INVESTIGATION ON THE EFFECTS OF HUMAN WEIGHT ON SPINAL FACET JOINTS USING FINITE ELEMENT ANALYSIS

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INVESTIGATION ON THE EFFECTS OF HUMAN WEIGHT ON SPINAL FACET JOINTS USING FINITE ELEMENT ANALYSIS

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A report submitted
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DECLARATION

I declare that this project report entitled “Investigation On The Effects of Human Weight On The Spinal Facet Joints Using Finite Element Analyis” is the result of my own work except as cited in the references.

Signature : ....

Name : ....

Date : ....

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APPROVAL

I hereby declare that I have read this project report and in my opinion this report is sufficient in terms of scope and quality for the award of the degree of Bachelor of Mechanical Engineering (Structure & Materials).

Signature :. ....

Name of Supervisor : ....

Date : ....

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DEDICATION

This project work is dedicated to my beloved mother and father for always been there to support and inspire me throughout my life.

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ABSTRACT

Obesity is a growing healthcare issue which always associated with different kind of diseases, such as hypertension, osteoarthritis (OA), intervertebral disc degeneration, body pain and difficulty in physical functioning. Since the obesity has a close relationship with the spinal disorders, it increase the risk and possibility in develop back pain at the lower lumbar spine, which bear the largest mechanical compressive load. Spinal facet joints which are one of the critical components in lumbar spine are thus subjected to high mechanical compressive load and highly exposed to the risk of joint damage and degeneration. Since the concern of facet joints studies with respect to mechanical compressive load is only being given attention recently in this few year due to the difficulty in the modelling of facet surfaces and lack of experimental references related to facet joints, hence, there is a need in the study and analysis of facet joints especially for the effects of human weight on the spinal facet joint. Finite element analysis (FEA) is a suitable method in the study of facet joint since FEA can provide a FE model for repeated simulation which can greatly save the time consuming and more cost-efficient. The main objectives of this study was to develop a verified FE lumbar spine model and investigate the effects of human weight on the spinal facet joint using finite element analysis. The FE lumbar spine was first developed and verified by comparing the intersegmental rotation of the FE model under pure moment of 7.5 Nm in extension and flexion motions with previous in vitro studies. The verified FE model was then subjected to compressive load of 700 N, 900 N and 1100 N, which represented the normal weight, overweight and obese weight to investigate the effect of human weight on the kinematics of lumbar spine and contact pressure on the facet surfaces. The results shown that as the compressive load increased, the intersegmental rotation of lumbar spine and contact pressure on the facet surfaces also increased. Besides, it appeared that L3-L4 region of lumbar spine experienced largest contact pressure compared to other region of lumbar spine in all three loading cases and the contact pressure on facet surfaces during extension motion was much higher than the contact pressure during flexion motion. Moreover, it also found that there was an effect of asymmetry behavior in the lumbar spine.

ABSTRAK

Obesiti adalah satu isu kesihatan yang semakin meningkat dan ia selalu dikaitkan dengan pelbagai jenis penyakit seperti tekanan darah tinggi, osteoartritis (OA), penyakit cakera degenatif, sakit badan dan kesukaran dalam fungsi fizikal. Sejak obesiti mempunyai hubungan yang rapat dengan sakit belakang, ia meningkatkan risiko dan kemungkinan dalam mencetuskan sakit belakang di tulang belakang lumbar yang berada di lokasi paling bawah tulang belakang, lokasi menanggung beban mampatan yang terbesar. Sendi facet tulang belakang merupakan salah satu komponen penting dalam tulang belakang lumbar. Oleh itu, ia tertakluk kepada beban mampatan mekanikal yang tinggi dan terdedah kepada risiko kerosakan dan degenerasi sendi. Sejak kebimbangan kajian sendi facet berkenaan dengan beban mampatan mekanikal hanya diberi perhatian baru-baru ini disebabkan daripada kesukaran untuk pemodelan permukaan sendi facet dan kekurangan rujukan eksperimen yang berkaitan dengan sendi facet. Oleh itu, kajian dan analisis sendi facet terutama bagi kesan berat badan manusia pada aspek sendi tulang belakang amat diperlukan. Analisis unsur terhingga (FEA) adalah satu kaedah yang sesuai dalam kajian sendi facet kerana FEA boleh menyediakan model FE yang boleh digunakan dalam simulasi berulang dan membantu dalam menjimatkan masa dan kos. Objektif utama kajian ini adalah untuk menghasilkan satu FE model tulang belakang lumbar yang sah dan menggunakan model FE tersebut dalam penyiasatan kesan berat badan manusia pada sendi facet dengan cara analisis unsur terhingga. FE lumbar tulang belakang dihasilkan dan disahkan dengan membandingkan putaran intersegmental model FE di bawah momen tulen 7.5 Nm dalam pergerakan bengkok ke depan dan belakang dengan kajian dalam vitro sebelum ini. Model FE yang disahkan kemudiannya dikenakan beban mampatan 700 N, 900 N dan 1100 N, mewakili berat badan biasa, berat badan berlebihan dan berat obesiti untuk mengkaji kesan berat badan manusia pada kinematik tulang belakang lumbar dan tekanan sentuhan pada permukaan sendi facet . Keputusan menunjukkan bahawa peningkatan beban mampatan meningkat putaran intersegmental tulang belakang lumbar dan tekanan sentuhan pada permukaan sendi facet. Selain itu, L3-L4 tulang belakang lumbar mengalami tekanan sentuhan terbesar berbanding dengan lokasi tulang belakang lumbar dan tekanan hubungan pada permukaan aspek semasa gerakan bengkok ke belakang adalah lebih tinggi daripada tekanan sentuhan semasa gerakan bengkok ke depan. lanjutan adalah lebih tinggi daripada tekanan sentuhan semasa gerakan akhiran. Selain itu, ia juga mendapati bahawa terdapat kesan asymmetry di tulang belakang lumbar.

ACKNOWLEDGEMENT

I would like to express my deepest appreciation to my supervisor Dr. Mohd Juzaila Bin Abd Latif for his guidance, advices and support throughout this project. Secondly, I am also want to take this opportunity to thank UTeM, especially Faculty of Mechanical Engineering for the support of facilities and utilities. I would like to thank my course mates for giving me their support, patience and encouragement. Finally, I would like to thank my family for their support.

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LIST OF ABBREVIATIONS

3-D	Three-dimensional
FE	Finite element
FEA	Finite element analysis
FEM	Finite element method
FSU	Functional spinal unit
IVD	Intervertebral disc
L1	The first lumbar vrtebra
L2	The second lumbar vertebra
L3	The third lumbar vertebra
L4	The fourth lumbar vertebra
L5	The fifth lumbar vertebra
LBP	Low back pain
IDP	Intradiscal pressure
MRI	Magnetic resonance imaging
OA	Osteoarthritis
ROM	Range of motion

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LIST OF SYMBOL

F	-	Force
v	-	Poisson’s ratio
E	-	Yong’s modulus
C_1, C_2	-	Material constant characterising the deviatoric deformation of material
M	-	Moment

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CHAPTER 1

INTRODUCTION

1.1 Background

Low back pain (LBP) is a health disorder which defined as pain and discomfort at the lumbar region of the spine (Koes et al. 2006). A previous studies in United Kingdom reported that LBP is the most common cause of work absenteeism which accountable for about 12.5% of all the sick days and it was estimated that about 80% of adult population will experience LBP at least once during their lifetime (Baliga et al. 2015).

There are many factors that contribute to the LBP, such as obesity, poor posture, lack of exercise, aging, genetic factors and joint injuries. These factors responsible for the osteoarthritis (OA) at the lumbar facet joint which have been recognized as the potential cause of low back pain (Gellhorn et al. 2013; Manchikanti et al. 2016). OA also known as degenerative joint disease can affect any joint and in case of low back pain, OA affects the lumbar facet joint. Facet joint is a synovial joint which is surrounded by a capsule of connective tissue and the joint surface is coated with a layer of cartilage. Facet joint play an important role in transfers load and restricts motion in spine. OA causes the cartilage in the joint to become stiff and lose its functionality as the shock absorber. This results in the joint pain, joint swelling and reduced mobility.

Finite element method (FEM) is commonly use in the study of biomechanics behaviours since it allows repeatable simulation, comprehensive result, rapid time calculation, cost-saving and ethical concerns. FEM is a computational method that

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discretizes a structure into several elements and describes the behaviour of each element based on the input loading and boundary condition.

1.2 Problem statement

Previous studies demonstrated that obesity is strongly associated with low back pain and the cases of obesity is expected to increase further in the future (Shiri et al. 2009). From the previous study, it was found that the body weight is related to the facet joint stress which is always a potential cause of facet joint degeneration (Vincent et al. 2012). Therefore, it is essential to understand the relationship between human weight and the biomechanical effects of the lumbar facet joint. Thus, the aim of the study is to investigate the effects of human weight on spinal facet joint using finite element analysis (FEA).

1.3 Objectives

The objectives of this study are:

1) To develop finite element model of lumbar spine.
2) To investigate the biomechanical effects of human weight on the spinal facet joint using finite element analysis.

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1.4 Scopes of project

The original three dimensional (3-D) L1-L5 lumbar spine model was obtained from the Faculty of Biosciences & Medical Engineering University Technology Malaysia. Three compressive loads which represent normal weight, overweight and obese weight were applied to the finite element (FE) model in this study. The results on the intersegmental rotations of the lumbar spine and contact pressure within spinal facet joint were simulated and presented in this study.

CHAPTER 2

LITERATURE REVIEW

2.1 Overview

This chapter reviews the background of LBP and human lumbar spine. Besides, the FEA is also outlined in this chapter. Throughout this study, the directions of the human body are referred to the anatomic terms shown in Figure 2.1.

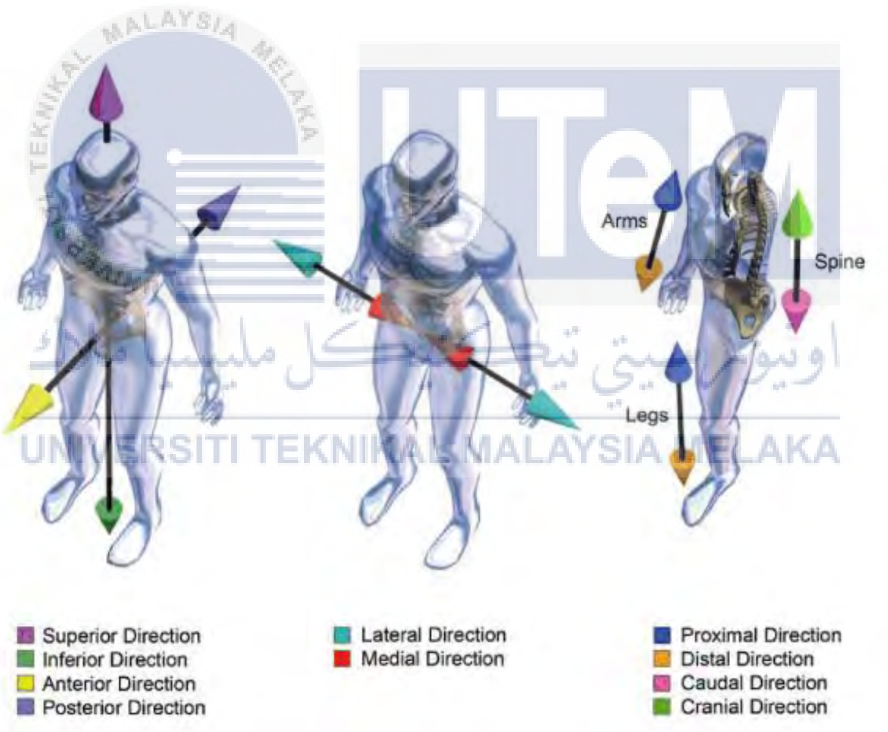

Figure 2.1: Anatomic reference directions. Adapted from Kurtz and Edidin (2006).

2.2 Anatomy of the human spine

Human spine is a column of vertebrae which is extended from the neck to the pelvis. The human spine is consisted of 24 individual vertebrae and 9 fused vertebrae (Aleti & Motaleb, 2014). The human spine functions as the support for the body weight, permits limited motion and protects the spinal cord. The human spine has a curvature shape and this curvature is divided into 4 regions, which are cervical, thoracic, lumbar and sacral regions as shown in the Figure 2.2. These curves contribute to the flexibility, stability and shock-absorbing capacity in the spine (Kurtz and Edidin, 2006).

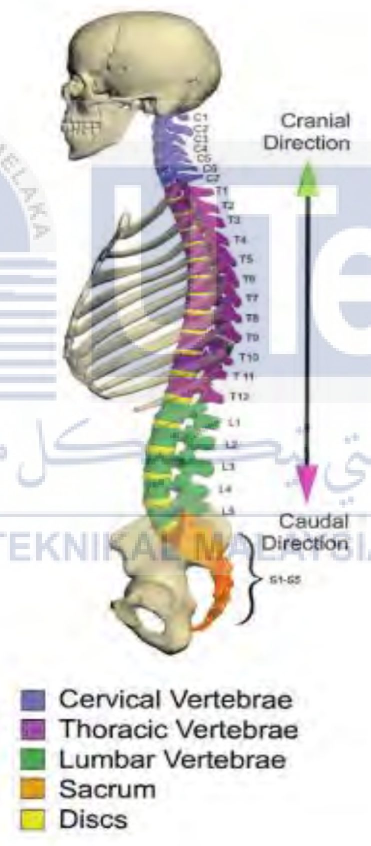

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Figure 2.2: Curvature of the spine. Adapted from Kurtz and Edidin (2006).

The cervical spine is located in the neck area and consisted of seven vertebrae which are numbered by C1-C7. These vertebrae supports the weight of the head, allows side to side and nodding motion of the head. The thoracic spine is located in the middle back and consists of twelve vertebrae which are numbered by T1-T12. The thoracic vertebrae is attached to the rib cage and functions as the protection to the heart and lung. The lumbar spine made up of five vertebrae in the lower back, which are numbered by L1-L5. The last region is the sacrum which are located below the lumbar spine. Sacrum is composed of five fused vertebrae which are numbered by S1-S5. The coccyx commonly called as the tail bone and consists of four fused rudimentary vertebrae. Coccyx is located at the terminal portion of the spinal column and it provides attachment for ligaments and muscles of the pelvic floor (Vaccaro, 2005).

2.3 Biomechanics of the spine

Biomechanics is the study of biological system using engineering science. It includes the study of mechanical principles, movement and structure of living organism. The biomechanics study of human spine is essential in order to provide more understanding about the spinal stability which played an important role in the support of body weight, allows upright posture and protects nervous structure in the body. These roles place a great strain on the spine and may cause the accelerated aging and symptomatic degeneration (Vaccaro, 2005). The kinematic motion of human spine can be simplified into extension, flexion, lateral bending, axial torsion, traction and compression as shown in the Figure 2.3.

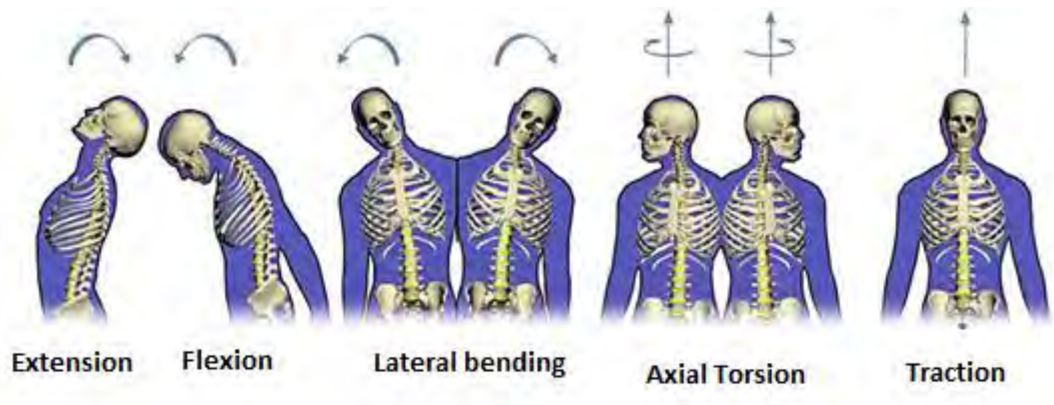

Figure 2.3: The kinematic motions of the spine. Adapted from Kurtz and Eidin (2006).

2.4 Human lumbar spine

Human lumbar vertebrae is larger compared to the vertebrae in other spinal region because lumbar spine carry more weight than other spinal region. Lumbar spine functions to absorb the stress of lifting, carries any weight superior to the lumbar spine and constrain certain motions of the spine (Vaccaro, 2005). The vertebrae form a 3-joint complex structure with an intervertebral disc and two pair of articulating facet joints as shown in the Figure 2.4.

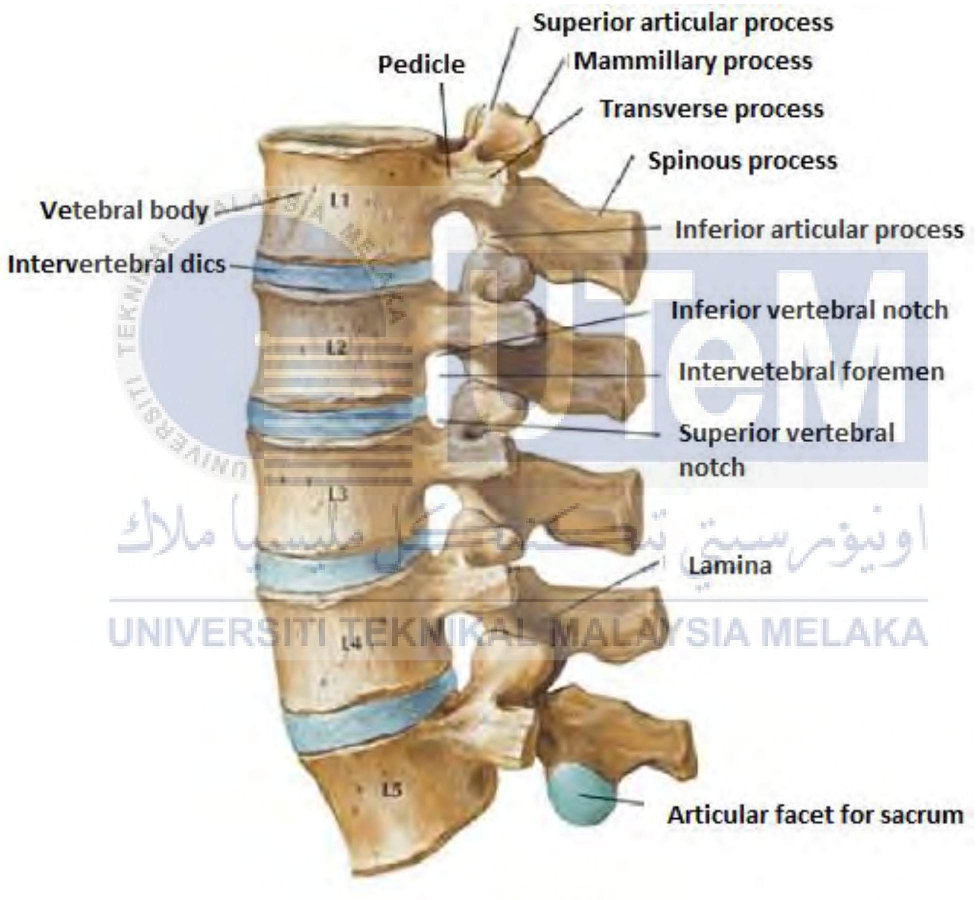

Figure 2.4: The left lateral view of the lumbar spine. Adapted from Netter (2006).

2.5 Lumbar vertebra

Each individual vertebra has its own unique features based on its size, shape and orientation of the facet joint, with the largest and heaviest vertebrae in the spine located in the lumbar region. Each vertebra consists of two main parts, which are the anterior part which includes the vertebral body, and the posterior part which includes the vertebral arc. The vertebral arch encloses the vertebral foramen and consists of two pedicles, two laminae, and seven processes, which are four articular facet, two transverse process and one spinous process as shown in the Figure 2.5.

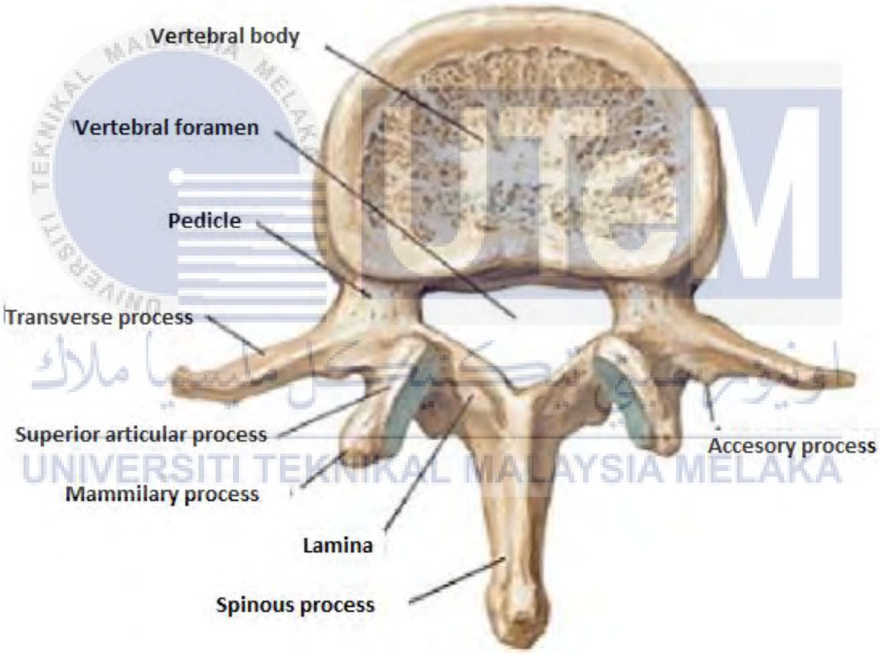

Figure 2.5: The superior view of L2 vertebra. Adapted from Netter (2006).

2.5.1 Vertebral Body

The vertebral body is elliptically cylindrical in shape and composed mainly of cancellous bone at the inside layers and a relatively thin layer of strong cortical bone at the outside layer as shown in the Figure 2.6 (Netter 2006). The cancellous bone are less dense and weaker than the cortical bone, but it is much more flexible compared to the cortical bone and this flexibility enables the bone to resist much greater changes of the unit shapes (White & Panjabi 1990).

Figure 2.6: The structure of the normal vertebra bone. Adapted from Netter (2006).

2.5.2 Facet Joint

Facet joint also known as zygapophyseal joint is a biomechanical structure in the posterior side of the spine and it is formed by the inferior facet of the superior vertebra meeting the superior facet of the inferior vertebra. In each vertebrae level, there are one pair of superior facets that connected to the top vertebrae and one pair of inferior facets that connected to the bottom vertebrae (Cramer & Darby, 2005).

Facet joint is considered as synovial joint and the joints surface are covered by a layer of cartilage which function to ensure smooth articulation between the facet surfaces. The facet joint is also covered by the ligamentous capsules which constraint the relative translations and rotations of adjacent vertebrae. The connective tissue within capsule will produce the synovial fluid to nourish and lubricate the joints (Jaumard et al 2011). Figure 2.7 shows the posterior and transverse view of facet joint.

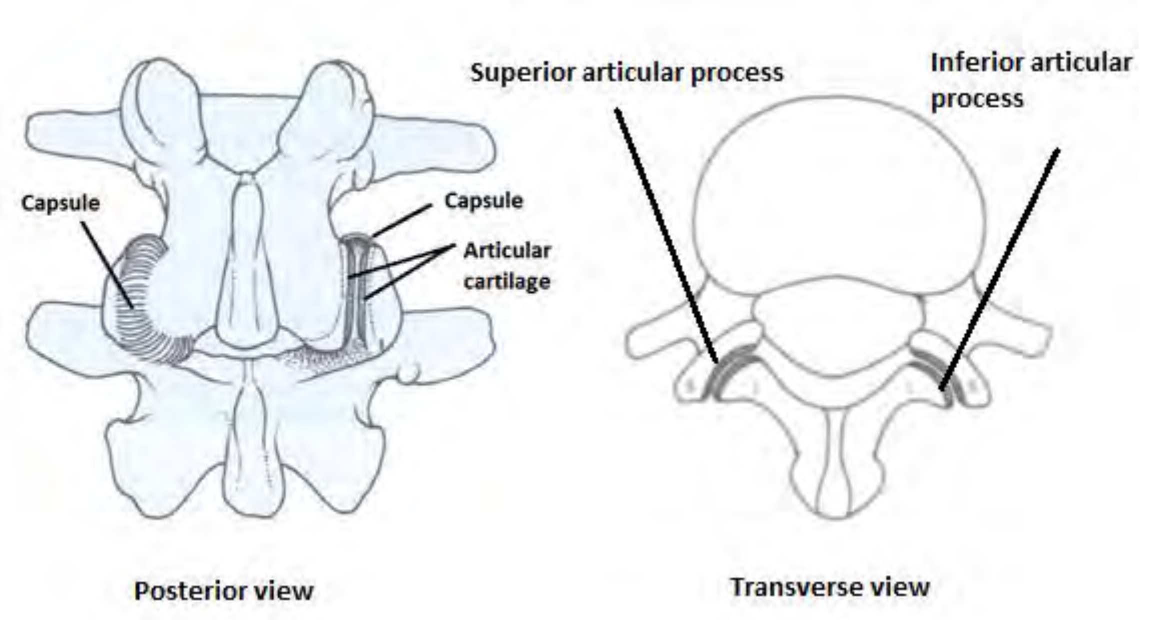

Figure 2.7: L3-L4 view of facet joint. Adapted from Bogduk (2005).

Facet joint plays an important role in maintain the mechanical stability of the spine. Previous studies had shown that 30 % of the compressive load applied to the spine is carried by the posterior side of the spine. In the cervical and upper thoracic region of spine, the facet articular surfaces are more horizontally-oriented to allow great degree of coupling of axial rotation and lateral bending. The facet surfaces become more vertically-oriented in the lower thoracic and lumbar spinal region and these limits the lateral bending and axial rotation in these regions to prevent injury to the intervertebral discs and spinal cord due to the excess kinematics motions (Jaumard et al. 2011; Jaffar et al. 2010).

2.5.3 Intervertebral disc

The intervertebral disc (IVD) is a fibrocartilaginous cushions between two vertebral bodies and it functions as the spine's shock absorbing system. IVD is considered as one of the main components of the spinal column since it responsible for one-third of the spinal column height (Urban & Roberts, 2003). IVD consists of two components, which are nucleus pulposus and annulus fibrosus. The top and bottom of IVD are covered by the vertebral endplates as shown in the Figure 2.8.

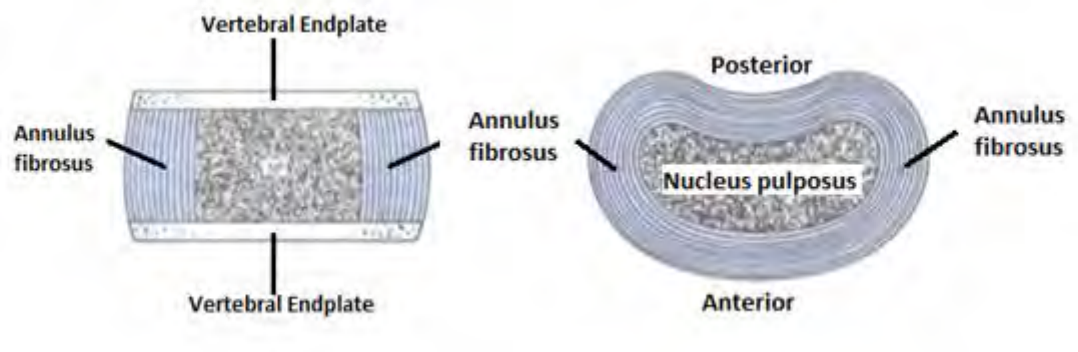

Figure 2.8: The basic structure of intervertebral disc. Adapted from Bogduk (2005).

Nucleus pulposus is a semi-fluid mass of mucoid material located at the centre of the IVD and it can be deformed under pressure due to its fluid porperties. Due to its fluid properties, when one side of the nucleus pulposus was subjected to compression, the nucleus will deforms to tranfers the pressure to all direction since the fluid volume cannot be reduced.

Annulus fibrosus is made up of 10 to 20 layers of concentric lamellae which surround the nucleus pulposus, with collagen fibres lay parallel in between each layers of lamellae. Annulus fibrosus function to protect the soft nucleus pulposus, while the function of nucleus pulposus is to provide shock absorption for the spine (Bogduk, 2005).

2.5.4 Ligament

Ligaments are the fibrous connective tissue attached to the spine, which are function to hold the vertebrae together. Human spine consisted of seven types of ligaments as shown in the Figure 2.9. The major ligaments of spine are anterior longitudinal ligament, posterior longitudinal ligament and ligamentum flavum, interspinous ligament, supraspinous ligament, intertransverse ligament and capsular ligament. Ligaments restraint excessive movements of the vertebral bone in order to prevent spine injuries and stabilize spinal segment motion.

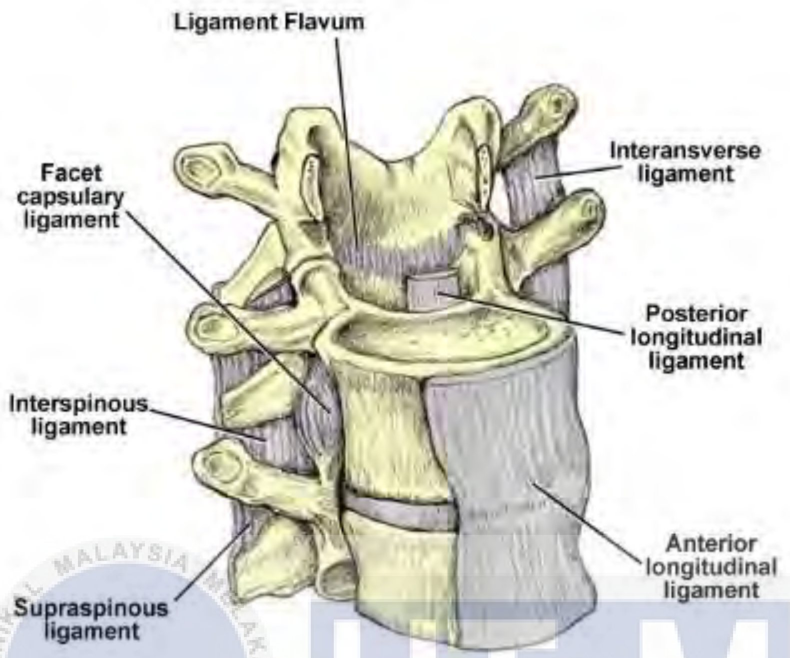

Figure 2.9: Ligaments attached to lumbar spine. Adapted from Netter (2006).

2.6 Low back pain

LBP is defined as pain, muscle tension and discomfort experienced at the lumbar region of the spine (Koes et al. 2006). LBP can either be acute, subacute or chronic depending on the duration of pain experience. There are many causes to the LBP and one of the most common cause of LBP is the facet joint degeneration, also known as osteoarthritis (Manchikanti et al. 2016). Degenerative change in the facet joint may lead to the lumbar instability and result in the occur of LBP. Since lumbar vertebrae located at the bottom region of the spine, thus they carry most amount of the body weight compared to the other regions of the spine and this causes the facet joint in the lumbar region more prone to degenerate (Gellhorn et al. 2013; Du et al. 2016; Koes et al. 2006).

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2.6.1 Obesity and low back pain

Obesity is believed to be associated with the occurrence of LBP and previous study has shown that obese people are more often to have LBP compared to the people who are not obese due to the increased of mechanical load on the lumbar spine (Djurasovic et al. 2008; Shiri et al. 2009). Obesity could causes some of the spine disorders such as joint degeneration and disc degeneration which contribute to the reduced of spinal mobility. Previous studies has shown that obesity could lead to joint cartilage defect and 5 kg of weight gain is corresponding to 36 % of increased possibility of OA developed (Vincent et al. 2012; Shiri et al. 2009).

2.6.2 Facet joint degeneration

Facet joint degeneration is a degenerative condition of facet joint which results in the wearing away of the articular cartilage at the facet surface. The damaged cartilage can result in higher stresses at the surrounding of the damaged cartilage area, and lead to the osteophyte formation. Osteophyte is the bony spurs formed at the edge of the cartilage surfaces. Besides, facet joint degeneration also leads to the hardening of joint structure such as cartilage and capsular ligament (Jaumard et al. 2011; Gellhorn et al. 2013).

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2.6.3 Diagnosis

There are a few methods use in the diagnosis of LBP. The most basic of diagnostic methods are the physical examination and medical history research. For the more serious cases, joint aspiration and imaging test will be used to diagnose the patients. Joint aspiration is the withdrawal of fluid from the infected joint to check the condition of the joint. There are two common types of imaging test which are X-rays and Magnetic resonance imaging

(MRI). X-rays shows the bone spur around joint but does not show the cartilage, while MRI reveals a better image of bone and cartilage.

2.7 Computational method

FEM is a mathematical technique used to obtain the approximate solutions to the complex problem by discretized a structure into several elements and described the behaviour of each element based on the material properties, input loading and boundary condition. Since 1950s, FEM is used for structural analysis in engineering sectors, especially in the structural analysis of aircraft and later implemented in the orthopaedic biomedical research (Babuška & Strouboulis, 2001; Brekelmans et al. 1972). Due to the increasing availability of affortable and powerful computers, FEM is widely used in the biomedical field to study the complex structure of living being, including human spine since FEM allows repeatable simulation, rapid time calculation, cost-saving, comprehensive results and ethical concerns.

2.7.1 Finite element model of lumbar spine

Generally, the lumbar spine FE model modelled in the previous studies was consisted of three main components of lumbar spine, which were lumbar vertebrae, intervertebral discs and facet joints. Besides, the lumbar spine FE model was refined with the attachment of ligaments to the vertebrae. The seven types of ligaments included anterior longitudinal ligament, posterior longitudinal ligament, intertransverse ligament, ligament flavum, capsular ligament, interspinous ligament, and supraspinous ligament. (Park et al. 2013; Kim et al., 2014).

2.7.2 Material properties of the human lumbar spine

In previous studies, isotropic elastic material was mostly used in define the mechanical behaviour of the vertebral body (Park et al. 2013; Kim et al., 2014; Schmidt et al. 2007).

Since the nucleus pulposus and annulus fibrosus are nearly incompressible and unable to react linearly to the loading, Mooney-Rivlin constitutive model was mostly used in previous studies to model the hyper-elastic and incompressible material properties of annulus fibrosus and nucleus pulposus, while criss-cross pattern truss element was always used in construct the annulus fiber (Park et al. 2013; Kim et al. 2014; Schmidt et al. 2007).

In order to imitate the synovial fluid effect at the facet articular process, frictionless sliding surface contact was commonly used in model the cartilage layer of facet joint (Schmidt et al. 2007; Kim, et al., 2014), while the space between two cartilage surface was defined using gap elements (Rohlmannm et al. 2007), spring elements (Kim, et al. 2014) and surface-to-surface contact elements (Schmidt et al. 2007).

Truss element (Kim, et al. 2014) or spring element (Schmidt et al. 2007) are popular method in the modelling of ligaments attached at the lumbar spine model with non linear material properties which exhibit similar behaviour with stress-strain curve as shown in Figure 2.10 (Samandouras, 2010). From the Figure 2.10, AB is the neutral zone, BC is elastic zone and CD is the plastic zone. Permanent deformation occurs past the yield point C and failure occurs after point D.

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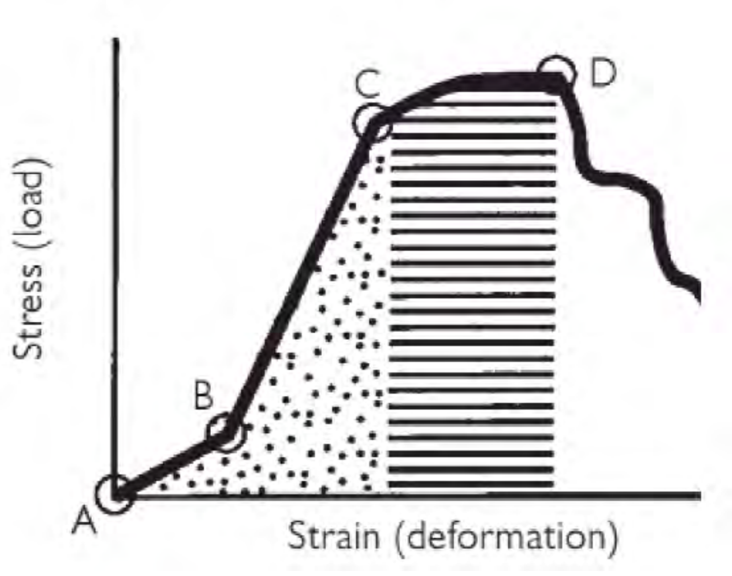

Figure 2.10: A typical stress-strain curve for biological tissue. Adapted from Samandouras (2010).

The material properties used in the previous studies is summarized and shown in the Table 2.1:

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Table 2.1: The material properties used to model lumbar spine in previous studies.

Source	Cortical bone	Cancellous bone	Annulus fibrosus	Nucleus Pulposus	Annulus fibers	Ligament	Cartilage of Facet Joint	Solver
Park et al., (2013)	E = 12000 MPa ν= 0.3	E = 100 MPa ν= 0.2	Hyperelastic Mooney-Rivlin $C_1 = 0.18$ $C_2 = 0.045$	Incompressible fluid-filled cavity	Non-linear dependent on distance from disc centre, 6 layers criss-cross pattern	Non-linear stress strain curve	Hard frictionless contact, E = 11 MPa, $\nu = 0.4$, Initial gap = 0.5 mm	Abaqus 6.10
Schmidt et al., (2007)	$E_1 = 22000$ MPa $E_2 = 11300$ MPa $\nu_1 = 0.484$ $\nu_2 = 0.203$	$E_1 = 200$ MPa $E_2 = 140$ MPa $\nu_1 = 0.45$ $\nu_2 = 0.315$	Hyperelastic Mooney-Rivlin $C_1 = 0.56$ $C_2 = 0.14$	Incompressible fluid-filled cavity	Non-linear stress- strain curve, 16 layers criss-cross pattern	Non-linear stress strain curve	Hard frictionless contact, E = 35 MPa, $\nu = 0.4$, Initial gap = 0.4 mm	Abaqus 6.10
Rohlmann et al., (2007)	E = 10000 MPa $\nu = 0.3$	$E_1 = 200$ MPa $E_2 = 140$ MPa $\nu_1 = 0.45$ $\nu_2 = 0.315$	Hyperelastic Neo-Hookean $C_1 = 0.3448$ $C_2 = 0.3$	Incompressible fluid-filled cavity	Non-linear dependent on distance from disc centre, 14 layers criss-cross pattern	Non-linear stress strain curve	Soft frictionless contact, Initial gap = 0.5 mm	Abaqus 6.10

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Table 2.1: The material properties used to model lumbar spine in previous studies.

Source	Cortical bone	Cancellous bone	Annulus fibrosus	Nucleus Pulposus	Annulus fibers	Ligament	Cartilage of Facet Joint	Solver
Kim et al., (2014)	E_1 = 22000 MPa E_2 = 11300 MPa ν_1 = 0.484 ν_2 = 0.203	E_1 = 200 MPa E_2 = 140 MPa ν_1 = 0.45 ν_2 = 0.315	Solid E = 4.2 MPa ν = 0.45	Incompressible fluid-filled cavity	Non-linear stress- strain curve, 8 layers criss-cross pattern	Non-linear stress strain curve	Hard frictionless contact, E = 35 MPa, ν = 0.4, Initial gap = 0.5 mm	Abaqus 6.6
Du et al., (2016)	E = 14000 MPa ν = 0.3	E = 100 MPa ν = 0.2	Mooney-Rivlin C_1 = 0.18 C_2 = 0.045	Mooney-Rivlin C_1 = 0.12 C_2 = 0.003	Calibrated stress- strain curve	Calibrated deflection-force curve	Neo- Hookean, C_{10} = 2	Abaqus 6.11
Kurutz & L.Oroszva´ry, (2010)	E = 12000 MPa ν = 0.3	E= 150 MPa ν = 0.3	E = 4 MPa ν = 0.45 For tension, E = 0.4 MPa	E = 1 MPa ν = 0.499 For tension, E = 0.4 MPa	Tension only E = 500 MPa for external fibre, 400 MPa for middle fibre and 300 MPa for internal fibre,	Linear elastic tension element, E = 8 MPa for ALL E = 10MPa for PLL E = 5 MPa for other ligaments, ν = 0.35 for all ligaments,	-	ANSYS Classic

*E: Young's modulus; ν: Poisson's ratio; C_1 and C_2: Material constant characterising the deviatoric deformation of material

2.7.3 Finite element studies of mechanical behaviour for human lumbar spine

Spine individual segment level motion is known as functional spinal units (FSU), which consisted of two adjacent vertebrae, the intervertebral disc, the facet joints, and the ligaments that connect the vertebrae. Each FSU has six degree of freedom which are three rotational motions and 3 translational motions (Ferguson, 2008). The three rotational motions are flexion-extension, lateral bending and axial rotation, while the translational motions include axial displacement, anterior shear, and lateral shear. The ROM at each level of the spine is influenced by the geometry of the spine's posterior elements, which are shape and orientation of the facet joints as shown in Figure 2.11(Hall, 1991).

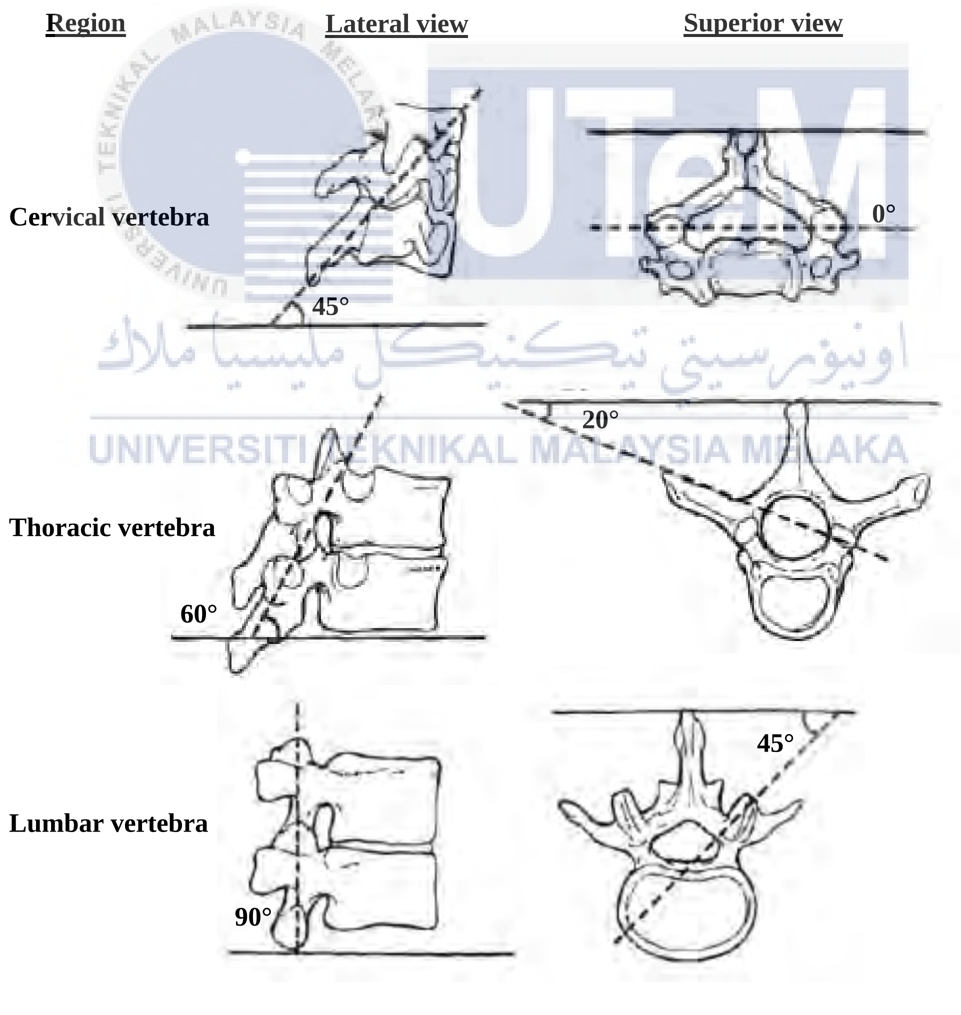

Figure 2.11: Orientation of the facet joints. Adapted from Hall (1991).

2.7.4 Range of motion

The range of motion (ROM) is the measurement of movement around a joint in the form of angle or displacement (Clarkson, 2013). In the kinematics of lumbar spine, extension-flexion motion accounted for the largest ROM compared to other types of motions. For the extension-flexion motion of lumbar spine, only the angle within the sagittal plane was considered and this parameters is very important in the examination of instability in lumbar spine (White & Panjabi, 1990).

2.7.5 Intradiscal pressure of the intervertebral disc

In the previous study, excessive mechanical load on the spine may induce high pressure within the nucleus pulposus and eventually leads to the disc degeneration and the happen of back pain (Urban & Roberts, 2003; Schmidt et al. 2007). Flexion motion produce higher intradiscal pressure compared to extension motion, while erect standing induced higher intradiscal pressure than sitting (Wilke et al. 1999; Schmidt et al. 2007).

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2.7.6 Stress analysis of the facet joint

In several previous studies, facet joint degeneration is strongly related to the LBP (Jaumard et al. 2011; Gellhorn et al. 2013; Manchikanti et al. 2016). Based on the previous study, 30% of the compressive load on the spine is sustained by the facet joint and excessive loading on the facet joint may lead to the facet joint degeneration. The degeneration of facet surface will eventually cause the mechanical instability of vertebral motion, which increases the mechanical stresses supported by the disc. The present of articulating cartilage on the facet joint suface allows frictionless motion between facet joint and high stress may wear away the cartilage layer and cause damage the facet surfaces. Previous FEA studies have shown less interest in the research related to the contact force on the lumbar facet joint during flexion motion since the upper facet surface move away from the lower facet surface during flexion motion and this lead to a contact force much more lower than the extension motion. (Kuo, et al. 2010; Jaumard et al. 2011; Jaffar et al. 2010). Besides, previous FEA studies also shown that the facet force are highest at the middle level of lumbar region which is L3-L4 of lumbar spine under extension motion (Du et al. 2016).

2.8 Summary

The review of literatures have shown that the mechanical behaviour of facet joints played a very crucial role in ensure the kinematics stability of lumbar spine. Previous studies shown that about 30 % of the compressive load subjected to lumbar spine will be transmitted throught the spinal facet joints, thus, high stress will be induced at the spinal facet joint and this may lead to the cause of LBP. Since the extension and flexion motions accounted for the largest ROM of lumbar spine, thus it is very important to investigate the effect of human weight on the kinematics of spinal facet joints under these two motions.

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CHAPTER 3

METHODOLOGY

3.1 Overview

This chapter describes the methodologies used in the study. FE model was developed and verified by compared the ROM and intradiscal pressure in this study with previous studies. The verified FE model were then used for the case studies of human weight effects on the spinal facet joints. The intersegmental rotation of lumbar spine and contact pressure within spinal facet joints under extension and flexion motions were analysed and presented at the end of this chapter. The flow process for this research is shown in Figure 3.1.

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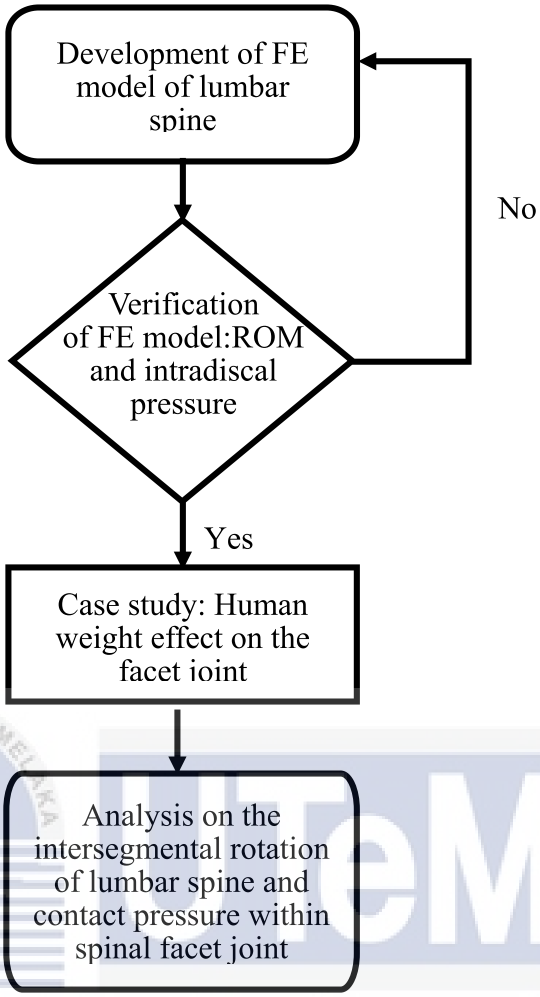

Figure 3.1: Flow chart of methodology

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3.2 Three-dimensional geometrical model of the lumbar spine

The original L1-L5 lumbar spine 3-D geometrical model was obtained from the Faculty of Biosciences & Medical Engineering University Technology Malaysia. The lumbar spine consists of five vertebrae and four intervertebral discs in between each vertebrae. The vertebrae is composed of hard cortical bone at the outside layer and less dense cancellous bone at the inside of the vertebrae. According to the data retrieved from the 3-D geometrical model, the thickness of the cortical bone was set to 1 mm and the thickness of the cartilage tissue was set to be 2 mm.

3.3 Finite element model of the lumbar spine

The 3-D model was meshed and optimized automatically using Abaqus. Figure 3.2 shows the 3-D FE model of the lumbar spine. In total, there were 1756884 tetrahedral elements and 355028 nodes in the model of the lumbar spine.

Figure 3.2: 3-D FE model of the lumbar spine

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3.4 Material properties

In this study, linear and isotropic material properties were adopted for the both the cortical and cancellous bone (Park et al. 2013). The facet surfaces were made smooth to imitate the frictionless motion between facet surfaces and the initial gap between the cartilage layers was assumed to be 0.5 mm with linear elastic and isotropic material properties (Kim, et al., 2014). IVD is composed of three components, which are nucleus pulposus, annulus fibrosus and annulus fiber. Mooney-Rivlin constitutive model was used to model the incompressible nucleus and hyperelastic characteristic of the annulus fibrosus, while annulus fibers were model by tension-only truss element with non-linear behaviour

(Schmidt et al. 2007; Du et al. 2016). Table 3.1 shows all the material properties used in the FE model of the lumbar spine model.

Table 3.1: Material properties used in the FE lumbar spine model

Element Set	Material Properties	Reference
Cortical bone	E= 12000 MPa, v= 0.3	(Park et al. 2013)
Cancellous bone	E= 100 MPa, v= 0.2	(Park et al. 2013)
Articular cartilage	E= 35 MPa, v=0.4	(Kim, et al., 2014)
Nucleus pulposus	E≈ 1.0 MPa, v= 0.4999	(Schmidt et al. 2007; Du et al. 2016)
Annulus ground substance	E≈ 4.2 MPa, v= 0.45	(Schmidt et al. 2006; Du et al. 2016)

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3.5 Loading and boundary conditions

The contact surfaces between the vertebral bodies and the IVDs are set to perfectly connect to ensure the IVDs lies between the vertebral bodies. The articulating facet surface were modelled as surface to surface contact with normal contact stiffness of 200 N/mm and friction coefficient of zero (Kim, et al., 2014). All nodes of the inferior surface of L5 vertebral body were constrained to fixed position with no translation and rotation motions. A moment of 7.5 Nm were applied to L1 superior surface using force couple as shown in the Figure 3.3. A force couple with equal and opposite forces +F and –F were applied to the anterior and posterior of L1 vertebral body respectively. The force couple for two motion, which are flexion and extension were calculated and presented in the Table 3.2. The value of force couple were calculated using equation of moment as shown in the equation (3.1) (Nordin & Frankel, 2001).

$$M = F_R \times d \tag{3.1}$$

Where M is the moment loads, F_R is the resultant force and d is the distance between anterior and posterior of the L1 vertebral body.

$$F_R = 7.5 \text{ Nm} / 0.03 \text{ m}$$

$$F_R = 250 \text{ N}$$

$$F_R = \sqrt{F_y^2 + F_z^2}$$

Let $F_y = 0.3\ F_s$ (30% of force acting on y-axis);

$F_z = 0.7\ F_s$ (70% of force acting on z-axis)

Thus,

$$250 \text{ N} = \sqrt{(0.3\ F_s)^2 + (0.7\ F_s)^2}$$

$$F_s = 328 \text{ N}$$

Therefore,

$$F_y = 0.3\ F_s = 0.3 \times 328 \text{ N} = 98 \text{ N}$$

$$F_z = 0.7\ F_s = 0.7 \times 328 \text{ N} = 230 \text{ N}$$

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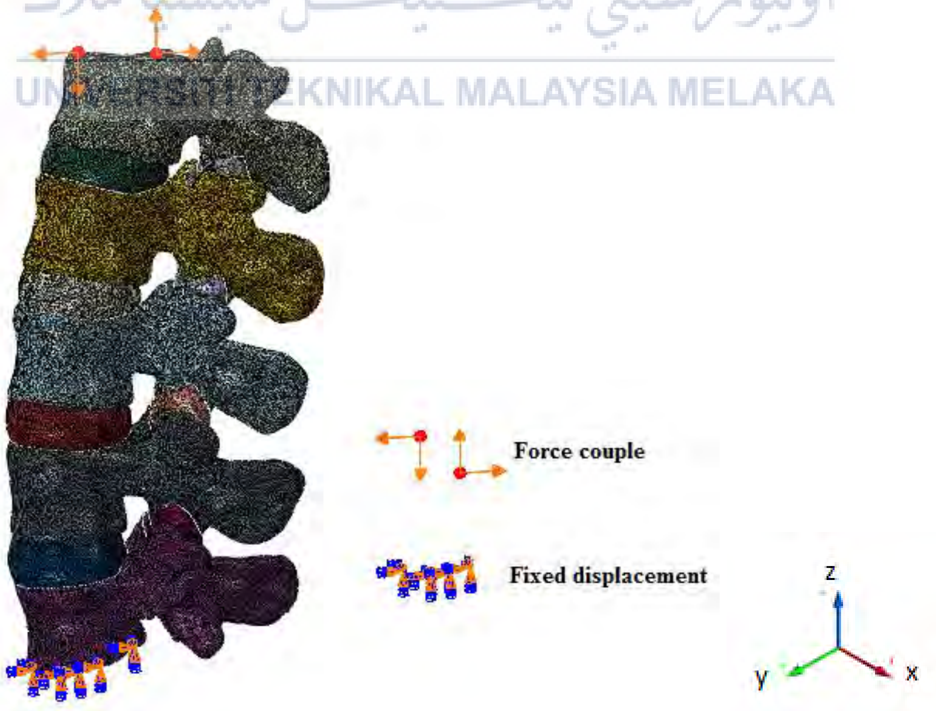

Figure 3.3: Loading and boundary condition on FE lumbar spine model.

Table 3.2 : Magnitude of moment loading applied on the lumbar spine

Loading Direction	Anterior Point		Posterior Point	
	F_y (N)	F_z (N)	F_y (N)	F_z (N)
Flexion	-98	-230	98	230
Extension	98	230	-98	-230

3.6 Verification of the finite element model

Verification and validation of FE model is a significant step in the development of a reliable FE model for the use of computational biomechanics study. The reliability of the FE model is very important since an invalidated model can lead to false conclusions (Henninger et al. 2010). A verification study on the intersegmental rotations of the lumbar spine and the IVD stresses were carried out in this study to verify the 3-D lumbar spine FE model. The verified model will be used in this study.

3.6.1 Intersegmental rotation of the lumbar spine

The FE model of the 3-D lumbar spine was compared with previous in vitro study under pure moments of 7.5 Nm for two types of motions, which are flexion and extension motions (Panjabi et al. 1994).

3.6.2 Intervertebral disc

The IVD was verified by comparing the vertical intradiscal pressure (IDP) and displacement to the previous in vitro studies (Markolf & Morris, 1974; Ranu, 1990). A vertical compressive load of 1100 N was applied to the L4 superior vertebral body as

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pressure distribution and the inferior surface of the L5 vertebral body was restrained to fixed displacement in rotation and translation as shown in Figure 3.4.

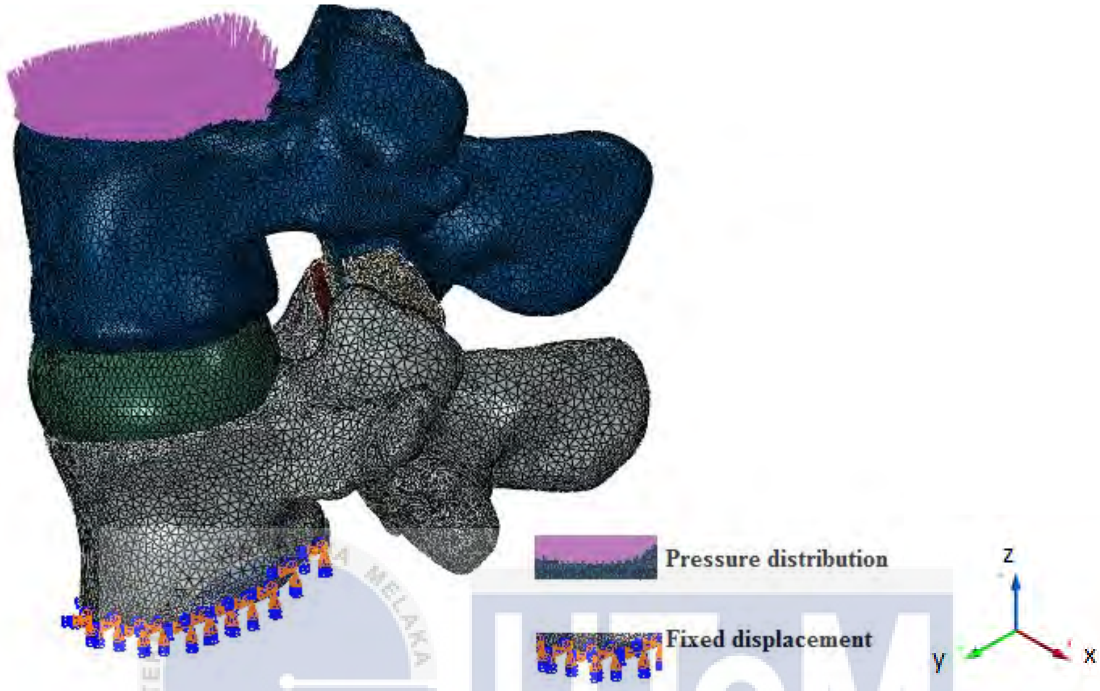

Figure 3.4: Loading and boundary conditions on L4-L5 FE lumbar spine model.

3.7 Finite element analysis of the lumbar spine model at various groups of human weight

A combination of 7.5 Nm pure moment and compressive follower load path were applied to the L1-L5 lumbar spine FE model in order to imitate the real flexion and extension physiological activities (Du et al. 2016). In this study, the compressive load used represents the upper body weight of the subjects. Three weights which are 55 kg, 80 kg and 110 kg were used in this study to represent the normal weight, overweight, and obese weight respectively (Walpole et al. 2012). The compressive load carried by the lumbar vertebrae is equal to 58 % to 60% of human weight with an additional of musle force approximately equal to the upper weight (Rohlmann et al 2006; Kurutz & Oroszva´ry, 2010). Therefore, the total compressive loads represent the normal weight, overweight, and obese weight which were applied to the lumbar vertebrae in this study are 700 N, 900 N and 1100 N respectively. The inferior body of L5 vertebral body was be fixed to zero motion in both translation and rotation. Intersegmnetal rotation of lumbar spine and contact pressure within lumbar facet joint when subjected to the three compressive loads during flexion and extension motion were conducted.

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CHAPTER 4

RESULTS AND DISCUSSION

4.1 Overview

This chapter reports and discusses the results obtained from the lumbar spine model simulation. The L1-L5 FE lumbar spine model was verified based on the range of motion, intradiscal pressure and axial displacement of intervertebral disc. Intersegmental rotation of lumbar spine and facet contact pressure under the effect of human weight during flexion and extension motion were studied.

4.2 Verification of the L1-L5 lumbar spine model

The verification of the FE lumbar spine model was carried out to on the intersegmental rotations of the lumbar spine model, IVD axial displacement and IDP of IVD. The verification is important in order to ensure the reliability of the lumbar spine model.

4.2.1 Range of motion

The FE model of lumbar spine model was verified by comparing the ROM with previous in vitro study for flexion and extension motions at pure moment of 7.5 Nm (Panjabi et al. 1994). Figure 4.1 shows the deformation of lumbar spine under flexion motion and extension motions. Lumbar spine bend forward during flexion motion and bend backward during extension motion. Figure 4.2 shows the comparison of total ROM of the lumbar spine under pure moments up to 7.5 Nm between FE model in this study and previously published in vitro result (Panjabi et al. 1994). From Figure 4.2, although there was a notable difference of 36.71% at 2.5 Nm in extension motion, but the percentage difference became smaller at 2.157% when the moment reached 7.5 Nm. The percentage difference at 2.5 Nm during flexion motion was 45.26% and the percentage difference at 7.5 Nm was 23.66%. Since the value of ROM in this FEA study was still within the accepted range reported in the literature, the FE model in this study could imitate the real lumbar spine's flexion and extension motion.

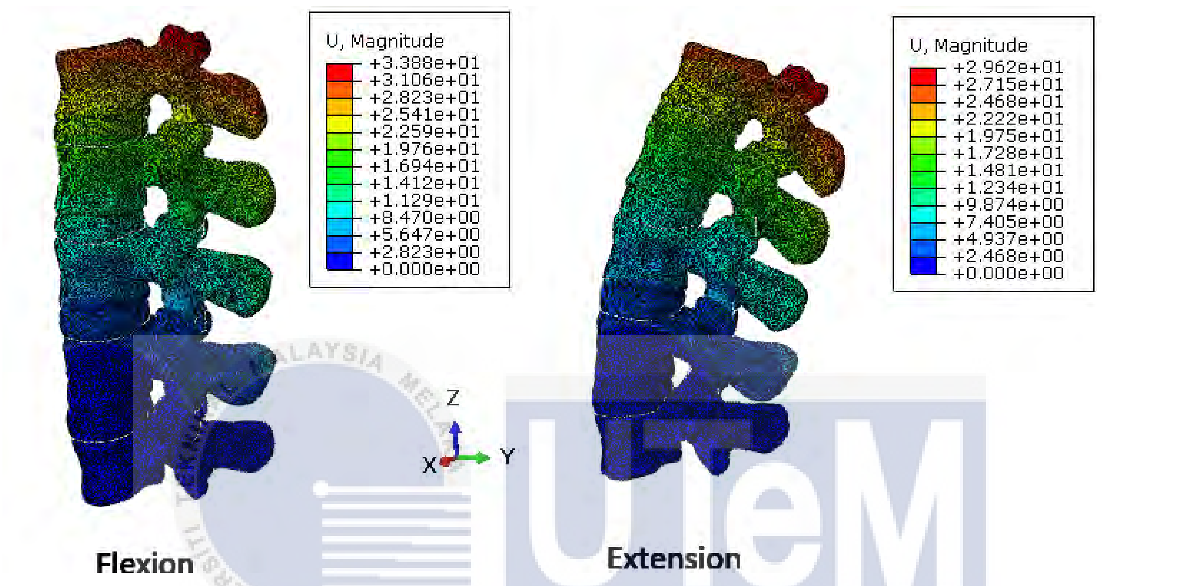

Figure 4.1: Deformation of lumbar spine under flexion and extension motion.

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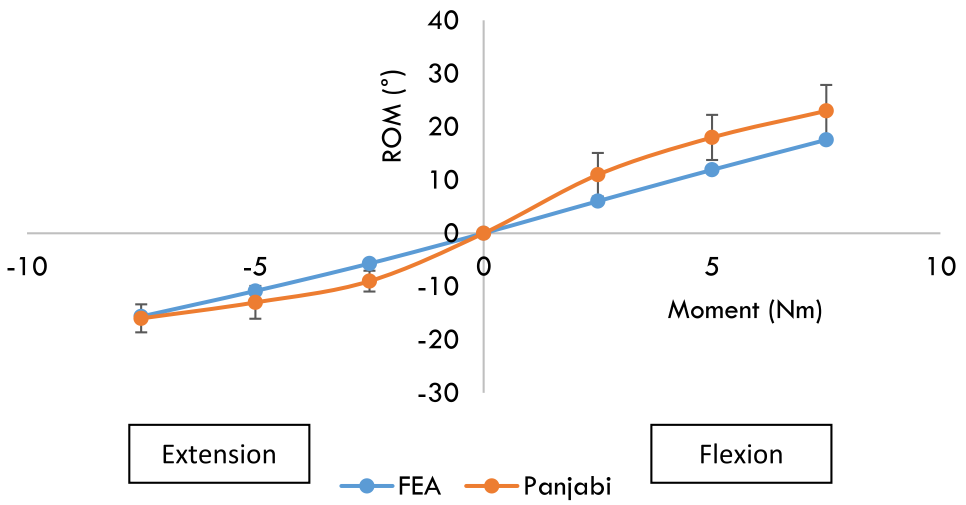

Figure 4.2: Comparison of total ROM of the lumbar spine under pure moments up to 7.5 Nm between FE model in this study and previously published in vitro result (Panjabi et al. 1994).

4.2.2 IVD verification

IVD was verified by comparing the axial displacement and IDP of the FE lumbar spine model in this study with previous vitro studies at shown in Figure 4.3 and Figure 4.4. From Figure 4.3 and Figure 4.4, the percentage difference between this study and previous in vitro studies for axial displacement and IDP of the IVD at 1100 N compression load were 18.03% and 0.01% respectively. The result of intradiscal pressure in this FEA study demonstrated an almost similar trend compared with the previous in-vitro study, while the percentage difference for the verification of axial displacement at 1100 N compressive load in this FEA study was still within the accepted range reported in the literature (Markolf & Morris 1974; Ranu 1990). These results show that the FE model in this study could produce a reliable results for the FE analysis.

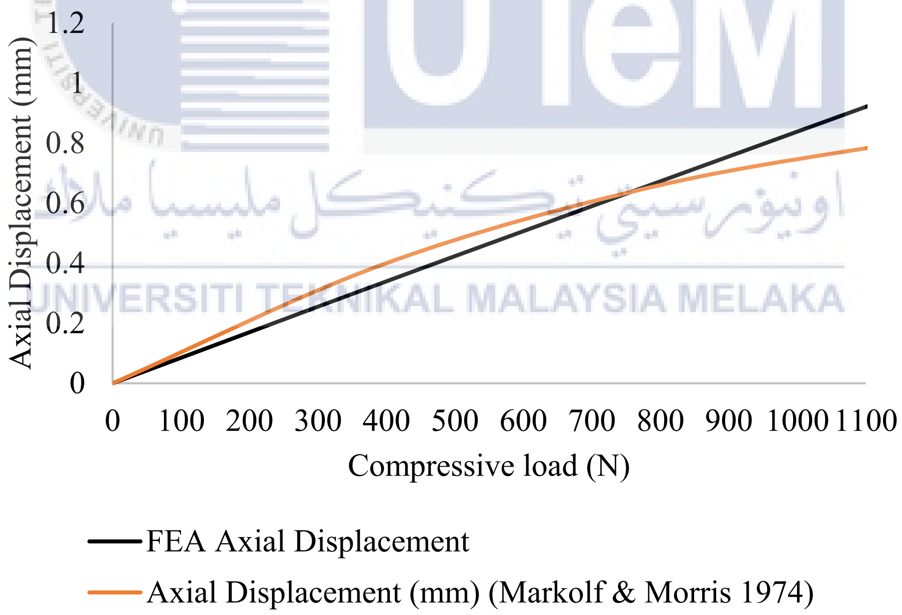

Figure 4.3: Comparison of axial displacement of FE lumbar spine model in this study with previous in vitro study (Markolf & Morris, 1974).

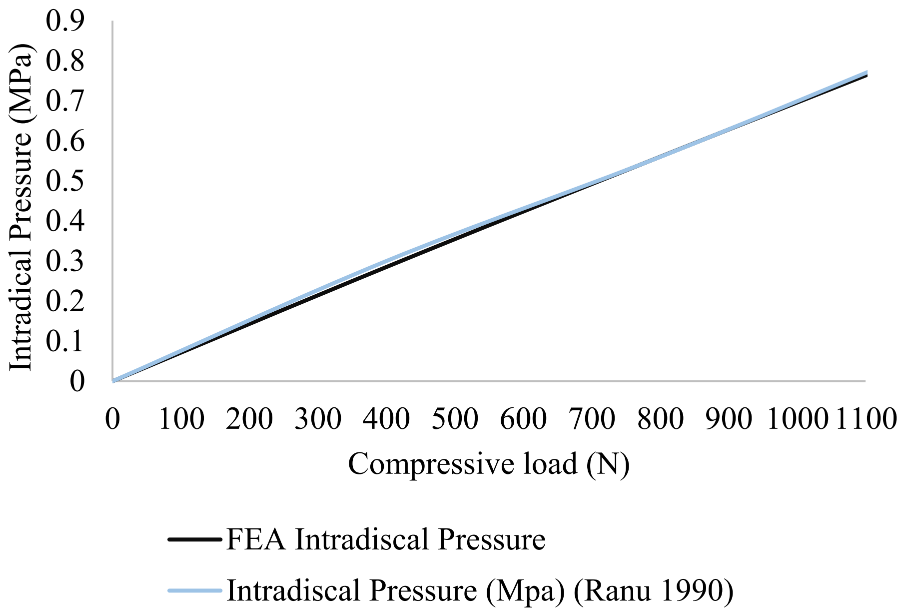

Figure 4.4: Comparison of axial intradiscal pressure of FE lumbar spine model in this study with previous in vitro study (Ranu, 1990).

4.3 Biomechanical effects of human weight on FE lumbar spine model

The FE lumbar spine model was subjected to three different types of loadings, which were 700 N, 900 N, and 1100 N to represent the effects of normal weight, overweight and obese weight on the lumbar spine and facet joint under flexion and extension motions.

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4.3.1 The effects of human weight on the kinematics of lumbar spine

The results of the simulation shown that the kinematics of lumbar spine were affected by the human weight. The intersegmental rotation of lumbar spine for different loading cases under flexion and extension motions were calculated and presented in Figure 4.5 and Figure 4.6 respectively.

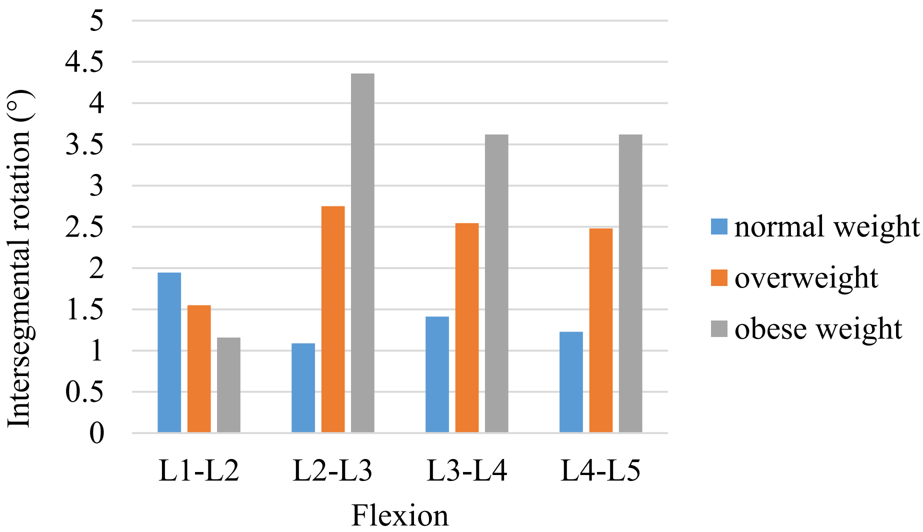

Figure 4.5: Intersegmental rotation for normal weight, overweight and obese weight under flexion motion.

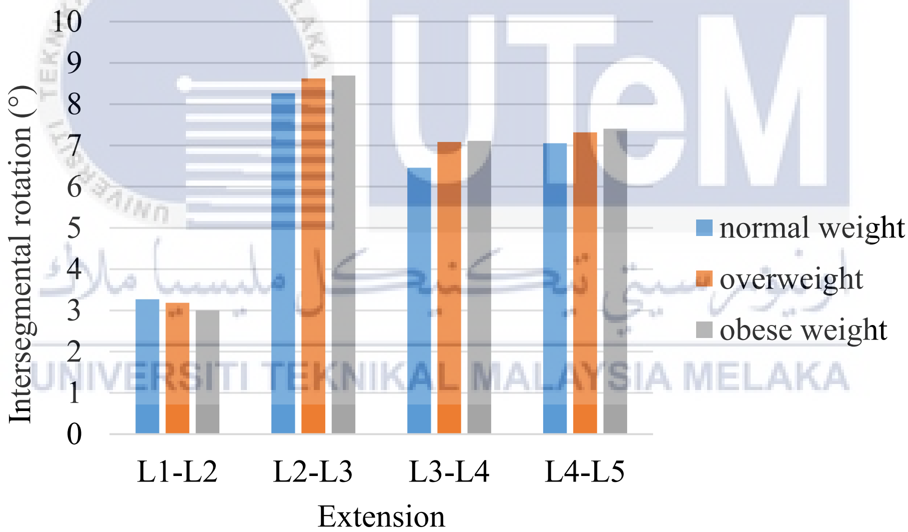

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Figure 4.6: Intersegmental rotation for normal weight, overweight and obese weight under extension motion.

Figure 4.5 and Figure 4.6 shown that the intersegmental rotations of lumbar spine were increased as the compressive load was increased except for the L1-L2 which shown an opposite trend compared to other level of lumbar spine in both flexion and extension motion. Although L1-L2 of lumbar spine exhibited a different trend compared to other level of lumbar spine, but the results still shown that the increase of human weight will affected the kinematics of lumbar spine.

From the results, it was found that the percentage difference of intersegmental rotation of lumbar spine between normal weight and obese weight under flexion motion was much higher than the intersegmental rotation of lumbar spine between normal weight and obese weight under extension motion. The maximum percentage difference of intersegmental rotations between normal weight and obese weight under flexion motion was 300 % at L2-L3 of lumbar spine, while the maximum percentage difference of intersegmental rotations between normal weight and obese weight under extension motion was 10.1 % at L3-L4 of lumbar spine. This can be explained by the role of facet joint which restrain the motion of lumbar spine in extension motion. In extension motion, the two adjacent facets moved toward each other, while the two adjacent facets moved away from each other in the flexion motion. Since there was a contact between two adjacent facet surfaces, the opposite reaction force from the adjacent facet surface can greatly reduce the motion of the lumbar spine. These results indicated the importance of facet joint in ensure the kinematics stability of lumbar spine.

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4.3.2 The effects of human weight on the facet contact pressure

Contact pressure on the facet surfaces for all three loading cases under flexion and extension motions were presented in Figure 4.7, Figure 4.8, Figure 4.9, and Figure 4.10, while the pressure distribution on the facet surfaces were presented in Figure 4.11 and Figure 4.12.

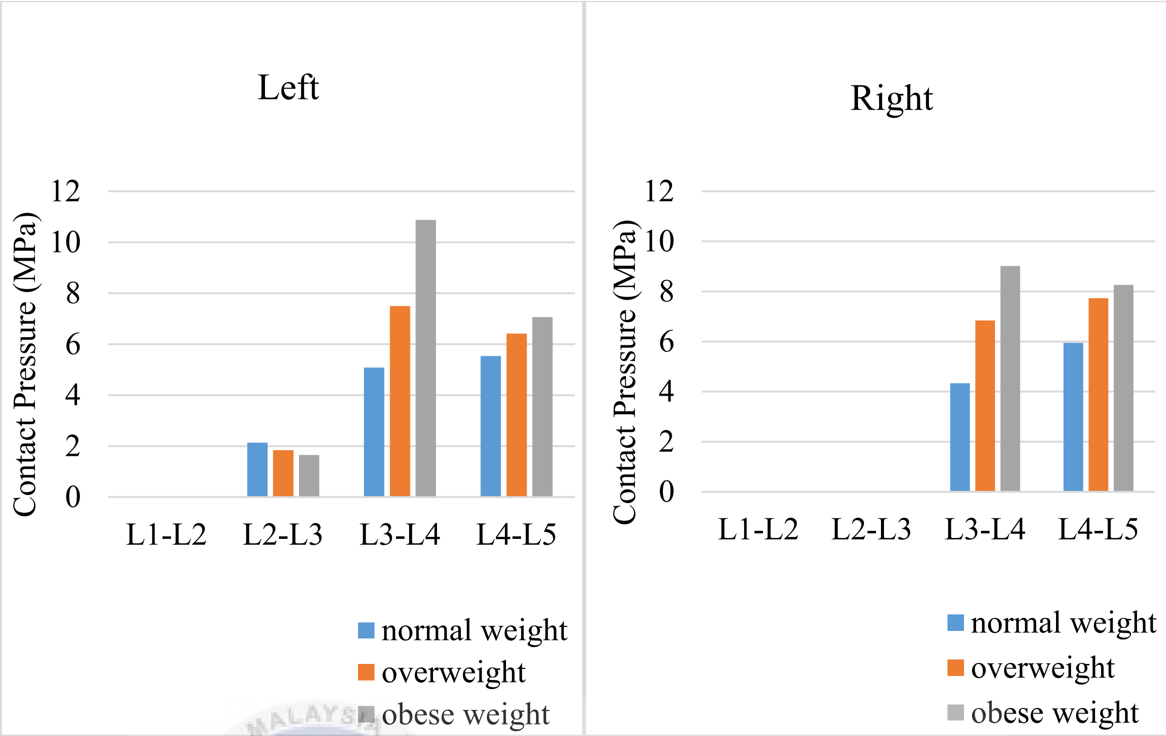

Figure 4.7: Left and right superior Facet Contact Pressure under flexion motion.

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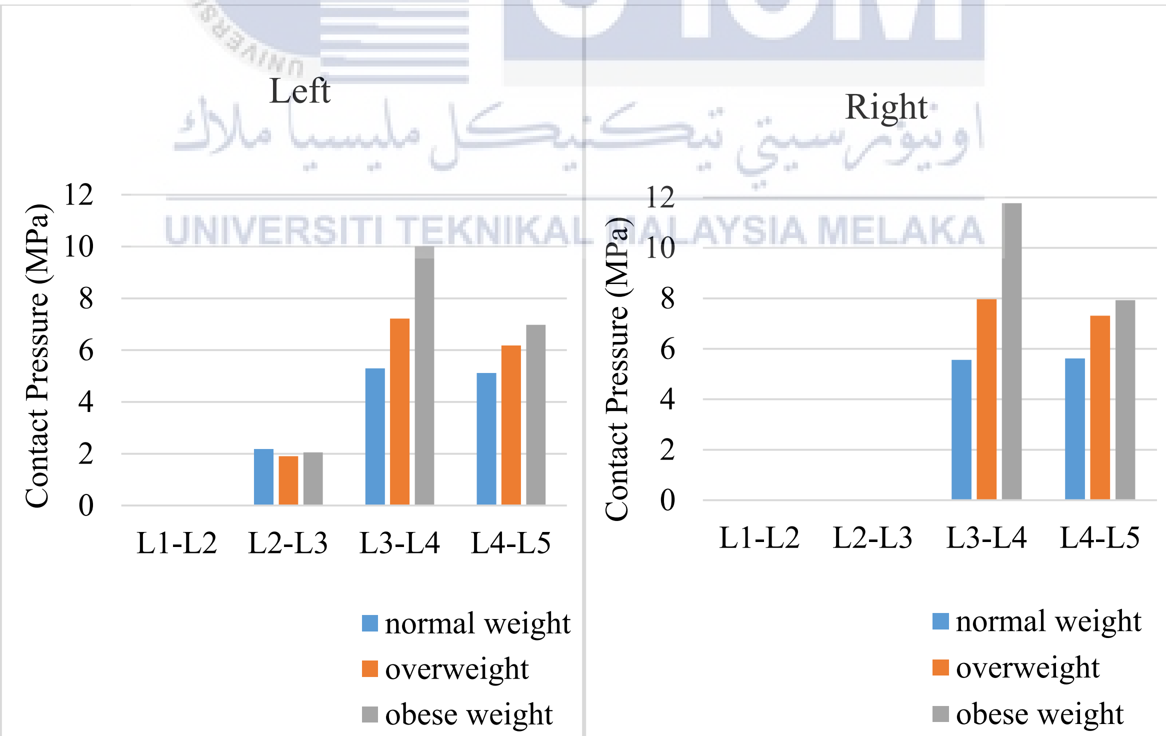

Figure 4.8: Left and right inferior Facet Contact Pressure under flexion motion.

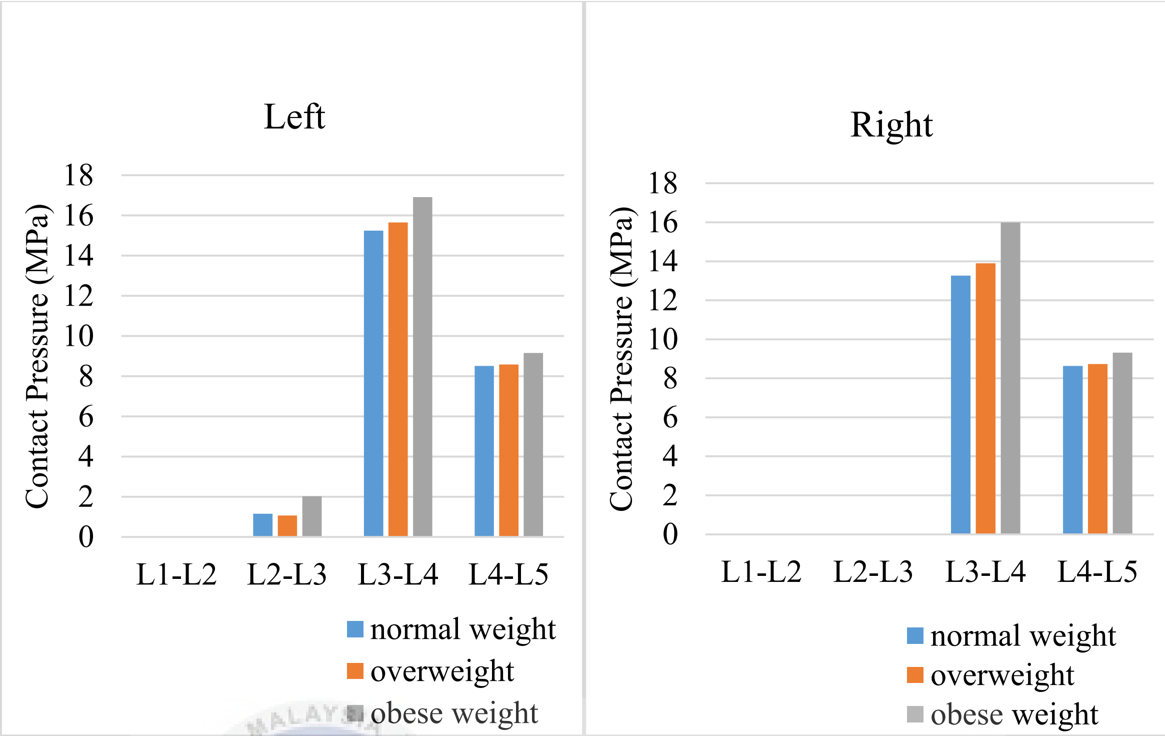

Figure 4.9: Left and right superior Facet Contact Pressure under extension motion.

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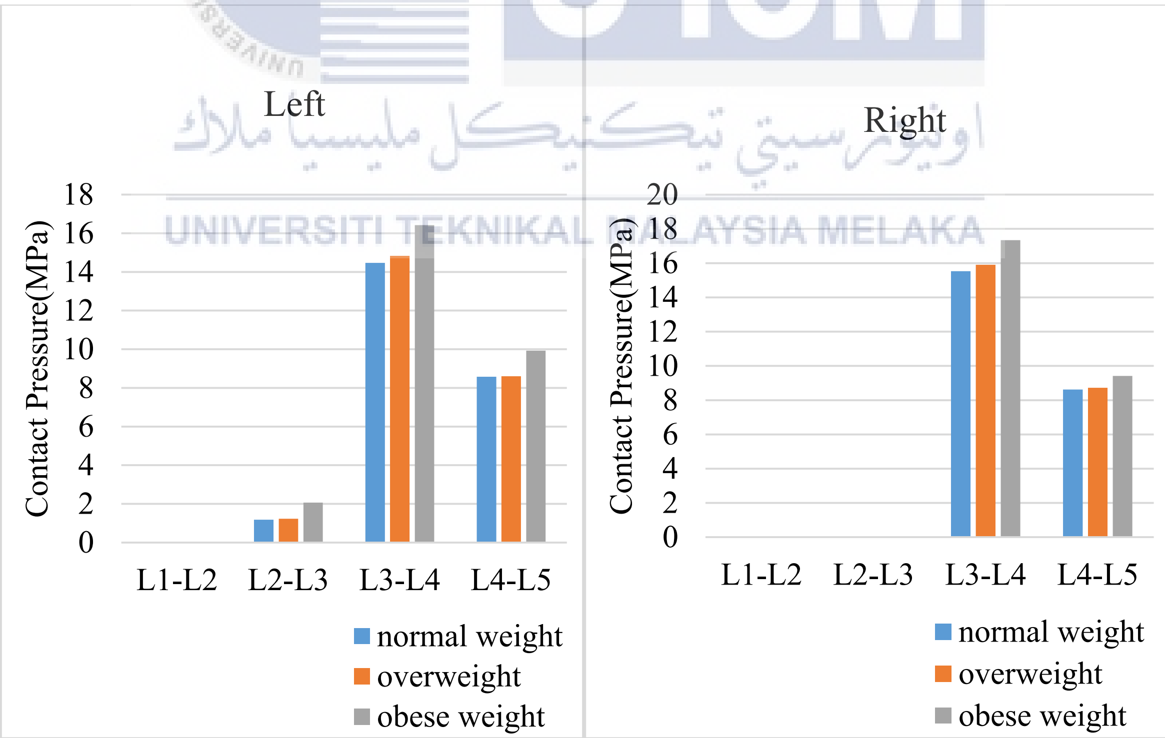

Figure 4.10: Left and right inferior Facet Contact Pressure under extension motion.

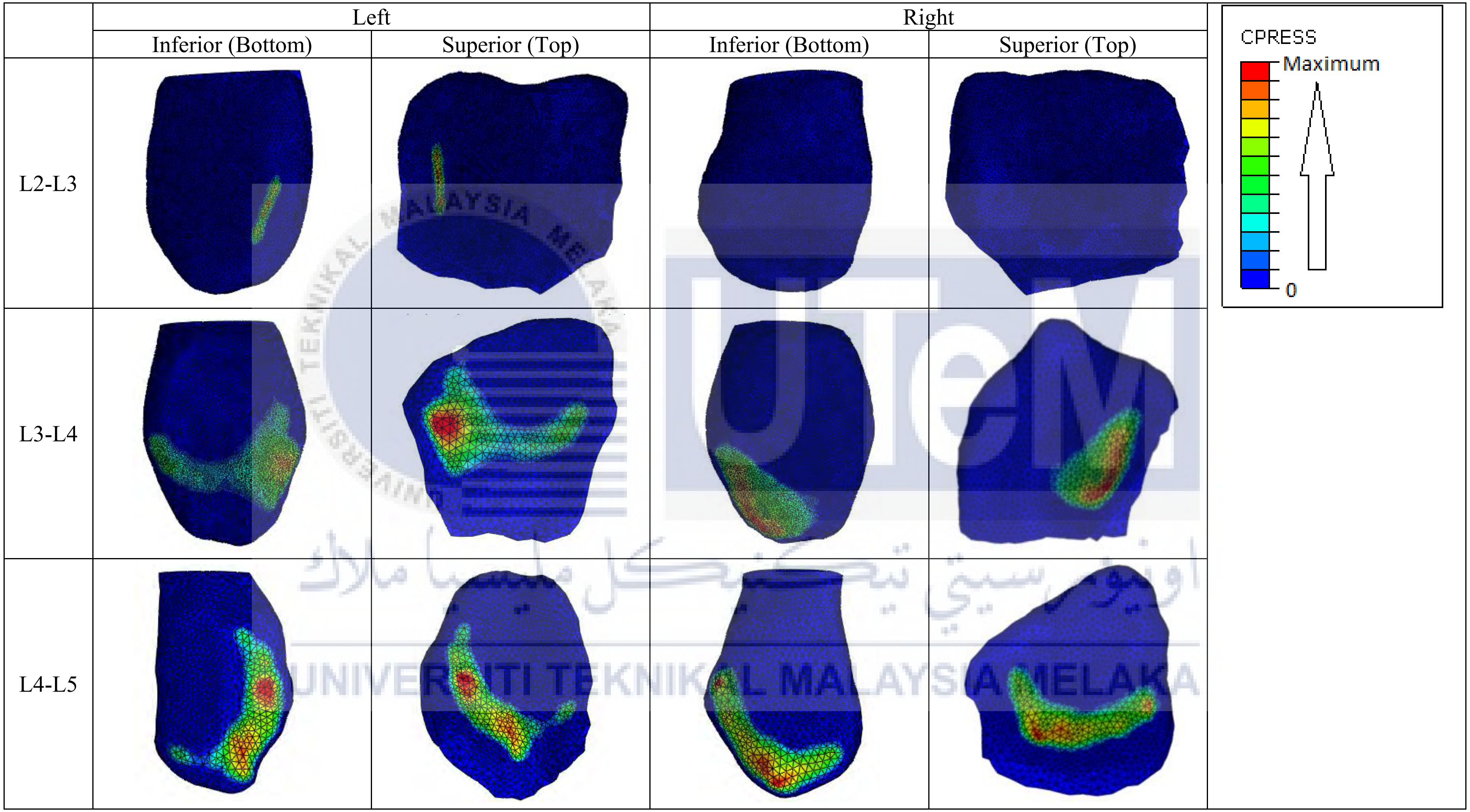

Figure 4.11: Pressure distribution on lumbar facet joint under normal weight during flexion motion.

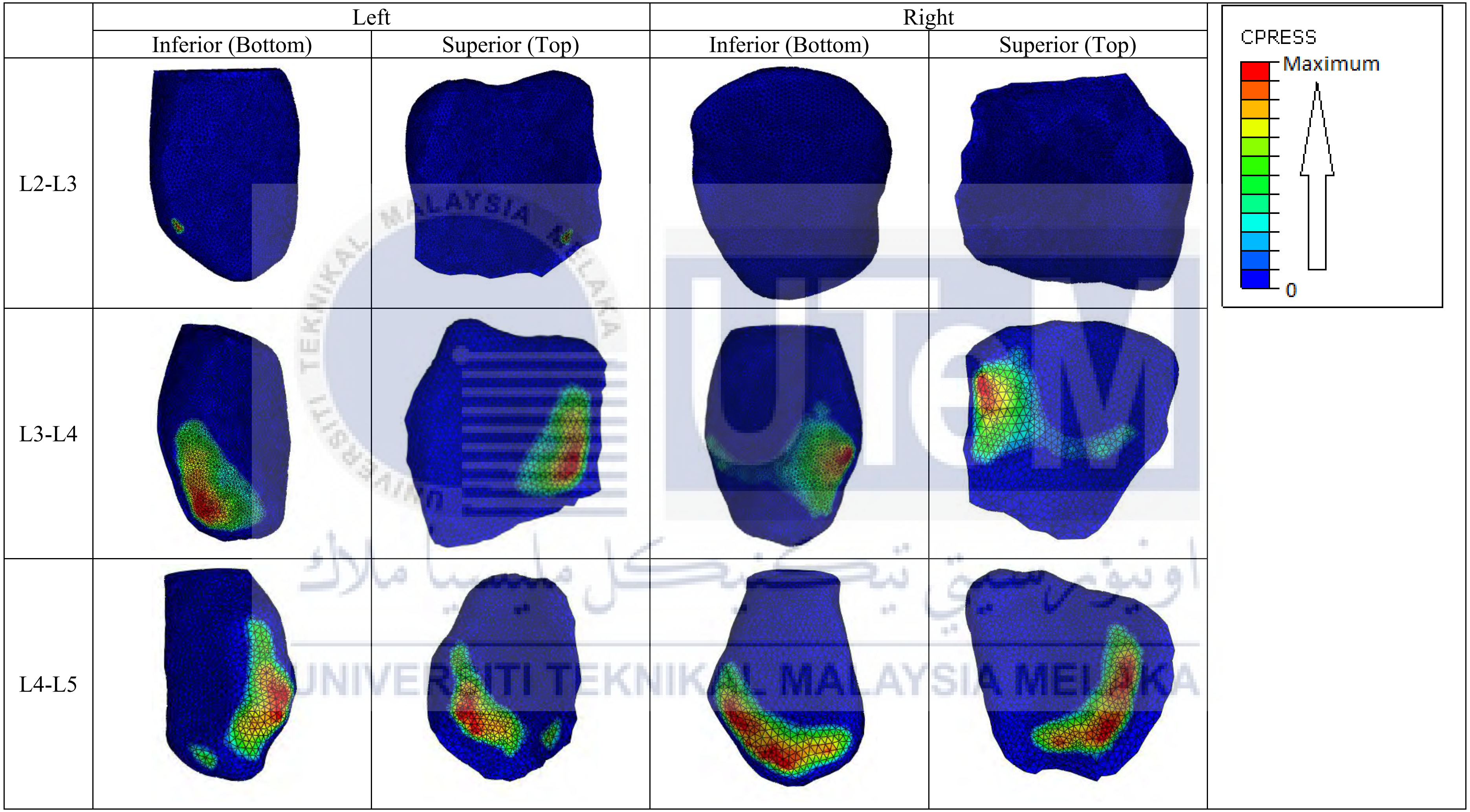

Figure 4.12: Pressure distribution on lumbar facet joint under normal weight during extension motion.

Based on the Figure 4.7, Figure 4.8, Figure 4.9, and Figure 4.10, the results shown that there were a presence of contact pressure on the left facet surfaces at L2-L3, while the contact pressure were absence on the right facet surfaces at L2-L3. These results indicated that there were no contact on the right facet surfaces at L2-L3 of lumbar spine and the difference of contact pressure between right facet and left facet shown an asymmetry behaviour of facet joint. This asymmetry problem also found to be appear in many of the previous studies (Du et al. 2016; Kuo, et al. 2010). Figure 4.11 shown that just a small area in left facets at L2-L3 were in contact when subjected to flexion motion and extension motion. Since the right facets at L2-L3 were not in contact, the contact at the left side of L2-L3 may due to the asymmetry factors of lumbar spine.

It was found that the facet contact pressure increased as the compressive load increased for most of the cases except for L2-L3 of lumbar spine which shown a different trend. For both flexion and extension motions, L3-L4 of lumbar spine experienced highest contact pressure compared to other level of FE lumbar spine model. These results exhibited the same trend as the previous studies which also indicated that L3-L4 level of lumbar spine experienced the highest contact pressure among 4 levels at lumbar region (Du et al. 2016). At L3-L4, the highest contact pressure recorded during flexion motion was 11.77 MPa under obese compressive loading and this value was 82.56 % and 32.62 % higher than the contact pressure at L2-L3 and L4-L5 respectively under the effect of same compressive loading. For the extension motion, the highest contact pressure was 17.34 MPa at L3-L4 and this value was 88.09% and 45.68 % higher than the contact pressure at L2-L3 and L4-L5 respectively under the effect of same compressive loading. These results can be related to the intersegmental rotation of lumbar spine, where the intersegmental rotation started to decrease at L3-L4 and increased back at L4-L5 of lumbar spine. The reduced of intersegmental rotation at L3-L4 was due to the increase of contact between the L4 superior facet and L3 inferior facet restrained the motion of lumbar spine and this eventually lead to the increase of contact pressure on the facet surfaces at L3-L4 of lumbar spine.

The contact pressure on the facet surfaces was highest during extension compared to the flexion motion of lumbar spine. The highest contact pressure when subjected to normal weight load, overweight load and obese weight load during extension motion were 15.53 MPa, 15.9 MPa and 17.34 MPa respectively while the highest contact pressure when subjected to normal weight load, overweight load and obese weight load during flexion motion were 5.563 MPa, 7.965 MPa and 11.77 MPa respectively. This phenomena was due to the reaction of facet joint in different types of motions. As explained in the earliest of this chapter, the inferior facets moved toward the superior facet during extension, while the inferior facets moved away from superior facet during flexion motion. Thus, the increase of contact during extension motion lead to a higher contact pressure on the facet surface.

The percentage difference of contact pressure on facet surfaces between normal weight and obese weight was higher in flexion motion compared to the extension motion. The maximum percentage difference of contact pressure on facet surfaces between normal weight and obese weight in flexion motion was 114% at the left L4 superior facet surface, while the largest percentage difference of contact pressure on facet surfaces between normal weight and obese weight in extension motion was 75.6% at the left L3 superior surface. This result shown that the increment of human weight affected the change of contact pressure more significantly in flexion motion than in extension motion.

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CHAPTER 5

CONCLUSIONS AND RECOMMENDATIONS FOR FUTURE RESEARCH

5.1 Conclusion

This project has developed a verified FE lumbar spine model for the investigation of human weight effects on the kinematics of human lumbar spine and facet joint. The FE model of human lumbar spine can be used for repeated simulation which is more cost-efficient, time saving and ethical concern.

The increasing of human weight has proved to impact the kinematics of human spine and lead to a higher contact pressure on the lumbar facet surfaces. The results has shown that higher contact pressure on the facet surfaces during extension motion compared to the flexion motion due to the biomechanical behaviour of facet joint. During extension motion, the highest contact pressure was 17.34 MPa and this value was 32.21 % higher than the contact pressure during flexion motion, which recorded a value of 11.77 MPa. These result has proved the roles of facet joint in restrict motion of lumbar spine in extension motion.

Since the increasing of human weight resulted in a larger intersegmental rotation and contact pressure, these can increase the risk of facet degeneration which will eventually lead to the low back pain. Results have found that the increasing of human weight from normal weight to obese weight increased the contact pressure for a maximum of 114% at left superior L4 facet surface during flexion motion and a maximum contact pressure of 75.6 % at the left L3 superior surface.

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5.2 Recommendation for future research

The existence of geometry asymmetry in the FE lumbar spine model is a factor which will influence the accuracy of the simulation. The asymmetry of lumbar spine may lead to some of the torsion motion or lateral bending motion. Thus, future studies should pay attention to the asymmetry problem of the lumbar spine and try to eliminate this problem for a more accuracy result. Since every human being have a different lifestyle and undergo different daily activities tasks, thus, body components will undergo minor changes in order to adapt itself to the living environment. Factors such as age, occupational effects, gender, health condition and genetic factors should also take in consideration in the future studies.

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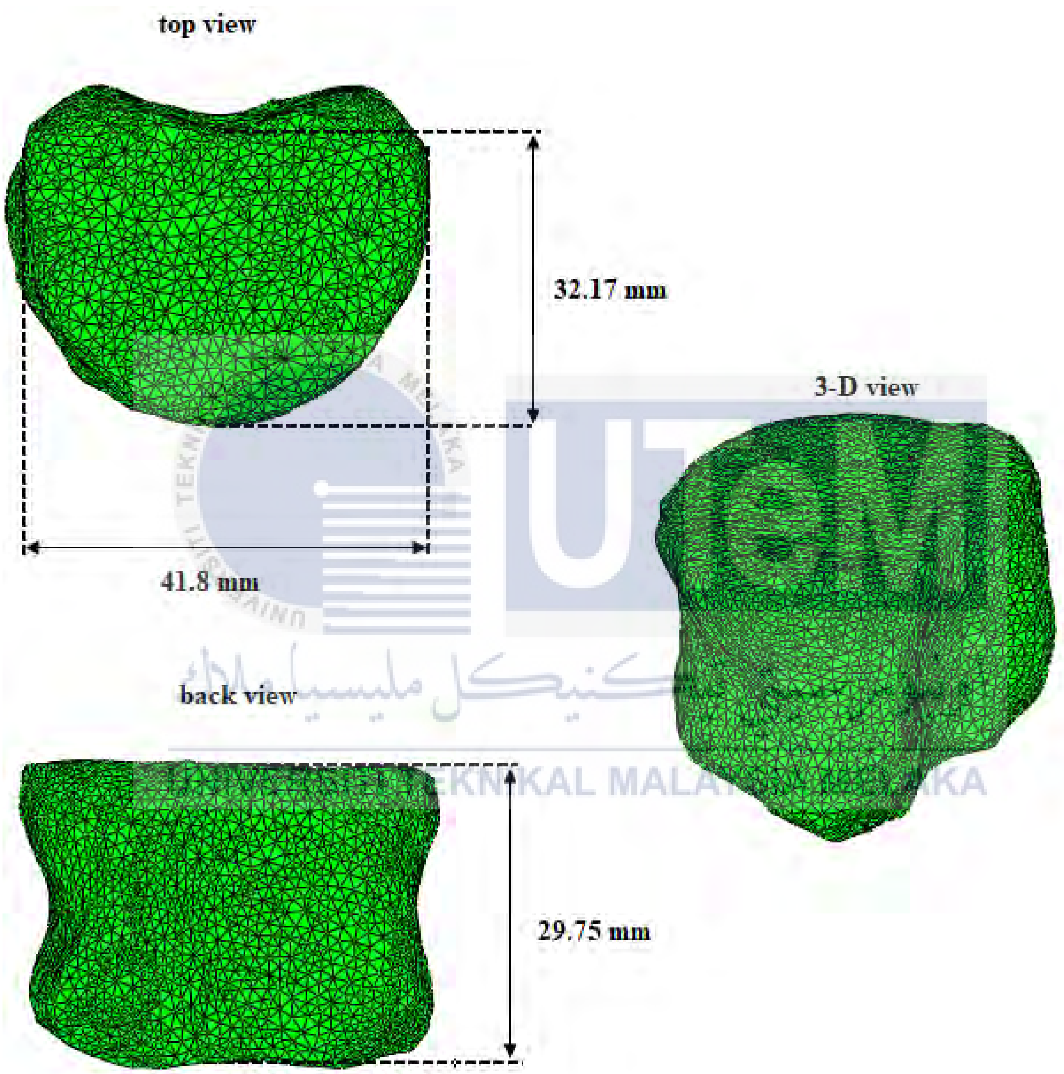
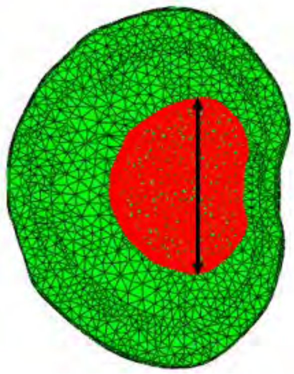
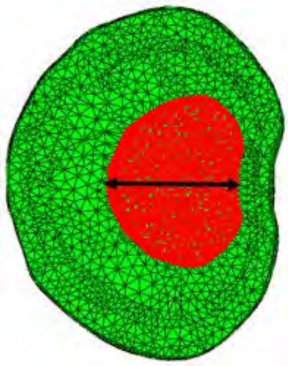
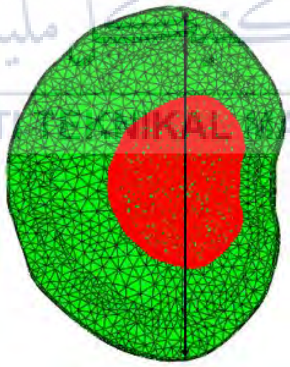
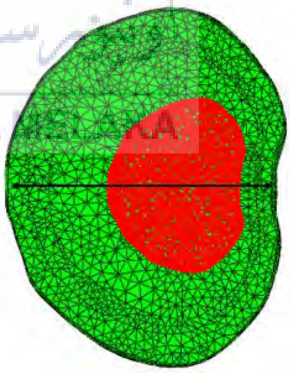
APPENDIX A1

L1 anterior vertebrae dimension details



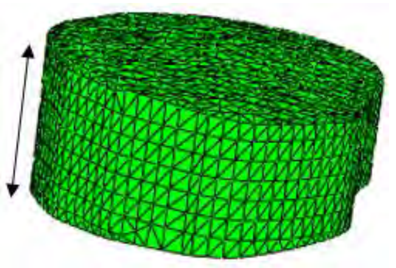
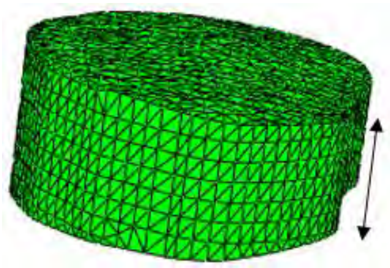
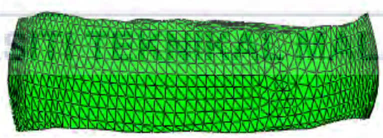
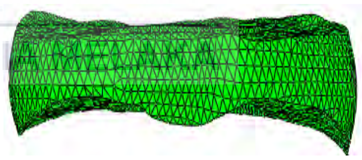
APPENDIX A2

Disc Specification (Length and Width)

		Length (mm)	Width (mm)
Nucleus	Nucleus location		
	L1/L2	20.20	16.16
	L2/L3	20.82	16.66
	L3/L4	21.35	16.73
	L4/L5	21.42	16.76
Disc		Length (mm)	Width (mm)
	Disc location		
	L1/L2	44.49	34.96
	L2/L3	45.04	33.48
	L3/L4	47.27	33.10
	L4/L5	51.83	48.99

APPENDIX A3

Disc Specification (Thickness)

		Anterior	Posterior
Nucleus	**Nucleus location**		
	L1/L2	8.31	5.67
	L2/L3	9.47	7.51
	L3/L4	10.17	8.18
	L4/L5	10.20	6.8
Disc		**Anterior**	**Posterior**
	Disc location		
	L1/L2	9.39	7.77
	L2/L3	10.25	8.05
	L3/L4	11.43	10.04
	L4/L5	13.42	6.83

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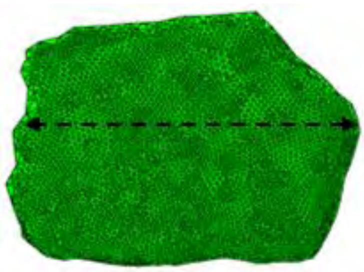
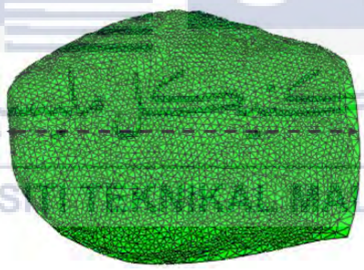
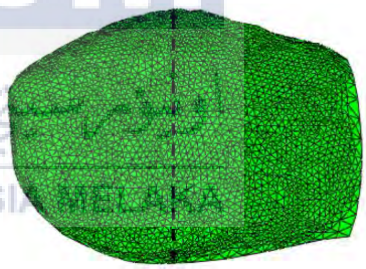
APPENDIX A4

Superior and Inferior Facet Joint Specification

<table>
<tr><td rowspan="6">Superior</td><td rowspan="2">Facet location</td><td>Length (mm)</td><td>Width (mm)</td></tr>
<tr><td></td><td></td></tr>
<tr><td>L2</td><td>15.25</td><td>12.23</td></tr>
<tr><td>L3</td><td>16.59</td><td>12.82</td></tr>
<tr><td>L4</td><td>16.71</td><td>16.09</td></tr>
<tr><td>L5</td><td>18.63</td><td>16.86</td></tr>
<tr><td rowspan="6">Inferior</td><td rowspan="2">Facet location</td><td>Length (mm)</td><td>Width (mm)</td></tr>
<tr><td></td><td></td></tr>
<tr><td>L2</td><td>13.92</td><td>10.73</td></tr>
<tr><td>L3</td><td>14.29</td><td>12.22</td></tr>
<tr><td>L4</td><td>17.44</td><td>13.19</td></tr>
<tr><td>L5</td><td>18.57</td><td>11.86</td></tr>
</table>

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APPENDIX B1

Case study- simulation data results

Flexion	Contact Pressure (Mpa)											
	Superior						Inferior					
	Left			Right			Left			Right		
Load	700	900	1100	700	900	1100	700	900	1100	700	900	1100
L1-L2	0	0	0	0	0	0	0	0	0	0	0	0
L2-L3	2.14	1.84	1.656	0	0	0	2.183	1.908	2.053	0	0	0
L3-L4	5.082	7.496	10.88	4.335	6.845	9.016	5.296	7.216	10.01	5.563	7.965	11.77
L4-L5	5.534	6.416	7.065	5.947	7.732	8.261	5.116	6.182	6.981	5.619	7.314	7.93

Extension	Contact Pressure (Mpa)											
	Superior						Inferior					
	Left			Right			Left			Right		
Load	700	900	1100	700	900	1100	700	900	1100	700	900	1100
L1-L2	0	0	0	0	0	0	0	0	0	0	0	0
L2-L3	1.156	1.068	2.03	0	0	0	1.183	1.227	2.066	0	0	0
L3-L4	15.25	15.65	16.91	13.27	13.9	15.99	14.47	14.83	16.42	15.53	15.9	17.34
L4-L5	8.511	8.588	9.155	8.633	8.732	9.316	8.581	8.606	9.925	8.617	8.722	9.419

Flexion	ROM (°)		
Load	700	900	1100
L1-L2	1.945522	1.550565	1.157617
L2-L3	1.088802	2.750918	4.358688
L3-L4	1.412292	2.545405	3.620174
L4-L5	1.227982	2.480931	3.619548

Extension	ROM (°)		
Load	700	900	1100
L1-L2	3.268152	3.185132	3.002972
L2-L3	8.263506	8.626074	8.696018
L3-L4	6.46112	7.082135	7.112883
L4-L5	7.052692	7.315937	7.404271