A THREE-DIMENSIONAL MULTIBODY COMPUTATIONAL MODEL OF LUMBAR SPINE

BY

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Abstract

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A Three-Dimensional Multibody Computational Model of Lumbar Spine
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The lower back is the most sensitive part of the human spine and over loading and bad posture during lifting can damage this area of the body. The lumbar spine consists of five vertebrae, which are responsible for carrying the weight of the upper body and loads. Intervertebral discs allow articulation between vertebrae. These discs are primarily made of non-homogeneous soft tissue, which allows the vertebrae to move and flex in all directions.

Biomechanical models have been developed in the past decades to model and to predict the behaviour of the spine in response to different loads. With the advances in computer modeling technology, analytical methods have become more popular in modeling the spine. These models are more cost effective and practical compared to the early models and use of human volunteers and cadavers. Unfortunately, due to the complexity of the spine, most of the models failed to offer an accurate estimation of reaction moments and forces. Most models also use proprietary and custom-made software which makes it difficult for other researchers to use and modify them.

This thesis reports the development and verification of a multi-body computational model of the lumbar spine. The model comprises five lumbar vertebrae (L1 to L5) and pelvis (S1). The vertebrae are connected to each other by intervertebral discs, which consist of an anatomically correct kinematic and dynamic constraints. This combination represents a six degree-of-freedom mobility and enables the model to accommodate flexion, lateral bending, and axial rotation.

The model is validated by carrying out a series of case studies including experimental motion studies. It is also used for preliminary evaluation of an ergonomical device called the dynamic trunk support (DTS), developed at Ryerson, School of Occupational and Public Health, in conjunction with the Mechanical and Industrial Engineering department. The results are in good agreement with the experimental results.

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CHAPTER 1

Introduction and definition

1.1 Introduction

Biomechanics is the application of mechanical principles to living organisms. The investigation of the forces that act on the limbs and cause movement is part of biomechanics. The human body is an advanced skeletal structure, which is capable of various activities. Like any other structure, it can be analyzed and the forces and moments within its joints and links can be predicted. Biomechanics has many applications in medicine, sports, and professions that deal with physical movements of the body. The information obtained through biomechanics helps maintaining healthy postures with minimum stress on the bones, joints, and soft tissues. Biomechanics has been also used for designing equipments that are ergonomically useful for many professions.

The skeleto-muscular system is responsible for the movement of limbs with its contractile tissue. The function of muscle is to produce force that causes motion. The spine is a flexible column made of 32 vertebrae that are stacked on top of each other and extended from the skull to the pelvis. It is divided into five regions. Cervical (7 vertebrae), thorax (12 vertebrae), lumbar (5 vertebrae), sacrum (5 vertebrae fused), and coccyx (3 vertebrae tailbone). The spine has several curves in each region when viewed from a lateral angle. These curves are essential and provide resistant and elasticity in distributing the body's weight and loads during movement.

There are more than 30 muscles and tendons responsible for the mobility of the spine. These muscles and tendons also provide the balance and stability of the spine. Muscles control the movement of the spine with their contraction and relaxation. Based on the pattern of the spine, they have been divided in four types of vertebral muscles; forward flexors, lateral flexors, rotators, and extensors.

The complexity of the spine's functions makes it prone to injuries. Injuries happen mostly within the low back vertebrae of the lumbar region. The most common cause of jobrelated disabilities is low back disorders (LBD). The incidence of LBD in the U.S., Germany, Norway and Sweden, has been reported to be 56%, 59%, 61%, and 70%, respectively [1-3]. A review of epidemiology studies concludes that one of the major documented risk factors for LBD is lifting [4-6]. According to the U.S. national institute of health, Americans spend at least $50 billion each year on low back pain (NIH Publication No. 03-5161). Studies have been conducted in the field of ergonomics and biomechanics to analyse and predict the behaviour of lower back and internal forces to prevent LBP in work places. Non-surgical and preventative treatments would benefit from more accurate estimations of muscle forces and loads. This will enhance the productivity at work places and reduce the health care costs.

The modelling of the spine has improved the development of spinal care in the past 50 years. The early models were made of masses representing rigid body segments attached to each other by springs and dampers representing muscles and ligaments.

Due to the complex structure of the human body, complex models are required to have a realistic simulation. In spite of all the progress in the field of biomechanical models, the demand for more sophisticated and detailed models is still evident. A popular approach to develop a biomechanical model is the analytical method. This technique incorporates the finite element method to build a realistic model based on the quantitative anatomy of rigid segments. The use of analytical methods in developing biomechanical models has increased along with the progress of computers and the known limitation in using human volunteers or dummies. Computer models combine the kinematical and analytical approach to predict and simulate forces and postures of the spine that leads to estimating the loads and stresses.

1.2 Research objective

The objective of this research is to examine a biomechanical model of the lumbar spine capable of predicting internal moments acting on the vertebrae. This computer model will be developed with the help of general software that could be easily worked with and maintained. The present study will try to combine mechanical calculation and solid modeling based on anatomical information into the development of a versatile model.

The model's rigid body elements will remain realistic in terms of quantitative predictions. No muscles will be used in this model to avoid redundancy. Instead of muscles, actuators will be responsible for body movement. To verify the model, case studies which reflect the typical body motion will be examined.

1.3 Thesis organization

This thesis is presented in six chapters organized as follow:

- Introduction, brief remarks regarding biomechanics, spine, simulation models, and methods.
- Literature review, discussion about early models, existing finite element and rigid body models and their applications.
- Methodology, this chapter deals with the analytical and mechanical approach to develop a model based on data from literature and experiments.
- Experimental evaluation, comparison between analytical results and experimental results.
- Application of model in ergonomics, dynamic trunk support (DTS) is analysed with the current model.
- Result and discussion.
- Conclusion.
- Appendix.

CHAPTER 2

Literature review

2.1 Introduction

The spine is one of the most complex structures in the human body. The flexibility and resilience of the spine column enables humans to take various action and movements, which are necessary for life. It is also very sensitive with regards to heavy loads and bad posture, which makes it prone to injuries and physical damage. In fact, low back injuries and low back pain are among the most common injuries in modern societies and cost billions of dollars in compensation money, treatment and time off of work. During the past few decades, many attempts have been made to model the spine and simulate its reactions under different situations. The motivation has been to get a better understanding of the spine and be able to predict its behaviour under various loads and configurations.

Efforts to build a universal model of the spine have lead to much progress in this field of research and the help of computers has brought a breakthrough in this regard.

This chapter reviews some of the previous attempts in modelling the spine and introduces the latest innovations in computational models used in biomedicine of the lumbar spine.

2.2 Early Models

Most early models represented a two-dimensional system of mass and springs. The application of these models in simulating the spine is more analytical than practical. One of the attempts to build a model to investigate the kinematics of lumbar spine was a two-dimensional model developed by Keller and Colloca [7]. This model is an analytical model, and is based on the mathematical equations of vibration of a mechanical system. It is assumed that all the rigid elements are homogeneous bodies. The result of calculation could predict the posterior-anterior (PA) motion of the lumbar spine. The motions are the response of the spine subjected to PA forces. This model had 5-degree of freedom, in which five lumps of masses represent L1 to L5. The masses were attached to each other in a series of mass-less spring and dampers (Figure 2.1)

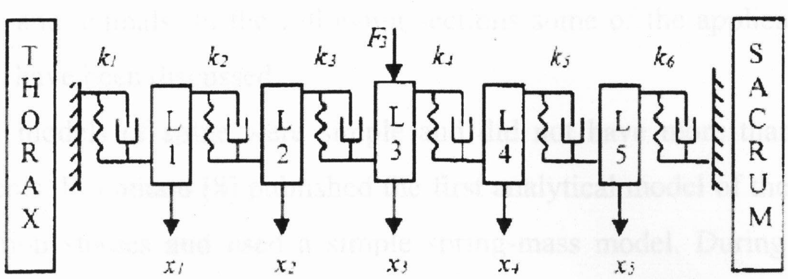

Figure 2.1: Model lumbar spine with five degrees of freedom [7]

Thorax and sacrum are assumed to be immobile. Differential equations in a matrix form were solved and the results showed the motion of lumbar spine in response to PA forces. The spring and damper joints allowed the users of this model to define mechanical properties for the soft tissues of intervertebral discs.

The immobility of thorax and sacrum deprives the ability of the model to predict rotational motions. However, this problem could be solved by adding more degrees of freedom to the model and make it more realistic. The other shortcoming is the fact that a three-dimensional structure, i.e., the lumbar spine is basically modelled as a one-dimensional series of mass and springs. The results would not be representative of the loads experienced by the spine.

2.3 Finite Element analysis of the spine

Finite element (FE) is a powerful technique that has been widely used in biomechanical research for nearly a quarter of century. This method employs a numerical technique to simulate a variety of clinical situations. The recent developments have elevated the FE method to a fully complementary relation with experimental approaches during investigation of clinical spine problems. The accuracy of the mathematical model, however, depends on the accuracy of the experiments and the measurements. Since the analytical method goes hand in hand with experimental method, lack of experimental data may make it difficult to validate the model predictions or even to evaluate a FE model.

Finite element modeling provides a variety of applications and physiological information for biomedical researchers and shows an acceptable accuracy compared to experimenting with cadavers and animals. In the following sections some of the applications of FE in spine research have been discussed.

The early FE models of spine were simple and did not have more than two or three motion segments. F. Latham [8] published the first analytical model of the spine in 1957 for pilot ejection studies and used a simple spring-mass model. During 1960's, more complex spring-mass models were developed using dampers and more degrees-of-freedom to represent the vertebrae and intervertebral discs [9]. Orne and Liu [10] proposed a more realistic model in 1970. In their model the internal stress and forces of shear and bending that act on the disc were taken into account and the viscoelastic behaviour of intervertebal elements were introduced.

Belytschko et al. [11] developed a three-dimensional FE model of spine in 1978. The goal was to evaluate the mechanical response of a pilot during ejection from cockpit. This model consisted of head, cervical, thoracic, and lumbar vertebrae, and pelvis (sacrum). Ligaments and muscles were modeled as springs and it was assumed that all the skeletal components are rigid body. To simulate the ejection, a non-linear force was applied to the base of the model.

Belytschko et al. [12] also developed the first FE model of intervertebral disc in 1974 and presented a detail analysis of the properties of the disc and its adjacent vertebrae. Shirazi et al. [13] developed a breakthrough model of intervertebral disc in 1984. They developed a three-dimensional no-linear model of L2-L3 disc. This model consisted of all the disc components and represented the non-linear properties of its elastic elements.

In the study of Chen et al. [14] FE method has been used to compare between three models, intact lumbar spine (INT model), a lumbar spine inserted with a pair of bilateral posterior lumbar interbody fusion (PLIF) supplemented with pedicle screw implants at L3-L4, and finally lumbar spine implanted with ProDiscII artificial disc (ADR) at L3-L4. All models were constrained at the end of L5 vertebra and subjected to 150 N preload and a moment of 10 Nm. The loading condition was similar to in vitro studies of Yamamoto et al. [15].

The FE analysis software ANSYS 9.0 (ANSYS INC., Canonsburg, PA) was used for reconstruction of the intact model of the spine base on the Initial Graphic Exchange

Specification (IGES) file. The IGES file was obtained from a CT scan of L1-L5 lumbar spine of a middle-aged male. All the material properties of bone, disc, cartilaginous endplate, and ligaments used for the intact model were taken from previous studies [14]. This model was validated and compared to models of Chen et al. [16], and Shirazi-Adl et al. [17], Table 2.1 shows the results.

Torsion	Loading Condition	L1-L2	L2-L3	L3-L4	L4-L5
Chen et al.'s study (2009) [14]	10 Nm with 150 N axial force	121	130	125	127
Chen et al.'s study (2001) [16]	10 Nm with 150 N axial force	121	157	161	155
Shirazi-Adl's study (1994) [17]	10 Nm	107	123	117	78

Table 2.1: Comparison between Chen and Shirazi-Adl [14]

The study of Chen et al. showed lower forces. This model was verified based on cadaveric in vitro data for more simulations.

Zhang and Teo [18] investigated the application of FE models in the case of using implants for treatment of low back pain. FE analysis was used to predict the value of stresses in the disc, vertebrae, and ligaments and provide the researchers with a detailed motion data of the segments. They investigated the potential effect of implant design parameters i.e. shape, material, and position for the stability of a biological fusion. Results of several studies conducted by Rohlmann et al., [19-20] on the effects of different fixation styles showed that the diameters and stiffness of the longitudinal rod of an internal spinal fixation device strongly influenced the stresses in the vertebral endplates [19]. In another study differences caused by changing the position of implants were investigated. Implants were fixed posteriorly and anteriorly in L4-L5 model [20]. Model was loaded with body weight and muscle forces to simulate standing position, flexion, extension and axial rotation. Result showed both implant types reduced the intersegmental rotation for flexion, extension, and axial rotation.

Disc degeneration is a natural part of aging, however, it could be intensified by a simple lifting accident. Dynamic stabilization is considered the best treatment for slight degenerations while a fusion and disc replacement is recommended for severe cases. Fusion is a common surgical procedure used to treat a painful vertebral segment. This technique involves adding bone graft in conjunction with the body's natural bone formation process. The formation of bone between the vertebrae will eliminate the abnormal motions and stabilizes that segment hence eliminating the pain. Many of the published FE studies of lumbar spine implants are related to spine fusion, and cover the study of implant design, study of bone material, and the fixation techniques.

The interbody implants also called cages or spacers are three-dimensional devices that are attached to the skeletal segments of the spine for stabilization and support. The role of cage is to enhance the fusion procedure in order to achieve better results. Fantigrossi et al. [21] studied different kinds of implants. Using FE analysis Zhong et al. [22] optimized the size and shape of the implants by using optimization methods. The FE models estimated the amount of volume reduction that could maintain the same biomechanical performances.

A three dimensional FE model of lumbar spine was developed by Lavaste et al. [23] based on six geometrical parameters of lumbar vertebrae. These six parameters were chosen to make it possible to calculate all the other dimensions of the vertebrae.

In order to reconstruct the lumbar vertebrae morphological analysis was performed on lumbar vertebrae of eight cadavers. X-ray and digitizers were employed to extract necessary geometrical data to describe the lumbar vertebrae.

To reconstruct the lumbar segment the position of each vertebrae must be identified in relation with the global coordinate system. The coordinate of each vertebra was defined earlier by associating a coordinate system at the center of each vertebra. The vertebrae can then be connected to each other using the intervertebral discs and the ligaments. Connection lines that attach points on two adjacent vertebrae define the ligaments. Figure 2.2 shows a single vertebrae and the 3D lumbar segment reconstructed by putting together the vertebral bodies.

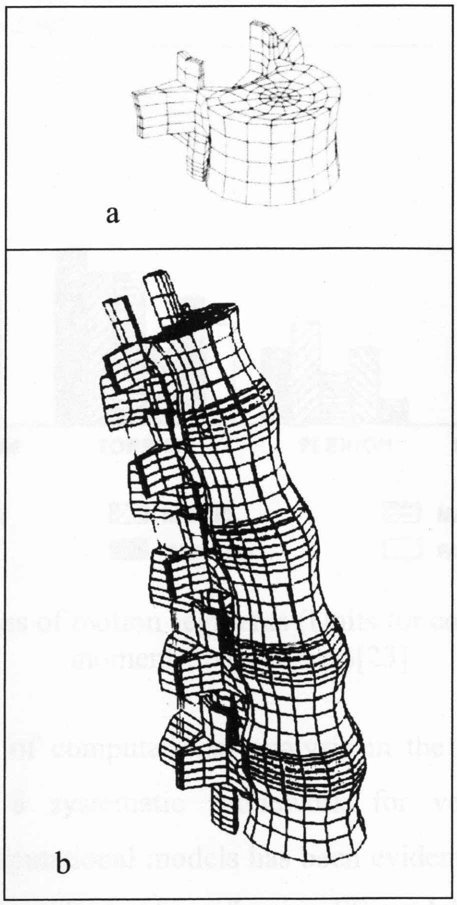

Figure 2.2: a) Lumbar vertebra b) FE model of lumbar spine segment by Lavaste et al. [23]

The lumbar spine is a dynamic structure and has flexion, lateral bending, and axial rotation. The physical properties of spine components must be taken into account. A mean stiffness has been used after comparing results from various published experiments. Figure 2.3 shows the information extracted from different sources and used to acquire a mean stiffness to apply on intervertebral parts of the FE model. Interverteral parts constrain the rigid parts and limit their motions to a certain degree. Other elements in rigid body models include springs, dampers, muscles, ligaments, etc. The physical properties of the motion segments can be taken either from literature or experimental data.

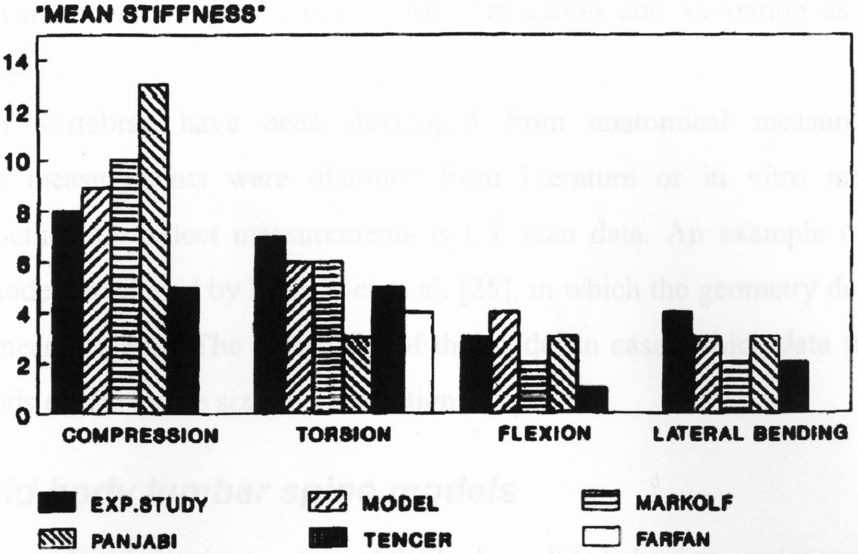

Figure 2.3: Mean stiffness of motion segments (Units for compression 10E5 N/m; for moments, Nm/degree)[23]

With the increase in use of computational analysis in the field of spine research the necessity of employing a systematic framework for verification, validation, and sensitivity analysis of computational models has been evident more than ever. In a study by Jones and Wilcox [24], methods of verification FE models have been examined. The methodologies that lead to biomechanical evaluation of treatment options are; in vitro or in vivo. In vitro methodologies involve experimental measurement on spinal component that is tested in laboratory and is limited to number of samples. In vivo methodologies provide information of the spine in its natural state.

In recent years FE analysis has become the first choice by many research groups for this purpose. Models of the spine are developed to investigate the risk factors in vertebrae degeneration and they are applied to optimize and evaluate the treatment options. The validation of a model could justify the results generated by the model and show that it corresponds to the results from a real world scenario. Validating the FE model against in vitro experiments can prove the integrity of the results.

The study of Jones and Wilcox [24] focused on three major models: model of vertebra, model of intervetebral disc, and segment models. In their research they concluded as a result that advance in experimental methods should be parallel to progress in computational technology. A robust model development should exploit both the technological innovations and methods for verification and validation as a minimum requirement.

Models of vertebrae have been developed from anatomical measurements. The anatomical measurements were obtained from literature or in vitro measurements. Another method to collect measurements is CT scan data. An example of that is the vertebra model developed by Liebshner et al. [25], in which the geometry data was taken from CT measurements. The sensitivity of the model in cases which data was from CT scan depends on the image size and resolution.

2.4 Rigid body lumbar spine models

Spine is a complex dynamic structure of the body and its behaviour under various loading situations is hard to model. Many research groups have tried to model different sections of the spine that could help predict the response of the spine to external forces and loads. Rigid body elements are used in FE models representing motion segments [26].

A rigid body model consists of several rigid segments representing the bony parts of the spine, i.e., the vertebrae. The rigid segments are attached to each other by joints. Joints are elements that represent the soft tissues like intervertebral discs, facia, ligaments, and muscles. However, because of the nature of soft tissues, it is hard to measure properties like elasticity and stiffness. The geometry and anatomical data of both rigid and soft parts are contained in [27], [28] and will be reviewed in a subsequent section.

Stokes et al. [27] developed a rigid body model consisting of five lumbar vertebrae, the sacrum/pelvis, the thorax, and sixty-six symmetric pairs of multi-joint muscles. Published data was used for the construction of the model, the position of the vertebrae and the attachment and sizes of symmetric and multi-joint muscles.

Two different models with the same geometry were used: (a) stiffness model, in which the motion segments were represented as either beams having predetermined stiffness properties (Figure 2.4a), and (b) static model, with a simple 'ball-and-socket' joint between the vertebrae (Figure 2.4b).

These two models were used to determine maximal load transmission through the spine with the requirement that muscle forces be compatible with equilibrium at all six joints of the lumbar spine.

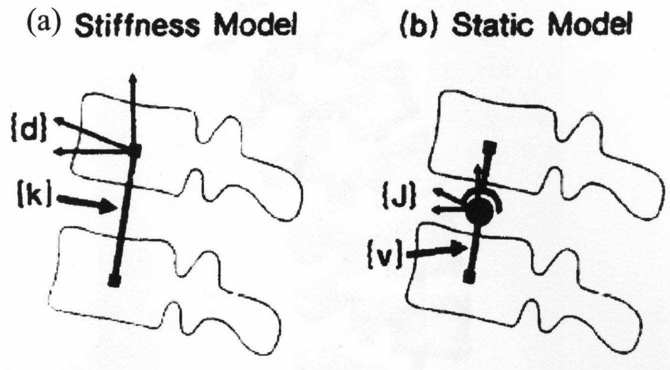

Figure 2.4: Two representation of the motion modeled by Stokes et al. [27], a) model with stiffness, b) model without stiffness

The two models were tested during flexion, extension, lateral bending, and axial torque. The results confirmed that the model with stiffness (Figure 2.4a) had a greater moment transferred through the spine and also more muscle activation.

Mark de Zee et al. [28] developed a detailed lumbar spine model with seven rigid segments and 18 degrees-of-freedom and 154 muscles. Inverse dynamics was used to predict muscle forces and joint reaction forces. The anatomy of lumbar spine and muscle data was taken from literature (Hansen et al. [29] and Bogduk [30]). The muscles are modelled as one-dimensional elements. It has been tried to describe the muscle path close to real muscle. The model consists of seven rigid segments; the pelvis, the five lumbar vertebrae and the thoracic part. Dimensions of lumbar vertebrae are taken from the studies of Nissan and Gilard, [31], which is based on a person with a height of 1.75 m and a weight of 72 kg. The mass moment of inertia of lumbar vertebrae is assumed zero. The joint between each vertebra is modelled as three degrees of freedom ball-and-socket joint. An important parameter is the center of rotation and is shown in Figure 2.5. The center of rotation was taken from the studies of Pearcy and Bogduk [32].

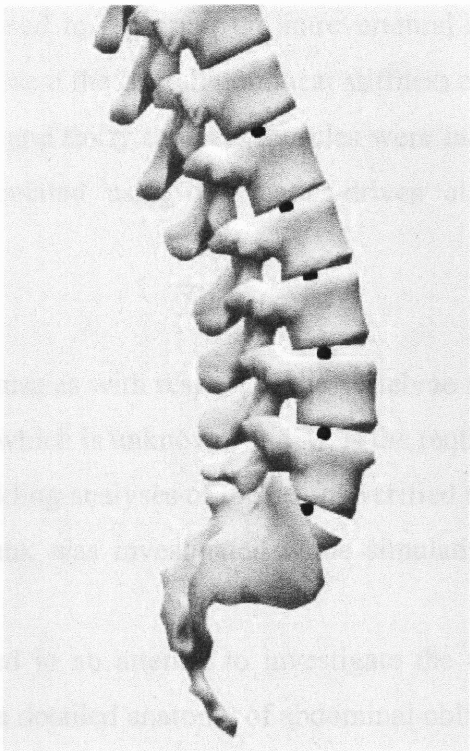

Figure 2.5: The dots show the center of rotation [28]

A comparison was made between the model and in vivo intradiscal experiments performed by Wilke et al. The in vivo experiments measured the pressure in the L4-L5 disc while the subject (body mass: 70 kg, body height: 1.74 m) was carrying a weight of 19.8 kg about 60 cm away from his chest. The result was 18 MPa pressure between L4-L5. The model was used to simulate the same situation. The estimated pressure was 18.9 MPa, which is a good match considering that the body mass and height of the model was slightly different. The main limitation with this and similar models, where inverse dynamic approach is used, is that there are many more unknowns than there are equilibrium equations. This redundancy can only be overcome by making some assumptions such as optimal recruitment of muscles. This, in many cases, is not a realistic assumption.

Shirazi-Adl et al. [33] developed a model using finite elements and a kinematics-driven approach. The model predicted the trunk response to maximal isometric axial torque. A nonlinear finite element model of thoracolumbar spine along with a kinematics-driven algorithm was used to overcome the trunk redundancy. Six sagittaly symmetric deformable beams were used to represent the intrevertebral segments between T12-S1 vertebrae. The beams represent the overall nonlinear stiffness of T12-S1.

Forty-six lumbar muscles and thirty thoracic muscles were incorporated into the model. Muscle forces were calculated using kinematics-driven algorithm. The equilibrium equation is:

$$\sum r \times f = M \qquad \text{Equation 2.1}$$

where r, is lever arm of muscles with respect to the vertebrae to which they are attached, f, is total force in muscle, which is unknown, and M, is the required moment.

Linear and non-linear buckling analyses of the beams verified the spinal stability and also torque strength of the trunk was investigated while simulating the experimental work with this model.

This model was developed in an attempt to investigate the trunk stability under axial torque exertions. It needs a detailed anatomy of abdominal oblique muscles for predicting the trunks response and too many parameters are involved. The main shortcoming of this model is the fact that it only allows for bending in one direction. The sliding of the vertebrae is not considered either.

A multi-body computational model of head and neck was developed by Van Lopik et al. [34]. The model represents the natural lordosis of the neck of an adult male. It consists of nine rigid bodies representing the head, seven cervical vertebrae, and the first thoracic vertebrae T1. Soft tissues are also included as the intervertebral discs, ligaments and muscles. Intervertebral discs are modeled as non-linear viscoelastic material with the required number of degrees of freedom. Ligaments are modeled as non-linear viscoelastic spring-damper elements. Virtual muscles support the realistic muscle properties. Published anatomical data was used to develop the bony elements of cervical spine and the skull. The vertebrae T1 is located at the origin of the global coordinate system. In addition to the anatomical geometry other parameters are used to develop the model. These include the vertebral body width, pedicle angles, and spinal canal dimensions, and facet joints, based on Panjabi et al. Studies [35].

Disc stiffness reported by Moroney et al. [36] and tension and compression values by Yoganandan et al. [37], were used for the stiffness characteristic of the disc and modelled with MSC.visualNastran4D. The intervertebral discs are modeled as a bushing constraint and located approximately at the disc center. A non-linear viscoelastic bushing constraint allows all translation and rotational degrees of freedom restricted by spring and damper relationships. A summary of disc biomechanical properties is shown in table 2.2

Loading direction	Stiffness k (N/mm)						Damping b (Ns/m)
	C2-C3	C3-C4	C4-C5	C5-C6	C6-C7	C7-T1	C2-T1
Anterior shear	62	62	62	62	62	62	1000
Posterior shear	50	50	50	50	50	50	1000
Lateral shear	73	73	73	73	73	73	1000
Tension	63.5	69.8	66.8	68.0	69.0	82.2	1000
Compression	637.5	765.3	784.6	800.2	829.7	973.6	1000

Table 2.2: Intervertebral physical properties [34]

Head and neck ligaments are also included into the model as shown in Figure 2.6 ligaments are models as non-linear viscoelastic spring elements in VisualNastran4D. Data from Panjabi et al. [35, 38] was used to determine the attaching points of ligaments to cervical spine.

This model has been validated with regard to experimental results and has shown a good agreement to those results. However, since it does not represent the lumbar spine there is still a gap left for back injury researchers to fill. To develop a lumbar spine model in accordance with this model a comprehensive understanding is required about the lower back anterior/posterior muscles and ligaments. The model also assumes that all muscle forces are at maximum at all times.

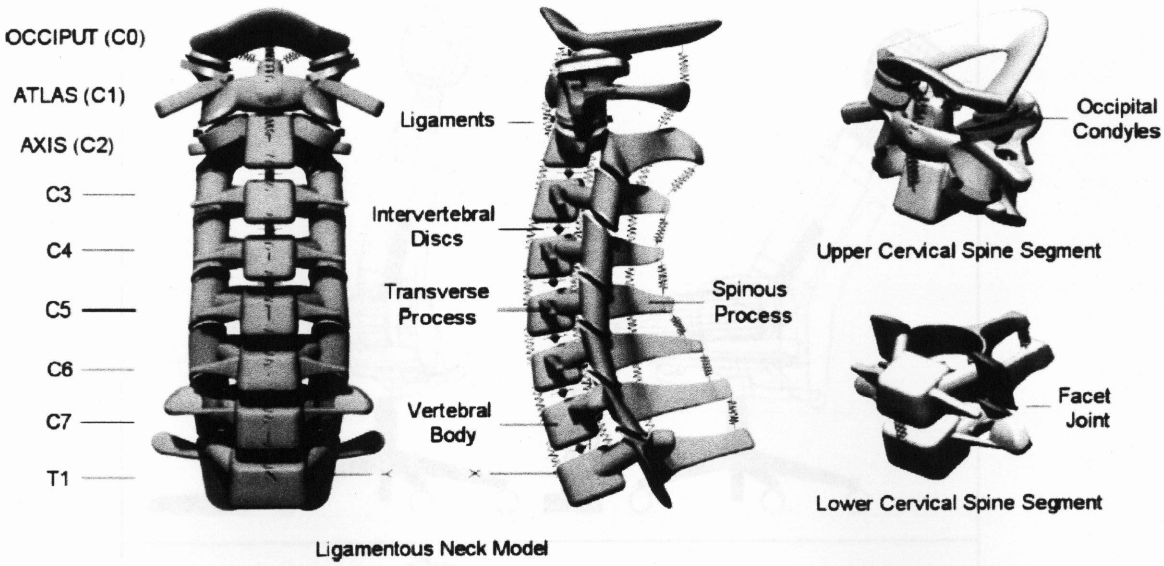

Figure 2.6: Ligaments replaced with linear springs by Van Lopik et al. [34]

Lengsfeld et al. [39] modeled lumbar spine curvature during prolonged office chair sitting. The model was developed with the ADAMS software (MDI Inc., Ann Arbor, USA) and consists of 15 rigid segments. The graphical display and the wire frame of the model is shown in Figure 2.7. The model represents an adult male, 1.78 m in height and weighing 78 kg. A right hand coordinate system established the origin of the coordinate system, which coincides with the position of the right hip joint. Positive x direction stretches through medial or midline of the body, positive y values refer to cranial, or pointing toward the head and positive z values the anterior directions represented in the model. The hip is modelled as a spherical joint, which gives it the required rotational motions. Vertebrae L1 to L5 are modeled according to the geometrical data from literature [40, 41].

The model was developed to test the lumbar support of two different models of office chair. The adaptability of the model to different chair concepts was due to its flexible joints between rigid segments. The model shows that during a backrest with 19° (Figure 2.7a right) the lumbar lordosis disappears completely. When the chair has been replaced with a more ergonomic model the lumbar lordosis maintains even though the backrest angle has been increased to 21° (Figure 2.7b right)

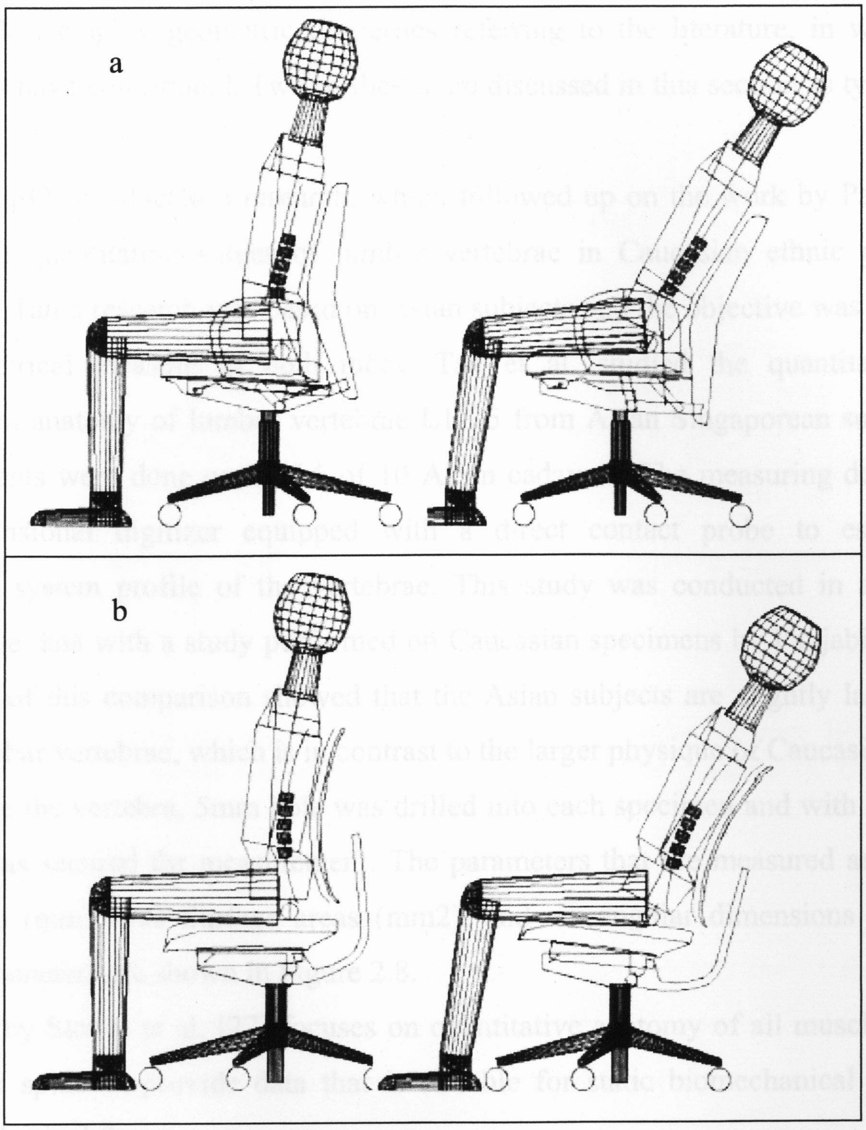

Figure 2.7: (a) Graphical display and wire frame of the model sitting on a chair with lesser lumbar support. (b) Same model on a chair with stronger lumbar support [39]

2.5 Lumbar data

The geometrical data of vertebrae is necessary to model it properly. The common method for obtaining this data is to measure them right from a cadaver and generalize the result base on the gender and race. The information about lumbar vertebrae can be looked at from two different views, first the geometrical description of vertebra and second the quantitative anatomy of the vertebra or the position of vertebra. A valid model should incorporate both notions to maintain the quantitative properties of spine.

Many studies employ geometric properties referring to the literature, in which many researchers have contributed. Two of theses are discussed in this section as typical of the field.

Tan et al. [42] conducted a research, which followed up on the work by Panjabi et al. [43] where quantitative values of lumbar vertebrae in Caucasian ethnic group were quantified. Tan's research was based on Asian subjects and the objective was to compare the geometrical measure in both races. Tan et al. studied the quantitative three-dimensional anatomy of lumbar vertebrae L1-L5 from Asian Singaporean subjects. The measurements were done on L1-L5 of 10 Asian cadavers. The measuring device was a three-dimensional digitizer equipped with a direct contact probe to establish the coordinate system profile of the vertebrae. This study was conducted in an effort to compare the data with a study performed on Caucasian specimens by Panjabi et al. [43]. The result of this comparison showed that the Asian subjects are slightly larger, in the size of lumbar vertebrae, which is in contrast to the larger physique of Caucasians.

To measure the vertebra, 5mm hole was drilled into each specimen and with a screw the vertebra was secured for measurement. The parameters that are measured are 14 linear dimensions (mm), five Surface areas (mm2), and six angular dimensions (deg) [42]. Theses parameters are shown in Figure 2.8.

The study by Stokes et al. [27] focuses on quantitative anatomy of all muscles crossing the lumbar spine to provide data that is suitable for static biomechanical analysis of muscle and spinal forces.

Four adults average-sized subjects were chosen. Radiographs were taken from their spine in a standing posture. In addition to the living subjects two cadavers were dissected and their abdominal muscles were examined. A global coordinate system was established. Center of each vertebra (T1 to S1) was coordinated and were normalized to the mean length of the spine of each subject. The three-dimensional coordinate of the center of each vertebra is shown in Table 2.3.

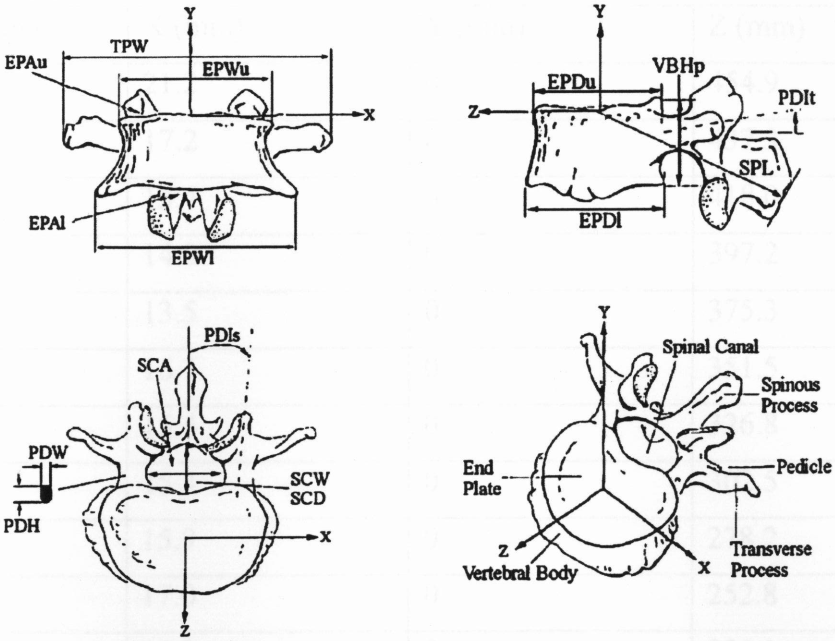

Figure 2.8: Four views of a lumbar vertebra. EPAl, lower end-plate area; EPAu, upper end-plate area; EPDl, lower end-plate depth; EPDu, upper end-plate depth; EPWl, lower end-plate width; EPWu, upper end-plate width; PDH, pedicle height; PDIs, left ped

The information of table 2.3 is useful when it comes to modeling the lumbar spine. A three-dimensional body segment of vertebrae can be modeled and maintain a lordotic posture with the realistic curve of lumbar spine. The origin of coordinate system for this purpose will be on S1 (0,0,0).

The results of literature review can be divided in two groups. First the quantitative anatomical information, which will be incorporated into the solid modeling of rigid body segments. Second the mechanical properties of soft tissues i.e. elasticity, stiffness, which effects the accuracy of simulation of muscles, ligaments, and discs.

Most of the models studied in the literature review represent custom made software. It must be considered that custom-made software is not user friendly especially for non-advanced uses. The model should be easy to operate and the user must able to easily change the parameters. Such a model could be versatile and used for many different loading situations.

Vertebra name	X (mm)	Y (mm)	Z (mm)
T1	21.2	0	454.9
T2	17.2	0	437.3
T3	13.4	0	418.0
T4	14.0	0	397.2
T5	13.5	0	375.3
T6	16.1	0	351.5
T7	15.4	0	326.8
T8	15.4	0	302.5
T9	15.9	0	278.2
T10	17.0	0	252.8
T11	16.9	0	225.1
T12	14.4	0	196.0
L1	8.3	0	166.3
L2	12.6	0	133.9
L3	13.0	0	100.6
L4	13.1	0	64.8
L5	4.2	0	31.7
S1	0	0	0

Table 2.3: The position of vertebral body centers in a standing posture [27]

CHAPTER 3
Methodology

3.1 Introduction

This chapter describes the methodology to construct a three dimensional multi-body model that represents the human lumbar spine. The model is constructed using general purpose solid modeling software generally available and as such can be easily evaluated and shared with the research community and other users.

Once constructed, the model will be able to simulate lumbar spine motions to investigate the reaction moments created between each lumbar vertebra under various loading conditions.

3.2 Kinematics of Lumbar spine

As shown in Figure 3.1, the lumbar spine is made of five lumbar vertebrae L1 to L5. L1 is the upper most vertebrae of the lumbar spine and is attached to T12 of thoracic spine. L5 is the lowest of five lumbar vertebrae and is attached to the sacrum. Lumbar vertebrae are articulated to each other by intervertebral discs, ligaments, and muscles.

Three fundamental movements of the spinal column are flexion-extension, lateral bending, and axial rotation. The greatest amount of movement takes place in cervical (neck area) and lumbar regions. Spinal muscles either anterior or posterior are responsible for the movements of the spinal column. The anterior to the spinal column are muscles that flex or rotate the spine. Those posterior to the spinal column are muscles that extend, laterally flex, or axially rotate.

In most biomechanical models of the human body, the skeleton is considered as a system of linked levers. In this system, bones are the links which are normally loaded in compression. Joints become fulcrums and muscles and ligaments are tension forces around each joint or fulcrum, resulting in moments or torques. These kinds of models basically treat the lumbar joints as one-dimensional rotational joints.

An important component of the spine is the intervertebral disc. Intervertebral discs are composed of annulus fibrosus and nucleus pulposus, located in the center of the annulus. This soft and flexible tissue changes in shape with movements of the spine. The result is a six-degree of freedom joint between each pair of vertebrae. The anatomical studies have

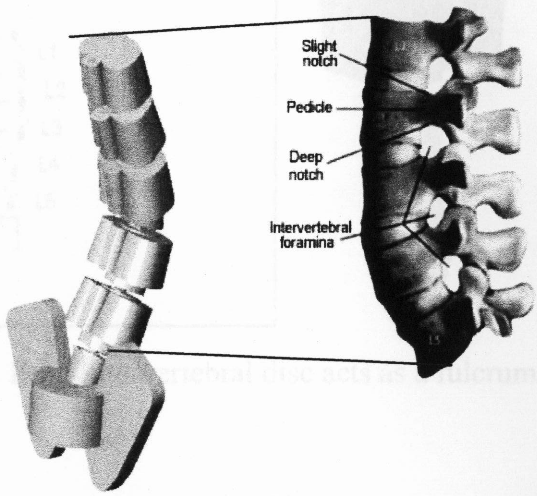

Figure 3.1: Three-dimension model of the lumbar spine was developed

shown the maximum segmental angle to be around 5° [39]. The compression and sliding movements between a pair of vertebrae although small is very important.

The model developed in this chapter (Figure 3.2) can represent all six degrees of freedoms described and therefore provide a versatile model of loads experienced by the lumbar spine.

3.3 Model components

The developed model consists of two basic components: vertebrae and intervertebral discs. The proper assembly of these components allows modeling of complex movements and interactions of the lumbar spine. Trunk and pelvis are shown as simplified models to complete the representation (see Figure 3.3)

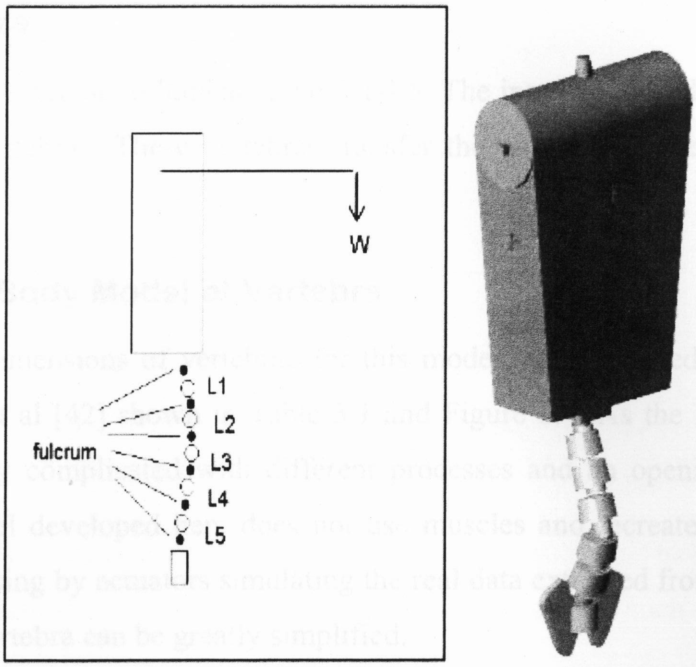

Figure 3.2: Each intervertebral disc acts as a fulcrum

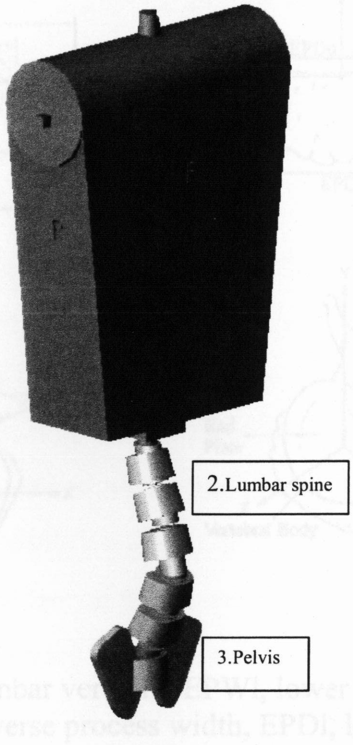

Figure 3.3: Current model components

3.4 Vertebra

There are five vertebrae in lumbar spine, L1-L5. The intervertebral discs are sandwiched between the vertebrae. These vertebrae transfer the motion and the forces through the spine.

3.4.1 Rigid Body Model of Vertebra

The physical dimensions of vertebrae for this model were obtained from Panjabi et al [43] and Tan et al [42] shown in Table 3.1 and Figure 3.4. As the Figure indicates the vertebra is very complicated with different processes and an opening for spinal cord. Since the model developed here does not use muscles and recreates the kinematics of lumbar spine using by actuators simulating the real data extracted from a motion capture, the model of vertebra can be greatly simplified.

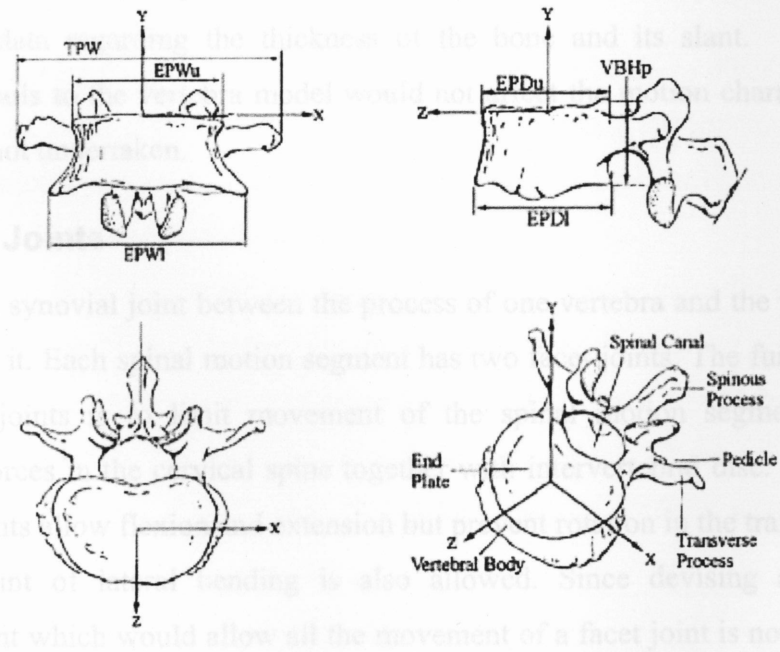

Figure 3.4: Four views of a lumbar vertebra. EPWl, lower end-plate width, EPWu, upper end-plate width, TPW, transverse process width, EPDl, lower end-plate depth, EPDu, upper end-plate depth, VBHp, posterior vertebral body height. [42, 43]

Linear Dim (mm)	L1	L2	L3	L4	L5
EPWu	42.68 +/-.44	44.90 +/-.48	46.96 +/-.39	49.35 +/-.22	48.89 +/-.40
EPWl	46.16 +/-.59	48.66 +/-.41	51.19 +/-.39	53.34 +/-.57	51.42 +/-.49
EPDu	32.32 +/-.52	33.27 +/-.60	35.15 +/-.30	36.26 +/-.23	35.82 +/-.57
EPDl	33.59 +/-.56	34.35 +/-.58	35.55 +/-.47	35.62 +/-.73	33.75 +/-.51
VBHp	26.37 +/-.49	27.15 +/-.38	25.97 +/-.46	25.42 +/-.40	23.51 +/-.71
TPW	63.05+/-1.71	75.64+/-1.81	83.99+/-2.20	79.68+/-1.66	83.92+/-2.89

Table 3.1: Mean value of measured parameters [42]

To this end, the information in each column of Table 3.1 was used to create a simple solid model of each vertebra so that the essential kinematics details such as center of gravity and body geometry are kept.

The resulting solid model for one of the vertebra is shown in Figure 3.5. The data from literature is used in order to generate the sketch on the left. The solid model is developed by using the data regarding the thickness of the bone and its slant. Adding more anatomical details to the vertebra model would not affect the motion characteristics and was therefore not undertaken.

3.4.2 Facet Joints

Facet joint is a synovial joint between the process of one vertebra and the process of the vertebra above it. Each spinal motion segment has two facet joints. The function of each pair of facet joints is to limit movement of the spinal motion segment and resist compressive forces in the cervical spine together with intervertebral disc. In the lumbar spine, facet joints allow flexion and extension but prevent rotation in the transverse plane. A small amount of lateral bending is also allowed. Since devising an equivalent mechanical joint which would allow all the movement of a facet joint is not possible, the function of these joints were modeled using an articulated disc which will be described in following sections. This will not affect the usefulness of the spine model since the joints' moments and not forces are of interest.

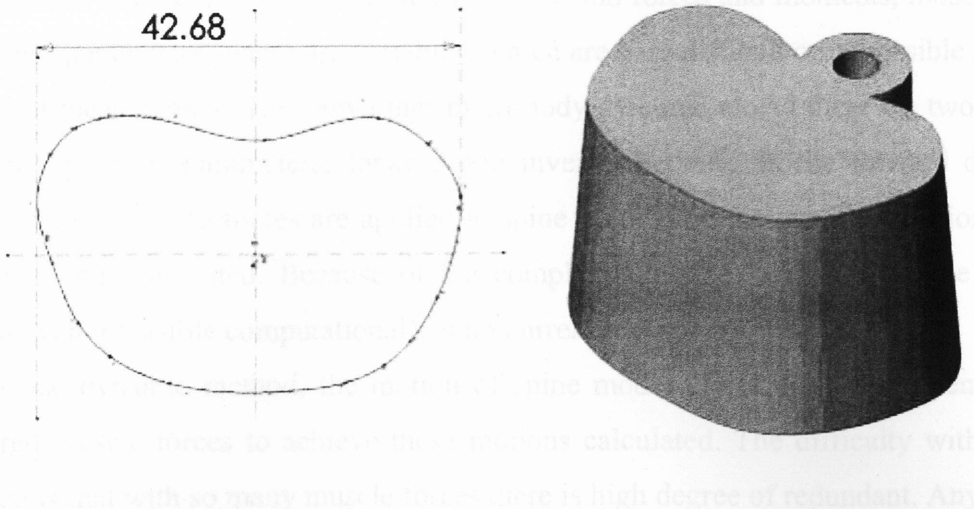

Figure 3.5: Sketch of L1 cross section created in Solidworks

3.5 Ligaments

Ligaments are fibrous bands or sheets of tissue that connect bones to other bones to form joints. Ligaments have several functions including providing stability during rest and movement, restricting excessive movements such as hyper-flexion or hyperextension, and preventing movement in certain directions.

Lumbar spine has three important ligaments, which are Supraspinous located in lumbar and thoracic region and limits the flexion, Interspinous located in lumbar region limits the flexion, and Intertransverse located in lumbar region and limits lateral bending. In the present model ligaments are replaced with torsional springs. Three of these springs stabilize rotation around x, y, and z-axis. The values used for stiffness were chosen from literature [23, 44].

3.6 Muscle Forces

In human body muscles and ligaments are responsible for the movements of spinal column. There are only 18 muscle groups involved in the motion of cervical spine [34]. These are attached to various points and create a complicated mechanical system.

The goal of a spine model is to predict joint reaction forces and moments, muscle forces and other parameters. These are variables, which are very difficult or impossible to obtain by direct measurement. Like any other multi-body dynamic model there are two ways of determining these parameters; forward and inverse methods. In the forward dynamics approach, the muscle forces are applied to spine model and the resulting motions, loads and moments calculated. Because of the complexity of the lumbar spine the forward method is not feasible computationally with current technology.

In inverse dynamic method, the motion of spine model components are given and the required muscle forces to achieve these motions calculated. The difficulty with inverse method is that with so many muscle forces there is high degree of redundant. Any attempt in solving the dynamic equations requires restrictive assumptions such as optimality and symmetry. These assumptions make the models useful in limited cases.

The model presented in this chapter, gets around this problem by focusing on reaction moments instead of reaction forces. In this case, instead of muscle forces, the resultant moment generated by the muscle forces are considered. By only considering the effect of muscle interaction in generating moments, the problem of redundancy is avoided.

The moments caused by muscle forces are implemented by using moment actuators between pairs of vertebrae. The moments' magnitudes represent the load carried by joints although it could not be used to identify compressive and shear forces. Still, the model can be of great use in many applications as will be discussed in upcoming chapters.

3.7 Intervertebral Discs

In the human spine, intervertebral discs are resilient discs that provide a number of functions including resisting compressive and shear loads, and relative motion between adjacent vertebrae, dampening the vibration and ensuring a wide range of motions. Two adjacent vertebrae have all six degrees of freedom, i.e. translation along x, y, and z-axis, and rotation about x, y, z-axis.

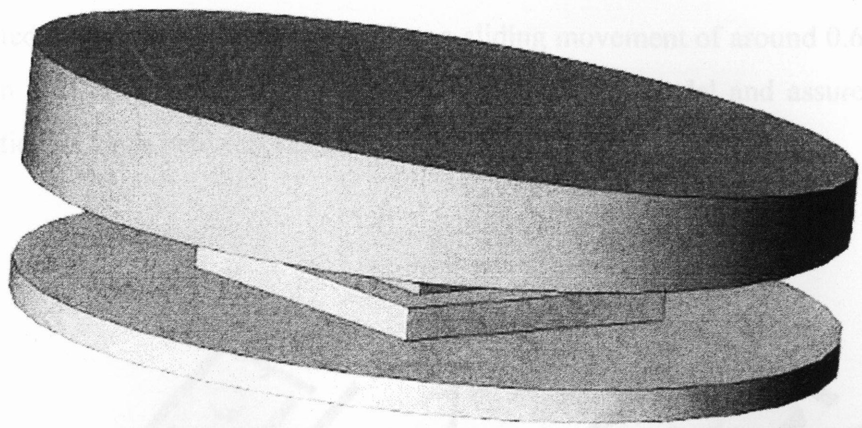

Figure 3.6: Intervertebral disc modeled in Solidworks

In the model presented here, the intervertebral discs are represented as an articulated multi-body subassembly. To ensure that this model accurately reflects the dynamics of the lumbar spine the components have to be selected carefully. Figure 3.6 shows one of the discs. The size of each disc varies depending on its location but they all have similar components.

The height of intervertebral discs varies according to the space between two adjacent discs, in order to fill up that space. The upper disc slides on top of the lower disc and four linear springs and dampers restrain its movement. The ball and socket joint enables the disc to rotate. Three torsional springs and dampers act as a substitute for ligaments and control the rotation of the disc. The stiffness and damping properties of intervertebral disc for both translational and torsional dynamics were found from literature. [23, 45]

3.7.1 Bottom plate and slider

Bottom plate shown in Figure 3.7 is where disc will be attached to one of the vertebra in the model. This plate includes a square feature that allows the modeling of translation in two planes. The motion is modeled by using a square slider. Proper kinematic restrictions are imposed such that only the two intended translation are allowed. Springs and dampers are added to these translational joints to model the behavior of intervertebral discs. The properties of springs and dampers i.e. stiffness and damping coefficient are a mean value matching those properties of the disc and have been extracted from literature [23, 45]. It is reported that on average the discs allow a sliding movement of around 0.6 mm in each direction [12]. This can be accurately represented in the model and assures the proper distribution of loads between vertebrae.

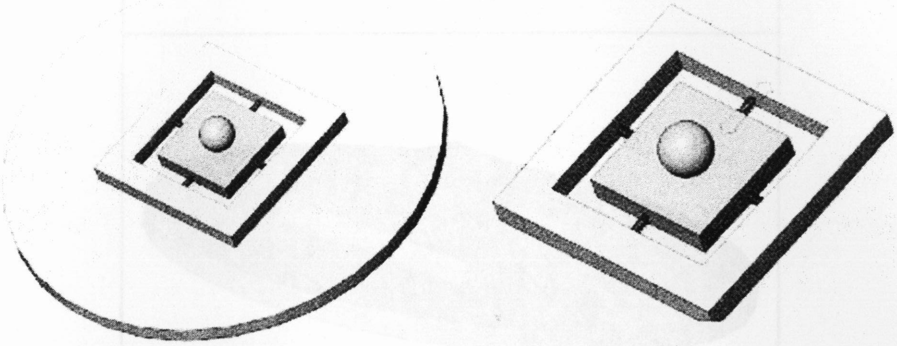

Figure 3.7: Bottom plate restricted on lower disc using mates

3.7.2 Ball and socket joint

As mentioned before the slider will provide the two-degree of freedom related to translation in the horizontal plane. In order to enable the lumbar spine to flex in sagittal plane a rotating joint is required. This is made possible by a ball and socket joint with the ball feature included in the slider as shown in Figure 3.8a. The translational degrees of freedom are transmitted to the upper part of the disc model using a top slider shown in Figure 3.8b. The combination of bottom plate, slider and top slider has a total of five degrees of freedom including two translational and three rotational degrees of freedom.
In order to model the effect of ligaments in holding the vertebrae in place torsional springs and dampers were added to the rotational degrees of freedom as shown in Figure 3.9. These are axial, horizontal, and sagittal planes. The fundamental motions of spine occur in one of these planes at a time. Lateral flexion occurs in axial plane, rotation occurs in horizontal plane, and flexion-extension occurs in sagittal plane. However, the combination of motion will occur in space and not a plane.

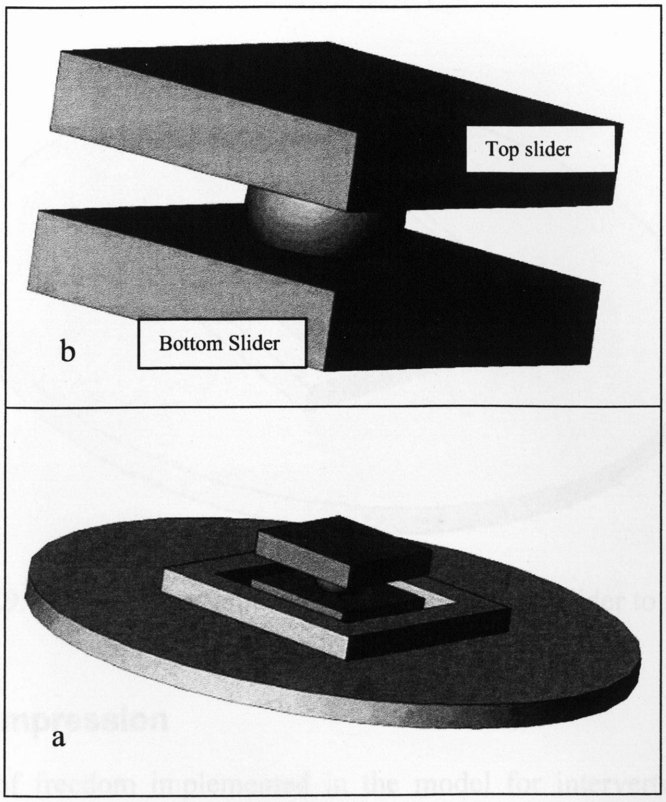

Figure 3.8: a) Joint mechanism with six degrees of freedom b) Ball and socket joint

Due to its non-homogenous properties, it is hard to come up with an accurate estimate of the stiffness and damping characteristics of the ligaments and discs. Interveltebral discs can endure compression forces up to a certain limit. Too much load will cause damage to the discs. Herniated discs are usually observed at lower levels (L4-L5 and L5-S1) of intervertebral discs because of larger range of motion.

The estimates of stiffness are shown in Table 3.2 are obtained from the works of Amirouche at al., [45] and Lavaste et al., [23].

Motion	Mean stiffness
Flexion-extension	2.6 Nm/rad
Lateral bending	2.8 Nm/rad
Axial rotation	5.2 Nm/rad

Table 3.2: Mean stiffness of a motion segment [23]

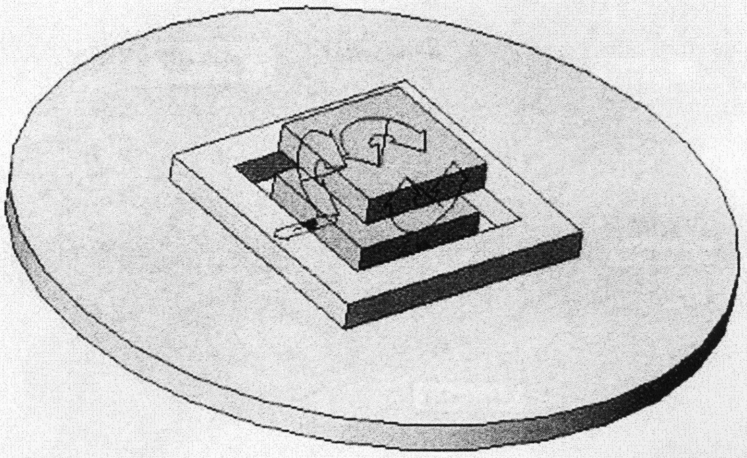

Figure 3.9: Spiral springs and dampers attach bottom slider to top slider

3.7.3 Axial compression

The last degree of freedom implemented in the model for intervertebral disc is axial deformation. This models the compression of disc and vertebral body under axial load. The axial movement is modeled using a sliding joint (see Figure 3.10). As in the case of other DOFs, stiffness and damping are added to this DOF with magnitudes gleaned from literature [23, 45].

The completed intervertebral disc is an assembly as shown in Figure 3.6. Five individual discs built on the same principle are sandwiched between vertebral bodies to create a model of lumbar spine.

3.8 Lumbar Assembly

After constructing each pair of vertebrae and the intervertebral disc in between, it is important to attach them to each other so that the curvature of spine maintains its convex and concave properties. To do so, the location of each vertebra for a healthy, typical male was obtained from Stokes et al, [27]. The resulting spine curve is shown in Figure 3.11a. The convex curve of lumbar region is crucial for the body to maintain its balance and endure loads. Spines curvature is only visible from a side view (sagittal plane) and a normal spine should look straight from a front view.

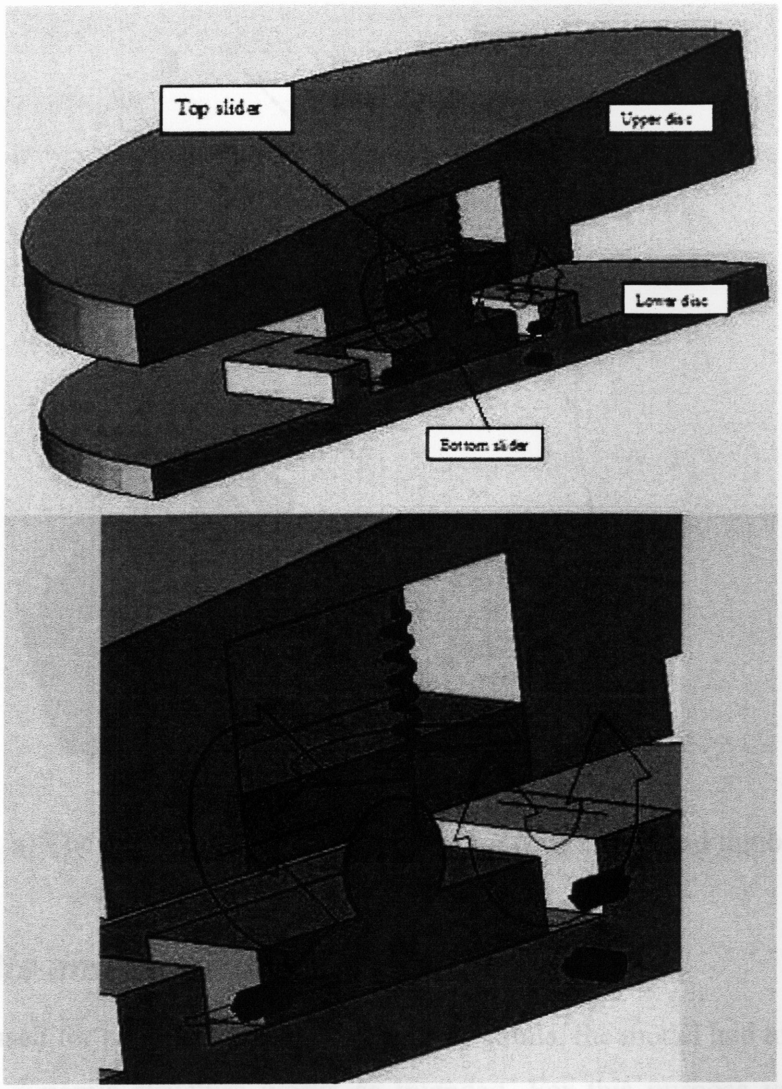

Figure 3.10: Linear spring applied in three planes

In order to position the individual vertebrae L1-L5 on this curve small cylinder were inserted at each point (see Figure 3.11b). These cylinders were used to assemble each individual model of vertebrae. Figure 3.11b also shows the model for pelvis and sacrum used as the base for lumbar spine.

Once the individual vertebrae models assembled, the spacing between them was used to properly model the intervertebral discs according to kinematics described in section 3.7. The resulting lumbar assembly can be seen in Figure 3.3.

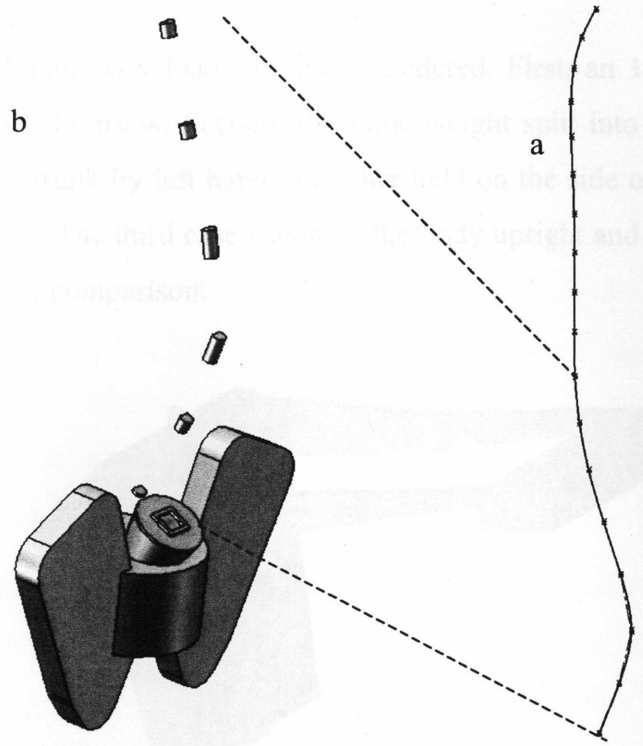

Figure 3.11: a) The curvature of spine from side view b) Pelvis and lumbar curvature

3.9 Results and verification

Before being used for predicting the loads at L1-L5 joints, the model had to be verified to make sure its dynamic structure works as intended. For this purpose various loading scenarios were tested starting with simple stationary loading to a dynamic lifting motion. The motion response for the joints between L5-S1, L4-L5, L3-L4, L2-L3, and L1-L2 were determined and validated in each loading and lifting condition.

3.9.1 Stationary Loading

The first test involved holding a load away from the body and keeping the upper body stationary as shown in Figure 3.12. There is no movement in this case and the upper body remains stationary, i.e., the body will show no roll, yaw, or pitch rotations. The muscles act so as to maintain the balance of the body. Using inverse dynamics method and knowing the load the inner forces and moments of joints from L1 to L5 can be determined.

Three cases of stationary loading were considered. First, an 11.5 kg weigh held away about 70 cm off the trunk. Second, the same weight split into two 5.5 kg portions one held away from trunk by left hand, the other held on the side of the trunk by right hand (see Figure 3.13). The third case was with the body upright and no weight held. This last case was used for comparison.

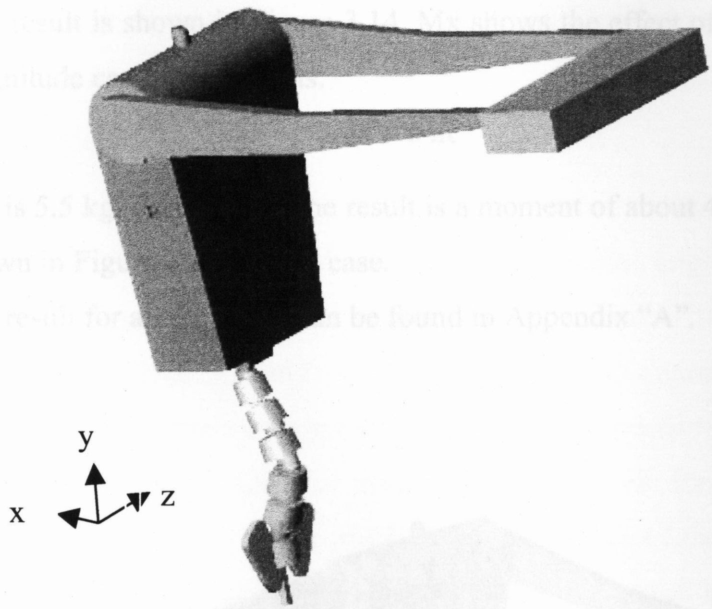

Figure 3.12: Weight 11.5 kg applied

Figure 3.14 shows the moments between L4-L5 for all three cases. Initially, the moment shows dynamic behavior by oscillating and then settling at a set value. This is due to the springs and dampers embedded in the joint and indicate an over damped system.

A simple calculation based on Newton's second law, will determine the magnitude of moment:

$$M = F.R \qquad \text{Equation 3.1}$$

Where weight is F= 11.5 kg, 5.5 kg or zero and the distance R is 70cm which will remain constant until the end of the 3sec simulation. The weight of the arms were considered zero and in this case, it is only the weight of the load acting on lumbar spine joints. The result for F=11.5 kg is a moment of about 80 Nm. This matches the result shown in Figure 3.14 for all three cases. The same moment is transmitted through all the joints as expected. This indicates that all the joints are behaving correctly in transmitting the loads. In the split case the weight has been split into two halves one held away from the body in front and the other one held at the right side of the body as shown in Figure 3.13 This change in the location of the loads will change the moment direction in a way to increase the moment in x direction. The same calculation as for the weight in front can be done for this case. The result is shown in Figure 3.14. Mx shows the effect of the load held to the side. The magnitude can be verified as:

$$M = F.R$$

where weight is 5.5 kg, R is 70 cm. The result is a moment of about 40 N.m. this matches the result shown in Figure 3.14 for Mx case.

The complete result for all the joints can be found in Appendix "A".

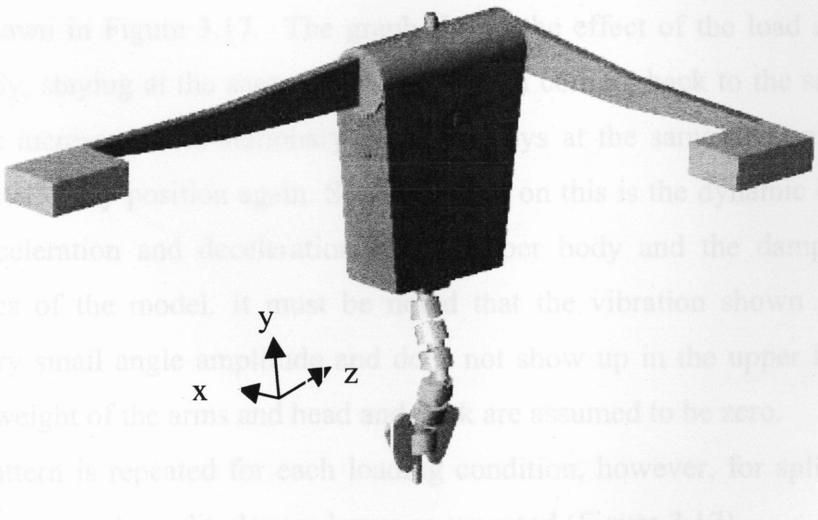

Figure 3.13: Half weight 5.5kg on right side

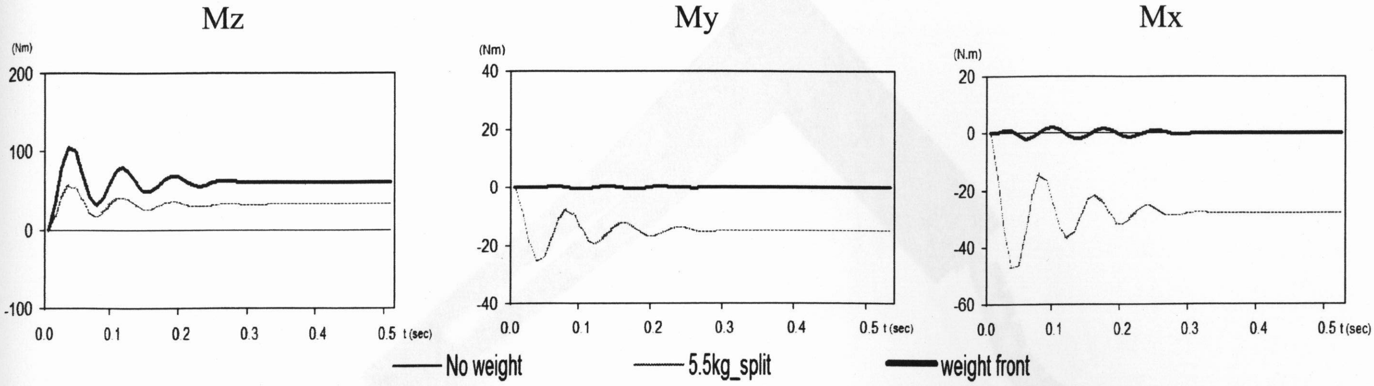

Figure 3.14: Moments x, y, z in Stationary loadings between L4-L5

3.9.2 Modeling of Flexion-Extension movement

In the flexion and extension case the upper body rotation occurs in sagittal plane alone. Figure 3.15 shows the model at the end of flexion. The rotation angles of the upper body in three directions of yaw, pitch, and roll are shown in Figure 3.16. Yaw, the rotation around z-axis will start from zero and get to a maximum of 45 degrees and return to zero again. Pitch and roll will remain constant on zero degrees, respectively.

The moments along z-axis required for moving the upper body through this prescribed motion is shown in Figure 3.17. The graph shows the effect of the load moving away from the body, staying at the same distance and then coming back to the same position. The moment increases from stationary position, stays at the same level and then goes back to the stationary position again. Superimposed on this is the dynamic behavior as a result of acceleration and deceleration of the upper body and the damped vibration characteristics of the model. It must be noted that the vibration shown in the figure represent very small angle amplitude and does not show up in the upper body rotation angles. The weight of the arms and head and neck are assumed to be zero.

The same pattern is repeated for each loading condition, however, for split weight and no-weight the moment amplitudes are lower as expected (Figure 3.17)

The complete results for flexion in all the joints can be found in Appendix "B".

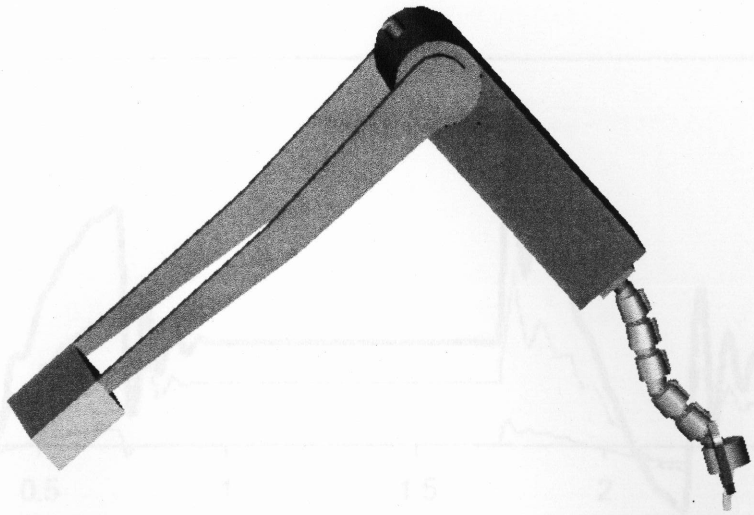

Figure 3.15: Flexion of upper body

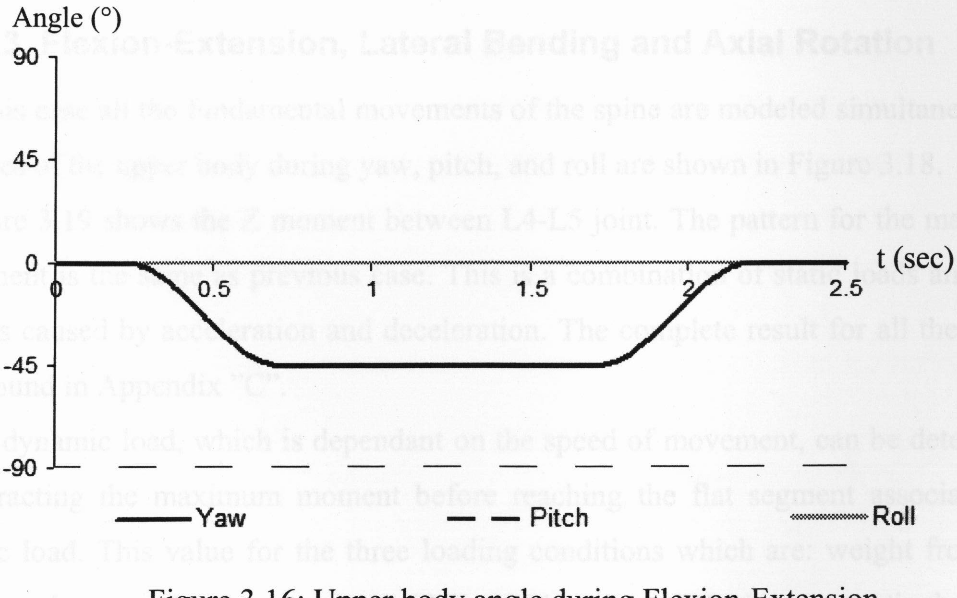

Figure 3.16: Upper body angle during Flexion-Extension

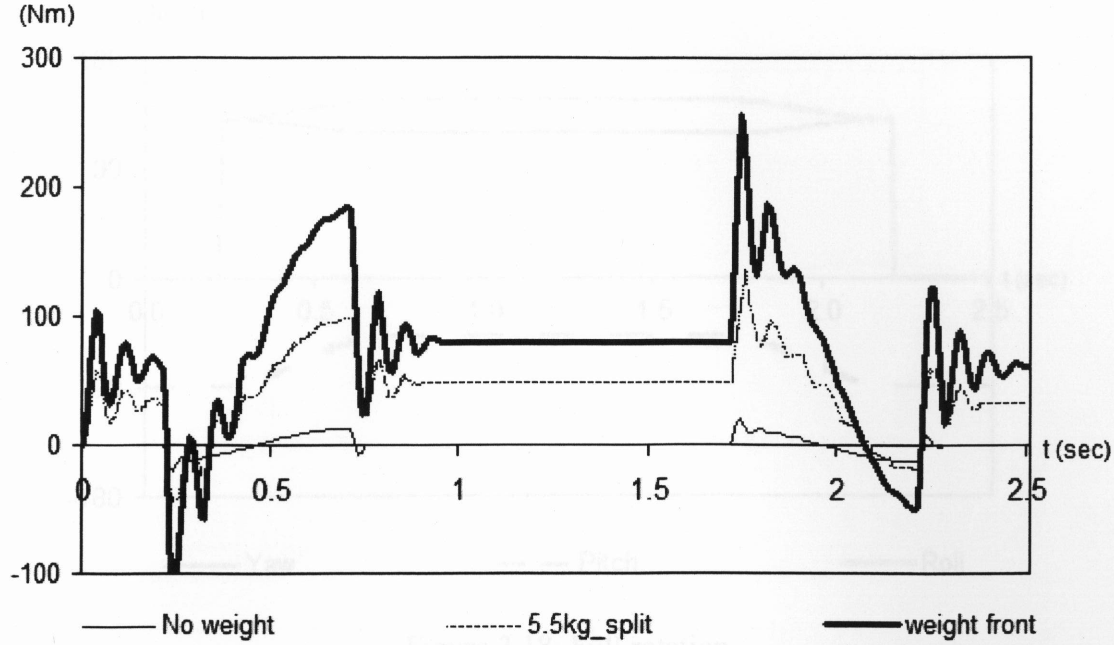

Figure 3.17: Moment Z in Flexion-Extension

3.9.3 Flexion-Extension, Lateral Bending and Axial Rotation

In this case all the fundamental movements of the spine are modeled simultaneously. The angles of the upper body during yaw, pitch, and roll are shown in Figure 3.18.

Figure 3.19 shows the Z moment between L4-L5 joint. The pattern for the magnitude of moment is the same as previous case. This is a combination of static loads and dynamic loads caused by acceleration and deceleration. The complete result for all the joints can be found in Appendix "C".

The dynamic load, which is dependant on the speed of movement, can be determined by subtracting the maximum moment before reaching the flat segment associated to the static load. This value for the three loading conditions which are: weight front, weight split, and no-weight respectively is: 49 N-m, 54 N-m, and 10 N-m, respectively.

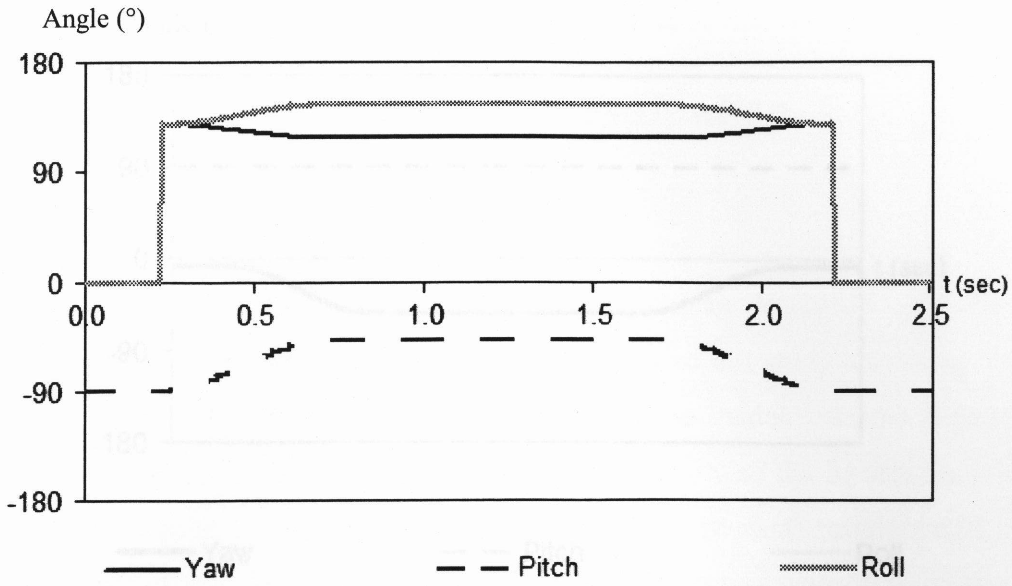

Figure 3.18: Full rotation

3.9.4 Lifting a Load

In this case the lumbar spine model is used to simulate a maneuver involving bending over, picking up a load of 11.5 kg weight and then straightening up to the standing position. Figure 3.20 shows the angles of upper body during this movement.

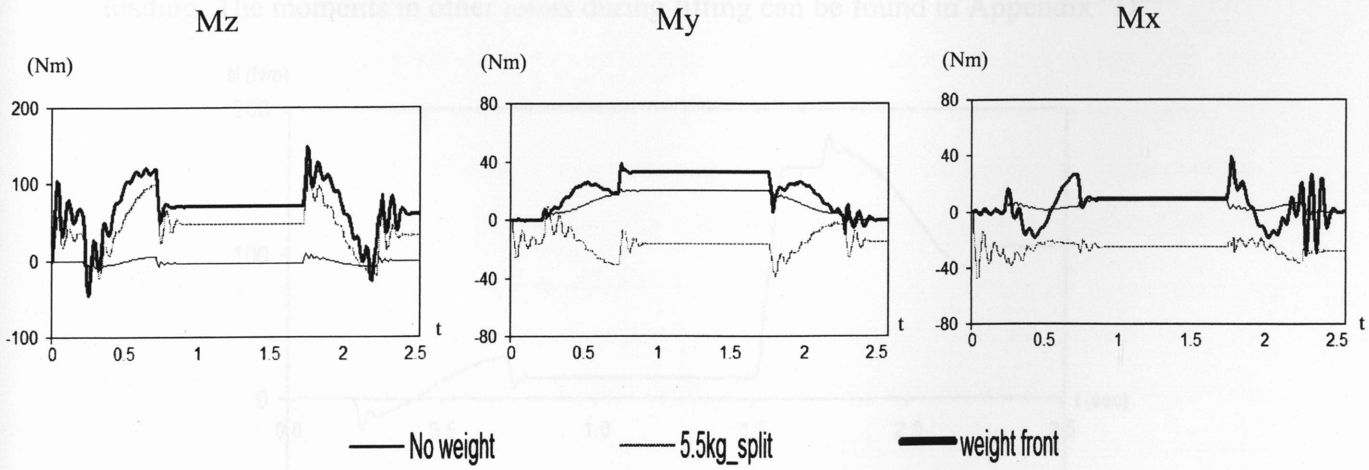

Figure 3.19: Moment between L4-L5 when model has a combined rotation (flexion-extension, lateral bending, and axial rotation)

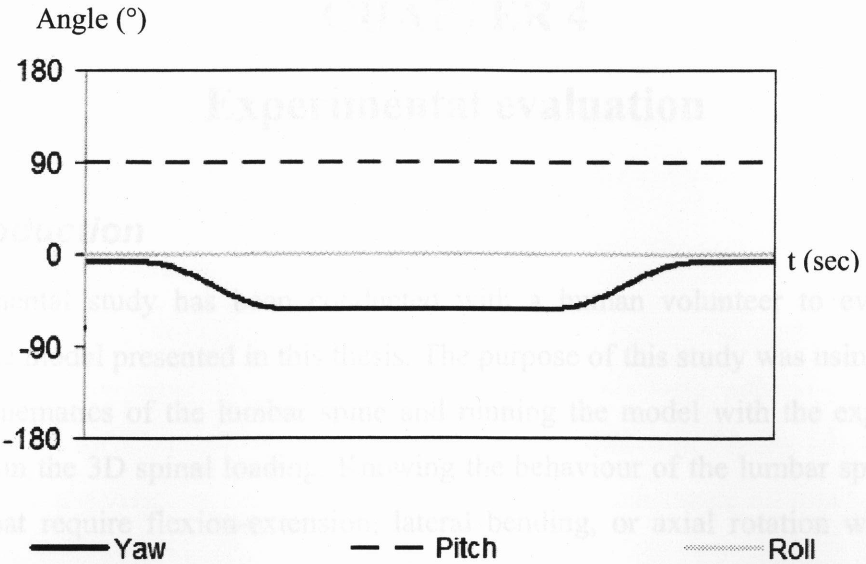

Figure 3.20: Upper body movement angles in lifting scenario

The loading starts at the 1.5th sec of the simulation. As shown in Figure 3.21 before that time the force and moment reaction of the lumbar spine is almost zero and it can be considered as the no-weight stationary condition. The damped vibration is again because of the spring damper system in the joints. After the load is picked up and lifted from the ground (at 1.5sec) an increase in the moment Z is observed. By reaching the standing position, the moments will be stabilized close to 100 Nm, which is close to stationary loading. The moments in other joints during lifting can be found in Appendix "D".

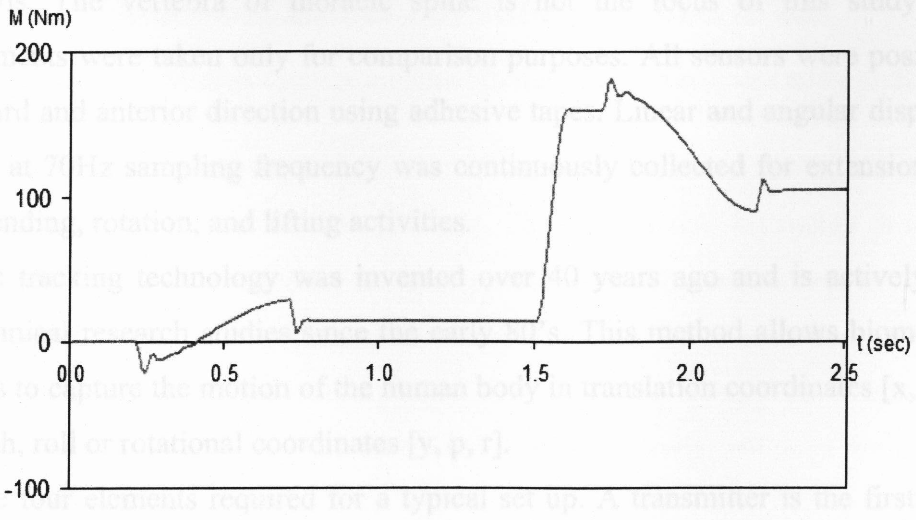

Figure 3.21: Changes of moment between L4-L5 during lifting

CHAPTER 4
Experimental evaluation

4.1 Introduction

An experimental study has been conducted with a human volunteer to evaluate the lumbar spine model presented in this thesis. The purpose of this study was using real data from the kinematics of the lumbar spine and running the model with the experimental data to obtain the 3D spinal loading. Knowing the behaviour of the lumbar spine during activities that require flexion-extension, lateral bending, or axial rotation will lead to more accurate results. For this purpose, electromagnetic sensors are used to track and record the three dimensional orientation of each vertebrae and pelvis according to the selected coordinate system. The intradiscal forces and moments were then reported and interpreted accordingly.

4.2 Experimental Setup

An electromagnetic human motion capture system, Fastrak^{TM} (Polhemus, Colchester, VT, USA), was used to determine changes in lumbar spine vertebra orientation and position in the sagittal, lateral and planar plane for the low back and pelvis. Five sensors were placed on the lumbar spine vertebra from L1 to L5 and two sensors were located on the trunk and pelvis. The vertebra of thoracic spine is not the focus of this study and the measurements were taken only for comparison purposes. All sensors were positioned in the upward and anterior direction using adhesive tapes. Linear and angular displacement (6 DOF) at 70Hz sampling frequency was continuously collected for extension-flexion, lateral bending, rotation, and lifting activities.

Magnetic tracking technology was invented over 40 years ago and is actively used in biomechanical research studies since the early 80's. This method allows biomechanical engineers to capture the motion of the human body in translation coordinates [x, y, z] and yaw, pitch, roll or rotational coordinates [y, p, r].

There are four elements required for a typical set up. A transmitter is the first element, which will create a magnetic field. The transmitter is usually installed on a wooden stand around the waist height. The second element is one or more sensors. Sensors are attached to the moving part. The third element is an interface device also called a filter. This device accepts inputs from sensors and uses a standard serial port (e.g. RS-232) to interface with a computer. And finally the last element is the computer.

4.2.1 Data Collection

A 35 years old male volunteer with no history of low back pain participated in this experiment. Sensors were attached to the body segments with adhesive tape giving enough freedom for a variety of spinal movements. As shown in Figure 4.1, the sensors were placed on the vertebra, pelvis, and upper torso. The cables were all fixed on the skin to avoid cable artefacts.

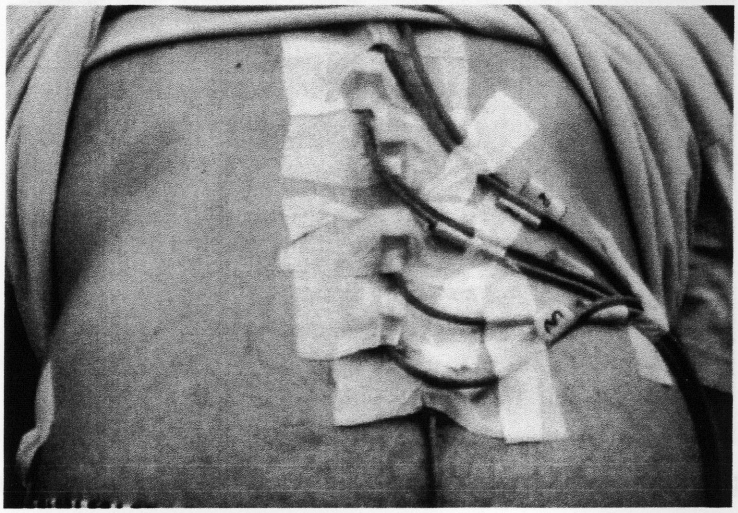

Figure 4.1: Motion sensors attached to the lumbar vertebrae and pelvis

A custom made software using Labview 8.2 (National Insturment, 2008) was applied to collect and save the data in to the computer. The subject was instructed to stand in a relax-soldier posture while holding the arms on the sides. The orientations of the sensors at this posture were set to zero and all the consequent movements were found with respect to the coordinate system accordingly. Three different conditions of trunk flexion-extension, rotation, and lateral bend as well as a lifting activity without a weight and the data were collected for simulation.

4.3 Flexion-extension

The first set of experiments was flexion-extension. The movement was initiated from an upright angle. The subject was asked to flex steadily from a straight standing position (Figure 4-2). Trunk angle in this posture was holding about 90 degrees from upright position while the lumbar spine was hold within a 45-degree in sagittal plane. Pitch and roll angles remained roughly zero.

To better match the kinematics of motion, hip joint was implemented as a spherical joint. The hip joint allows L5, which is attached to S1 and consequently to the hip, move relative to its previous vertebra.

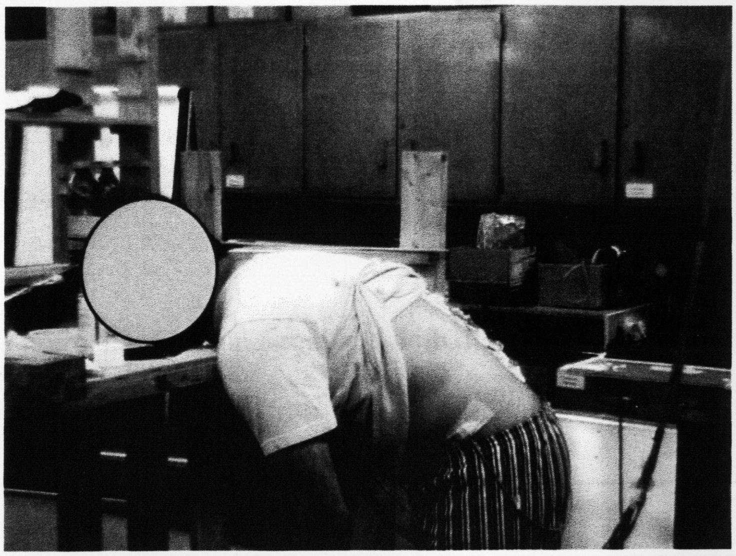

Figure 4.2: Flexion with motion sensors attached to lumbar spine

Figure 4.3 shows the position of the model after using the experimental data as input. The movement of each vertebra in the model can be controlled by its properties while uploading its coordinates from experimental data collected.

The trunk angle recorded by the Fastrak is shown in Figure 4.4. Comparing this angle to Figure 4.2 shows that upper body around T12 and L1 has reached a 55 degree angle. Figure 4.5 shows the rotation angle for each vertebra. The required moments for three sets of vertebra are shown in Figure 4.6. The noise in the result reflects the source of input, which is the fastrack system.

Figure 4.3: Flexion simulated with data from experiment

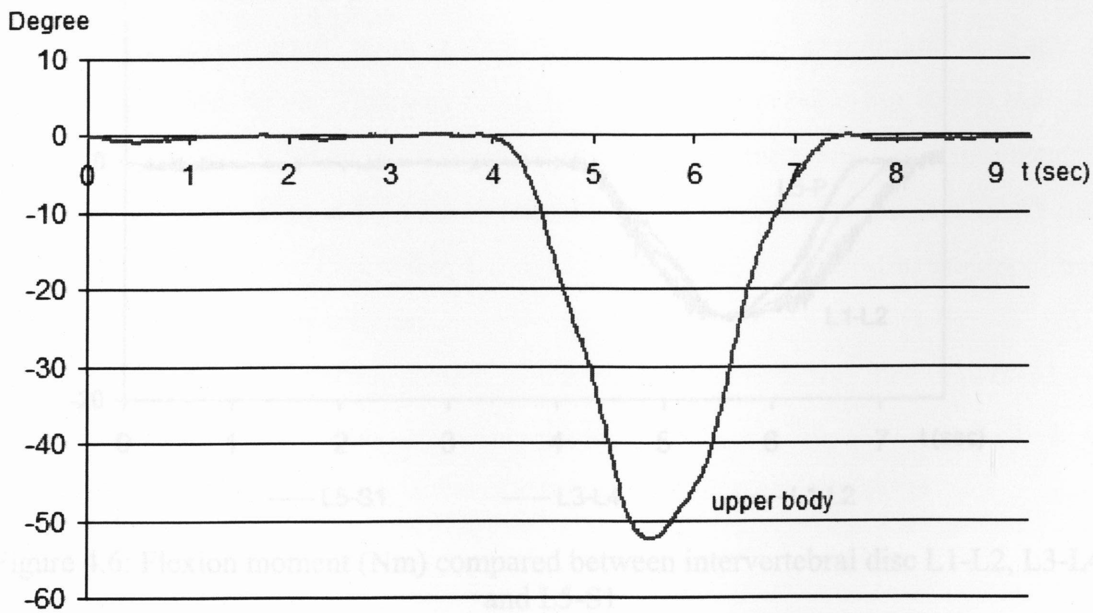

Figure 4.4: Angular displacement of upper body during flexion

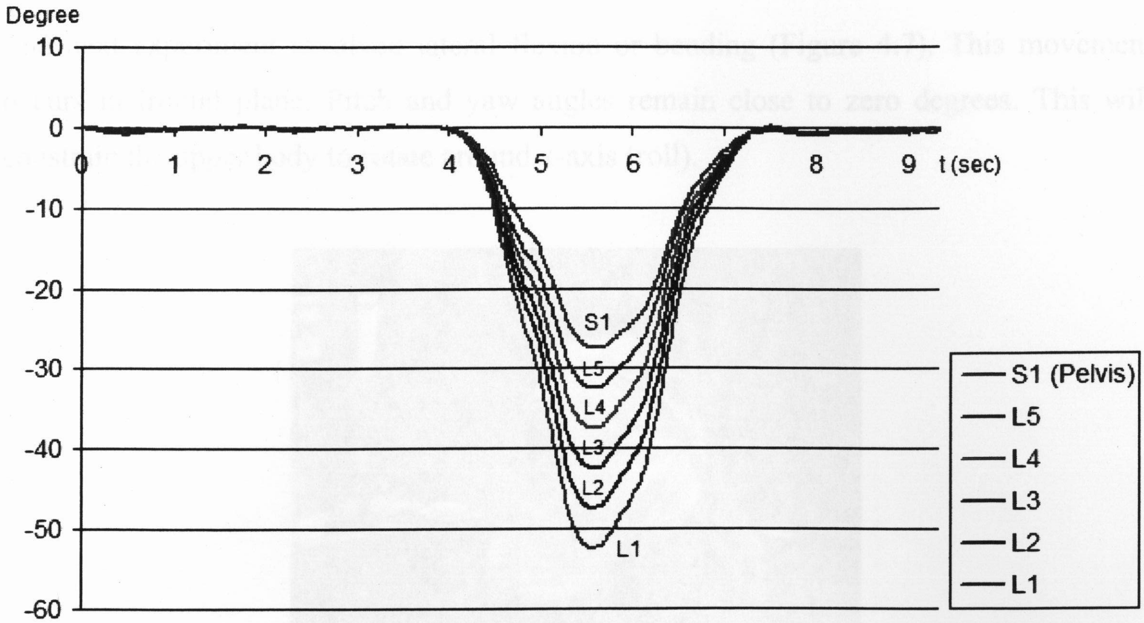

Figure 4.5: Global angular displacement from L1 to S1 (Pelvis) during flextion

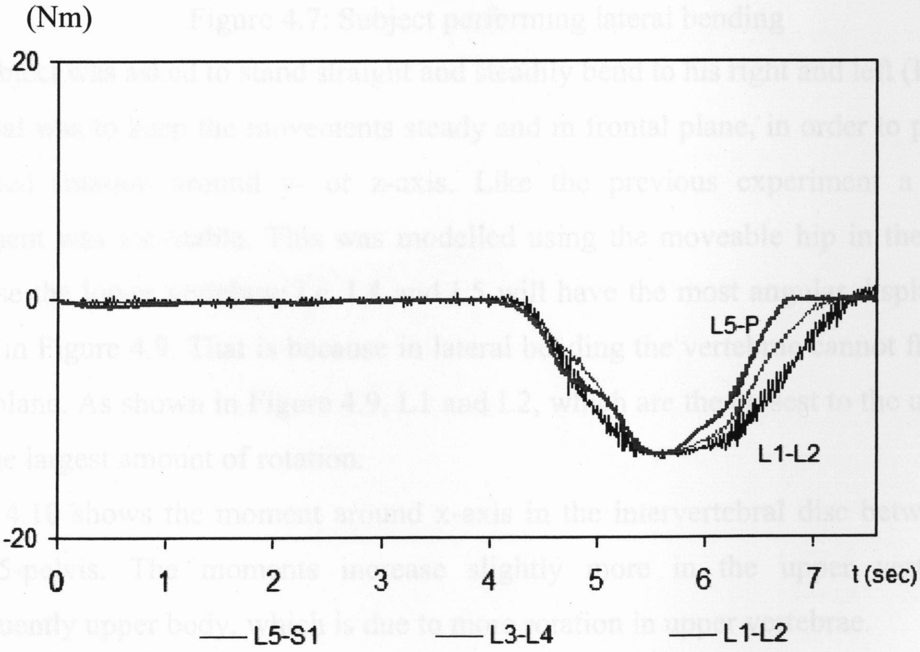

Figure 4.6: Flexion moment (Nm) compared between intervertebral disc L1-L2, L3-L4, and L5-S1

4.4 Lateral bending

The next experiment involved lateral flexion or bending (Figure 4.7). This movement occurs in frontal plane. Pitch and yaw angles remain close to zero degrees. This will constrain the upper body to rotate around x-axis (roll).

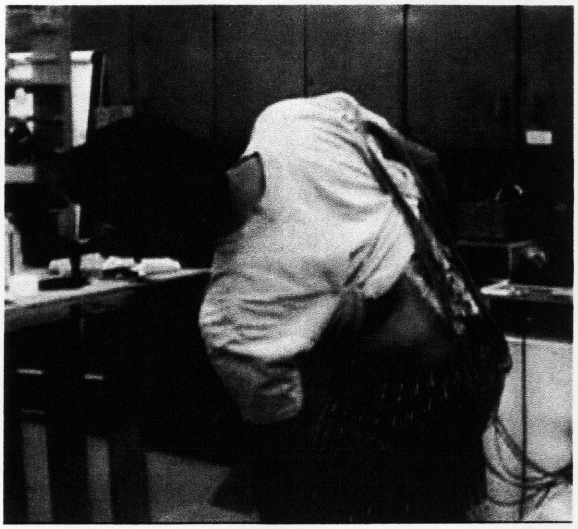

Figure 4.7: Subject performing lateral bending

The subject was asked to stand straight and steadily bend to his right and left (Figure 4.8). The goal was to keep the movements steady and in frontal plane, in order to prevent any unwanted rotation around y- or z-axis. Like the previous experiment a slight hip movement was inevitable. This was modelled using the moveable hip in the model. In this case the lower vertebrae i.e. L4 and L5 will have the most angular displacement as shown in Figure 4.9. That is because in lateral bending the vertebrae cannot freely rotate in the plane. As shown in Figure 4.9, L1 and L2, which are the closest to the upper body, have the largest amount of rotation.

Figure 4.10 shows the moment around x-axis in the intervertebral disc between L1-L2 and L5-pelvis. The moments increase slightly more in the upper vertebrae and consequently upper body, which is due to more rotation in upper vertebrae.

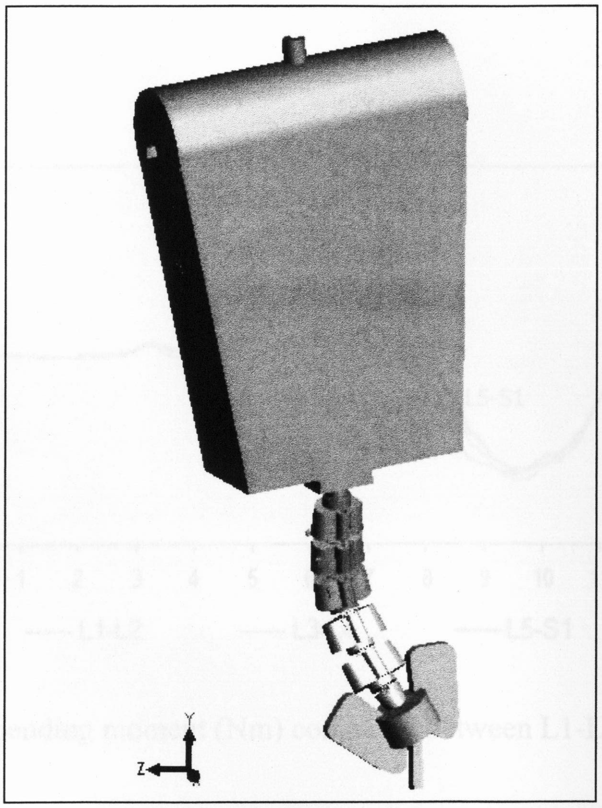

Figure 4.8: Lateral bending in the model using the experiment data as input

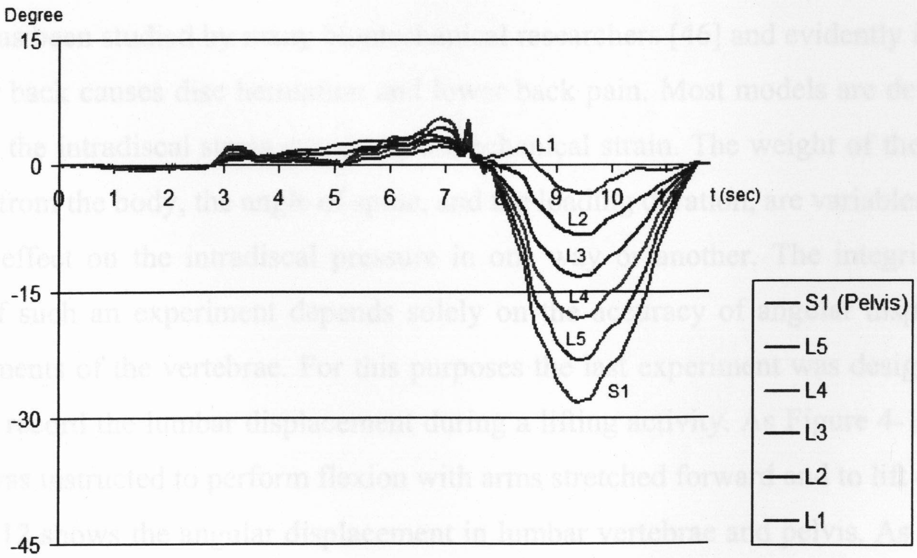

Figure 4.9: Global angular displacement from L1 to S1 (pelvis) during lateral bending

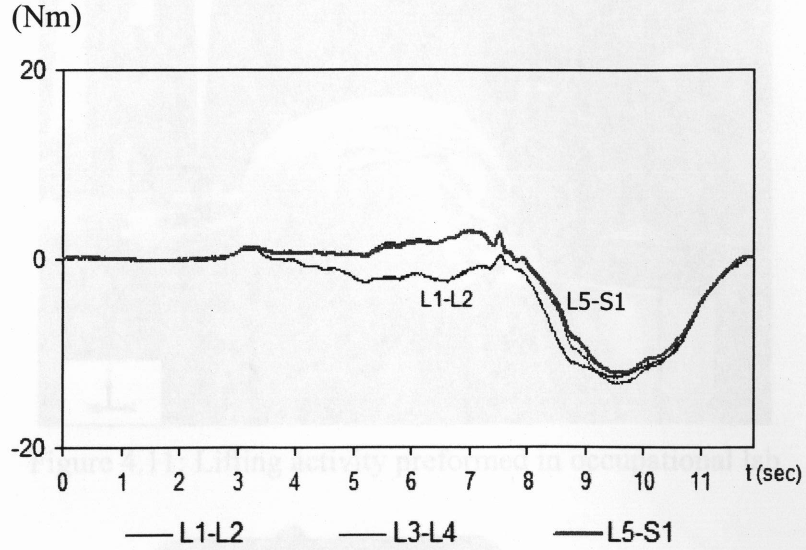

Figure 4.10: Lateral bending moment (Nm) compared between L1-L2 and L5-S1 (Pelvis)

4.5 Lifting

Lifting has been studied by many biomechanical researchers [46] and evidently its effects on lower back causes disc herniation and lower back pain. Most models are designed to calculate the intradiscal stress caused by a mechanical strain. The weight of the load, its distance from the body, the angle of spine, and the loading duration, are variables that can have an effect on the intradiscal pressure in one way or another. The integrity of the results of such an experiment depends solely on the accuracy of angular displacement measurements of the vertebrae. For this purposes the last experiment was designated for lifting to record the lumbar displacement during a lifting activity. As Figure 4-11 shows, subject was instructed to perform flexion with arms stretched forward and to lift a weight. Figure 4.12 shows the angular displacement in lumbar vertebrae and pelvis. As expected most of the rotation happens in upper vertebrae i.e. L1 and L2

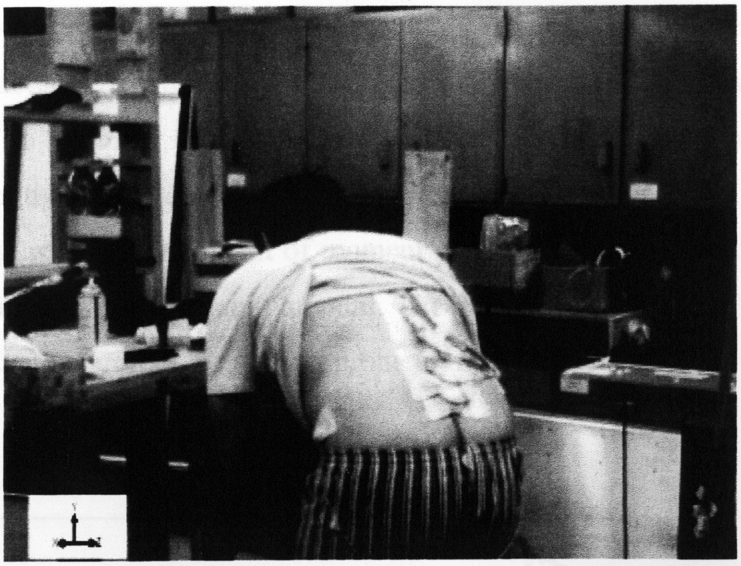

Figure 4.11: Lifting activity preformed in occupational lab

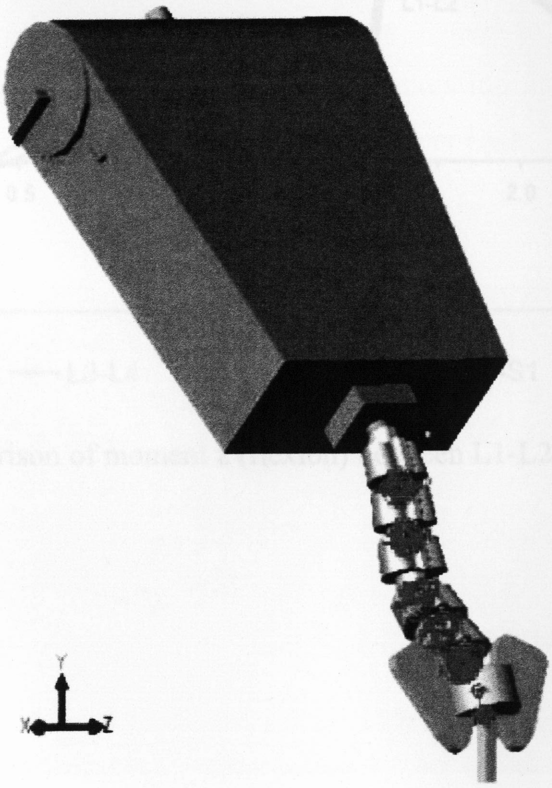

Figure 4.12: Lifting activity simulated with experimental data

Figure 4.13 shows the changes in moment z during lifting with respect to time. The moments shown are between L1-L2, L3-L4, and L5-S1 (Pelvis). The moment increases at

1.5sec when the force is applied. When the weight is at its furthest distance from the body the moment is about 170 Nm. The moment gradually sets back to a lower amount and stabilizes close to 100 Nm, when the body is back to the standing position and the distance between the weight and body is at its lowest.

As shown in Figure 4.13 the amount of moment increases as it goes further down to the pelvis where it reaches its maximum amount. The disc between L5-S1 (Pelvis) bares the most of the load and therefore is the most prone to injury.

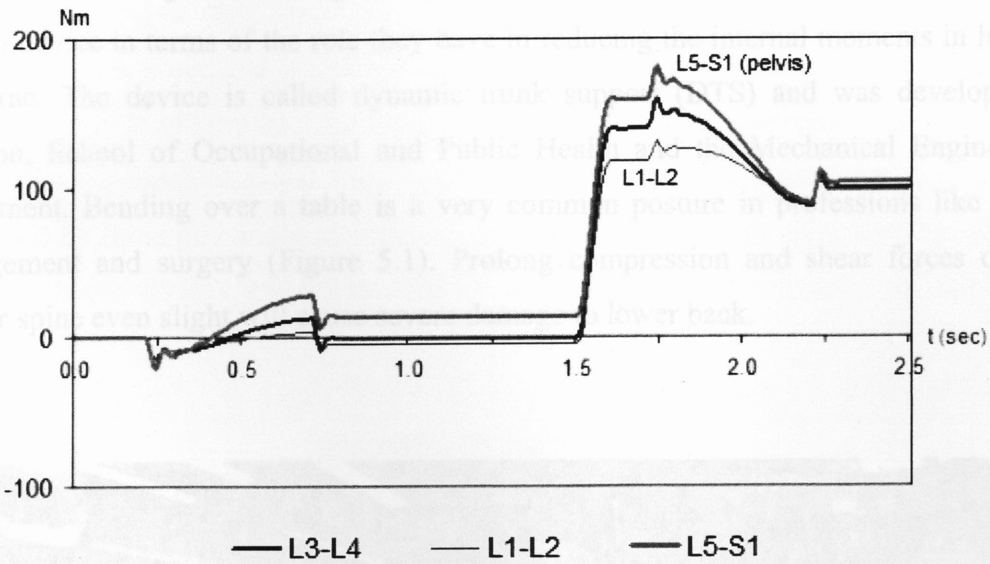

Figure 4.13: Comparison of moment z (flexion) between L1-L2, L3-L4, and L5-S1

CHAPTER 5

Application in evaluating ergonomic devices

5.1 Introduction

In this chapter, one of the applications of the present model in human factors is discussed. This application refers to a device that is designed to be ergonomically useful in daily activities like lifting and bending. The present model is able to analyse and evaluate this type of device in terms of the role they have in reducing the internal moments in lumbar vertebrae. The device is called dynamic trunk support (DTS) and was developed at Ryerson, School of Occupational and Public Health and the Mechanical Engineering department. Bending over a table is a very common posture in professions like waste management and surgery (Figure 5.1). Prolong compression and shear forces on the lumbar spine even slight will cause severe damage to lower back.

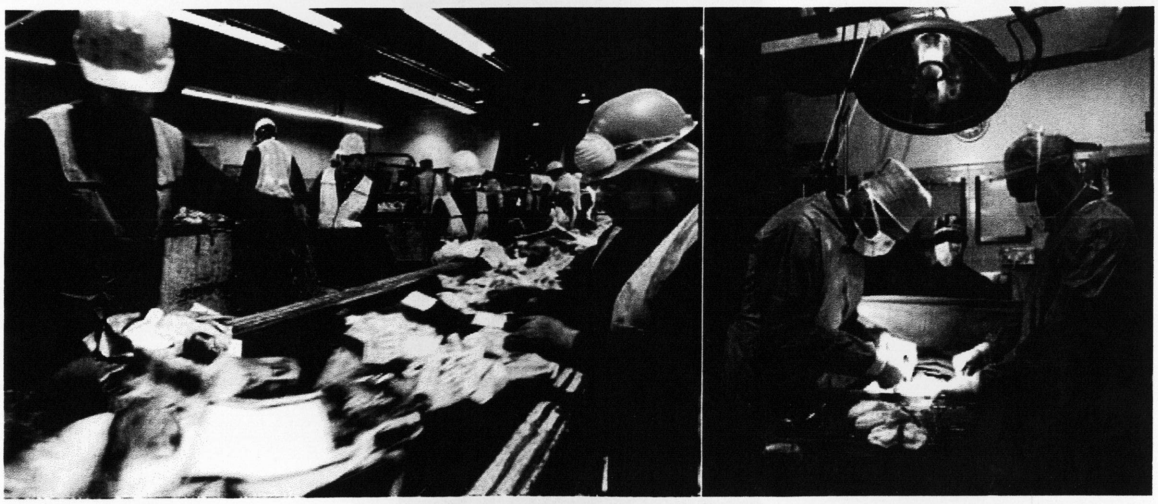

Figure 5.1: Professions like waste management and operation room are the focus of this study

5.2 *Dynamic Trunk Support (DTS)*

DTS was developed to reduce spinal loading and is based on the same principles as a backrest on an office chair except the support is continuous through movement and is provided to the anterior chest. The device has three main components; a support plate, post and a mechanism that provides attachment to the workstation (see Figure 5.2). Since leaning through the anterior chest is a new idea, the shape of the support plate was determined based on anatomical factors including transferring the load through the sternum (Abdoli et al. [48]). The device is designed to partially support the trunk. This has two advantages; first static positioning of the lumbar spine joints and associated tissue creep is avoided and the device can be used with dynamic work where the trunk is repeatedly recruited into the reaching movement pattern (Damecour et al. [50]). The mechanism has been changed from a single plane hinge to a multidirectional joystick controlled by three guy wires and springs replicating human back muscles (Damecour et al. [51]). These springs are responsible for providing enough resistance to support the trunk.

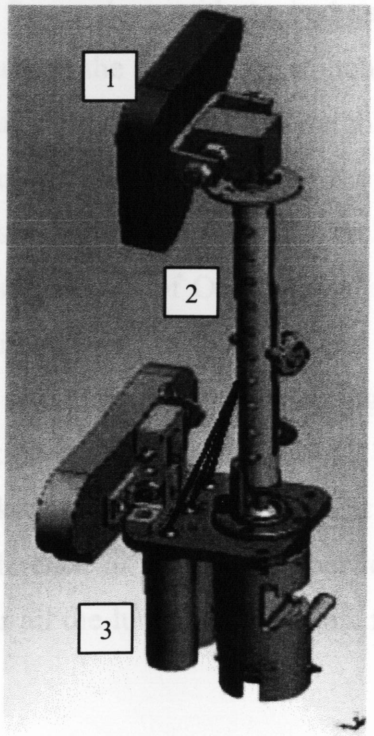

Figure 5.2: DTS consists of (1) support plate (2) post (3) fixture mechanism

5.3 *Flexion and lifting with DTS*

To study the effect of DTS on lumbar spine, three-dimensional model of DTS was imported into the model. It should be noted that this did not require any additional modeling effort. Basically, the solid models used during the design of DTS were used for the study. To evaluate the device, the 3-D model of the DTS is connected to the lumbar spine model. The effect of using DTS was then investigated by simulating typical movements including flexion-extension.

By using the lumbar spine model, it is possible to investigate the effect of various parameters on the way DTS and similar devices interact with lumbar spine. Comparing the results from simulation runs with and without DTS will also help determine the efficacy of the DTS in reducing the load on lumbar joints.

5.3.1 Flexion

The model including the DTS was run for simulating the flexion movement. Figure 5.3.a shows how DTS is attached to the chest while the upper body is in a standing position. The effect of DTS on the body is zero in a standing position. As the trunk starts to flex forward, (Figure 5.3.b), DTS flexes along with the body and allows the operator to lean on it (Figure 5.4). In this situation the springs are stretched and the reaction of springs creates a counter moment that can act against the weight of the trunk and eliminate a portion of the load on lumbar vertebrae. The size of the load taken up from vertebrae should depend on the stiffness of the springs used. The actual experimenting with the DTS was performed at Ryerson School of Occupational and public Health. Figure 5.4 shows two volunteers. DTS is attached to the table and the volunteer leans on it and performs manual activities. A sample result is shown in Figure 5.5. This Figure shows the moment around z-axis for flexion with DTS and without it between L5 and L4. The effect of DTS can be seen by a marked reduction in the moment load carried by the joint. This amounts to about 50 Nm difference in the moment between L5 and L4 in the interval 0.75-1.75 s. Similar results for all the lumbar spine joints from L1 to pelvis are included in Appendix “E”.

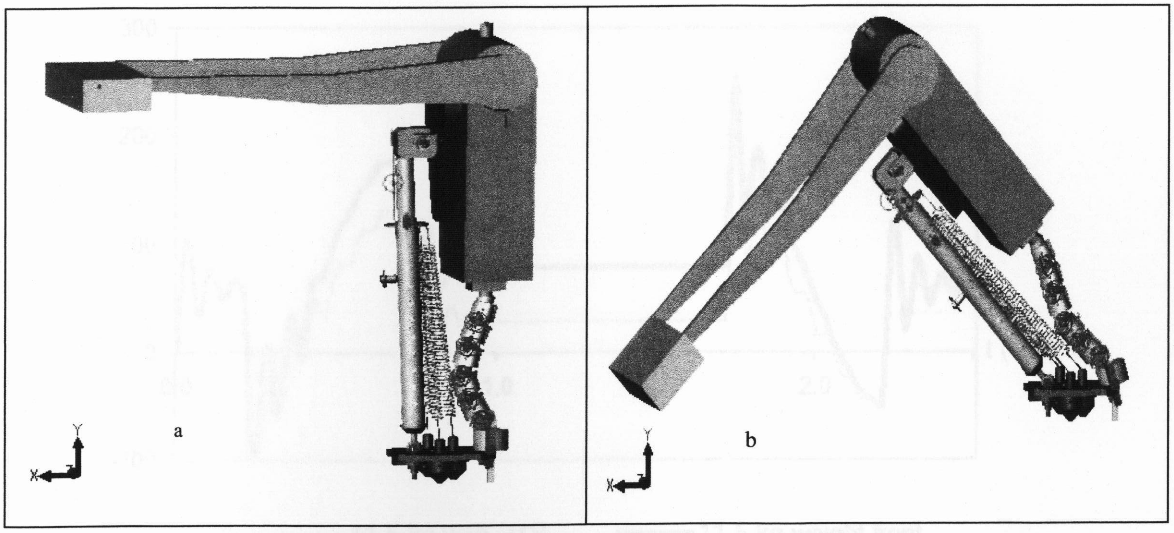

Figure 5.3: a) DTS attached to the body in standing position b) Body leaning on DTS during flexion

Figure 5.4: Symmetrical bend and side reach with two different versions of DTS

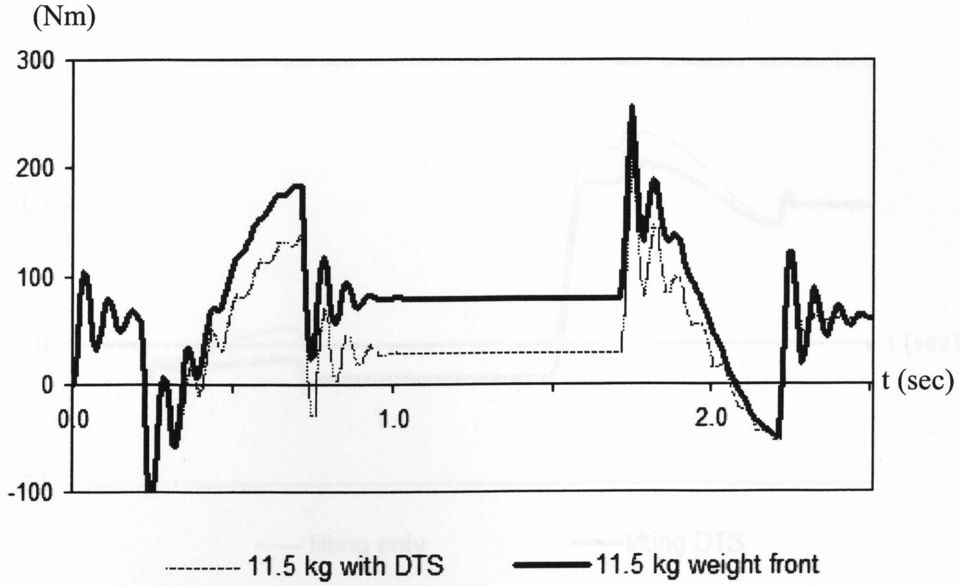

Figure 5.5: Moment –Z comparison between flexion with and without DTS for L5-L4

5.3.2 Lifting

As a second investigation the effect of DTS for a lifting scenario was simulated. The DTS and body will follow the same path as flexion. The model starts from the stationary position without any weight. Lifting the 11.5 kg weight starts at the time of maximum flex, which occurs at 1.5 seconds from the start (Figure 5.6). The difference in moment around z between L5 and L4 in this case is 25 Nm. The reduction in uptake of the load is a result of using different stiffness values for DTS. The spring used in DTS for flexion case had a stiffness of 10000 N/m, and for lifting case the stiffness was changed to 5000 N/m. by changing the stiffness of the springs used in DTS, we can control the amount of push back on the trunk. It can also vary based on the weight of the user. The complete result of lifting activity with and without DTS for all the joints can be found in Appendix "F".

5.4 Results from experimental data

The effect of DTS was also investigated in case of data collected experimentally. This should clarify whether a device such as DTS can show beneficial results.

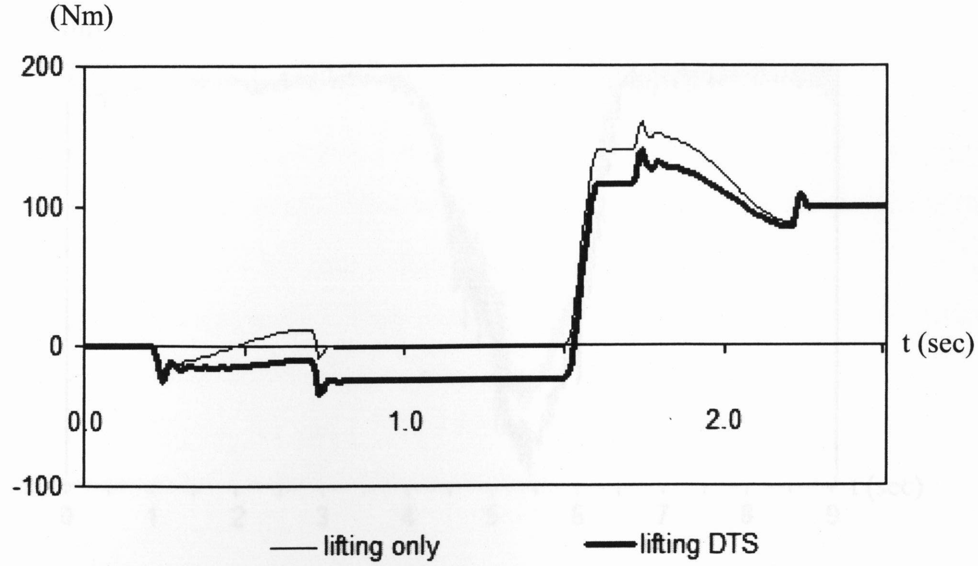

Figure 5.6: Lifting activity with and without DTS between L5-L4. DTS could reduce 25 Nm from original moment

5.4.1 Flexion with experimental data

For this round of simulation, the experimental data collected, using the setup described in Chapter 4 was used. DTS was added to the lumbar spine model and the model was run for between 9 to 11 seconds.

Figure 5.7 shows the flexion moment of L5–S1 vertebra during the flexion. During this motion the upper body bends over and then straightens up. While the sensors have picked up a lot of noise, the effect of DTS is obvious. The result in Figure 5.7 shows an improvement in the internal forces after using the DTS. The moment shows 50 Nm reduction, which could be improved by increasing the stiffness of the DTS springs. The springs can be easily changed over both in the model and on the DTS machine as shown in Figure 5.8. The stiffness for all the three springs (left, right, and center) of DTS was set to 10000 N/m in the model.

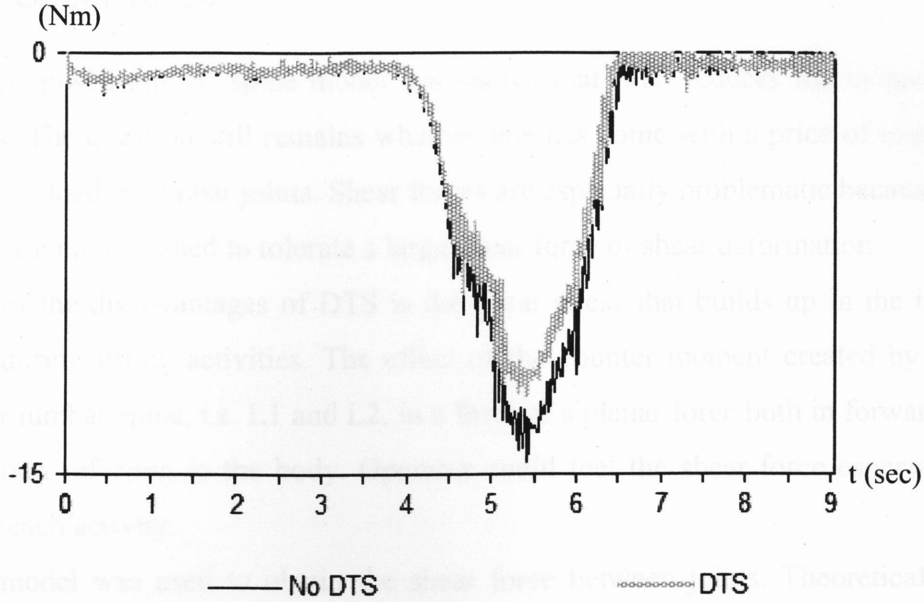

Figure 5.7: Flexion simulated with experimental data with and without DTS in L5-S1

Figure 5.8: The stiffness of the spring could be changed very quickly on DTS

5.5 *Shear force*

The proposed lumbar spine model has shown that DTS reduces the moments in spinal joints. The question still remains whether this has come with a price of increasing other types of load in lumbar joints. Shear forces are especially problematic because invertebral discs are not designed to tolerate a large shear force or shear deformation

One of the disadvantages of DTS is the shear stress that builds up in the intervertebral disc during lifting activities. The effect of the counter moment created by DTS on the upper lumbar spine, i.e. L1 and L2, in a form of a planar force both in forward and lateral direction referring to the body. Operator could feel the shear force especially during a side reach activity.

The model was used to obtain the shear force between joints. Theoretically the stress should happen where the top and bottom slider coincident, therefore reading the forces that develop in each linear spring should lead to finding the shear stress.

Referring back to the model anatomy there are three linear springs involved in the movements of each disc. The orientation of these spring are, forward, lateral (side), and vertical. Since the vertical spring is involved in compression and not shear, force will only distribute in forward and lateral (side) directions. The force magnitude in these springs can be determined by running the model. The total shear is calculated by adding the forward and lateral (side) force vectors as:

$$F = \sqrt{(f^2 + s^2)} \qquad \text{Equation 5.1}$$

Where, F is the magnitude of force, f is force in forward direction, and s is force in lateral direction.

The results shown in Table 5.1 indicate an increase in the shear stress between L1-L2 and L2-L3 while a decrease in other joints. Table 5.2 shows that shear force in Side direction could be neglected because of its low value.

Vertebrae	Shear force (with DTS)	Shear force (without DTS)
L1- L2	116	9
L2- L3	106	20
L3- L4	26	90
L4- L5	15	99
L5- S1	23	94

Table 5.1: Shear force in lumbar intervetebrae. All the units are Newton (N)

	DTS		NO DTS	
	F	S	F	S
L1-L2	116	3	9	3
L2-L3	106	3	20	3
L3-L4	26	3	90	3
L4-L5	15	3	99	3
L5-S1	23	3	94	3

Table 5.2: Shear force broken down in two directions "Forward" and "Side"

It should be noted that in a perfect lifting or bending activity, soft tissue counterbalance the load and body weight by applying continuous force. In this ideal situation, shear stress experienced by joints should be zero. The idea of using the DTS is to help the soft tissue a) apply less force to counterbalance shear and compression forces and therefore b) prolong the period of applying the force. The stiffness of the DTS should be adjusted so that it partially supports the trunk. In this case, the device acts like the soft tissue within the spine column. According to a study by Abdoli et al. [49] the maximum acceptable rate of the applied support on the chest for seven hours intermittent leaning activities is about 10 to 15 percent of the upper body weight. When this force is applied to the chest, the lumbar muscle activity is reduced about 30 to 55 percent (Damecour et al. [50]). Although the current simulation shows that the anterior posterior shear stress on the spine is increased, but biological evaluations during symmetrical (Damecour et al. [50]) and asymmetrical (Damecour et al. [51]) leaning forward activities suggests that the resultant force on the disc should be reduced accordingly. Currently another model is under investigation (Fairman et al. [52]) to investigate this assumption.

CHAPTER 6

Conclusion, contribution, and future work

6.1 Contribution

- Development of a simple, fast rigid body lumbar spine model
- The model can be used for quantitative assessment of joint moments and qualitative assessment of joint forces
- Preliminary analysis of an ergonomic device was demonstrated
- The model uses widely available engineering software facilitating sharing and comparison of results
- The model can easily interface with other CAD models and devices with no further modifications

6.2 Conclusion

The present model has been used for different tasks. First, it was used to calculate the internal loads on lumbar vertebrae. The data have been plotted and compared between each vertebrae and disc. The model could successfully predict the moment created between L4-L5 under various load conditions. An experiment helped to validate the application of DTS in ergonomics. Data were acquired from experiments in the laboratory and the results were used as input to the model. The earlier results from trials with volunteers proved that DTS could improve the amount of pressure on the lower back. Results from the model confirm the earlier trials and showed reduction in the moment on each vertebra during different activities.

The model also revealed that DTS causes increase in shear stress between L1-L2 and L2-L3 discs. The lack of muscles in the model makes it difficult to suggest whether the shear force could be eliminated or not.

6.3 Future work

This project can be expanded in a number of directions. Other parts of the spine including the cervical spine can be added to the model to create a more complete picture of the way loads are transmitted.

Muscles can also be added to the current model in order to evaluate forces in the joints. This step will require the modeling of the muscle behaviour and accurate mapping of attachment points.

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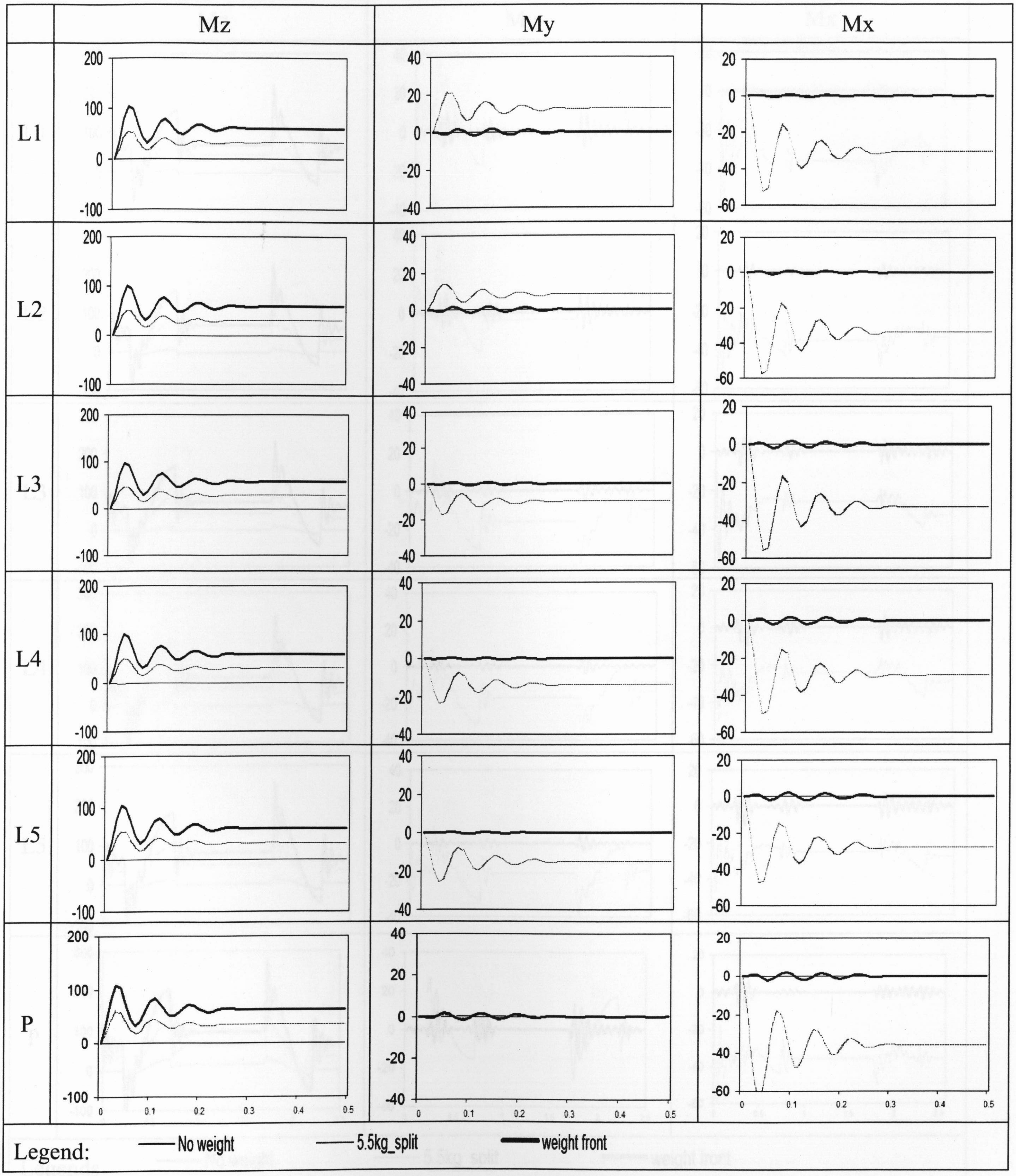
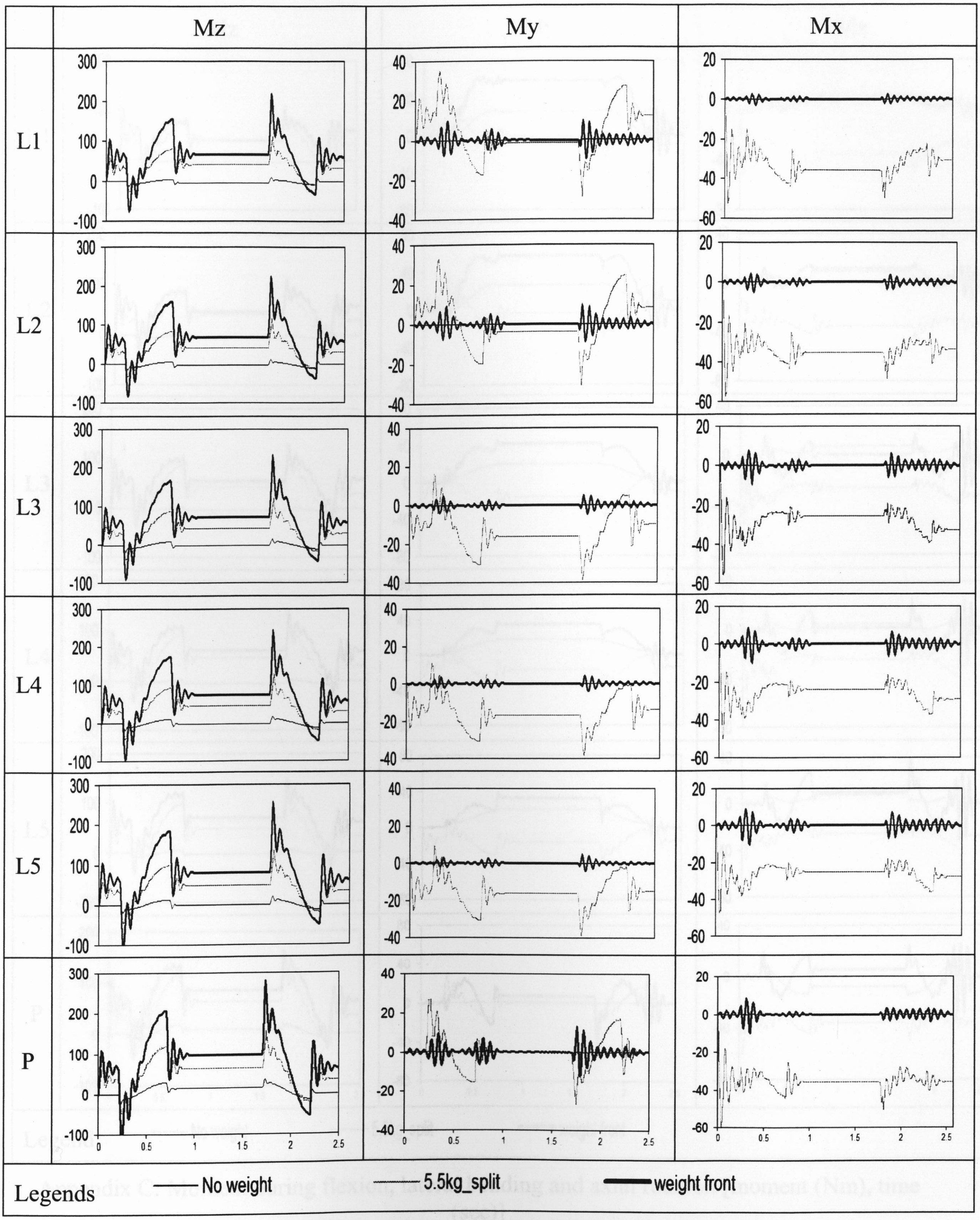
Appendix "A"



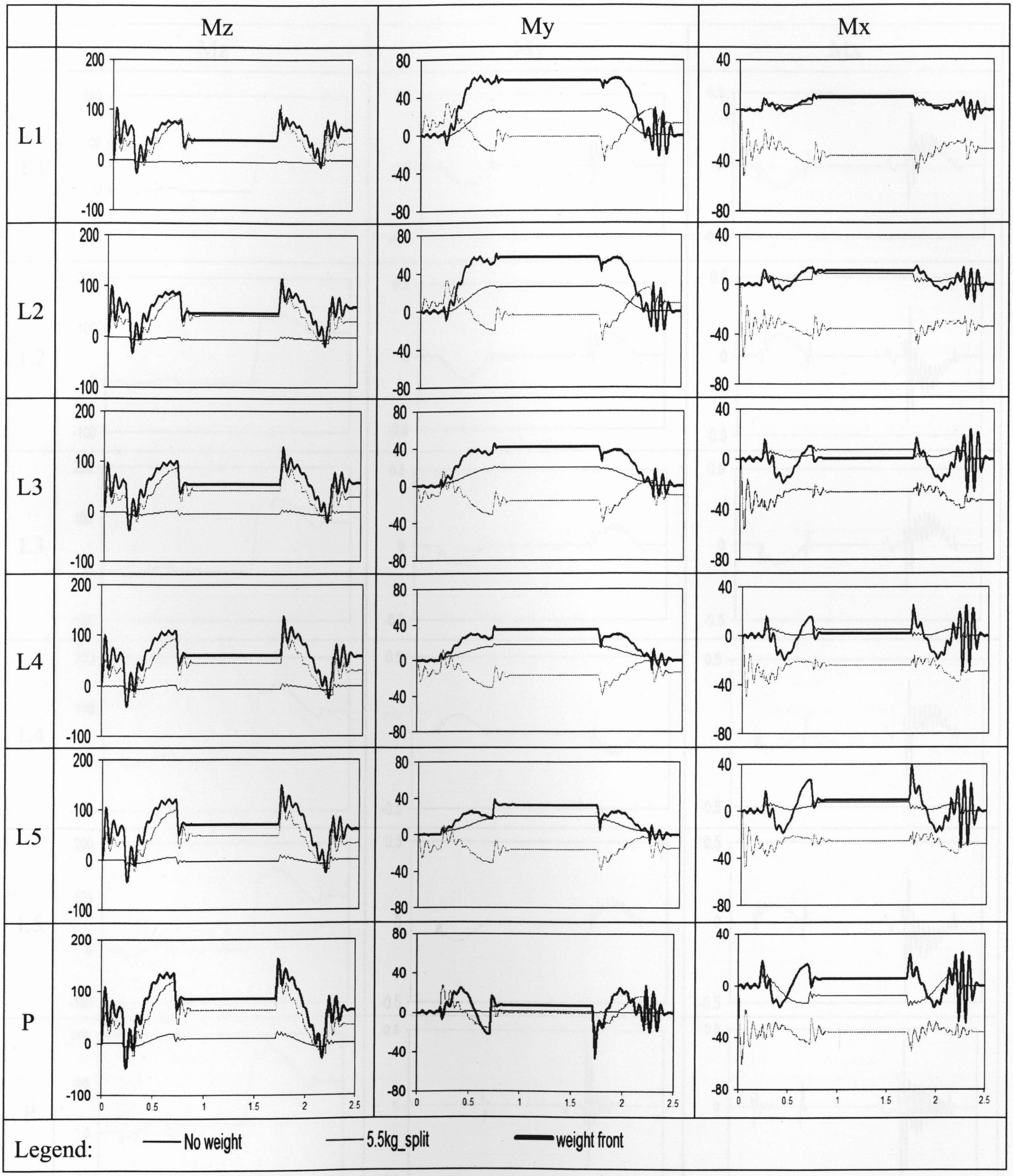
Appendix A: Moment during stationary position [Moment (Nm), Time (sec)]

Appendix “B”



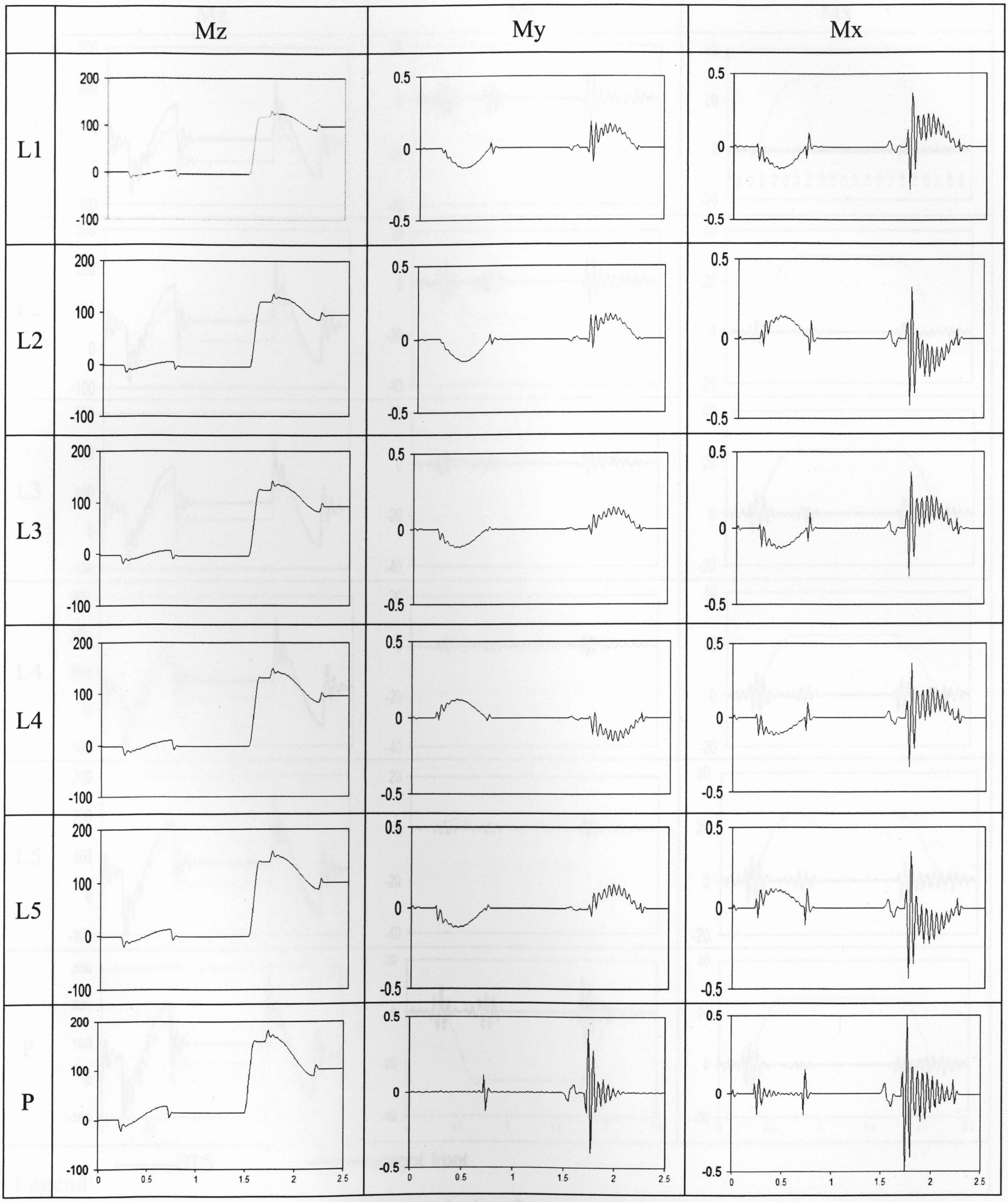
Appendix B: Moment during Flexion [Moment (Nm), Time (sec)]

Appendix "C"



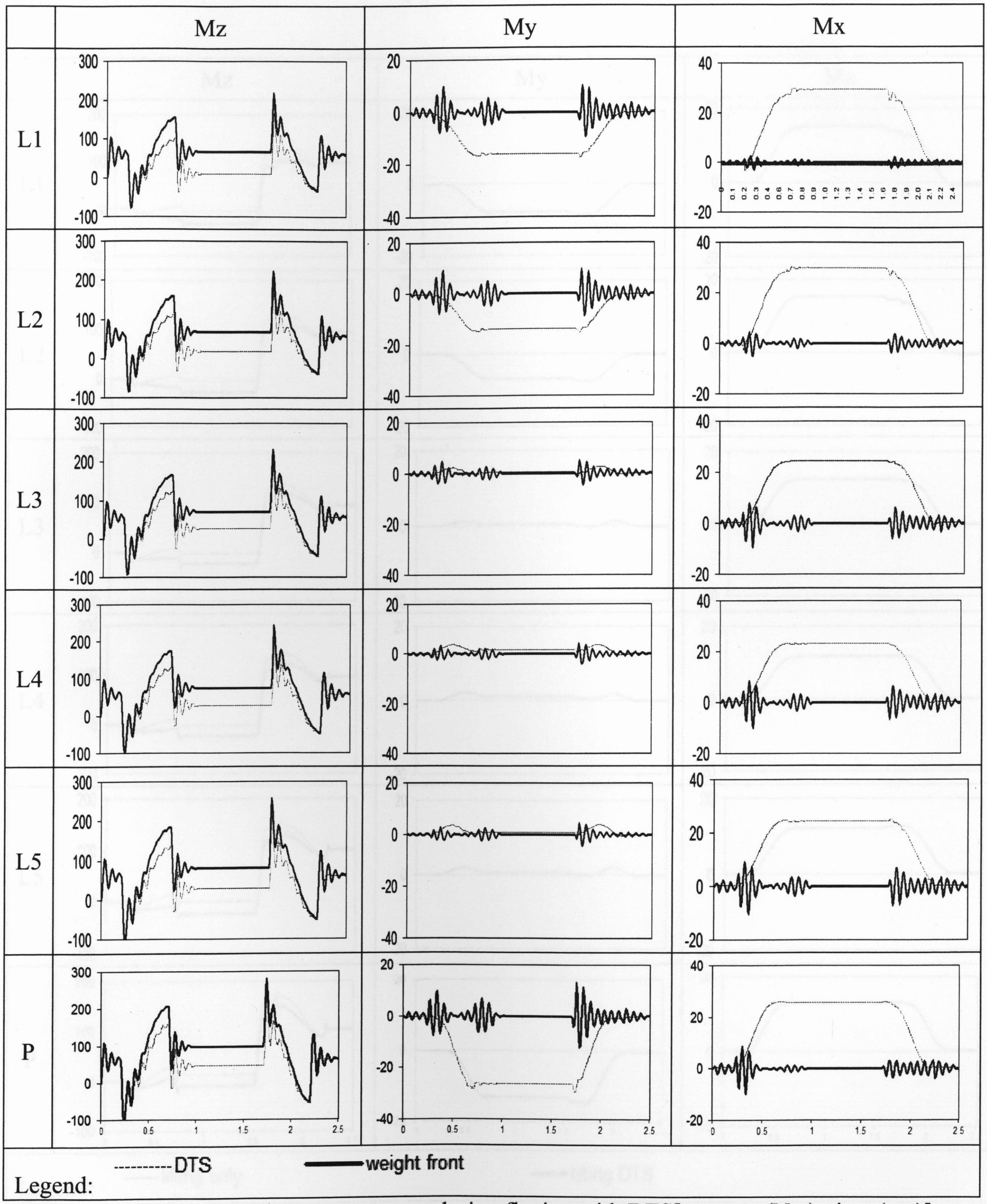
Appendix C: Moment during flexion, lateral bending and axial rotation [moment (Nm), time (sec)]

Appendix “D”



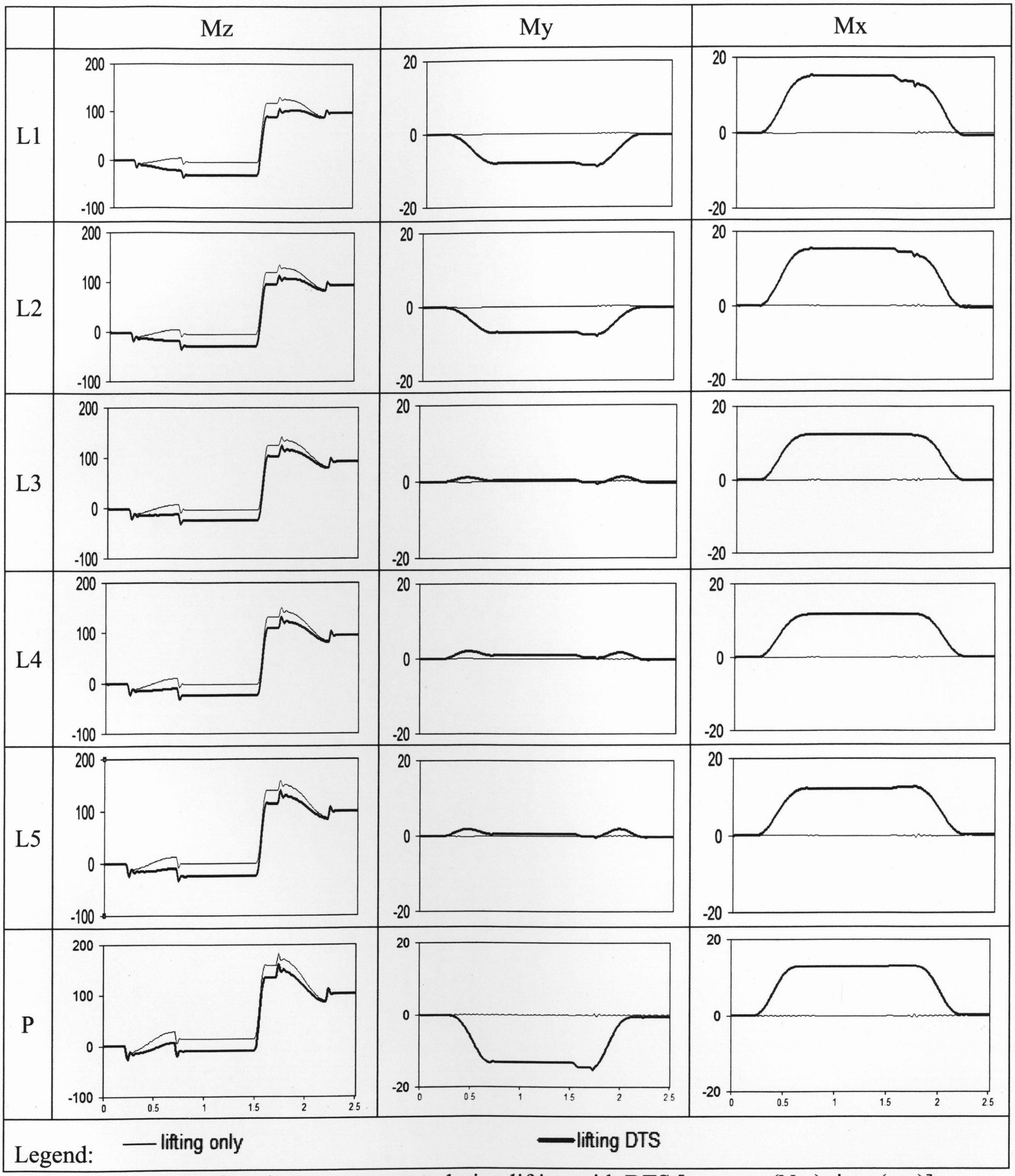
Appendix D: Moment during lifting [Moment (Nm), Time (sec)]

Appendix “E”



Appendix E: Comparison between moment during flexion with DTS[moment (Nm), time (sec)]

Appendix “F”



Appendix F: Comparison between moment during lifting with DTS [moment (Nm),time (sec)]