

2.3.1 Morphological Changes in DDD

The intervertebral discs lie between the vertebral bodies, consisting of the outer ring of annulus fibrosus, the central nucleus pulposus and the cartilage end-plates which sandwich the nucleus pulposus inferiorly and superiorly, as shown in Figure

A-5. During growth and skeletal maturation, the boundary between annulus and nucleus becomes less obvious, and with increasing age the nucleus generally becomes more fibrotic and less gel-like (Buckwalter 1995). With ageing and degeneration the disc changes in morphology, becoming more and more disorganized. The annular lamellae become irregular, bifurcating and interdigitating, and the collagen and elastin networks also appear to become more disorganized (Urban et al. 2003). In an *in vitro* study on 135 lumbar discs, Osti et al. (1992) reported defects of annulus fibrosus associated with degeneration of the disc, which were classified as peripheral rim tears, circumferential tears and radiating fissures. The cartilage endplates of the disc, by which the majority of disc nutrition is supplied, undergo extensive mineralization soon after maturity (Robert 2004). Boos et al. (2002) demonstrated an age-associated change in morphology, with discs from individuals as young as 2 years of age having some very mild cleft formation and granular changes to the nucleus. With increasing age comes an increased incidence of degenerative changes, including cell death, cell proliferation, mucous degeneration, granular change and concentric tears. It is difficult to differentiate changes that occur solely due to ageing from those that might be considered ‘pathological’ (Urban et al. 2003).

2.3.2 Biochemical Changes in DDD

The mechanical properties and physiologic functions of the disc are directly influenced by the extracellular matrix. The main components present in the intervertebral disc are collagens and proteoglycans (PGs), the latter of which have a high anionic charge that creates a strong swelling pressure, attracting water to form

a gel, which is retained within a fibrous network of collagen and elastin (Yu et al. 2002). The collagen network, making up approximately 70% of the dry weight in the outer annulus fibrosus and 20% of the central nucleus pulposus, provides tensile strength to the disc and anchors the tissue to the bone. The PG concentration is very high in the center of the disc, where the amount of swelling is constrained by tension arising in the collagen fibers in the matrix. This combination provides a composite structure highly suited to carrying and redistributing compressive loads yet allowing movement in all planes (Robert 2004).

During disc degeneration, the PG population of the disc becomes more heterogenous and fragmented. Loss of PG is the most significant biochemical change in degenerate discs, which lose possibly up to 80% of their total glycosaminoglycan (GAG) content compared to apparently “normal” discs of a similar age. Degeneration is associated with decreased hydration, especially in the nucleus. The water content in the nucleus pulposus drops to less than 70% of the tissue wet weight in the elderly (Antoniou et al. 1996). The collagen population of the disc also changes with degeneration of the matrix, although the changes are not as obvious as those of the PGs. There are little changes of the absolute quantity of collagen but the types and distribution of collagens can alter (Urban et al. 2003; Robert 2004). In addition, other components can change in disc degeneration and disease in either quantity or distribution.

2.3.3 Effect of degenerative changes on disc function and pathology

The main functions of the discs are mechanical; the discs serve both to transmit load and to act as the joints of the spinal column. In vivo, the discs are always under loads in various modalities, both from body weight and from the activity of the spinal muscles (Bibby et al. 2004). The loss of PG, one of the major compositional changes in disc degeneration, has a major effect on the disc's load-bearing behavior. With the loss of PG, the osmotic pressure of the disc falls and the disc reduces its capability to maintain hydration under load. The water content of degenerate discs is lower than that of normal age-matched discs (Lyons et al. 1981), and degenerate discs lose height and fluid more rapidly under load (Frobin et al. 2001). The degenerate disc is also less stiff and tends to bulge more than a normal disc when loaded. The loss of hydration following the loss of PG and matrix disorganization leads degenerated discs to lose their hydrostatic property under load (Adams et al. 1996), undergoing a transition from fluid-like to solid-like behavior. Loading may thus lead to inappropriate stress concentrations along the endplate or in the annulus, which could adversely affect matrix structure or cellular activity. The stress concentrations seen in degenerate discs have also been associated with discogenic pain during discography (McNally et al. 1996). In the annulus fibrosus, there is a significant increase in compressive modulus (Iatridis et al. 1998) and decrease in radial permeability (Gu et al. 1999), as well as a moderate increase in shear modulus of the tissue with grade of disc degeneration. These alterations may be due to the loss of water content and increase in tissue density. With degeneration and ageing, there is a decrease in the number of layers of the annulus and an increase in the thickness and spacing of the collagen fibers

(Marchand et al. 1990; Tsuji et al. 1993), which directly increases the interlaminar shear stresses (Iatridis et al. 2004).

Such major changes in disc behavior have a strong influence on other spinal structures, and may affect their function and predispose them to injury. For example, owing to the rapid loss of disc height under load in degenerate discs, facet joints adjacent to such discs may be subject to abnormal loads (Adams et al. 1990) and eventually develop osteoarthritic changes. Loss of disc height can also reduce the tensional forces on the ligamentum flavum and hence may cause remodeling and thickening, which will increase the tendency of the ligament to bulge into the spinal canal, leading to spinal stenosis.

These compositional and structural changes in degenerated discs alter the macroscopic behaviors as seen in the disc flexibility (Mimura et al. 1994) and intradiscal pressure (Adams et al. 1996). The relative contributions of the annulus fibrosus, nucleus pulposus, and posterior elements can also be substantially changed by degenerative changes (Dunlop et al. 1984; Adams et al. 1996).

2.4 Surgical treatments for Disc Degeneration

The biologic changes of disc degeneration are associated with back pain and other spinal disorders, such as disc herniation, spondylolisthesis, facet arthropathy, and spinal stenosis. Currently, there is no effective method that can reverse or even retard disc degeneration. However, many different strategies are used for the treatment of the degenerated disc, which are classified into two broad areas:

nonsurgical and surgical. Surgical treatment is normally performed only after a specific pathoanatomic condition has been identified as the cause of the patient's symptoms and is an option for patients who have failed to respond to conservative treatment.

2.4.1 Fusion

Since the procedure was first introduced by Albee and Hibbs in 1911, fusion has been one of the most important and frequently employed operations of the spine. The ideal fusion is to achieve the necessary therapeutic goals with the minimal effective decrease in motion and minimal disruption of normal structure and function of the spinal column (Panjabi et al. 1990). A spinal fusion surgery is designed to stop the motion at a painful vertebral segment, which in turn should decrease pain generated from the joint. All lumbar spinal fusion surgery involves adding bone graft to an area of the spine to set up a biological response that causes the bone graft to grow between the two vertebral elements and thereby stabilizes that segment. Choices for fusion include posterolateral fusion with or without instrumentation; posterior lumbar interbody fusion; anterior lumbar interbody fusion; combined anterior and posterior fusion; and fusion performed by placing interbody cages either anteriorly open, or posteriorly, among which posterolateral fusion is a most time-honored method, which is postulated to relieve discogenic pain by eliminating motion through the disc (Passuti et al. 2004).

With recent developments of the fusion techniques, successful fusion rate has approached nearly 100%, but this has failed to reflect a comparable success in

relieving back pain (Boos et al. 1997). In a prospective study of a consecutive series with long-term follow-up performed by Jackson et al. (1985), a fusion rate of 87% has been reported, but only 58% of their patients with disc degeneration diseases experienced pain relief following posterolateral fusion with pedicle screws. In a review of 871 lumbar fusion procedures performed within 8 years, Zucherman et al. (1992) found that the theoretical advantages of lumbar fusion with posterolateral pedicle screw were not borne in discogenic disease. Although their successful fusion rate was 89%, they achieved clinical success in only 67%. Wetzel et al. (1994) examined the clinical efficacy of lumbar arthrodesis by a retrospective review of 48 patients who suffered discogenic pain and were treated by lumbar arthrodesis with positive discography results. Overall, only 46% had a satisfactory clinical outcome at final follow-up.

In addition, complications in lumbar fusion, especially its adverse effect on the spinal segments immediately adjacent to the fused level, have been frequently reported. Wiltse et al. (1994) examined 110 patients, all of who had been treated a posterolateral fusion, with an average of a 22.6 years follow-up. Significant loss of disc height at adjacent levels was observed, which may result in abnormal loads on facet joints adjacent to these discs. Kumar et al. (2001a) examined degenerative changes above the level of fusion in a very long term (30 years). In their study, the incidence of radiographic changes was 50% higher in the fusion group (changes of instability in 14.2% of patients and disc space narrowing in 35.7% of patients who had fusion, as against a 7.4% incidence of instability and 18.5% incidence of disc

space narrowing in the non-fusion group), which was statistically significant. In another adjacent segment degeneration study, Kumar et al. (2001b) reviewed 83 consecutive patients who underwent lumbar fusion for degenerative disc disease in a mean follow-up period of 5 years. The results showed that with radiographic evidence, 31 patients out of the 83 patients (36.1%) experienced adjacent level degenerative changes above the level of fusion. Guigui et al. (2002) obtained the similar conclusion as Kumar et al. (2001a). By examining 127 patients who underwent posterolateral arthrodesis extending from L4 to the sacrum with a mean 9-year follow-up, they found at L3-L4 level, disc narrowing was observed and intervertebral slipping also worsened, which led to the conclusion that the arthrodesis increases the frequency and severity of degenerative lesions, deteriorating the functional outcome. In 1994, Dekutoski et al. investigated the changes of adjacent segment motion after lumbar fusion with a canine model in vitro; an increase in motion at the adjacent segment (L2-L3) was found for all motions after immobilization of the L3-L7 motion segments.

Some other researchers analyzed effects of fusion on the stress distribution of the adjacent disc using FE models. Chen et al. evaluated stress alteration of the disc adjacent to the fused level by using a validated five-level FE model of the lumbar spine in 2001 and 2002. They reported stress increase in the adjacent disc under all loading conditions and pedicle fracture was observed due to the change of load-sharing between different vertebral components after fusion. Totoribe et al. (1999) studies the biomechanical effect of lumbar fusion using the FE model of a mono

motion segment (L4-L5). Their results suggested that posterolateral fusion yielded elasticity.

With fusion, there is increased intradiscal pressure, increased facet loading, and hypermobility, which in longitudinal animal studies have caused disc degeneration at the adjacent level, the development of which can in turn necessitate further surgical intervention and adversely affect functional outcome (Park et al. 2004).

2.4.2 Dynamic Stabilization Devices in lumbar spine

Dynamic stabilization of the lumbar spine, also known as soft stabilization or flexible stabilization, is currently a new concept as an alternative treatment of fusion for degenerate disc diseases, which was initiated from the dissatisfaction with the outcomes of fusion surgery for the treatment of back pain. Dynamic stabilization may be defined as a system, which would favorably alter the movement and load transmission of a spinal motion segment, without the intention of fusing segment. The hypothesis behind such a system is that the control of abnormal motions and more physiological load transmission would relieve pain, and prevent adjacent segment degeneration, and a remote expectation for dynamic stabilization is that, once normal motion and load transmission is achieved, the damaged disc may repair itself, unless the degeneration is too advanced (Sengupta 2005a).

The various dynamic stabilization systems described in the literature are all posterior implants, and can be classified into following two categories.

(A) The interspinous distraction devices

The interspinous distraction devices are to be implanted between two adjacent spinous processes at the pathological level. They are not rigidly connected to the vertebrae, which avoids the possibility of loosening, a major concern for any implant which would have to survive against motion over time. The main objective of the interspinous devices is to hold the spine or the segment into flexion and restrict extension of the segment by the interspinous distraction.

The Wallis

The first-generation implant of the non-rigid interspinous device was developed in 1986. The device's original design included a titanium interspinous blocker and an artificial ligament made of Dacron. After a successful clinical trial initiated in 1988 on more than 300 patients who were treated for degenerative lesions, a second generation implant called the "Wallis" implant (Figure 2-11) was developed by changing the material of interspinous block to polyetheretherketone (PEEK), which has much greater elasticity and is therefore less rigid than the previously used titanium (Sénégas 2002).

Figure 2-11: (A) The non-rigid fixation "Wallis" implant; (B) Schematic view of the Wallis implant in place. (Sénégas J, 2002)

The interspinous spacer restricts extension of the segment, and by distraction between the spinous processes, it holds the segment in relative flexion, a posture known to relieve neurogenic claudication pain in spinal stenosis. Two Dacron tapes were wrapped around the spinous processes and fixed under tension to the blocker. A randomized clinical trial and an observational study of the second generation implant are currently underway. The author recommended the Wallis system can be used for lumbar disc disease in the following indications: (1) discectomy for voluminous herniated disc leading to substantial loss of disc material; (2) a second discectomy for recurrence of herniated disc; (3) discectomy for herniation of a transitional disc with sacralization of L5; (4) degenerative disc disease at a level adjacent to a previous fusion; (5) isolated Modic I lesion leading to chronic low-back pain.

The X-STOP

Another titanium interspinous distraction device that is currently undergoing investigation for FDA approval is the X-STOP (St. Francis Medical Technologies, Concord, Calif., USA). This device consists of an oval titanium spacer that is positioned between the two symptomatic spinous processes. The lateral wing is then attached to prevent the implant from migrating anteriorly or laterally out of position, as shown in Figure 2-12.

Figure 2-12: A sagittal, transverse, posterior, and orthogonal view of the X-STOP implant (Zucherman et al. 2004)

Initial published reports examined the biomechanical effects of the implant in cadaveric models. Swanson et al. (2003) investigated the intradiscal pressure changes both at the implanted level and at adjacent levels under flexion, neutral and extension in a cadaver model. They found that the pressures at the adjacent discs were not significantly affected by the interspinous implant insertion, while there was a significant decrease in intradiscal pressure at the treated level in the posterior annulus and nucleus in the neutral and extended positions. Based on their findings, they concluded that the implant would be unlikely to cause accelerated disc degeneration at the adjacent levels. Lindsey et al. (2003) tested the kinematics of the instrumented and adjacent levels due to the insertion of this interspinous implant in seven lumbar spines (L2-L5) under different loading modalities. They reported that at the instrumented level (L3-L4), the flexion-extension range of motion was significantly reduced from 7.60 for the intact specimens to 3.10 for the implanted specimens, and ROMs under axial rotation and lateral bending were not altered (for lateral bending, the difference of rotation between the intact and the implanted specimens was 0.60; for torsion, the difference was even smaller with a

value of 0.10). The ROM under all kinds of loading configurations at the adjacent segments was not significantly affected by the device (the difference of ROM between the intact and the implanted specimens for all loading modalities was less than 0.60). Therefore, it was concluded that the implant did not affect the kinematics of the adjacent motion segments. To evaluate the safety and efficacy of the X-STOP interspinous implant, Zucherman et al. (2004, 2005) conducted a prospective randomized multi-center study with a 1-year follow-up and 2-year follow-up, respectively, and reported the corresponding trial results. In their study, 191 patients were treated, among which 100 received the X-STOP and 91 received non-operative therapy. Using the Zurich Claudication Questionnaire as the primary outcome measurement, they observed a significant improvement in clinical symptoms in the X-STOP group in comparison with traditional non-operative management. The authors also identified the primary indication for the use of the X-STOP system is in patients whose symptoms are exacerbated in extension and relieved in flexion. Richards et al. (2005) and Wiseman et al. (2005) examined the effect of the X-STOP implant on the dimensions of spinal canal and neural foramina during flexion and extension and on the facet loading during extension in cadaver specimens, respectively. They found that the X-STOP interspinous process implant prevented narrowing of the spinal canal and foramina during extension and significantly decreases the facet loading at the treated level but does not significantly affect the dimensions of the canal and foramina and the facet loading conditions at the adjacent levels.

(B) The ligament based dynamic stabilization systems

Graf ligament

The Graf ligament system was one of the first relatively widely practiced methods of soft stabilization. The concept was that, once the facet joints were locked, it would stop rotation. This instrumentation system consists of 5- to 7-mm titanium pedicle screws and looped 8-mm braided polyester bands (Figure 2-13).

Figure 2-13: Graf artificial ligament instrumentation (Kanayama et al. 2005)

Quint et al. (1998) performed an *in vitro* test in six fresh frozen human lumbar spine specimens to determine the stabilizing effect of Graf ligament following laminectomy. They found that this ligamentous system significantly reduced the range of motion with the highest stabilization effect observed during flexion. Graf ligament has been used by several independent surgeons and its mid- to long-term clinical results have been reported by several authors (Grevitt et al. 1995; Brechbühler et al. 1998; Hashimoto et al. 2001; Rigby et al. 2001; Gardner et al. 2002; Askar et al. 2004; Kanayama et al. 2005) and the clinical results remain controversial. In 1995, Grevitt et al. reported the results of the first 50 consecutive patients using the Graf stabilization with an average follow-up period of 24 months.

They found the early clinical results of Graf ligament were encouraging with “excellent” or “good” results in 72% of the patients. In another relatively longer follow-up, Brechbuhler et al. (1998) also reported the positive effect of Graf ligament with “excellent” or “good” results in 74% of the patients. By reviewing the clinical results of a selected patient population with degenerative lumbar disc disease and consecutive instability of the affected motion segments, they demonstrated that overall lumbar lordosis as well as regional lordosis of the instrumented segment was maintained after a mean period of observation of 50 months. Although there was a lightly decrease of disc height at treated segments, the changes were not statistically significant. They concluded that the Graf technique showed the protective effect on the degenerative cascade of a destabilized motion segment. The clinical results reported by Hashimoto et al. (2001) indicated that the Graf system is a successful alternative to spinal arthrodesis for mild and early lumbar degenerative disease with minimum flexion instability of less than 10°. In a 7-year follow-up, Gardner et al. (2002) also suggested the long-term sustained beneficial effects of Graf ligamentoplasty with a 62% rate of “excellent” or “good” subjective results in 31 patients. In contrast, the results reported by Rigby et al. (2001) were not as encouraging as earlier results. In their clinical study, they retrospectively reviewed 51 patients out of 69 who were treated with Graf ligament stabilization following failure to response to conservative management of back pain with a mean follow-up time of 51.7 months (varying from 23 to 84 months). They found post-operative Oswestry Disability Index scores showed no appreciable improvement in comparison with the pre-

operative scores and the overall re-operation rate was 21%. Similar results were recently reported by Askar et al. (2004). More recently, Kanayama et al. (2005) evaluated the clinical effects of Graf system by reviewing 64 consecutive patients using this device with a mean follow-up period of 67 months (36-112 months). They found that the Graf system maintained lordosis and preserved segmental motion in 80% of patients and concluded that it was an effective alternative to spinal arthodesis in the treatment of symptomatic degenerative spondylolisthesis (Grade I) with minimal disc space narrowing and coronal facet tropism. In their retrospective study, they also clearly defined surgical indication for Graf ligamentoplasty: symptomatic degenerative spondylolisthesis with (1) <25% of vertebral slip, (2) coronal facet tropism, and (3) minimal disc space narrowing.

Fulcrum Assisted Soft Stabilization (FASS) System

The FASS system is developed based on the concept that the primary mechanism of chronic low back pain is abnormal load distribution across the disc space following disc degeneration rather than the abnormal motion in a degenerative disc. In fact, this system is only a modification of normal ligament instrumentation by introduction of a fulcrum in front of the ligament connecting the pedicle screws (Figure 2-14), instead of the facet joints or the posterior edges of the vertebral bodies as a fulcrum in the Graf system. The ligament and the fulcrum combination are expected to maintain or restore the lordosis, as well as to act as a load-sharing device with the disc and the facets.

Figure 2-14: Schematic diagram of FASS system. (Sengupta et al. 2005)

Sengupta et al. (2005b) conducted an experimental study on cadaver spine and spine model to evaluate biomechanical effects of the FASS system. In their experiment, effects of FASS system with different fulcrum diameters on the range of motion (ROM), lordosis and changes of intradiscal pressure during flexion-extension were explored on the selected spine model, whose load-deformation characteristics were validated against that of the cadaver spines. They found that the FASS system further reduced ROM significantly on the basis of reduction due to the application of ligaments in all loading configuration and could maintain the lordosis achieved by the ligament alone under a certain length of the fulcrum. Following application of the fulcrums, the disc pressure was reduced at maximum flexion while not affected in maximum extension. The authors explained that this was because the extension was not produced by a compression load in their experimental set-up, different from the normal condition under which both flexion and extension were produced by a compression load. They also proposed that it might be possible to achieve any desired degree of disc unloading by using a suitable combination of ligament and fulcrum sizes and the ideal degree of disc unloading would be different from one individual disc to the other, which probably

was located somewhere between 30 to 50%. Studies of a clinically applicable design of the FASS system on cadaveric spine are still under way.

Dynamic Neutralization System (Dynesys)

The Dynesys (Zimmer GmbH, Winterthur, Switzerland) is another frequently used dynamic stabilization device and was developed by Gilles Dubois in 1994, aiming at pathological conditions with some form of segmental instability and various forms of sequelae. It is a non-fusion pedicle screw system for the stabilization of the lumbar spine, consisting of two titanium alloy screws connected by a polyester cord (PET) and a polycarbonateurethane (PCU) cylindrical spacer (Figure 2-15). The spacer withstands compressive loads, while the PET cord stabilizes the system and sustains tensile loads as well as flexion moments.

Figure 2-15: (A) Components of the dynesys system; (B) Schematic view of dynesys system in place. (Adapted from the following website whose copyright is reserved by Zimmer Inc.:

<http://www.zimmerspine.eu/web/enUS/pdf/dynesysA5AW1.pdf>)

In vitro biomechanical investigations of the Dynesys system were carried out by several independent researchers. By testing the Dynesys on four cadaveric lumbar

spine segments from L3-L4 or L4-L5 level with a new spine simulator, Freudiger et al. (1999) found that this system significantly reduced the mobility of the instrumented motion segment level. In another *in vitro* study of the Dynesys system, Schmoelz et al. (2006) investigated the influence of this device on load bearing of a bridged disc on six fresh frozen human lumbar spines and the results showed that the Dynesys device significantly unloaded the disc in extension and lateral bending while restored the load distribution similar to an intact segment in flexion and axial rotation. They also found that the effects of the stabilization system on the intervertebral disc pressure of the adjacent discs were negligible. A more comprehensive *in vitro* study of the kinematic behavior of the Dynesys dynamic stabilization system was recently performed by Niosi et al. (2006). The main conclusions obtained from their study included that the Dynesys system significantly reduced the magnitude of the ROM in all loading directions with least significant effect in axial rotation, it restored the neutral zone to a magnitude less than that of the intact spine, and the helical axis of motion was significantly shifted posteriorly in flexion-extension and axial rotation. In addition, they found that the kinematic behavior of the Dynesys system was affected by its spacer length: the longer spacer resulted in a range of motion and motion pattern which was much closer to that seen in an intact specimen.

Several retrospective reviews of clinical results with a mean follow-up period varying from 24 months to 38.1 months are also available in the literature and the reported outcomes remain controversial. Stoll et al. (2002) performed a multi-

center study of the Dynesys dynamic stabilization system by reviewing the clinical results on 83 patients with a mean follow-up time of 38.1 months and reported that at the follow-up, mean pain and function scores significantly improved with Oswestry Disability Index from 55.4% to 22.9%. Similar results were reported by Schmake et al. (2003) and by the same authors who examined the clinical results on 39 patients with a mean follow-up period of 35.3 months in 2004. The clinical results from these studies proved that Dynamic neutralization is a safe and effective alternative to fusion in the treatment of degenerative lumbar disease. However, such improvement was not observed by other authors in their retrospective studies. Grob et al. (2005) conducted a retrospective study on a consecutive series of 50 patients instrumented with Dynesys over the preceding 40 months and found that only 40% of the patients reported an improvement in their ability to perform physical activities and just 50% an improvement in their quality of life. In 2002, Eberlein et al. numerically evaluated the effect of the Dynesys implant using a FE model and observed that considerable stiffening for flexion and extension and extremely stiff in torsion happened.

All foregoing researches demonstrate that dynamic stabilization devices appear to be promising. The few soft stabilization systems that have had clinical applications for degenerative lumbar diseases so far yielded a clinical outcome comparable to that of fusion while the surgical procedure was less invasive. However, the biomechanical compatibility and corresponding indications for these devices are far from clear, therefore further investigations are required.

2.5 Discussion on the need for the current project

From the foregoing review, it can be noted that degenerate disc diseases are responsible for most of the low back pain and the morphological and biochemical changes occurring during disc degeneration will cause changes in biomechanical behavior of the disc and even the whole spinal column. The use of developed dynamic stabilization devices is expected to overcome drawbacks in the traditional treatment of fusion for degenerative disc diseases since the surgical procedure for dynamic implants is less invasive and less possible to accelerate the degeneration at motion segments adjacent to the spinal fusion site while similar clinical outcomes are achieved. Although dynamic stabilization implant for degenerative disc diseases has been becoming more and more popular, its exact mechanism remains unclear.

In addition, with the increasing of computer power and the availability of commercial FE software, FE method has become a useful tool in spinal research and an indispensable complement to *in vivo* or *in vitro* studies due to its reproducibility and repeatability and the difficulty to get human spinal specimens. It can be also used to evaluate biomechanical effects of a new implant and optimize its design before the implant is manufactured. A FE model can be used to predict the biomechanical behavior of the human spine or the efficacy of a new implant in improving biomechanical characteristics of spinal motion segments only after it is validated. The geometrical data for most of developed FE models to date were

extracted from CT scans, which cannot clearly separate the bone structure from the soft tissue for the reconstruction of the FE model.

Therefore, the current study is concentrated on the development of a geometrically accurate, 3D nonlinear FE model of the lumbar spinal motion segment (L3-S1) with geometrical data obtained using a high-definition digitizer with accuracy up to 0.1mm. The created FE model is then validated against experimental results available in the literature. The FlexPLUS and the rigid rod system were then incorporated into the validated model, and their biomechanical influence (i.e., ROM, load transmission) on the treated lumbar motion segment and adjacent levels were studied and compared against that of some existing products.

Chapter 3: Materials and Methods

3.1 Introduction of FE Method

The FE method is a numerical analysis technique for obtaining approximate solutions to many types of engineering problems. It originated from the needs for solving complex elasticity, structural analysis problems in civil engineering and aeronautical engineering. The basic concept of FE method is that a body or structure may be divided into smaller elements of finite dimensions called as “Finite Elements”. The original body or structure is then considered as an assemblage of these elements connected at a finite number of joints or “Nodes”. The properties of the elements are formulated and combined to obtain the properties and geometries of the entire structure.

The equations of equilibrium for the entire structure are then obtained by combining the equilibrium equations of each element such that the continuity is ensured at each node. The necessary boundary conditions are then imposed and the equations of equilibrium are solved to obtain the required unknown quantities such as stress, strain, temperature distribution or velocity flow depending on the application. Thus instead of solving the problem for the entire structure in one operation, in the FE method, attention is mainly paid to the formulation of properties of the constituent elements.

A common procedure is adopted for combining the elements, solution of equations and evaluation of the required variables in all fields. Thus the modular structure of the FE method is well exploited in various disciplines of engineering.

3.2 FE Application in Biomechanics

The FE method is but one of numerous analytical techniques used in conjunction with experimental techniques to solve problems in biomechanics. Although it arose from the fields of mathematics and engineering in the 1950s, the FE method was not applied to solve problems in biomechanics until the late 1960s.

The FE method enjoys a distinct advantage over many other mathematical modeling techniques (particularly those involving closed-form equations) in that it can handle, with relative ease, the complex geometries and material and geometrical nonlinearities encountered in the modeling of spine (Gilbertson et al. 1995). In FE modeling, the spine can be modeled with a wide variety of element types (e.g. bars, plates, blocks, etc.) in varying degrees of complexity. This enables the researcher to obtain preliminary estimates rather quickly with a simple model while developing a more realistic, complex model. The method also possesses other advantages, which contribute to its popularity in biomechanics. First, a FE model can be analyzed parametrically to investigate different loading conditions, pathological states and material properties with no new specimens involved. Second, the approach can deliver additional information which is difficult or even impossible to acquire experimentally. Finally, as computer-processing equipment becomes more sophisticated and available, the overall time and costs are reduced.

The FE method has been used extensively and will be continuously used to predict the biomechanical performance of various implant designs as well as the effect of clinical factors on implant success.

3.3 Intact FE Model

In this section, the FE model of L4-L5 motion segment was first developed. Based on this mono-segmental model, the L3-S1 model was further created. The details for the development of L4-L5 model and L3-S1 model are given below. The validation study of the FE models is presented in section 4.1 and 4.2.

3.3.1 Development and validation of L4-L5 model

The same digitizing and FE modeling technique developed by senior members (Teo et al. 1994; Teo et al. 2001; Lee et al. 2003) of the group is also employed in the current project. For the sake of completion, the procedure is described again here. To develop a realistic, geometrically accurate FE model of a lumbar motion segment (L4-L5), the high-definition multi-axis digitizer (FaroArm, Bronze Series, Faro Technologies, Inc.) was adopted to extract detail geometrical data from the specimen.

3.3.1.1 Specimen

In this project, the 3D digitized point data of the lumbar vertebrae (L4 and L5) were obtained from an embalmed lumbar spine acquired from the Singapore General Hospital. Examination of the specimen chosen showed no gross physical abnormalities. The geometrical dimensions of the specimen were found to be within the quantitative range of anatomical lumbar vertebra from Asian subjects

based on 60 lumbar vertebrae (Tan et al. 2002). The dimensional details for this specimen are tabulated in Table 3-1:

Specimen	Age	Sex	Race	Weight(kg)	Height(m)	Cause of Death
1	59	M	Chinese	64	1.68	Liver Cancer

Parameter	Specimen		Tan et al. (2002)	
	L4	L5	L4	L5
EPW _u	49.15	48.52	49.35 ± 0.22	48.89 ± 0.4
EPW _l	52.91	51.03	53.34 ± 0.57	51.42 ± 0.49
EPD _u	36.05	35.31	36.26 ± 0.23	35.82 ± 0.57
EPD _l	34.99	33.43	35.62 ± 0.73	33.75 ± 0.51
VBH _a	25.47	26.34	25.36 ± 0.68	25.83 ± 0.66
VBH _p	25.10	23.35	25.42 ± 0.40	23.51 ± 0.71
PHD _l	15.33	20.62	15.29 ± 0.42	20.78 ± 0.51
PDH _r	15.82	19.87	15.72 ± 0.41	20.08 ± 0.39
PDW _l	10.46	13.02	10.71 ± 0.52	13.34 ± 0.51
PDW _r	9.95	13.57	10.07 ± 0.46	13.90 ± 0.61
EPA _u	1350.21	1312.52	1355.10 ± 23.87	1326.22 ± 28.90
EPA _l	1400.82	1251.11	1410.99 ± 41.74	1240.08 ± 24.74

Table 3-1: Specimen Data (in mm). See Figure 3-1 for the interpretation of parameters

Figure 3-1: Four orthogonal views (front, side, top and isometric) of a lumbar vertebra. EPWu-upper end-plate width; EPWI-lower end-plate width; EPDu-upper end-plate depth; EPDI-lower end-plate 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 end-plate area; EPAI-lower end-plate area

3.3.1.2 Procedure for the FE model development of lumbar vertebrae

Figure 3-2 shows the main steps to develop a FE model using the coordinate data extracted from the specimen. First the essential cross-sections of the specimen were marked and digitized using FaroArm. Second, the files containing the 3D coordinate data from the digitizer were translated to a format with an extension of IGES by Caliper 3D. The obtained IGES file was then processed to Surfacr 7.1 (Imageware, Inc., Michigan, USA) to create the planar coordinate points of the cross sections of the lumbar vertebrae. Finally the file including planar coordinate

points generated in Surfer 7.1 was imported to commercially available FE software, ANSYS 10.0 (ANSYS, Inc. Pennsylvania, USA), for 3D modeling and meshing generation. The details for each step were indicated as follows.

Figure 3-2: Flowchart for FE model generation of L4 and L5

3.3.1.3 Capturing of 3D geometrical data using the digitizer

To extract the detail 3D coordinate data for the lumbar vertebrae L4 and L5 respectively, their essential cross sections were marked and digitized using FaroArm. Since the procedures of data extraction for L4 and L5 are same, here L4 has been chosen as example.

Before the specimen was processed to the digitizer, a small hole was drilled into its body, and then a threaded rod was passed through and secured by two nuts. The rod was in turn clamped in a vice clamp, as shown in Figure 3-3. The assembly was then fixed within the reach of the digitizer's arm.

Figure 3-3: Fixation of the specimen in the vice clamp

To ensure the accuracy of the extracted coordinate data, calibration was undertaken before any digitization was carried out. Figure 3-4 illustrates some of the geometrical point data from the digitizer. The 3D coordinate point data from the digitizer were translated into a universal graphics data format (IGES) in Caliper 3D and later processed to Surfacar 7.1 to obtain the sequential planar coordinate data of cross sections of the vertebra, which was also saved as an IGES file for following exportation to a commercially available FE modeling software, ANSYS 10.0 (ANSYS, Inc. Pennsylvania, USA) for the 3D model and FE mesh construction.

Figure 3-4: Captured coordinate point data for L4 from digitizer

3.3.1.4 Intact Solid Model Developing

After the creation of planar coordinate points of cross sections, the corresponding IGES file was next imported into ANSYS 10.0 for reconstruction of the solid model of the vertebra. There are mainly two approaches for FE model generation in ANSYS: solid modeling and direct generation. With solid modeling, the geometrical boundaries of the model are defined first, and then controls over the size and shape of the elements are established. Consequently, all the nodes and elements are created automatically. By contrast, with direct generation, the locations of all nodes and the size, the shape, and connectivity of every element are determined before the entities of the model are defined. In this project, the former method was employed because it possesses several advantages over the latter one. The advantages enjoyed by solid modeling method include aptness for large or complex models (especially 3D models of solid volumes), allowance for geometric operations (such as dragging and rotations) that cannot be done with nodes and elements, support for the use of Boolean operations, allowance for modifications to geometry, and facility to change element distribution.

In ANSYS, a bottom-up approach was adopted to rebuild the solid model of the vertebra. The planar 3D coordinate points imported to ANSYS as IGES file from Surfacr 7.1 was used to define the “lowest order” solid model entities known as keypoints, and all “higher order” solid model entities (i.e., lines, areas, and volumes) were created based on these keypoints. The solid model was developed in two stages: surface creation and solid volume generation. At first, closed planar curves were created by using “splines” connecting these keypoints and each curve

was divided into four line sections which were used to form an area. Each area created in this way was used to represent one cross section outline. Next, by using all corresponding line sections in each cross section outline with “skinning”, the four-sided surfaces of a solid were generated. Thus, a series of surfaces were created. The surfaces of whole model were carefully re-examined against the specimen for accuracy. A series of volumes were then generated by enclosing its six surfaces. The solid model for the entire structure was thus developed following this procedure.

After the solid models for L4 and L5 were developed, the next step was to generate the solid model for the intervertebral disc between them. The intervertebral disc was modeled as three layers from superior to inferior. The superior and inferior layers were both 0.5mm in thickness for endplates. The middle layer consisted of nucleus pulposus surrounded by annulus fibers. The lumbar intervertebral discs tend to be of greater height anteriorly than posteriorly, which is responsible for the lordosis of the lumbar spine. Therefore, a thickness of 13.65mm and 9.35mm was taken for the anterior and posterior disc heights respectively, with a mean height of 11.5mm, from the literature (Gilad et al. 1986; Inoue et al. 1999).

The solid model of the motion segment L4-L5 was finally created by connecting the two lumbar vertebrae with the intervertebral disc. Figure 3-5 illustrates the volume rendering of the lumbar motion segment L4-L5.

Figure 3-5: Volume rendering of L4-L5 motion segment

3.3.1.5 FE Model Generating

After the solid model was developed, generation of the mesh was carried out. The FE mesh was obtained by discretizing the solid model using a mapped mesh technique. The mapped mesh method in ANSYS allows the automatic creation of finite elements, which conform exactly to the shape of the defined volume. But mapped mesh requires a volume to be regular, that is, it must meet certain criteria. Therefore, before meshing the model, element type and element size were identified depending on the requirement for mapped mesh, the complexity of the geometry, desired problem size and analysis nature of the FE model.

In the present project, eight-noded isoparametric hexahedral solid elements, instead of four-noded tetrahedron solid elements, were chosen to model the motion segment L4-L5 with finite element. Quadrilateral and hexahedral elements are generally considered to have superior performance to triangle and tetrahedral

elements when comparing an equivalent number of degrees of freedom. By using hexahedral elements, the number of elements can be significantly reduced, and consequently analysis and post-processing times can be reduced. Moreover, quadrilateral and hexahedral elements are more suitable for non-linear analysis as well as situation where alignment of elements is important to the physics of the problems, such as in the intervertebral disc where the fibers are oriented at $\pm 30^\circ$ to the horizontal plane, and the articulating facets.

Referring to the anatomy of lumbar vertebrae, the vertebral body was modeled as a cancellous core with a cortical shell whose thickness was assumed to be 1.0mm. Two cartilaginous endplates with a thickness of 0.5mm were assumed at the superior and inferior surfaces of the intervertebral disc, respectively. For the intervertebral disc, the nucleus pulposus portion was assumed to occupy approximately 48% of the disc's cross sectional area, which fell in the range reported in the literature (Panjabi et al. 1990) and simulated to be inviscid incompressible. The annulus fibrous was modeled as a composite with a matrix of homogeneous ground substance reinforced by annulus fibers. It was divided into 3 consecutive layers to simulate its laminar structure. In each layer, collagen fibers were inclined on average at $\pm 30^\circ$ with respect to the circumferential direction of the disc (Figure 3-6). A volume ratio of 19% was assumed for the fiber content by referring to the literature (Sharma et al. 1995). The average fiber cross sectional areas of each layer were then calculated based on this assumed volume fraction, the number of fiber elements modeled, the length of the fibers and the volume of each

annulus layer. Thus, the geometrical nonlinearity of the intervertebral disc was taken into account.

Figure 3-6: Collagen fiber arrangement in the annulus fibrosus laminar

The seven ligaments incorporated in the present model were the anterior and posterior longitudinal ligaments, ligamentum flavum, the intertransverse ligament, the capsular ligament, the interspinous ligament and the supraspinous ligament (Figure 3-7). They were modeled as a collection of uniaxial elements oriented in the direction of their collagenous fibers and sustained tensile force only. The cross sectional area of each type of ligaments was chosen based on the values reported in the literature (Chazal et al. 1985; Shirazi-Adl et al. 1986; Goel et al. 1990; Pintar et al. 1992), and their insertion sites were determined according to anatomical books (Crouch 1985; Marieb 2004). The contact type used in current FE model is surface-to-surface contact and assumed to be frictionless, as the facet joints are anatomically covered with a cartilage layer. The contact stiffness is automatically calculated in ANSYS using the material properties of the underlying elements. According to the rule for designation of contact and target surfaces, the large concave surface of superior facet of L5 is defined as target surface, while the nearly

flat small surface of inferior facet of L4 is defined as contact surface. When the motion segment is subjected to external loading, the contact surface is expected to come into contact with the target surface. Figure 3-8 shows the complete FE model of L4-L5 motion segment.

Figure 3-7: FE model of ligaments

Figure 3-8: FE model of L4-L5 motion segment

After convergence test (see Appendix B for detail), Table 3-2 summarizes the type and number of elements used to model the various component of the motion segment L4-L5.

Component Name	Element Type	Number of Element
Cortical Bone	8-node brick (solid 45)	560
Cancellous Bone		2480
End plate		304
Bony Posterior Element		4210
Annulus Ground Substance		336
Nucleus Pulposus		340
Annulus Fiber	3D-Cable (link 8)	896
Intertransverse Ligaments		6
Capsular Ligaments		14
Supraspious Ligaments		3
Interspinous Ligaments		9
Ligamentum Flavum		7
Anterior Longitudinal Ligaments		28
Posterior Longitudinal Ligaments		20
Articular Facets	3D surface to surface contact (targe 170, conta 174)	56

Table 3-2: Types and number of elements used in FE Model of L4-L5

3.3.1.6 Material Properties

Material properties of different structural components of the motion segment are the next important consideration following the generation of FE model. The nucleus pulposus portion of the disc is modeled to be nearly inviscid incompressible. Table 3-3 summarizes the material properties assigned to each part in the present project.

All the significant components of the motion segment were assumed to be homogenous, isotropic and linear elastic except that the ligaments were considered nonlinear. The nonlinear stress-strain curves for the ligaments and their cross-sectional areas are illustrated in Figure 3-9 and Table 3-4, respectively.

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

Figure 3-9: Stress-strain curve for the lumbar spine ligaments

Component name	Young's Modulus E (MPa)	Poisson's Ratio ν	Reference
Cortical Bone	12,000	0.3	Goel et al. 1988 ; Lavaste et al. 1992
Cancellous Bone	100	0.2	Goel et al. 1988; Shirazi-Adl et al. 1984
Bony Posterior Elements	3500	0.25	Goel et al. 1988 ; Shirazi-Adl et al. 1986
Endplate	500	0.3	Lavaste et. al. (1992)
Annulus Ground Substance	2	0.45	Shirazi-Adl et al. 1984; Goel et al. 1988
Nucleus Pulposus	1	0.4999	Lavaste et al. 1992
Annulus fiber	500	0.3	Lavaste et al. 1992
Ligaments	Material nonlinearity (as in Figure 3-9)		Chazal et al. 1985 ; Panjabi et al. 1990 ; Goel et al. 1990; Pintar et al. 1992;

Table 3-3: Material properties adopted in this project

Ligament	ALL	PLL	ITL	ISL	SSL	LF	CL
Area(mm ²)	49	17.7	12	25.5	23	67	36

Table 3-4: Cross-sectional areas of lumbar spine ligaments used in this study

3.3.2 Development of L3-S1 model

The FE models of the L3 and S1 vertebrae from the same lumbar spine have been developed by the senior members (Lee et al. 2004; Guo et al. 2005). Based on the validated L4-L5 model, the L3-L4 disc and vertebra L3 were added superiorly, and the L5-S1 disc and vertebra S1 were added inferiorly to form the FE model of L3-S1 lumbar spine. The anterior and posterior disc heights for L3-L4 disc were 12.2mm and 9.0mm; for L5-S1, the corresponding values were 14.5mm and 7.5mm, respectively, which were within the range reported in literature (Gilad et al. 1986). The mid-plane of L3-L4 disc was kept horizontal. The L3-S1 model had a total of 20328 elements and 21048 nodes. The element type and material properties for each component are the same as those used for L4-L5 model (refer to Tables 3-2, 3-3, 3-4 and Figure 3-9). The intact L3-S1 model was shown in Figure 3-10.

Figure 3-10: FE model of L3-S1 lumbar spine

3.3.3 FE model of nucleotomized spine with implant

In the current project, two types of posterior implants were simulated: the FlexPLUS system and the rigid rod system. The IGS file of FlexPLUS was obtained from the French company, and imported and meshed in ANSYS. The nucleus pulposus of L4-L5 disc was removed to simulate the most extreme case of disc degeneration. To the nucleotomized model, FlexPLUS and rigid rod were inserted in the same location at L4-L5 level, respectively. A solid volume with a cross-section area of 26.91mm^2 (28.26mm^2 for the pedicle screw) was first created between pedicle and FlexPLUS or rigid rod, and then meshed with hexahedral solid element (solid 45), thus the dynamic stabilization device was linked to the pedicle, as shown in Figure 3-11. Both of the devices were implanted bilaterally and the rigid screws were simulated by defining corresponding elements with a material of titanium. The details of simulation of each device are given below:

Figure 3-11: Detail view for the link between rod and vertebrae (left side)

FlexPLUS

The structure of FlexPLUS is shown in Figure 3-12. It is composed of an external cylinder and seven internal titanium wires. The external cylinder includes the upper rigid part and the lower rigid part and the middle flexible part. The rigid parts are made of titanium and supposed to provide strong fixation between the FlexPLUS and the vertebrae, and the flexible part is made of polycarbonate urethane (PCU): ChronoFlex® C 80A. Among the seven titanium wires, the middle one is straight, while the other six wires surround the middle one in 480 ° of twist outside, as shown in Figure 3-13. The flexible part with PCU and the internal wires are not rigidly fused and allow the treated motion segment to move flexibly within a certain range. In clinical application, this FlexPLUS was supposed to be linked to the two adjacent pedicles of L4 and L5 on each side by pedicle screws.

Figure 3-12: The structure of FlexPLUS

Figure 3-13: The ISO view of a FlexPLUS structure

The whole structure of FlexPLUS was discretized with 8-noded brick elements (solid 45), and the contact between wires was simulated using contact pairs (target 170 and conta 174). Different material properties were assigned to different parts. The element type, element number and material property of each component are summarized in Table 3-5, and Figure 3-14 and Figure 3-16a illustrate the FE models of FlexPLUS and its incorporation in L3-S1 lumbar spine, respectively.

Figure 3-14: Exploded FE model of FlexPLUS

In order to check that the FlexPLUS is closely modeled, the FE analysis of the FlexPLUS under a compressive force of 500N was performed. This loading condition was also used in the experimental study carried out by the company (SpineVision). Under a 500N compressive force, the behavior of the FlexPLUS, including the wire bundle, was found to be similar to what was observed in the experimental study. Furthermore, the obtained force-displacement curve was also in good agreement with the experimental result, as shown in Figure 3-15. Therefore, it can be confirmed that the FlexPLUS structure was closely modeled in the current FE model.

Figure 3-15: Comparison of FE result with experimental result for FlexPLUS under compression

Rigid Rod

The rigid rod was simulated to have the same dimensions as that of FlexPLUS, 6mm in diameter and 49mm in length, and given the material properties of titanium

(Table 3-5). Figure 3-16b shows the incorporation of rigid rod in nucleotomized L3-S1 lumbar spine model.

Component Name		Material Type	Young's Modulus E(MPa)	Poisson's ratio ν
FlexPLUS	Rigid part & Wires	Ti-6Al-4V	110,000	0.31
	Flexible part	PCU	8.62	0.47
Rigid Rod		Ti-6Al-4V	110,000	0.31

Table 3-5: Material properties used for different stabilization systems

(a)

(b)

Figure 3-16: Incorporation of different devices in nucleotomized L3-S1 lumbar spine model. (a) FlexPlus; (b) Rigid Rod

3.4 Boundary and Loading Conditions

The inferior surface of S1 vertebral body was constrained in all six degrees of freedom. The pure moment of 9Nm in various degrees of freedom, which are

flexion, extension, lateral bending and torsion, was applied to the superior surface of L3 vertebral body for all models (i.e. intact model, nucleotomized model, nucleotomized with FlexPLUS model and nucleotomized with rigid rod model), as shown in Figure 3-17. The maximum moment was achieved in five steps with an increment of 1.8Nm.

Figure 3-17: Boundary and loading conditions

3.5 Data Analysis

3.5.1 Rotational motion across motion segment

The rotational angles were calculated using the nodal deformations of a constant set of nodes, which lied in the middle of the vertebral bodies. This was accomplished by recording the original coordinates of two points on opposite sides of the vertebral body parallel to the plane of deflection, as shown in Figure 3-18. The change of each of the coordinates was measured and a rotational angle was

calculated. This measurement was taken in the sagittal plane for flexion and extension, the frontal plane for lateral bending, and the transverse plane for torsion.

Calculation of rotational angle θ :

Flexion/Extension:

$$\theta(L3-L4) = \text{atan}(\frac{Y1(235366) - Y1(234683)}{Z1(235366) - Z1(234683)}) - \text{atan}(\frac{Y0(235366) - Y0(234683)}{Z0(235366) - Z0(234683)})$$

$$\theta(L4-L5) = \text{atan}(\frac{Y1(203086) - Y1(202538)}{Z1(203086) - Z1(202538)}) - \text{atan}(\frac{Y0(203086) - Y0(202538)}{Z0(203086) - Z0(202538)})$$

$$\theta(L5-S1) = \text{atan}(\frac{Y1(205050) - Y1(204455)}{Z1(205050) - Z1(204455)}) - \text{atan}(\frac{Y0(205050) - Y0(204455)}{Z0(205050) - Z0(204455)})$$

Lateral Bending:

$$\theta(L3-L4) = \text{atan}(\frac{Y1(234732) - Y1(235033)}{X1(234732) - X1(235033)}) - \text{atan}(\frac{Y0(234732) - Y0(235033)}{X0(234732) - X0(235033)})$$

$$\theta(L4-L5) = \text{atan}(\frac{Y1(202565) - Y1(202825)}{X1(202565) - X1(202825)}) - \text{atan}(\frac{Y0(202565) - Y0(202825)}{X0(202565) - X0(202825)})$$

$$\theta(L5-S1) = \text{atan}(\frac{Y1(204759) - Y1(204439)}{X1(204759) - X1(204439)}) - \text{atan}(\frac{Y0(204759) - Y0(204439)}{X0(204759) - X0(204439)})$$

Torsion:

$$\theta(L3-L4) = \text{atan}(\frac{Z1(234732) - Z1(235033)}{X1(234732) - X1(235033)}) - \text{atan}(\frac{Z0(234732) - Z0(235033)}{X0(234732) - X0(235033)})$$

$$\theta(L4-L5) = \text{atan}(\frac{Z1(202565) - Z1(202825)}{X1(202565) - X1(202825)}) - \text{atan}(\frac{Z0(202565) - Z0(202825)}{X0(202565) - X0(202825)})$$

$$\theta(L5-S1) = \text{atan}(\frac{Z1(204759) - Z1(204439)}{X1(204759) - X1(204439)}) - \text{atan}(\frac{Z0(204759) - Z0(204439)}{X0(204759) - X0(204439)})$$

Note: 0-nodal coordinate before rotation; 1-nodal coordinate after rotation

Figure 3-18: Schematic showing the locations and numbers of the nodes and the equations used to determine the rotational angle of motion segments in the FE model

3.5.2 Facet forces

In the current FE models, facet joints were represented using contact elements (targe 170, conta 174). The facet forces were obtained by extracting the contact force using the postprocessing module in the FE software. To calculate the facet normal contact force (force normal to the spatial orientation of facet articulating

surface), it was calculated by adding up vectorial summation of three components (axial, sagittal, lateral) of the contact force on each facet. Then the facet forces on both sides at a given segments were added together as the total facet force at this segment. The total facet forces were calculated at all the segments for all the models in this study.

3.5.3 Von Mises stresses

The maximum von Mises stress and stress contours at annulus fibrosus and nucleus pulposus of L3-L4 and L5-S1 and annulus fibrosus of L4-L5 were extracted for all the models. These stress contours and the maximum stress values would indicate the change of loading patterns through the intervertebral disc for the intact, the nucleotomized and the implanted models.

Chapter 4: Results and Discussions

4.1 Validation of L4-L5 model

The L4-L5 model was first validated under compression, flexion, extension, lateral bending and axial torsion by directly compared with previous reported experimental data conducted on human cadavers as reviewed in section 2.1.1.2.

For axial compression, the nodes in the superior surface of L4 were coupled to create a master node, to which an axial force of 3000N was applied in five even steps. At each increment, the axial displacement of the master node was extracted and compared with reported experimental data (Virgin 1951; Brown et al. 1957; Markolf 1972), as shown in Figure 4-1. This axial displacement included the deformation of the disc as well as the local deformation of the vertebrae.

For rotation, the boundary and loading conditions was simulated according to the *in vitro* experimental study conducted by (Panjabi et al. 1984). A compressive preload of 150N was first applied to the top surface of the vertebral body of L4 in an increasing manner. With the preload maintained, flexion and extension moment, lateral bending moment, and axial rotating moment with a magnitude of 7.5Nm were uniformly applied to the same surface, respectively. The maximum moment was reached in five steps with an increment of 1.5Nm. The method shown in Figure 3-14 was employed to calculate the maximum rotations in sagittal plane, frontal plane and transverse plane. The resulting comparison of maximum rotation

in degree under flexion, extension, lateral bending and axial torsion between the current FE predictions and the experimental results was shown in Figure 4-2.

Figure 4-1: Displacement-force response curve under axial compression. Mean- the mean value of various lumbar segments, CA-cadaver A, CB-cadaver B, CC-cadaver C

Figure 4-2: Comparison of maximum FE rotation for rotatory loads with experimental results (Panjabi et al. 1984). F=flexion, E=extension, RLB=right lateral bending, LLB=left lateral bending, RAT=right axial torsion, LAT=left axial torsion

From Figure 4-1, it can be seen that the FE predicted displacement-force curve appeared to possess the general nonlinear character of the experimental results and were located in the mid-range of the experimental scatter. As observed in *in vitro* experiments, the FE result also predicted a stiffening behavior of the motion segment. The quantitative variation between the FE prediction and the experimental data may be caused by the simplified material properties used in the FE model and the differences between the specimens used in the current project and those used in the experimental studies. Biological materials are naturally non-homogeneous and anisotropic, but in the FE model, the material properties for all important structures were assumed to be homogeneous and isotropic, as other researchers did. In addition, the dimensions of the intervertebral disc were derived from literature rather than from an actual disc.

From Figure 4-2, it can be noted that the predicted ROM under different rotatory loading modalities fell within the standard deviation of the mean experimental data. Moreover, the FE prediction showed a similar stiffness trend as observed in the experiment study: the L4-L5 motion segment was most flexible in flexion and comparatively rigid in axial torsion.

4.2 Validation of L3-S1 model

Validation study is indispensable for a FE model to provide more realistic and reliable predictions of motion segments behavior regarding various pathological changes and load conditions. For validation of L3-S1 model, the boundary and loading conditions used in the cadaveric study conducted by Goel et al. (2005) was replicated. The intact FE model was also subjected to similar loading of 9Nm as the cadaveric testing. The comparison of range of motion at each motion segment between the experimental testing and the FE model were plotted in Figure 4-3. It was found that the FE predicted angular displacements across the segments fell within the standard deviation of the experimental data. The variation between the FE predictions and the experimental results should be mainly caused by the difference in the specimens used, such as geometrical dimensions, degeneration grade, and so on.

(a)

(b)

(c)

Figure 4-3: Comparison of experimental results and current FE predictions in response to 9Nm pure moment. (a) under flexion and extension; (b) under lateral bending; (c) under torsion.

The FE method is well suited for parameter studies and able to deliver additional results, which are difficult or even impossible to acquire experimentally. Therefore, validation is only possible for measurable parameters. Other parameters only indicate trends and do not necessarily represent the absolute values very precisely.

4.3 Results

This section provides the relative rotational motion, facet forces and intervertebral disc stresses for all the four models (a. L3-S1 intact; b. L3-S1 with nucleotomy at L4-L5; c. as b with FlexPLUS; d. as b with rigid rod) in flexion, extension, lateral bending and torsion.

4.3.1 ROM for all the models

The ROM for all the four models (a. L3-S1 intact; b. L3-S1 with nucleotomy at L4-L5; c. as b with FlexPLUS; d. as b with rigid rod) in flexion, extension, right and left lateral bending, and right and left torsion were evaluated. The predicted angular displacements at 9Nm for each model under different loading modalities were plotted in Figures 4-4, 4-5, 4-6, 4-7, 4-8 and 4-9. The comparison of the angular displacements and percentage changes with the intact case at 9Nm were given in Tables 4-1, 4-2 and 4-3.

Flexion/Extension

In flexion, for the nucleotomized model, the increase in motion at L4-L5 level was 17.53% with a corresponding reduction of 7.38% and 7.21% at L3-L4 and L5-S1 respectively. For the model implanted with FlexPLUS, the reduction of motion at L4-L5 was 60.39% with a corresponding increase of 24.18% at L3-L4 and 24.28% at L5-S1. For the rigid rod system, the decrease in motion at the L4-L5 motion

segment in flexion was 67.86% with an increase of 27.05% and 27.4% at L3-L4 and L5-S1 respectively.

In extension, with the removal of the nucleus pulposus, the changes of motion on all levels were minor, which were 1.69% at L3-L4, 4.55% at L4-L5, 1.25% at L5-S1, respectively. For the FlexPLUS model, the motion at L4-L5 was reduced by 50.83% with an increase of 13.14% and 11.72% at L3-L4 and L5-S1 respectively. For the rigid rod system, the motion at L4-L5 was reduced by 59.92% with a corresponding increase of 15.68% at L3-L4 and 13.72% at L5-S1.

Figure 4-4: Comparison of range of motion at all levels of the lumbar spine for all models in 9Nm pure flexion

Figure 4-5: Comparison of range of motion at all levels of the lumbar spine for all models in 9Nm pure extension

Flexion								
	L3-S1		L3-L4		L4-L5		L5-S1	
	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change
Intact	9.67	---	2.44	---	3.08	---	4.16	---
Nucleotomized	9.74	0.72	2.26	-7.36	3.62	17.53	3.86	-7.21
FlexPlus	9.42	-2.59	3.03	24.18	1.22	-60.39	5.17	24.28
Rigid Rod	9.39	-2.90	3.1	27.05	0.99	-67.86	5.3	27.40
Extension								
	L3-S1		L3-L4		L4-L5		L5-S1	
	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change
Intact	8.78	---	2.36	---	2.42	---	4.01	---
Nucleotomized	8.82	0.46	2.32	-1.69	2.53	4.55	3.96	-1.25
FlexPlus	8.34	-5.01	2.67	13.14	1.19	-50.83	4.48	11.72
Rigid Rod	8.26	-5.92	2.73	15.68	0.97	-59.92	4.56	13.72

Table 4-1: The maximum rotations (degrees) and percentage changes compared to the intact case for the nucleotomized model and instrumented models in 9Nm pure moment in flexion and extension. (Negative % change indicates a reduction in the motion)

Lateral Bending

In right lateral bending, with the removal of the nucleus pulposus, the motion at L4-L5 was increased by 18.97% with a decrease of 7.48% at L3-L4 level and 7.65% at L5-S1 level. For the FlexPLUS model, the reduction in motion at L4-L5 was 67.67% with a corresponding increase of 25.2% at L3-L4 and 26.29% at L5-S1. For the rigid rod system, the reduction of motion at L4-L5 was 76.72% with a 28.74% increase at L3-L4 and a 29.08% increase at L5-S1.

In left lateral bending, for the nucleotomized model, the motion at L4-L5 was increased by 19.67% with a decrease of 8.2% at L3-L4 level and 8.25% at L5-S1 level. Implanted with the FlexPLUS, the reduction in motion at L4-L5 was 68.44% with a corresponding increase of 27.87% at L3-L4 and 26.29% at L5-S1 level. For the rigid rod system, the reduction of motion at L4-L5 was 77.46% with a 31.15% increase at L3-L4 and a 29.90% increase at L5-S1.

Figure 4-6: Comparison of range of motion at all levels of the lumbar spine for all models in 9Nm pure right lateral bending

Figure 4-7: Comparison of range of motion at all levels of the lumbar spine for all models in 9Nm pure left lateral bending

Right Lateral Bending								
	L3-S1		L3-L4		L4-L5		L5-S1	
	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change
Intact	6.82	---	2.54	---	2.32	---	1.96	---
Nucleotomized	6.92	1.47	2.35	-7.48	2.76	18.97	1.81	-7.65
FlexPlus	6.4	-6.16	3.18	25.20	0.75	-67.67	2.46	25.51
Rigid Rod	6.34	-7.04	3.27	28.74	0.54	-76.72	2.53	29.08
Left Lateral Bending								
	L3-S1		L3-L4		L4-L5		L5-S1	
	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change
Intact	6.81	---	2.44	---	2.44	---	1.94	---
Nucleotomized	6.93	1.76	2.24	-8.20	2.92	19.67	1.78	-8.25
FlexPlus	6.33	-7.05	3.12	27.87	0.77	-68.44	2.45	26.29
Rigid Rod	6.27	-7.93	3.2	31.15	0.55	-77.46	2.52	29.90

Table 4-2: The maximum rotations (degrees) and percentage changes compared to the intact case for the nucleotomized model and instrumented models in 9Nm pure moment in lateral bending. (Negative % change indicates a reduction in the motion)

Torsion

In right torsion, the removal of the nucleus pulposus increased the motion by 28% at L4-L5 and reduced the motion by 6.62% at L3-L4 and 5.19% at L5-S1. With the

FlexPLUS implanted, the motion at L4-L5 was increased by 18.67% with a minor decrease at the other levels. For the rigid rod model, the motion at L4-L5 was decreased by 32% with an increase of 6.62% at L3-L4 and 4.55% at L5-S1.

In left torsion, for the nucleotomized model, the increase at L4-L5 was 28.48% with a corresponding reduction of 5.97% at L3-L4 and 6.71% at L5-S1. For the FlexPLUS model, the motion at L4-L5 was increased by 17.09% with a minor decrease at the other levels. For the rigid rod model, the motion at L4-L5 was decreased by 36.08% with an increase of 5.97% at L3-L4, and 6.71% at L5-S1.

Figure 4-8: Comparison of range of motion at all levels of the lumbar spine for all models in 9Nm pure right torsion

Figure 4-9: Comparison of range of motion at all levels of the lumbar spine for all models in 9Nm pure left torsion

Right Torsion								
	L3-S1		L3-L4		L4-L5		L5-S1	
	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change
Intact	4.56	---	1.51	---	1.5	---	1.54	---
Nucleotomized	4.79	5.04	1.41	-6.62	1.92	28.00	1.46	-5.19
FlexPlus	4.71	3.29	1.45	-3.97	1.78	18.67	1.48	-3.90
Rigid Rod	4.24	-7.02	1.61	6.62	1.02	-32.00	1.61	4.55
Left Torsion								
	L3-S1		L3-L4		L4-L5		L5-S1	
	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change	Max. Rotation (Degs)	% Change
Intact	4.55	---	1.34	---	1.53	---	1.64	---
Nucleotomized	4.82	5.93	1.26	-5.97	2.03	28.48	1.53	-6.71
FlexPlus	4.72	3.74	1.29	-3.73	1.85	17.09	1.57	-4.27
Rigid Rod	4.18	-8.13	1.42	5.97	1.01	-36.08	1.75	6.71

Table 4-3: The maximum rotations (degrees) and percentage changes compared to the intact case for the nucleotomized model and instrumented models in 9Nm pure moment in torsion. (Negative % change indicates a reduction in the motion)

4.3.2 Facet forces for all the models

The total facet forces were determined for each model subjected to 9Nm pure moment in extension, lateral bending and torsion, and were compared to the intact

case. Figures 4-10, 4-11, 4-12, 4-13 and 4-14, showed the total facet forces for the intact, the nucleotomized and different posterior stabilization systems. The facet forces in flexion were negligible and hence not reported. Tables 4-4, 4-5 and 4-6 showed the total facet forces and the percentage changes compared to the intact case for different stabilization systems.

Extension

With the removal of the nucleus pulposus of L4-L5, the facet forces was increased by 6.91% at L4-L5 with a corresponding decrease of 1.81% and 0.91% at L3-L4 and L5-S1 respectively. Implanted with the FlexPLUS, the facet force reduction at L4-L5 was 80.38% with an increase of 25.7% and 24.06% at L3-L4 and L5-S1 respectively. For the rigid rod system, the facet force reduction at L4-L5 was 84.16% with an increase of 29.86% and 30.97% at L3-L4 and L5-S1 respectively.

Figure 4-10: Total facet forces (N) in response to 9Nm pure moment in extension for all the models

Extension						
	L3-L4		L4-L5		L5-S1	
	Total Facet Force (N)	% Change	Total Facet Force (N)	% Change	Total Facet Force (N)	% Change
Intact	55.13	---	58.78	---	59.05	---
Nucleotomized	54.13	-1.81	62.84	6.91	58.51	-0.91
FlexPlus	69.3	25.70	11.53	-80.38	73.26	24.06
Rigid Rod	71.59	29.86	9.31	-84.16	77.34	30.97

Table 4-4: Total facet forces and the percentage changes compared to the intact case for the nucleotomized and instrumented models in 9Nm pure moment in extension. (Negative % change indicated a reduction in the force.)

Lateral Bending

In right lateral bending, with the removal of the nucleus pulposus the increase of the facet forces at L4-L5 was 50.87% with a decrease of 4.05% and 5.46% at L3-L4 and L5-S1 respectively. For the FlexPLUS system, the facet forces at L4-L5 were reduced by 27.66%, and increased by 26.01% at L3-L4 and 31.51% at L5-S1. For the rigid rod system, the reduction at L4-L5 was 43.26% with an increase of 26.01% and 31.51% at L3-L4 and L5-S1, respectively.

In left lateral bending, for the nucleotomized model, the values were similar to that for right lateral bending, which were 51.49% at L4-L5, 6.06% at L3-L4 and 3.27% at L5-S1 respectively. Instrumented with the FlexPLUS, the facet forces at L4-L5 were reduced by 29.21% with a corresponding increase of 24.25% at L3-L4 and 35.89% at L5-S1. For the rigid rod system, the reduction at L4-L5 was 49.75% with an increase of 32.9% and 37.61% at L3-L4 and L5-S1 respectively.

Figure 4-11: Total facet forces (N) in response to 9Nm pure moment in right lateral bending for all the models

Figure 4-12: Total facet forces (N) in response to 9Nm pure moment in left lateral bending for all the models

Right Lateral Bending						
	L3-L4		L4-L5		L5-S1	
	Total Facet Force (N)	% Change	Total Facet Force (N)	% Change	Total Facet Force (N)	% Change
Intact	16.03	---	13.27	---	24.56	---
Nucleotomized	15.38	-4.05	20.02	50.87	23.22	-5.46
FlexPlus	20.2	26.01	9.6	-27.66	32.3	31.51
Rigid Rod	21.05	31.32	7.53	-43.26	33.12	34.85
Left Lateral Bending						
	L3-L4		L4-L5		L5-S1	
	Total Facet Force (N)	% Change	Total Facet Force (N)	% Change	Total Facet Force (N)	% Change
Intact	15.50	---	12.12	---	23.24	---
Nucleotomized	14.56	-6.06	18.36	51.49	22.48	-3.27
FlexPlus	19.30	24.52	8.58	-29.21	31.58	35.89
Rigid Rod	20.60	32.90	6.09	-49.75	31.98	37.61

Table 4-5: Total facet forces and the percentage changes compared to the intact case for the nucleotomized and instrumented models in 9Nm pure moment in lateral bending. (Negative % change indicated a reduction in the force.)

Torsion

In right torsion, the removal of the nucleus pulposus slightly increased the facet force at L4-L5 with a valued of 1.51% and reduced the facet force at L3-L4 level by 5.85% and that at L5-S1 by 6.34%. For the FlexPLUS system, the facet force reduction at L4-L5 was 9.18% with a reduction of 3.03% at L3-L4 and 3.13% at L5-S1. For the rigid rod system, the facet force reduction at L4-L5 was 43.02% with a corresponding increase of 10.66% at L3-L4 and 9.75% at L5-S1.

In left torsion, for the nucleotomized model, the facet force at L4-L5 was increased by 3.64% with a reduction of 7.58% at L3-L4 and 5.14% at L5-S1. For the model implanted with FlexPLUS, the facet force reduction at L4-L5 was 9.96% with a

reduction of 5.54% at L3-L4 and 2.49% at L5-S1. For the rigid rod model, the facet force reduction at L4-L5 was 49.04% with a 5.31% increase at L3-L4 and a 9.05% increase at L5-S1.

Figure 4-13: Total facet forces (N) in response to 9Nm pure moment in right torsion for all the models

Figure 4-14: Total facet forces (N) in response to 9Nm pure moment in left torsion for all the models

Right Torsion						
	L3-L4		L4-L5		L5-S1	
	Total Facet Force (N)	% Change	Total Facet Force (N)	% Change	Total Facet Force (N)	% Change
Intact	93.92	---	88.98	---	88.21	---
Nucleotomized	88.43	-5.85	90.32	1.51	82.62	-6.34
FlexPlus	91.07	-3.03	80.81	-9.18	85.45	-3.13
Rigid Rod	103.93	10.66	50.7	-43.02	96.81	9.75
Left Torsion						
	L3-L4		L4-L5		L5-S1	
	Total Facet Force (N)	% Change	Total Facet Force (N)	% Change	Total Facet Force (N)	% Change
Intact	89.49	---	88.21	---	98.88	---
Nucleotomized	82.71	-7.58	91.42	3.64	93.8	-5.14
FlexPlus	84.53	-5.54	79.42	-9.96	96.42	-2.49
Rigid Rod	94.24	5.31	44.95	-49.04	107.83	9.05

Table 4-6: Total facet forces and the percentage changes compared to the intact case for the nucleotomized and instrumented models in 9Nm pure moment in torsion. (Negative % change indicated a reduction in the force.)

4.3.3 Stresses in the annulus fibrosus and nucleus pulposus for all the models

The von Mises stresses of the disc for the nucleotomized and instrumented models were compared to that of the intact. The von Mises stress plots for the annulus fibrosus at all levels and the nucleus pulposus at L3-L4 and L5-S1 for all the loading modes were illustrated in Figures 4-15, 4-16, 4-17, 4-18, 4-19, and 4-20. Tables 4-7 to 4-12 reported the maximum von Mises stress values in the annulus fibrosus and nucleus pulposus and their percentage changes with respect to the values for the intact, respectively.

Flexion/Extension

In flexion, with the removal of the nucleus pulposus, the maximum von Mises stress in the annulus fibrosus at L4-L5 was increased by 31.4% with a reduction of

6.15% at L3-L4 and 7.03% at L5-S1; the stress reduction in the nucleus pulposus was 6.81% at L3-L4 and 7.17% at L5-S1. For the FlexPLUS model, the reduction in maximum von Mises stress in the annulus was 18.24% at L4-L5 with an increase of 28.98% and 24.86% at L3-L4 and L5-S1 respectively; in the nucleus pulposus, the maximum von Mises stress was increased by 23.77% at L3-L4 and 25.35% at L5-S1. For the rigid rod system, the maximum von Mises stress in the annulus at L4-L5 was reduced by 33.33% with a corresponding increase of 32.06% at L3-L4 and 27.88% at L5-S1; the stress increase in the nucleus pulposus was 26.67% at L3-L4 and 28.48% at L5-S1.

In extension, for the nucleotomized model, the maximum von Mises stress in the annulus fibrosus at L4-L5 was increased by 7.1% with a minor reduction at L3-L4 and L5-S1; the reduction of the maximum von Mises stress in the nucleus pulposus was 1.32% at L3-L4 and 1.13% at L5-S1. For the FlexPLUS model, the maximum stress in the annulus was reduced by 13.03% at L4-L5 with a corresponding increased of 12.18% and 10.02% at L3-L4 and L5-S1 respectively. For the rigid rod model, the maximum von Mises stress in the annulus at L4-L5 was reduced by 27.23% with a corresponding increase of 13.81% at L3-L4 and 11.69% at L5-S1.

Figure 4-15a: von Mises stress plots in the annulus fibrosus at all levels for all the modes in 9Nm pure moment in flexion

Figure 4-15b: von Mises stress plots in the nucleus pulposus at the adjacent levels for all the modes in 9Nm pure moment in flexion

Figure 4-16a: von Mises stress plots in the annulus fibrosus at all levels for all the modes in 9Nm pure moment in extension

Figure 4-16b: von Mises stress plots in the nucleus pulposus at the adjacent levels for all the modes in 9Nm pure moment in extension

Flexion						
	L3-L4		L4-L5		L5-S1	
	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change
Intact	0.1853	---	0.1815	---	0.212	---
Nucleotomized	0.1739	-6.15	0.2385	31.40	0.1971	-7.03
FlexPlus	0.239	28.98	0.1484	-18.24	0.2647	24.86
Rigid Rod	0.2447	32.06	0.121	-33.33	0.2711	27.88
Extension						
	L3-L4		L4-L5		L5-S1	
	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change
Intact	0.1904	---	0.1803	---	0.2326	---
Nucleotomized	0.1879	-1.31	0.1931	7.10	0.2303	-0.99
FlexPlus	0.2136	12.18	0.1568	-13.03	0.2559	10.02
Rigid Rod	0.2167	13.81	0.1312	-27.23	0.2598	11.69

Table 4-7: Maximum von Mises stresses in the annulus fibrosus (MPa) and percentage changes at all levels with respect to the intact case in 9Nm pure moment in flexion and extension for the nucleotomized and stabilized models. (Negative % indicated a decrease in the stress)

Flexion				
	L3-L4		L5-S1	
	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change
Intact	0.069	---	0.099	---
Nucleotomized	0.0643	-6.81	0.0919	-7.17
FlexPlus	0.0854	23.77	0.1241	25.35
Rigid Rod	0.0874	26.67	0.1272	28.48
Extension				
	L3-L4		L5-S1	
	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change
Intact	0.0683	---	0.0884	---
Nucleotomized	0.0674	-1.32	0.0874	-1.13
FlexPlus	0.0785	14.93	0.0995	12.56
Rigid Rod	0.0801	17.28	0.1011	14.37

Table 4-8: Maximum von Mises stresses in the nucleus pulposus (MPa) and percentage changes at all levels with respect to the intact case in 9Nm pure moment in flexion and extension for the nucleotomized and stabilized models. (Negative % indicated a decrease in the stress)

Lateral Bending

In right lateral bending, with the removal of the nucleus pulposus, the maximum von Mises stress in the annulus at L4-L5 was increase by 7.73% with a reduction of 7.01% at L3-L4 and 7.15% at L5-S1; the stress in the nucleus pulposus was reduced by 7.59% and 6.55% at L3-L4 and L5-S1, respectively. For the model implanted with the FlexPLUS, the reduction of the maximum stress in the annulus was 68.47% with an increase of 29.56% at L3-L4 and 25.36% at L5-S1; the stress in the nucleus pulposus was increased by 25.74% at L3-L4 and 26.2% at L5-S1. For the rigid rod model, the maximum von Mises stress in the annulus at L4-L5 was reduced by 75.34% with a 32.71% increase at L3-L4 and a 28.66% increase at L5-S1; the maximum von Mises stress in the nucleus pulposus was increased by 29.04% and 29.09% at L3-L4 and L5-S1, respectively.

In left lateral bending, with the removal of the nucleus pulposus, the maximum von Mises stress in the annulus at L4-L5 was increase by 15.1% with a reduction of 8.45% at L3-L4 and 7.96% at L5-S1; the stress in the nucleus pulposus was reduced by 8.11% and 7.68% at L3-L4 and L5-S1, respectively. For the model implanted with the FlexPLUS, the reduction of the maximum stress in the annulus was 667.37% with an increase of 33.32% at L3-L4 and 26.53% at L5-S1; the stress in the nucleus pulposus was increased by 29.32% at L3-L4 and 26.56% at L5-S1. For the rigid rod model, the maximum von Mises stress in the annulus at L4-L5 was reduced by 77.04% with a 36.31% increase at L3-L4 and a 29.25% increase at

L5-S1; the maximum von Mises stress in the nucleus pulposus was increased by 32.79% and 29.69% at L3-L4 and L5-S1, respectively.

Figure 4-17a: von Mises stress plots in the annulus fibrosus at all levels for all the modes in 9Nm pure moment in right lateral bending

Figure 4-17b: von Mises stress plots in the nucleus pulposus at the adjacent levels for all the modes in 9Nm pure moment in right lateral bending

Figure 4-18a: von Mises stress plots in the annulus fibrosus at all levels for all the modes in 9Nm pure moment in left lateral bending

Figure 4-18b: von Mises stress plots in the nucleus pulposus at the adjacent levels for all the modes in 9Nm pure moment in left lateral bending

Right Lateral Bending						
	L3-L4		L4-L5		L5-S1	
	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change
Intact	0.2155	---	0.2328	---	0.2603	---
Nucleotomized	0.2004	-7.01	0.2508	7.73	0.2417	-7.15
FlexPlus	0.2792	29.56	0.0734	-68.47	0.3263	25.36
Rigid Rod	0.286	32.71	0.0574	-75.34	0.3349	28.66
Left Lateral Bending						
	L3-L4		L4-L5		L5-S1	
	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change
Intact	0.2272	---	0.2378	---	0.2612	---
Nucleotomized	0.208	-8.45	0.2737	15.10	0.2404	-7.96
FlexPlus	0.3029	33.32	0.0776	-67.37	0.3305	26.53
Rigid Rod	0.3097	36.31	0.0546	-77.04	0.3376	29.25

Table 4-9: Maximum von Mises stresses in the annulus fibrosus (MPa) and percentage changes at all levels with respect to the intact case in 9Nm pure moment in lateral bending for the nucleotomized and stabilized models. (Negative % indicated a decrease in the stress)

Right Lateral Bending				
	L3-L4		L5-S1	
	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change
Intact	0.0909	---	0.0794	---
Nucleotomized	0.084	-7.59	0.0742	-6.55
FlexPlus	0.1143	25.74	0.1002	26.20
Rigid Rod	0.1173	29.04	0.1025	29.09
Left Lateral Bending				
	L3-L4		L5-S1	
	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change
Intact	0.0863	---	0.0768	---
Nucleotomized	0.0793	-8.11	0.0709	-7.68
FlexPlus	0.1116	29.32	0.0972	26.56
Rigid Rod	0.1146	32.79	0.0996	29.69

Table 4-10: Maximum von Mises stresses in the nucleus pulposus (MPa) and percentage changes at all levels with respect to the intact case in 9Nm pure moment in lateral bending for the nucleotomized and stabilized models. (Negative % indicated a decrease in the stress)

Torsion

In right torsion, the removal of the nucleus pulposus increased the maximum von Mises stress in the annulus pulposus at L4-L5 by 52.14% and reduced those at L3-L4 and L5-S1 by 4.99% and 5.96% respectively; the reduction in the maximum von Mises stress in the nucleus pulposus was 5.3% at L3-L4 and 6.64% at L5-S1. For the FlexPLUS model, the maximum von Mises stress in annulus fibrosus at L4-L5 was increased by 22.89 with a minor reduction at the adjacent levels; the reduction in the maximum von Mises in the nucleus pulposus was 2.91% at L3-L4 and 3.9% at L5-S1. For the rigid rod system, the stress in the annulus fibrosus at L4-L5 was reduced by 28.27% with a 4.19% increase at L3-L4 and a 5.22% increase at L5-S1; and the stress in the nucleus pulposus was increased by 2.56% at L3-L4 and 6.2% at L5-S1.

In left torsion, with the removal of the nucleus pulposus, the maximum von Mises stress in the annulus pulposus at L4-L5 was increased by 69.03% with a corresponding reduction of 4.93% at L3-L4 and 6.58% at L5-S1; the reduction in the maximum von Mises stress in the nucleus pulposus was 4.55% at L3-L4 and 5.93% at L5-S1. For the FlexPLUS model, the maximum von Mises stress in annulus fibrosus at L4-L5 was increased by 27.83 with a reduction of 2.89% at L3-L4 and 4.67% at L5-S1; the reduction in the maximum von Mises in the nucleus pulposus was 2.36% at L3-L4 and 3.95% at L5-S1. For the rigid rod system, the stress in the annulus fibrosus at L4-L5 was reduced by 20.4% with a 5.14%

increase at L3-L4 and a 3.71% increase at L5-S1; and the stress in the nucleus pulposus was increased by 6.58% at L3-L4 and 4.71% at L5-S1.

Figure 4-19a: von Mises stress plots in the annulus fibrosus at all levels for all the modes in 9Nm pure moment in right torsion

Figure 4-19b: von Mises stress plots in the nucleus pulposus at the adjacent levels for all the modes in 9Nm pure moment in right torsion

Figure 4-20a: von Mises stress plots in the annulus fibrosus at all levels for all the modes in 9Nm pure moment in left torsion

Figure 4-20b: von Mises stress plots in the nucleus pulposus at the adjacent levels for all the modes in 9Nm pure moment in left torsion

Right Torsion						
	L3-L4		L4-L5		L5-S1	
	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change
Intact	0.1383	---	0.1638	---	0.2031	---
Nucleotomized	0.1314	-4.99	0.2492	52.14	0.191	-5.96
FlexPlus	0.1341	-3.04	0.2013	22.89	0.1953	-3.84
Rigid Rod	0.1441	4.19	0.1175	-28.27	0.2137	5.22
Left Torsion						
	L3-L4		L4-L5		L5-S1	
	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change
Intact	0.1419	---	0.1466	---	0.1777	---
Nucleotomized	0.1349	-4.93	0.2478	69.03	0.166	-6.58
FlexPlus	0.1378	-2.89	0.1874	27.83	0.1694	-4.67
Rigid Rod	0.1492	5.14	0.1167	-20.40	0.1843	3.71

Table 4-11: Maximum von Mises stresses in the annulus fibrosus (MPa) and percentage changes at all levels with respect to the intact case in 9Nm pure moment in torsion for the nucleotomized and stabilized models. (Negative % indicated a decrease in the stress)

Right Torsion				
	L3-L4		L5-S1	
	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change
Intact	0.0585	---	0.0693	---
Nucleotomized	0.0554	-5.30	0.0647	-6.64
FlexPlus	0.0568	-2.91	0.0666	-3.90
Rigid Rod	0.06	2.56	0.0736	6.20
Left Torsion				
	L3-L4		L5-S1	
	Max. Von Mises Stress (MPa)	% Change	Max. Von Mises Stress (MPa)	% Change
Intact	0.0593	---	0.0658	---
Nucleotomized	0.0566	-4.55	0.0619	-5.93
FlexPlus	0.0579	-2.36	0.0632	-3.95
Rigid Rod	0.0632	6.58	0.0689	4.71

Table 4-12: Maximum von Mises stresses in the nucleus pulposus (MPa) and percentage changes at the adjacent levels with respect to the intact case in 9Nm pure moment in torsion for the nucleotomized and stabilized models. (Negative % indicated a decrease in the stress)

4.4 Discussion

In this project, the biomechanical effects of the new dynamic stabilization device (FlexPLUS) on the treated and adjacent motion segments were studied in terms of the intersegmental ROM and the change of load transmission patterns (i.e., facet force changes, and von Mises stress distribution in the intervertebral discs) under various rotatory loading modes (flexion, extension, lateral bending and torsion). For the sake of comparison, the influence of nucleotomy and rigid rod system was also analyzed.

Spinal fusion is the most frequently used conventional surgical treatment for chronic low back pain due to degenerative disc diseases in the lumbar spine. It aims to restrict the motion at the instrumented segment, based on the hypothesis that abnormal range of motion is the cause for discogenic back pain. But many studies (Jackson et al. 1985; Zucherman et al. 1992; Wetzel et al. 1994; Boos et al. 1997) have shown that there was no necessary relation between clinical outcome and the success of the fusion. The pressure profilometry study conducted by McNally et al. (1992) on fifteen patients who were subjected to fusion showed that the discs, which were painful on discography, had particular patterns of stress profile, which led to the conclusion that loading transmission pattern, rather than absolute levels of loading was related to pain in a degenerated disc. Adjacent segment degeneration following lumbar and lumbosacral fusion has also been frequently reported by many researchers (Wiltse et al. 1994; Penta et al. 1995; Lazennec et al. 2000; Kumar et al. 2001a; Kumar et al. 2001b; Guigui et al. 2002).

Therefore, it may be reasonable to accept that the method in achieving relief of back pain is not to stop movement. Creating a normal loading pattern is more important for clinical success, which leads to the appearance of dynamic stabilization. The concept of dynamic stabilization is based on the hypothesis that discogenic back pain is caused by abnormal changes in the loading transmission patterns rather than abnormal motion, and its intension is to alter the load bearing pattern of the motion segment, as well as to control any abnormal motion at the segment.

Range of motion is usually the primary kinematic parameter reported in the biomechanical evaluation of fusion and non-fusion devices due to its relative feasibility to be quantified through vitro experimental studies. In degenerated conditions, postural related abnormal load transfer through the annulus happens causing an alteration in the disc stresses, which might lead to further disc degeneration.

The biomechanical functions of the facet joints in the lumbar spine are guiding segmental motion and sharing a portion of the global mechanical load applied to the spine with the intervertebral discs. The amount of load the facet joints carry varies with the posture. The facet loads bear a considerable load in extension and torsion. Alteration of the facet loads might indicate abnormal loading pattern in the segment. However, there is little information available in literature regarding the load in facet joints during physiologic motion.

The facet joints and the intervertebral disc form a three joint complex and share majority of the load on the spine. Any alteration in the intervertebral disc loading or the facet joint loading might instigate a degenerative cascade. In an instrumented spine the load sharing is altered such that the loads shared by the intervertebral disc and the facet joints are redistributed. Intervertebral disc stress, facet joint loading may give as a suggestion about the loading conditions that occur in the implanted spine. This information is especially relevant to the evaluation of dynamic stabilization devices in the spine, since an implicit goal of many of these systems is to restore the natural load sharing of the operated segment. In this regard, it would be of great interest to characterize and compare the facet joint forces and the disc stresses before and after implantation.

With the removal of the nucleus pulposus at L4-L5 level, the whole ROM of L3-S1 lumbar spine was slightly changed. The ROM at the pathological level was increased and that at the adjacent levels was slightly decreased in all the loading modes. The facet force at L4-L5 after nucleotomy was slightly increased in extension and torsion, while nearly doubled in lateral bending. This result demonstrated that extension and torsion moments were mainly resisted by the facet joints rather than the intervertebral discs. The effects of nucleotomy on the adjacent levels were minor in all the loading modes. The stress contours in the annulus fibrosus and the nucleus pulposus were not changed much at all levels in flexion and extension. In lateral bending and torsion, the disc stress was strengthened in

the load-applied side, that is, in right or left lateral bending and torsion, the disc stress at the right or left side was strengthened respectively. The effect of nucleotomy on the disc stress distribution at the adjacent levels was small. The increase in maximum von Mises stresses in the annulus fibrosus at the injured level was highest at torsion, followed by flexion, extension and lateral bending. At the adjacent levels, the maximum stresses were slightly reduced in the annulus and the nucleus in all the loading modes.

With the insertion of the FlexPLUS, the ROM at the implanted level was dramatically reduced compared to that for the intact in flexion, extension and lateral bending. In torsion, the ROM was less than that for the nucleotomized but still larger than that for the intact. At the adjacent levels, the ROMs were increased in all the loading conditions except for torsion, in which the motion was partially restored but still less than that for the intact. The FlexPLUS system is composed of a wire bundle inside and a plastic cylinder surrounding it. In flexion, the wire bundle acts like ligaments to sustain tensile stress and then restricts the motion. In extension and lateral bending, the cylinder takes an increasing load and restricts the motion. All these contribute to the large reduction of ROM in the loading modes at the instrumented level. The reduction of ROM at the treated level has to be taken over by the adjacent levels to keep a nearly unchanged overall ROM of the L3-S1 spine, which leads to the increase of ROM at the adjacent levels. As a consequence of reduction of ROM, the facet force at the treated segment was in turn significantly reduced especially in extension, in which the fulcrum was moved

from the posterior annulus or the facet joint to the implant. Hence, in extension, there is an unloading of the anterior annulus without compression of the posterior annulus with a corresponding stress increase at the adjacent levels. In lateral bending, the maximum stress at the instrumented level was most strongly reduced, from 0.2328MPa for the intact to 0.0734MPa. The maximum stresses in the annulus fibrosus and the nucleus pulposus at the adjacent levels were in turn increased. In torsion, the stress distribution was partially restored by the FlexPLUS at the treated level with minor effects on disc stress at the adjacent levels.

Fusion devices like the rigid rod system attempt to limit the motion of the instrumented segment. The results from this study indicated that the rigid rod system showed reductions in ROM, facet force and annulus stress at the treated level in all the loading modalities. The effects of the rigid rod system on the ROMs on all segments were more pronounced than that for the FlexPLUS in flexion, extension and lateral bending. In torsion, the rigid rod system greatly reduced the ROM at the treated level and slightly increased those at the adjacent levels in comparison with that for the intact. The reduction of the facet forces at the instrumented level in extension and lateral bending for the rigid rod was only a little more than that for the FlexPLUS. However, in torsion, the rigid rod system decreased the facet force by about 40% more than the FlexPLUS. This implied that the rigid rod is much stiffer than the FlexPLUS in torsion, therefore takes over more load. With the insertion of the rigid rod system, the stress in the annulus fibrosus at the instrumented level was greatly reduced in all the loading modes.

Especially in lateral bending, the annulus fibrosus was almost completely unloaded. The discs at the adjacent levels were correspondingly overloaded in all the loading modes except for torsion, in which, the stress distribution and the maximum stress in annulus fibrosus and the nucleus pulposus were only slightly affected.

From the present results, it can be found that the behavior of the L3-S1 lumbar spine with the FlexPLUS was similar to that with the rigid rod system in all the loading modalities except torsion, although the FlexPLUS can effectively unload the disc and the facet joint in extension and lateral bending and caused a disc stress distribution and a facet force closer to that for the intact model than the rigid rod did in all the loading configurations. The FlexPLUS overloaded the facet joints at the adjacent segments to a certain extent. But it should be noted that the current model only consisted of three segments not the whole lumbar spine, therefore the two adjacent segments instead of all the lumbar segments would have to take over the reduction at the implanted level. If more segments had been analyzed, the redistribution of load transmission may lead to the percentage of reduction at the instrumented level taken over by each adjacent segment lower than the current predictions.

Several other dynamic posterior spinal stabilization devices have been put into clinical application (Mulholland et al. 2002; Sengupta 2005a). The Graf ligament is one of the first used dynamic stabilization devices. Quint et al. (1998) conducted a biomechanical study to determine the stabilizing effect of the flexible Graf system

and rigid device. Their results showed that both the Graf system and rigid instrumentation significantly lowered the ROM following laminectomy at the instrumented level. For the Graf system, the decrease in the ROM was up to 85.7% in flexion, 71% in extension, 62.5% in lateral bending, and 22% in torsion. Higher reductions for the rigid fixator were found with a percentage of 89.5% in flexion, 87.2% in extension, 84.3% in lateral bending and 71.1% in torsion. This indicated that the Graf system was more flexible than the rigid fixator but the difference was small except for torsion.

Schmoelz et al. (2003) studied the degree of stabilization achieved by Dynesys and the effect of stabilization on the adjacent segment by conducting an in vitro experiment in six lumbar cadaver spines. Their results showed that the Dynesys stabilized the spine and was more flexible than an internal fixator. The motion in the adjacent segments was not influenced when pure moments were applied. More recently, the same authors (Schmoelz et al. 2006) investigated the influence of this device on load bearing of a bridged disc on six fresh frozen human lumbar spines and the results showed that the Dynesys device significantly unloaded the disc in extension and lateral bending while restored the load distribution similar to an intact segment in flexion and axial rotation. They also found that the Dynesys did not show substantial differences in intervertebral disc pressure of a bridged disc compared to the rigid fixator. The in vitro study conducted by Niosi et al. (2006) found that at the treated level the ROM with the Dynesys spacer was significantly less than in the intact condition in all directions of loading with an exception of

torsion. The reduction percentage ranged from 67% to 72% in flexion, extension, and lateral bending and 28% in torsion.

By comparing the kinematic effects of the Graf ligament, the Dynesys spacer and the FlexPLUS system on the ROM at the treated level, it can be seen that the results for different implants are comparable, while the stiffness of these devices differ greatly, from a very low value for the Graf ligament to about 2000N/mm for the FlexPLUS system. This implies that the stiffness property of the device might not significantly affect the stabilizing capability of a dynamic stabilization device in resisting motion under pure moments. The difference between various devices might lie in the degree of restoring a natural load transmission pattern. Unfortunately, to the best knowledge of the author, there are no results in terms of the facet joint force and disc stress distribution available for the Graf ligament and the Dynesys device. Therefore, the comparison between effects of different dynamic stabilization devices on the alteration of load transmission pattern can not be made so far.

In addition, Rohlmann et al. (2007) investigated the effects of posterior fixation devices by differentiating the stiffness to simulate dynamic and rigid ones using a FE model of L1-L5 lumbar spine with L3-L4 instrumented in flexion, extension and torsion. They found that the intersegmental rotation at implant level in flexion and extension was reduced and that at the level above slightly increased. The effect

of a posterior implant on ROM in torsion was less pronounced at all the levels. The present predictions were in agreement with their results.

However, in the same study of Rohlmann et al. (2007), the authors also compared the effects of posterior dynamic and rigid implants and relatively small differences were found. Thus, they suggested that the difference in the mechanical effect of a dynamic or rigid implant is smaller than often expected. In fact, similar phenomena were also observed in the current study. Therefore, further studies are needed to address the effectiveness of dynamic stabilization and its advantages over a rigid implant.

Chapter 5: Conclusion and Future Work

5.1 Conclusion

The current project examined the effects of a new dynamic stabilization device (FlexPLUS) on the intersegmental motion of the instrumented and adjacent segments. For the sake of comparison, the effect of a rigid rod system was also investigated. All these were achieved by employing the load-control protocols using a validated FE model of L3-S1 lumbar spine. The effects of the FlexPLUS and the rigid rod system on the range of motion, facet forces and the disc loading patterns were evaluated in various loading configurations, including flexion, extension, lateral bending and torsion.

The results indicated that at the instrumented segment, the FlexPLUS significantly reduced the ROM in flexion, extension and lateral bending; it strongly unloaded the facet joint and the annulus fibrosus in extension, lateral bending and torsion. At the adjacent levels, the ROMs were increased in flexion, extension and lateral bending. The facet joints and the discs were overloaded in extension and lateral bending with the insertion of the FlexPLUS. The effect of the FlexPLUS was minor on the adjacent levels in torsion. Compared to the rigid rod system, the FlexPLUS showed a slight advantage in improving the biomechanical behavior of the degenerated spine in all the loading modes.

In conclusion, the FlexPLUS is more flexible than the rigid system in various rotatory loading modalities but not flexible enough to say that they preserve motion and restore the natural load sharing of the operated segment.

5.2 Limitations of this study

The major limitation of this study is the lack of experimental data for the FlexPLUS system. In addition, although the present FE model is realistic in geometry, several approximations have been introduced in order to improve the computational performance. First of all, current analysis did not take into account the rate-dependent characteristic of the motion segment, which leads to the fact that the present predictions can be only interpreted in terms of static or quasi-static external loading conditions. Secondly, material properties for soft/hard tissues except ligaments were considered to be linear elastic, homogeneous and isotropic. This assumption was supported by the results presented previously, which indicated that a combination of linear material properties and geometric nonlinearity was adequate to characterize the nonlinear and stiffening effect in the mechanical responses of the intervertebral disc. Finally, in the current FE model, the pedicle screws were modeled by assigning a material property of titanium to the corresponding elements; therefore, a perfect bone screw interface was assumed. These simplifications might affect the absolute values but should not change the trends predicted by the current FE model.

5.3 Recommendation for Future work

The current study is restricted to being a FE analysis, therefore undertaking a cadavaric study of FlexPLUS would give us more confidence in the model predictions. Loading sharing characteristics of the FlexPLUS system as a function of variation in its location with regard to the vertebrae would be helpful in assessing the optimal surgical placement of the system. A FE study with the FlexPLUS adjacent to fusion or combined with other non-fusion technologies like the disc nucleus replacements may be helpful in understanding various biomechanical parameters and the load sharing characteristics of the dynamic stabilization systems with a prosthetic disc nucleus.

Further detail analysis of the changes in motion and load transmission between adjacent levels together with clinical follow-up of patients treated with stabilization devices may provide insights and information to explain the post augmentation of any further spinal deformity in terms of mechanical changes or natural properties of the underlying disease or ageing process.

Reference

- Abumi, K., Panjabi, M. M., et al. (1990). "Biomechanical evaluation of lumbar spinal stability after graded facetectomies." Spine 15(11): 1142-1147.
- Adams, M. A., Dolan, P., et al. (1990). "Diurnal changes in spinal mechanics and their clinical significance." J Bone Joint Surg Br 72: 266-270.
- Adams, M. A., McNally, D. S., et al. (1996). "'Stress' distributions inside intervertebral discs." J Bone Joint Surg Br 78-B: 965-972.
- Adams, M. A. and Roughley, P. J. (2006). "What is intervertebral disc degeneration, and what causes it." Spine 31(18): 2151-2161.
- Ahmed, A. M., Duncan, N. A., et al. (1990). "The effect of facet geometry on the axial torque-rotation response of lumbar motion segments." Spine 15(5): 391-401.
- Antoniou, J., Thomas, S., et al. (1996). "The human lumbar intervertebral disc. Evidence for changes in the biosynthesis and denaturation of the extracellular matrix with growth, maturation, ageing and degeneration." J Clin Invest 98(4): 996-1003.
- Ashman, R. B., Cowin, S. C., et al. (1984). "A continuous wave technique for the measurement of the elastic properties of the cortical bone." J Biomech 17: 349-361.
- Askar, Z., Wardlaw, D., et al. (2004). "Correlation between inter-vertebral disc morphology and the results in patients undergoing Graf ligament stabilisation." Eur Spine J 13: 714-718.
- Belytschko, T., Kulak, R. F., et al. (1974). "Finite element stress analysis of an intervertebral disc." J Biomech 7: 277-285.
- Bibby, S. R. S., Yu, J., et al. (2004). The physiology of intervertebral disc degeneration. In Degenerative Disc Disease (Eds. Gunzburg R, Szpalski M and Andersson GBJ), Lippincott Williams & Wilkins, Philadelphia, PA, USA.

- Boden, S. D., Davis, D. O., et al. (1990). "Abnormal magnetic-resonance scans of the lumbar spine in asymptomatic subjects-a prospective investigation." J Bone Joint Surg Am 72: 403-408.
- Boos, N. and Webb, J. K. (1997). "Pedicule screw fixation in spinal disorders: a European view." Eur Spine J 6: 2-18.
- Boos, N., Weissbach, S., et al. (2002). "Classification of age-related changes in lumbar intervertebral discs." Spine 27: 2631-2644.
- Brethbühler, D., Markwalder, T. H. M., et al. (1998). "Surgical results after soft system stabilization of the lumbar spine in degenerative disc disease- long term results." Acta Neurochir (Wien) 140: 521-525.
- Brinckmann, P. and Grootenboer, H. (1991). "Change of disc height, radial disc bulge, and intradiscal pressure from discectomy: an in vitro investigation on human lumbar discs." Spine 16: 641-646.
- Brown, T., Hansen, R. J., et al. (1957). "Some mechanical tests on the lumbosacral spine with particular reference to the intervertebral discs." J Bone Joint Surg 39-A(5): 1135-1164.
- Buckwalter, J. A. (1995). "Aging and degeneration of the human intervertebral disc." Spine 20: 1307-1314.
- Carter, D. R. and Hayes, W. C. (1977). "The compressive behaviour of bone as a two phase porous structure." J Bone Joint Surg 59A: 954-62.
- Carter, D. R. and Spengler, D. M. (1978). "Mechanical properties and composition of cortical bone." Clin Orthop Rel Res 135: 192-217.
- Chazal, J., Tanguy, A., et al. (1985). "Biomechanical properties of spinal ligaments and a histological study of the supraspinal ligament in traction." J Biomech 18(3): 167-176.
- Chen, C. S., Cheng, C. K., et al. (2002). "A biomechanical comparison of posterolateral fusion and posterior fusion in the lumbar spine." J Spinal Disord Tech 15(1): 53-63.
- Chen, C. S., Cheng, C. K., et al. (2001). "Stress analysis of the disc adjacent to interbody fusion in lumbar spine." Med Eng Phys 23: 483-491.

- Choi, K., Kuhn, J. L., et al. (1990). "The elastic moduli of human subchondral, trabecular, and cortical bone tissue and the size-dependency of cortical bone modulus." J Biomech 23(11): 1103-1113.
- Crouch, J. E. (1985). Functional Human Anatomy (4th edition), Lea &Febiger.
- Dekutoski, M. B., Schendel, M. J., et al. (1994). "Comparison of In vivo and in vitro adjacent segment motion after lumbar fusion." Spine 19(15): 1745-7451.
- Dunlop, R. B., Adams, M. A., et al. (1984). "Disc space narrowing and the lumbar facet joints." J Bone Joint Surg 66-b(5): 706-710.
- Eberlein, R., Holzapfel, G. A., et al. (2002). Assessment of a spinal implant by means of advanced FE modeling of intact human intervertebral discs. Fifth World Congress on Computational Mechanics, Vienna, Austria.
- Fagan, M. J., Julian, S., et al. (2002). "Finite element analysis in spine research." Proc Inst Mech Eng [H] 216(5): 281-298.
- Freudiger, S., Dubois, G., et al. (1999). "Dynamic neutralisation of the lumbar spine confirmed on a new lumbar spine simulator in vitro." Arch Orthop Trauma Surg 119: 127-132.
- Frobin, W., Brinckmann, P., et al. (2001). "Height of lumbar discs measured from radiographs compared with degeneration and height classified from MR images." Eur Radiol 11(2): 263-269.
- Galbraith, P. C. and Bryant, J. T. (1989). "Effects of grid dimensions on finite element modles of articular surface." J Biomech 22: 385-93.
- Gardner, A. and Pande, K. C. (2002). "Graf ligamentoplasty: a 7-year follow-up." Eur Spine J 11 (Suppl. 2): S157-S163.
- Gilad, I. and Nissan, M. (1986). "A Study of Vertebra and Disc Geometric Relations of the Human Cervical and Lumbar Spine." Spine 11(2): 154-157.
- Gilbertson, L. G., Goel, V. K., et al. (1995). "Finite element methods in spine biomechanics research." Crit Rev Biomed Eng 23(5-6): 411-473.
- Goel, V. K., Grauer, J. N., et al. (2005). "Effects of Charité Artificial Disc on the Implanted and Adjacent Spinal Segments Mechanics Using a Hybrid Testing Protocol." Spine 30(24): 2755-2764.

- Goel, V. K., Kim, Y. E., et al. (1988). "An analytical investigation of the mechanics of spinal instrumentation." Spine 13: 1003-1011.
- Goel, V. K., Monroe, B. T., et al. (1995). "Interlaminar shear stresses and laminae separation in a disc." Spine 20(6): 689-698.
- Goel, V. K. and Weinstein, J. N. (1990). Biomechanics of the spine:clinical and surgical perspective, Boca Raton, FL: CRC Press Inc..
- Goldstein, S. A. (1987). "The mechanical properties of trabecular bone: dependence of anatomic location and function." J Biomech 20: 1055-1061.
- Goto, K., Tajima, N., et al. (2002). "Mechanical analysis of the lumbar vertebrae in a three-dimensional finite element method model in which intradiscal pressure in the nucleus pulposus was used to establish the model." J Orthop Res 7: 243-246.
- Grevitt, M. P., Gardner, A. D. H., et al. (1995). "The Graf stabilisation system: early results in 50 patients." Eur Spine J 4: 169-175.
- Grob, D., Benini, A., et al. (2005). "Clinical experience with the Dynesys semirigid fixation system for the lumbar spine." Spine 30(3): 324-331.
- Gu, W. Y., Mao, X. G., et al. (1999). "The anisotropic hydraulic permeability of human lumbar annulus fibrosus. Influence of age, degeneration, direction, and water content." Spine 24: 2449-2455.
- Guigui, P., Wodecki, P., et al. (2002). "Long-term influence of associated arthrodesis on adjacent segments in the treatment of lumbar stenosis: a series of 127 cases with 9-year follow-up." J Bone Joint Surg Br 84(Supp. I, 23).
- Guo, L. X. and Teo, E. C. (2005). "Prediction of the modal characteristics of the human spine at resonant frequency using finite element models." Proc Inst Mech Eng [H] 219: 277-283.
- Haer, T. R., O'Brien, M., et al. (1994). "The role of the lumbar facet joints in spinal stability." Spine 19(23): 2667-2671.
- Hansson, T. H., Keller, T. S., et al. (1987). "A study of the compressive properties of lumbar vertebral trabeculae: effects of tissue characteristics." Spine 12(1): 56-62.

- Hashimoto, T., Oha, F., et al. (2001). "Mid-term clinical results of Graf stabilization for lumbar degenerative pathologies:a minimum 2-year follow-up." Spine J 1(4): 283-289.
- Haughton, V. M., Rogers, B., et al. (2002). "Measuring the axial rotation of lumbar vertebrae in vivo with MR imaging." AJNR Am J Neuroradiol 23: 1110-1116.
- Haut, R. C. and Little, R. W. (1972). "A constitutive equation for collagen fibres." J Biomech 5: 423-430.
- Hirsch, C. and Sonnerup, L. (1968). "Macroscopic rheology in collagen material." J Biomech 1: 16-18.
- Iatridis, J. C. and Gwynn, I. (2004). "Mechanisms for mechanical damage in the intervertebral disc annulus fibrosus." J Biomech 37: 1165-1175.
- Iatridis, J. C., Setton, L. A., et al. (1998). "Degeneration affects the anisotropic and nonlinear behaviors of human annulus fibrosus in compression." J Biomech 31: 535-544.
- Inoue, H., Ohmori, K., et al. (1999). "Radiographic evaluation of the lumbosacral disc height." Skeletal Radiol 28: 638-643.
- Jackson, R. K., Boston, A., et al. (1985). "Lateral mass fusion: a prospective study of a consecutive series with long-term follow-up." Spine 10: 828-832.
- Kabel, J., Rietbergen, B. v., et al. (1999). "The role of an effective isotropic tissue modulus in the elastic properties of cancellous bone." J Biomech 32: 673-680.
- Kanayama, M., Hashimoto, T., et al. (2005). "Non-fusion surgery for degenerative spondylolisthesis using artificial ligament stabilization." Spine 30(5): 588-592.
- Kempson, G. E., Spivey, C. J., et al. (1971). "Patterns of cartilage stiffness on normal and degenerative human femoral heads." J Biomech 4: 597-609.
- Kulak, R. F., Belytschko, T. B., et al. (1976). "Nonlinear Behavior of the Human Intervertebral Disc under Axial Load." J Biomech 9: 377-386.

- Kumar, M. N., Baklanov, A., et al. (2001a). "Correlation between sagittal plane changes and adjacent segment degeneration following lumbar spine fusion." Eur Spine J 10: 314-319.
- Kumar, M. N., Jacquot, F., et al. (2001b). "Long-term follow-up of functional outcomes and radiographic changes at adjacent levels following lumbar spine fusion for degenerative disc disease." Eur Spine J 10: 309-313.
- Lavaste, F., Skalli, W., et al. (1992). "Three-dimensional geometrical and mechanical modelling of the lumbar spine." J Biomech 25(10): 1153-1164.
- Lazennec, J. Y., Ramare, S., et al. (2000). "Sagittal alignment in lumbosacral fusion: relations between radiological parameters and pain." Eur Spine J 9: 47-55.
- Lee, K. K., E.C.Teo, et al. (2004). "Effect of facetectomy on lumbar spinal stability under sagittal plane loadings." Spine 29(15): 1624-1631.
- Lee, K. K., Teo, E. C., et al. (2003). "Finite element modeling of L2-L3 using digitizer." Int. J. Comput. Applicat. Technol. 20: 1-9.
- Lindsey, D. P., Swanson, K. E., et al. (2003). "The effect of an interspinous implant on the kinematics of the instrumented and adjacent levels in the lumbar spine." Spine 28: 2192-2197.
- Lorenz, M., Patwardhan, A., et al. (1983). "Loading-bearing characteristics of lumbar facets in normal and surgically altered spinal segments." Spine 8(2): 122-130.
- Lu, Y. M., Hutton, W. C., et al. (1996). "Can variations in intervertebral disc height affect the mechanical function of the disc?" Spine 21(19): 2208-2216.
- Lyons, G., Eisenstein, S. M., et al. (1981). "Biochemical changes in intervertebral disc degeneration." Biochim Biophys Acta 673(4): 443-453.
- Maniadakis, N. and Gray, A. (2000). "The economic burden of back pain in the UK." Pain 84: 95-103.
- Marchand, F. and Ahmed, A. M. (1990). "Investigation of the laminae structure of lumbar disc annulus fibrosus." Spine 15: 402-410.
- Marieb, E. N. (2004). Human Anatomy & Physiology (6th edition), Pearson Education Inc. Daryl Fox.

- Markolf, K. L. (1972). "Deformation of the thoracolumbar intervertebral joints in response to external loads-A biomechanical study using autopsy material." J Bone Joint Surg 54-A(3): 511-533.
- Markolf, K. L. and Morris, J. M. (1974). "The structural components of the intervertebral disc." J Bone Joint Surg 56-A(4): 675-687.
- McNally, D. S. and Adams, M. A. (1992). "Internal intervertebral disc mechanics as revealed by stress profilometry." Spine 17: 66-73.
- McNally, D. S., Shackelford, I. M., et al. (1996). "In vivo stress measurement can predict pain on discography." Spine 21: 2580-2587.
- Miller, J., Schmatz, C., et al. (1988). "Lumbar disc degeneration: Correlation with age, sex, and spine level in 600 autopsy specimens." Spine 13: 173-178.
- Mimura, M., Panjabi, M. M., et al. (1994). "Disc degeneration affects the multidirectional flexibility of the lumbar spine." Spine 19: 1371-1380.
- Mulholland, R. C. and Sengupta, D. K. (2002). "Rationale, principles and experimental evaluation of the concept of soft stabilization." Eur Spine J 11 (Suppl. 2): S198-205.
- Niosi, C. A., Zhu, Q. A., et al. (2006). "Biomechanical characterization of the three-dimensional kinematic behavior of the Dynesys dynamic stabilization system: an in vitro study." Eur Spine J 15: 913-922.
- Noren, R., Trafimow, J., et al. (1991). "The role of facet joint tropism and facet angle in disc degeneration." Spine 16(5): 530-532.
- Osti, O. L., Vernon-roberts, B., et al. (1992). "Annular tears and disc degeneration in the lumbar spine." J Bone Joint Surg Am 74-B: 678-682.
- Panjabi, M. M., Krag, M. H., et al. (1984). "Effects of disc injury on mechanical behavior of the human spine." Spine 9(7): 707-713.
- Panjabi, M. M. and White, A. A. (1990). Clinical Biomechanics of the Spine (2nd edition), Lippincott-Raven.
- Park, P., Garton, H. J., et al. (2004). "Adjacent segment disease after lumbar or lumbosacral fusion:review of the literature." Spine 29(17): 1938-1944.

- Passuti, N., Delecrin, J., et al. (2004). Posterolateral fusion. In Degenerative Disc Disease (Eds. Gunzburg R, Szpalski M and Andersson GBJ), Lippincott Williams & Wilkins, Philadelphia, PA, USA.
- Penta, M., Sandhu, A., et al. (1995). "Magnetic resonance imaging assessment of disc degeneration 10 years after anterior lumbar interbody fusion." Spine 20: 743-747.
- Pintar, F. A., Yoganandan, N., et al. (1992). "Biomechanical properties of human lumbar spine ligaments." J Biomech 25(11): 1351-1356.
- Quint, U., Wilke, H. J., et al. (1998). "Laminectomy and functional impairment of the lumbar spine: the importance of muscle forces in flexible and rigid instrumented stabilization-a biomechanical study in vitro." Eur Spine J 7: 229-238.
- Rao, A. A. and Dumas, G. A. (1991). "Influence of material properties on the mechanical behaviour of the L5-S1 intervertebral disc in compression: a nonlinear finite element study." J Biomed Eng 13: 139-151.
- Reilly, D. T. and Burstein, A. H. (1974). "The elastic modulus for bone." J Biomech 7: 271-275.
- Reilly, D. T. and Burstein, A. H. (1975). "The elastic and ultimate properties of compact bone tissue." J Biomech 8: 393-405.
- Reuber, M., Schultz, A. B., et al. (1982). "Bulging of lumbar intervertebral disks." J Biomech Eng 104: 187-192.
- Richards, J. C., Majumdar, S., et al. (2005). "The treatment mechanism of an interspinous process implant for lumbar neurogenic intermittent claudication." Spine 30(7): 744-749.
- Rigby, M. C., Selmon, G. P. F., et al. (2001). "Graf ligament stabilisation: mid- to long-term follow-up." Eur Spine J 10: 234-236.
- Robert, J. M. (2004). Morphologic changes of endplates in degenerative disc disease. In Degenerative Disc Disease (Eds. Gunzburg R, Szpalski M and Andersson GBJ), Lippincott Williams & Wilkins, Philadelphia, PA, USA.

- Rohlmann, A., Burra, N. K., et al. (2007). "Comparison of the effects of bilateral posterior dynamic and rigid fixation devices on the loads in the lumbar spine: a finite element analysis." Eur Spine J (Epub ahead of print).
- Sanjeevi, R., Somanathan, N., et al. (1982). "A viscoelastic model for collagen fibres." J Biomech 15: 181-183.
- Sato, K., Kikuchi, S., et al. (1999). "In vitro intradiscal pressure measurement in healthy individuals and in patients with ongoing back problems." Spine 24(23): 2468-2474.
- Schmake, K. J., Scharen, S., et al. (2003). "Dynamic stabilization for spinal stenosis with degenerative spondylolisthesis of the lumbar spine: a prospective clinical study." Spine J 3: 67S-171S.
- Schmoelz, W., Huber, J. F., et al. (2003). "dynamic stabilization of the lumbar spine and its effects on adjacent segments: an in vitro experiment " J Spinal Disord Tech 16: 418-423.
- Schmoelz, W., Huber, J. F., et al. (2006). "Influence of a dynamic stabilisation system on load bearing of a bridged disc: an invitro study of intradiscal pressure." Eur Spine J 15(8): 1276-1285.
- Schultz, A. B., Warwick, D. N., et al. (1979). "Mechanical Properties of Human Lumbar Spine Motion Segments - Part I: Responses in Flexion, Extension, Lateral Bending, and Torsion." J Biomech Eng 101: 46-52.
- Sénégas, J. (2002). "Mechanical supplementation by non-rigid fixation in degenerative intervertebral lumbar segments: the Wallis system." Eur Spine J 11(Suppl. 2): S164-S169.
- Sengupta, D. (2005a). "Dynamic stabilization device in the treatment of low back pain." Neurol India 53(4): 466-474.
- Sengupta, D. K. and Mulholland, R. C. (2005b). "Fulcrum assisted soft stabilization system: a new concept in the surgical treatment of degenerative low back pain." Spine 30(9): 1019-1029.
- Sharma, M., Langrana, N. A., et al. (1995). "Role of ligaments and facets in lumbar spinal stability." Spine 20: 887-900.

- Shirazi-Adl, A., Ahmed, A. M., et al. (1986). "Mechanical response of a lumbar motion segment in axial torque alone and combined with compression." Spine 11(9): 914-927.
- Shirazi-Adl, A., Shrivastava, S. C., et al. (1984). "Stress analysis of the lumbar disc-body unit in compression: a three-dimensional nonlinear finite element study." Spine 9(2): 120-134.
- Shirazi-Adl, S. A., Shrivastava, S. C., et al. (1984). "Stress analysis of the lumbar disc-body unit in compression." Spine 9(2): 120-134.
- Stoll, T., Dubois, G., et al. (2002). "The dynamic neutralization system for the spine: a multi-center study of a novel non-fusion system." Eur Spine J 11 (Suppl. 2): S170-S178.
- Swanson, K. E., Lindsey, D. P., et al. (2003). "The effect of interspinous implant on intervertebral disc pressures." Spine 28: 26-32.
- Tan, S. H., E.C.Teo, et al. (2002). "Quantitative three-dimensional anatomy of lumbar vertebrae in singaporean asians." Eur Spine J 11: 152-158.
- Teo, E. C. and Ng, H. W. (2001). "First cervical vertebra (atlas) fracture mechanism studies using finite element method." J Biomech 34: 13-21.
- Teo, E. C., Paul, J. P., et al. (1994). "Finite element stress analysis of a cadaver second cervical vertebra." Med Biol Eng Comput 32(2): 236-238.
- Thibodeau, G. A. and Paton, K. T. (2004). Structure & Function of the Body, Robin Carter.
- Totoribe, K., Tajima, N., et al. (1999). "A biomechanical study of posterolateral lumbar fusion using a three-dimensional nonlinear finite element method." J Orthop Sci 4: 115-126.
- Tsuji, H., Hirano, N., et al. (1993). "Structural variation of the anterior and posterior annulus fibrosus in the development of human lumbar intervertebral disc. A risk factor for intervertebral disc rupture." Spine 18: 204-210.
- Ueno, K. and Liu, Y. K. (1987). "A three-dimensional nonlinear finite element model of lumbar intervertebral joint in torsion." J Biomech Eng 109: 200-209.

- Urban, J. P. G. and Roberts, S. (2003). "Degeneration of the intervertebral disc." Arthritis Res Ther 5(3): 120-130.
- van-Tulder, W. M., Koes, B. W., et al. (1995). "A cost-of-illness study of back pain in the Netherlands." Pain 62: 233-240.
- Virgin, W. J. (1951). "Experimental investigations into the physical properties of the intervertebral disc." J Bone Joint Surg 33B(4): 607-611.
- Wetzel, F. T., LaRocca, S. H., et al. (1994). "The treatment of lumbar spinal pain syndromes diagnosed by discography. lumbar arthrodesis." Spine 19(7): 792-800.
- White, A. A. (1990). Your aching back (the revised, updated edition), Simon & Schuster/Fireside.
- Wiltse, L. L. and Hambly, M. F. (1994). Degenerative changes in the first two segments above a lumbosacral fusion: A 22.6 year(average) follow-up. "Instrumented spinal fusion"., Georg Thieme Verlag Stuttgart, New York; Thieme Medical Publishers, Inc., New York.
- Wiseman, C. M., Lindsey, D. P., et al. (2005). "The effect of an interspinous process implant on facet loading during extension." Spine 30(8): 903-907.
- Yamamoto, I., Panjabi, M. M., et al. (1989). "Three-Dimensional Movements of the Whole Lumbar Spine and Lumbosacral Joint." Spine 14(11): 1256-1261.
- Yoon, H. S. and Katz, J. L. (1976). "Ultrasonic wave propagation in human cortical bone-II. Measurement of elastic properties and microhardness." J Biomech 9: 459-464.
- Yu, J., Peter, C., et al. (2002). "Elastic fibre organization in the intervertebral discs of the bovine tail." J Anat 201(6): 465-475.
- Zucherman, J., Hsu, K., et al. (1992). "Clinical efficacy of spinal instrumentation in lumbar degenerative disc disease." Spine 17(7): 834-837.
- Zucherman, J. F., Hsu, K. Y., et al. (2005). "A multicenter, prospective, randomized trial evaluating the X STOP interspinous process decompression system for the treatment of neurogenic intermittent claudication: two-year follow-up results." Spine 30(12): 1351-1358.

Reference

Zucherman, J. F., Hsu, K. Y., et al. (2004). "A prospective randomized multi-center study for the treatment of lumbar spinal stenosis with the X-STOP interspinous implant: 1-year results." Eur Spine J 13: 22-31.

Appendix A: Human Lumbar Spine

A.1 Anatomical Terms and Definitions

A.1.1 Anatomical position

In order to keep the consistency of description and prevent misunderstanding, this section presents universally accepted anatomical terms which are frequently used in this report. To discuss the body posture and the relationship of one area to another, a standard position, called the **anatomical position**, is adopted as reference. The anatomical position is such a reference position in which the body is in an erect or standing posture with the arms at the sides and palms turned forward, as shown in Figure A-1.

Figure A-1: Anatomical position (Thibodeau et al. 2004)

A.1.2 Directions and planes of the body

The commonly used directional terms are defined in table A-1 and illustrated in Figure A-2.

Term	Definition
Superior (cranial)	Toward the head end or upper part of a structure or the body; above
Inferior (caudal)	Away from the head end or toward the lower part of a structure or the body; below
Anterior	Toward or at the front of the body; in front of
Posterior	Toward or at the back of the body; behind
Medial	Toward or at the midline of the body; on the inner side of
Lateral	Away from the midline of the body; on the outer side of
Proximal	Closer to the origin of the body part or the point of attachment of a limb to the body trunk
Distal	Farther from the origin of a body part or the point of attachment of a limb to the body trunk

Table A-1: Commonly used directional terms (modified from Marieb, 2004)

To facilitate the study of the body as a whole, three types of planes are most frequently used, which intersect at right angles to each other. As shown in Figure

A-2, a **sagittal plane** divides the body into right and left sides. The special sagittal plane which divides the body into two equal halves is called a **mid-sagittal plane**. A **frontal plane** (also called as **coronal plane**) divides the body into anterior and posterior parts, perpendicular to sagittal planes. Both sagittal and frontal planes are vertical planes. However, a **Transverse plane** is a horizontal plane at right angles to sagittal and frontal planes, dividing the body into superior and inferior parts.

Figure A-2: Directions and planes of the body (Thibodeau et al., 2004)

A.2 Vertebral Column

A.2.1 General Characteristics

The vertebral column, also known as the spinal column, extends from the skull to the pelvis and is composed of 33 individual bones termed vertebrae, between

which are fibrocartilaginous discs (called intervertebral discs). The spinal column lies in the mid-dorsal region and serves as the chief axial support of the body and is the key of the posture of the trunk (Crouch 1985). It also surrounds and protects the delicate spinal cord and provides attachment points for the ribs and for muscles of the back and neck (Marieb, 2004).

A.2.2 Divisions and Curvatures

Depending on the regions where they are located, these 33 vertebrae are divided into five categories. The first seven vertebrae located at the top of the spinal column are the **cervical vertebrae**; the next 12 are the **thoracic vertebrae**; following the thoracic vertebrae are the five **lumbar vertebrae** which support the lower back. Below the lumbar vertebrae are the five **sacral vertebrae**; the bottom of the spinal column are the four **coccygeal vertebrae**. The five sacral vertebrae and the four coccygeal vertebrae fuse together to form sacrum and coccyx, respectively.

When looked at from the side, the spinal column forms four curvatures that give it its sinusoid shape. The cervical and lumbar curvatures are concave posteriorly (lordosis), while the thoracic and sacral curvatures are convex posteriorly (kyphosis), as illustrated in Figure A-3. The spinal column exhibits such a curved structure that it has enough strength to support the weight of the rest of the body and helps to maintain the balance of the upper body. These curves increase the resilience and flexibility of the spine as well, enhancing the shock-absorbing capacity of the spinal column.

Figure A-3: The vertebral column (Marieb, 2004)

A.3 Lumbar Vertebrae

The lumbar region of the vertebral column is the third major region of the spine, containing five individual vertebrae. From top to bottom, the five vertebrae are subsequently numbered as L1, L2, L3, L4 and L5. A typical lumbar vertebra is illustrated in Figure A-4. The lumbar vertebra consists of an anterior body and a posterior vertebral arch. These two parts form the wall of the vertebral foramen.

Figure A-4: The structure of a typical lumbar vertebra (modified from Marieb, 2004)

The thickest part of the vertebra is the anterior body, which is massive and kidney-shaped. The body is wider from side to side than from anterior to posterior, and a little thicker in front than behind. The superior and inferior surfaces of the body are flattened and rough, giving attachment to the intervertebral discs. In its anterior surface, the body has a few small foramina for passage of nutrient vessels. In the central region of its posterior surface is located one or more larger irregular holes for the passage of the basivertebral veins from the body of the vertebra.

The posterior vertebral arch is composed of two pedicles and two laminae. The two pedicles are thick, short structures that attach to the posterolateral sides of the body. The laminae are broad, short and strong, extending posteriorly and medially from the ends of the pedicles to join on the midline. Projecting posteriorly and inferiorly from the point of junction of the laminae is the spinous process. The transverse process extends laterally from where the lamina and pedicle meet on each side. Each vertebra presents four articular processes, two superior and two inferior,

which also arise from the point of junction of the pedicles and laminae. Their articular surfaces are covered with a layer of hyaline cartilage known as the articular facet. The spinous and transverse processes provide attachment points for muscles and ligaments, and the articular processes serve for the joining of adjacent vertebrae.

The facets of the lumbar region are not planes, but have significantly curved mating surfaces; the inferior facets are convex, while the superior facets are concave.

Compared to the vertebrae in other spinal regions, the five lumbar vertebrae possess following distinguishing characteristics (Marieb, 2004):

- a. The pedicles and laminae are shorter and thicker than those of other vertebrae;
- b. The spinous processes are short, flat and hatchet shaped and are easily seen when a person bends forward. These processes are robust and project directly backward, adaptations for the attachment of the large back muscles;
- c. The vertebral foramen is triangular;
- d. The orientation of the facets of the articular processes of the lumbar vertebrae differs substantially from that of the other vertebra types. These modifications lock the lumbar vertebrae together and provide stability by preventing rotation of the lumbar spine.

The sturdier structure of the lumbar vertebrae reflects their enhanced weight-bearing function.

A.4 Intervertebral Disc

Intervertebral discs are found between each two adjacent vertebrae and account for about 25% of the vertebral column height. The discs act as shock-absorbers for the spine and allow the spine to bend and twist. It is also the major compression-carrying component of the spine. Each disc is a cushionlike pad consisting of three distinct parts: the annulus fibrosus, the nucleus pulposus and the vertebral endplates. Figure A-5 shows the basic structure of an intervertebral disc.

Figure A-5: (A) An intervertebral disc consists of a nucleus pulposus surrounded by concentric laminates of the annulus. (B) The fibers are oriented at about $\pm 30^\circ$ with respect to the horizontal plane.

Like most biological structures, intervertebral discs exhibit nonlinear mechanical property, which means that they become stiffer as they are more deformed. This nonlinearity can be due to the way collagen fibers stretch. Nonlinearity enables the disc to deform considerably under low loads without collapsing completely under sudden high loading. This property is also responsible for the disc to act as a shock absorber.

A.4.1 Nucleus Pulposus

The nucleus pulposus is the water rich gelatinous center of the disc, which generates hydrostatic pressure under applied external loads. The water content of the nucleus pulposus ranges from 70-90% (Panjabi et al. 1990) and tends to decrease with age. Being a fluid, the nucleus can be deformed under pressure without a reduction in volume. 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 lumbar nucleus fills 30-50% of the total disc area in cross-section. It is usually more posterior than central and lies at about the juncture of the middle and posterior thirds of the sagittal diameter (Panjabi et al., 1990).

A.4.2 Annulus Fibrosus

The annulus fibrosus forms the outer boundary of the disc, surrounding the nucleus pulposus. It is a composite structure with collagen fibers embedded in the concentric laminated bands, as shown in Figure 2-5. The fibers are arranged in a helicoid manner, running in about the same direction in a given laminate but in opposite direction in any two adjacent bands. The annulus fibrosus keeps the pressurized nucleus from exploding outward and holds the successive vertebrae together.

A.4.3 Vertebral End-plates

The vertebral end-plate is composed of hyaline cartilage that separates the other two components of the disc from the vertebral bodies. Starting with an active

growth cartilage, the age changes result in irregularly arranged growth cartilage, which disappears with time and is replaced by bone (Panjabi et al., 1990).

A.5 Spinal Ligaments

The spinal ligaments are uniaxial structures, like rubber bands, effective in carrying loads along the direction in which the fibers run. They readily resist tensile forces but buckle when subjected to compression. The individual ligaments provide tensile resistance to external loads by developing tension when the motion segment is subjected to different complex force and torque vectors. The ligaments are made up of densely packed collagen fibers and help to provide structural stability.

There are seven spinal ligaments, which can be grouped into two primary systems, the intrasegmental and intersegmental systems (Figure A-6). The intrasegmental system holds individual vertebra together, while the intersegmental system holds many vertebrae together.

Figure A-6: Ligaments of the spine (Panjabi et al., 1990).

A.5.1 Intrasegmental system

The intrasegmental ligament system includes the ligamentum flavum, interspinous ligament, intertransverse ligament and facet capsular ligament.

Ligamentum Flavum (LF)

The ligamentum flavum extends from the anteroinferior border of the laminae above to the posterosuperior border of the laminae below. The ligament is composed of a large amount of elastic fibers and represents the most pure elastic tissue in the human body (Panjabi et al., 1990).

Interspinous Ligament (ISL)

The interspinous ligament connects adjacent spines, extending from the root to the apex of each process. It is broader and thicker in the lumbar region (Panjabi et al., 1990).

Intertransverse Ligament (ITL)

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

Facet Capsular Ligament (CL)

The capsular ligament is attached just beyond the margins of the adjacent articular processes. It is generally oriented in a direction perpendicular to the plane of the face joints (Panjabi et al., 1990).

A.5.2 Intersegmental system

The intersegmental ligament system contains the anterior longitudinal ligament, the posterior longitudinal ligament and the supraspinous ligament (Panjabi et al., 1990).

Anterior Longitudinal Ligament (ALL)

The anterior longitudinal ligament is a fibrous tissue structure that arises from the anterior aspect of the basioccipital and is attached to the atlas and the anterior surfaces of all vertebrae, down to and including a part of the sacrum. It attaches firmly to the edges of the vertebral bodies but is not so firmly affixed to the annular fibers of the intervertebral disc (Panjabi et al., 1990).

Posterior Longitudinal Ligament (PLL)

The posterior longitudinal ligament arises from the posterior aspect of the basioccipital, and runs over the posterior surfaces of all the vertebral bodies down to the coccyx (Panjabi et al., 1990).

Supraspinous Ligament (SSL)

The supraspinous ligament originates in the ligamentum nuchae and continues along the tips of the spinous processes as a round, slender strand down to the sacrum (Panjabi et al., 1990).

A.6 Functional Spinal Unit (FSU)

The functional spinal unit, also called the motion segment, is the smallest segment of the spine that exhibits biomechanical characteristics similar to those of the entire spine. It consists of two adjacent vertebrae, including the intervertebral disc

between them and the ligaments that bind them together. For its biomechanical characterization, the lower vertebra is fixed while the loads are applied to the upper vertebra, and its displacements are measured. The stability of the motion segment is provided by ligaments, facet joints and the intervertebral disc. The contribution each component provides to the stability depends on the type of loading and its degenerated grade. A typical motion segment only with two vertebrae and the intervertebral disc from lumbar region is shown in Figure A-7.

Figure A-7: A motion segment from lumbar spine, showing two adjacent vertebrae and the intervertebral disc between them

Appendix B: Convergence Test of L4-L5 FE model

Convergence test is an important part of a finite element study to ensure the reliability of FE predictions due to mesh density. In this project, three FE models with different element number were created to evaluate the effect of mesh density. The element number was increased from 8291 (Model 1) to 9365 (Model 2) and finally to 10489 (Model 3) as shown in Figure B.1. For each model, the inferior surface and the spinous process of L5 were completely constraint, and then a uniform axial compressive displacement of 1.8mm was applied to the superior surface of L4. The reaction force was extracted from the constraint surfaces.

(a)

(b)

(c)

Figure B.1: Finite element model of L4-L5 motion segment. (a) model 1 with 8291 elements; (b) model 2 with 9365 elements; (c) model 3 with 10489 elements

The results were summarized in Table B.1. It can be noted that the effect of mesh density is very small. The predicted reaction forces for the three models were within the same order of magnitude, especially for model 2 and model 3, the difference of which was only 1.4%. Therefore, Model 2 was chosen for the subsequent analysis.

Model	Number of elements	Reaction Force (N)
1	7925	2763.53
2	8999	2670.88
3	10123	2634.26
1 and 2	% Difference = (the larger one - the smaller one)/the larger one	3.3%
2 and 3		1.4%

Table B.1: The effect of mesh density on the FE predictions

Appendix C : Publications

Zhou YL, Qiu TX, Zhang QH, Teo EC. The relationship between load transmission and disc degeneration in thoracolumbar spine: a finite element study. *Journal of Musculoskeletal Research* 10(3):121-130, 2006.

Zhou YL, Zhang QH, Teo EC. Effect of disc height and disc wedge angle on lumbar spine under axial compressive force. A paper will be presented at the 3rd WACBE World Congress on Bioengineering, Thailand, Jul 2007

Zhou YL, Qiu TX, Teo EC, Petit D. FE assessment of an interspinous implant (ISS). A paper presented at International Conference on Biomedical and Pharmaceutical Engineering, Singapore, Dec 2006

Zhou YL, Teo EC, Qiu TX. FE kinematics analysis of bilateral facetectomy of thoracolumbar (T11-L1) spine. A paper presented at 6th ASEAN ANSYS Conference, Singapore, Nov 2006.