FINITE ELEMENT ANALYSIS OF LUMBAR SPINAL CORD FOR WEARERS OF HIGHHEELED FOOTWEAR

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RESEARCH REPORT SUBMITTED IN PARTIAL FULFILLMENT OF THE REQUIREMENT FOR THE DEGREE OF MASTER OF ENGINEERING

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Abstract

It has already been reported by many researches about the possible negative effects due to wearing high heel shoes including, feeling pain in lower part of the back due to curvature increment of spinal, and feeling foot pain as consequent of placing more weight and finally more pressure on the toe feet. The objective of recent study is to investigate the stress distribution on lumber intervertebral disc of L4/L5 using a three dimensional finite element analysis during standing on high heels with three different heights. Therefore, a 3D finite element model of Functional Spine Unit (FSU) including L4, L5 and intervertebral disc was generated based on MRI scan files of a young female which was further modified in CATIA CAD modeling software and meshed and loaded in ANSYS finite element analysis software. Simulation results yielded a compression stress distribution of 3 MPa due to bending of forward flexion of FSU and downward displacement of 2 mm due to using elevated shoes with 8 cm height. Clinical concerns with prolonged standing on high elevated shoes with 8 cm height would be addressed to disc thinning observed in simulation results of the current study which can be resulted in low back pain. Therefore avoiding of prolonged standing on the high elevated shoes is advisable.

Banyak laporan kajian telah ditulis mengenai kesan negatif yang mungkin berlaku akibat memakai kasut tumit tinggi, termasuk rasa sakit di bahagian bawah belakang disebabkan meningkatnya kelengkungan tulang belakang, dan rasa sakit kaki akibat meletakkan lebih berat dan akhirnya lebih tekanan pada kaki-kaki. Objektif kajian baru ini adalah untuk menyiasat pengagihan tegasan pada cakera intervertebral lumber L4/L5 menggunakan analisis unsur terhingga tiga dimensi semasa berdiri di atas tumit tinggi dengan tiga ketinggian yang berbeza. Oleh itu, model unsur terhingga 3D Unit tulang belakang Fungsian (FSU) termasuk L4, L5 dan cakera intervertebral telah dijana

berdasarkan fail imbasan MRI wanita muda yang seterusnya diubah suai dalam perisian pemodelan CAD CATIA dan terjalin dan dimuatkan dalam perisian analisis unsur terhingga ANSYS. Keputusan simulasi menghasilkan pengagihan tegasan mampatan 3 MP disebabkan lenturan akhiran hadapan FSU dan anjakan ke bawah 2 mm kerana menggunakan kasut setinggi 8 cm. Kebimbangan klinikal terhadap berdiri yang berpanjangan pada kasut tumit tinggi dengan ketinggian 8 cm akan ditujukan kepada kemerosotan cakera secara beransur-ansur dan penipisan cakera yang menyebabkan sakit belakang. Oleh itu, mengelakkan berdiri dengan masa yang berpanjangan pada kasut tumit tinggi adalah dinasihatkan.

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List of Symbols and Abbreviations

CMB	Center of mass of the body
FEA or FEM	Finite element technique
IGES	Initial Graphics Exchange Specification
NURBS	Non-Uniform Rational B-Spinal
FSU	Functional Spine Unit
IVD	Intervertebral Disc
PDW	Pedicle cross-section width
PDIt	Pedicle axis inclination to the transverse plane
PDH	Pedicle cross-section height
PDls	Pedicle axis inclination to the sagittal plane
MRI	Magnetic Resonance Imaging
FL	Follower Load
DOF	Degree of Freedom
NED	Number of Element Divisions
MR	Mooney - Rivlin
CT	Computed Tomography

1. Introduction

1.1 High heels

High heel term is referred to the types of the shoes which the heel is placed indicatively higher than the toes generally with intention to aesthetic approaches of looking taller and slender (Benslimane 2012).

1.2 Effects of high heels on different body limbs

It has been reported that spine can be put out of its natural alignment due to wearing high heels which is result in more pinch of spinal nerves and finally feeling of numbness, pain and tingling (Opila et al., 1988). This can be referred to impose of more arches on lower back and consequently more strain of muscles which support the lower back.

1.2.1 Posture and Gait

The adopted posture is to compensate the spine misalignment in the form of forward leaning of the lower parts of body concurrent with leaning back of upper parts because of more pressure existence on forefoot in order to satisfy the mechanism of equilibrium and body balance. The existence pressure stems from a plantarflexed position of foot due to wearing high heels (Opila et al., 1988).

1.2.2 Spine and Low Back

On the other side because of more plantarflexion of foot, over work of hip flexors and forward flexion of knee are required to compensate the lack of power arisen in the moment of push off the ground (Esenyel et al., 2003). The S-curve is a normal back posture of individuals which operates as a shock absorber to reduce the vertebrae stress.

However wearing elevated shoes causes the lumbar flattening and it forces the thoracic and head to posterior displacement.

Therefore wearers have to reduce the forward curve of their lower back by leaning forward and consequence of overuse of muscle can cause back pain (Opila et al., 1988).

1.2.3 Hips and Knees

Permanently wearing high heels may result in chronically overused of hip flexors which are located upper front of thigh and consequently shortening and contracture of these muscles which can lead to lumber spin flattening that is a body misalignment and can cause ununiformed weight distribution on body skeletal system (Ebbeling et al., 1994).

Wearing the high elevated shoes causes the knee to maintain flexed and also inward turning of tibia which results in generation of more compressive force on medial of the knee. On the other side due to increase in the height from the floor, extent of knee torque is increased as well. All these interpretation can intensify the risk of osteoarthritis (Kerrigan et al., 2005; Kerrigan et al., 1998).

1.2.4 Ankles and Feet

The freedom of Ankle motion, Ankle power and length of calf muscles are restricted by wearing high elevated shoes and the loss of power due to shortened muscles is often appeared during push off (heel raise) part of stance phase of gait cycle. Another controversial concern is discussed in the risk of insertional Achilles tendonitis (J.-Y. Lee et al., 2010).

This problem may be caused by pulling of the Achilles tendon at the point of attached to the back of calcaneus due to shortening of Achilles tendon because of the position adopted by the ankle (Ebbeling et al., 1994). Downward position of foot due to wearing high heels can lead to placing more pressure on forefoot and consequent deformities like bunions, neuromas. Furthermore this situation results in Haglund deformity as well due to alters in pulling direction of achilles tendon as the foot will be more supinated (Snow and Williams, 1994).

High heels cannot offer proper support during walking and affiliate extra movements including sloshing and wobbling besides teetering over the toes that can be transferred up the leg toward the spine vertebra and disks. Therefore push the body alignment to be changed to provide the stability and balance by forward movement of center of gravity (Snow and Williams, 1994).

Indeed, foot pain, foot deformities such as valgus hallux (Al-Abdulwahab and Al-Dosry, 2000) are mentioned as concerns against using high heels. Other researchers investigated concerns such as ankle sprain, calluses and corns, tight heel cords and plantar forefoot pain (Stephens, 1992; Ebbeling et al., 1994).

1.3 Objective of the current study

Current study investigates the stress distribution on the lumber spine due to wearing different elevated heels with more concentrate on medial to inferior structure of the spine includes lumbar intervertebral discs. Therefore, the objective of recent study is to investigate the stress distribution on lumber intervertebral disc of L4/L5 using a three dimensional finite element analysis during standing on high heels with three different heights.

All the past literatures has just concluded the increase of low back stress due to wearing high heels based on conceptual dependency and interrelation between increment of force and subsequently stress increment. Due to the limitations of experimental attitudes including individualized features of participants spine shape and properties and aggression with attaining data using cadaver studies and lack of directly engineered approaches to implement a comprehensive stress analysis of the effects of high heels on low back structure, simulation of stress distribution over the lumbar disc structure can promote the better understanding of vertebra spine reactions against wearing high heels. Therefore, the significance of the study is to use an accurate computerized analysis for every small element of the problem and providing a realistic simulation for spine behavior under investigated circumstances.

2. Literature review

The literature relevant to the recent research is covered under two sections. The first part offers a review of the literature concerning the biomechanics of wearing high heels and it focuses on reactions and positioning of body limbs including ankle, spine, foot, knee against the changes and misalignments imposed by using elevated shoes. The second section looks at research conducted on finite element analysis of vertebra spine and details provided about the recent findings in this area.

2.1 High heels effects on Spine

The early time of interest in the use of wearing high heel shoes by women was referred to middle of 16 century to early of 17 century in France in order to fulfill the passion of looking more beautiful (Linder and Saltzman, 1998).

It has already been reported by many researches about the possible negative effects due to wearing high heel shoes including, feeling pain in lower part of the back due to curvature increment of the whole spinal, and feeling of foot pain as a consequence of placing more weight and finally more pressure on the toe feet (Hyun and Kim, 1997;

Jang and Kim, 1998; Yoe, 1994), increase in possibility of ankle sprain generation (Nieto and Nahigian, 1975), reduction in the length of Achilles tendon (Scholl, 1946) and more anterior movement of centre of mass of the body (CMb) due to wearing high heel (C.-M. Lee et al., 2001).

On the other side, some studies have been considered the clinical aspects of the issue because of wearing high heel shoes such as, raise in consumption of oxygen extent (Mathews and Wooten, 1963; Ebbeling et al., 1994) and also increase in potential of knee osteoarthritis (Kerrigan et al., 1998) but those who have previously studied the issue of back pain, they more focused on electrical activity of the lower parts.

Furthermore, Most of the studies have only focused on the gait deficiencies imposed by elevated shoes including alters in velocity of walking and mobility of individuals (Murray et al., 1970; Alexander, 1992) and reduction of the length of stride (Nieto and Nahigian, 1975). These groups of studies mostly focused on motion analysis of individuals gait and their performances due to wearing high heels during walking or while static standing.

It was also used a compound method to more focus on biomechanical effects of wearing high heels on low back changes and alteration angel of lumbar disks of L4 and L5 (C.-M. Lee et al., 2001). The study used combination of (Yoo, 1997) technics to determine the changes of the lumbar flexion angle besides applying the motion analysis technique to simulate the walking phase of wearing high heels for comparing the alteration angles of the lumbar spine during static phase and dynamic phase and to measure the pressure on the lumbar spine. It was also investigated the vertical movement of center of body mass (BM) during walking as well as stationary posture.

Therefore, it was found that for each 1 cm higher heel the flexion angle of the truck is reduced 1 degree as well (Lee et al. 2001) in order to satisfy equilibrium condition of the body by utilizing a compensatory mechanism. This compensatory mechanism restitutes the body from anterior movement of the CBM and counteracts the resultant feeling of falling forward (Snow and Williams, 1994). Additional compensation reactions by the joints of lower extremities and consequently changes imposed in parameters of gait (Gastwirth et al., 1991) such as reduction in stride length (de Lateur et al., 1991) were already reported as well. It was also hypothesised that the upward movement of the CBM due to wearing elevated shoes results in increment of the lumbar lordosis and finally raise in magnitude of compressive forces on low back (Yu et al., 2008).

2.2 Finite element analysis of spine

The importance of lumbar stability which is offered and controlled by surrounding muscles and nervous respectively, have been reported to be a critical factor in specifying the response of the truck to abrupt loading (Panjabi, 1992; Gardner-Morse et al., 1995;Cholewicki and McGill, 1996). Falls and slips are two frequently incidents which are caused low back injuries by subjecting the lumber spine to the abrupt cycling load (Troup et al., 1981; Bigos, Spengler et al., 1986; Omino and Hayashi, 1992).

The two existed hypotheses express that low back injuries are caused either by improper response of the muscles of truck (Radeboldet al., 2000; Panjabi 1992) or insufficient stabilization of the lumbar before an abrupt loading is applies (Panjabi, 1992; Cholewicki and McGill, 1996). Regulation of the spine muscle force as a linear approximation of muscle stiffness is resulted in control of the spine endurance (Cholewicki and McGill, 1996; Crisco and Panjabi, 1991).

Therefore it has been reported that reinforcement of the truck muscles including antagonistic and agonistic causes the lumbar spine to stay stabilized (Cholewickiet al. 1997; Gardner-Morse et al., 1995). Hitherto, many studies developed the finite element model of different parts of vertebral and spine for further understanding the stress distribution and concentrations due to different situations. One study specifies that the central surface of the vertebral besides inferior region of the facet joints undergoes higher stress concentration as well as superior region (Nabhani and Wake, 2002).

A Finite element model of the entire spine was produced to find the lumbar respond during the human activities which is offered the realistic load distribution (Ezquerro et al., 2004).

1. Methodology

As it was already mentioned current study is a directional research, meaning that predictions have taken place based on previous literatures and other researcher's experience that is the main characteristic of all applied researches. Following steps of using required softwares are presented as instrumentation and procedure. Finite element technique (FEA or FEM) enables us to get the distribution of stress and strain over the entire body in terms of displacement and force applied on each element (Nabhani and Wake, 2002) and (Ezquerroet al., 2004). Following flow chart represents the current of analysis and each parts of research procedures are detailed followed by that as well.

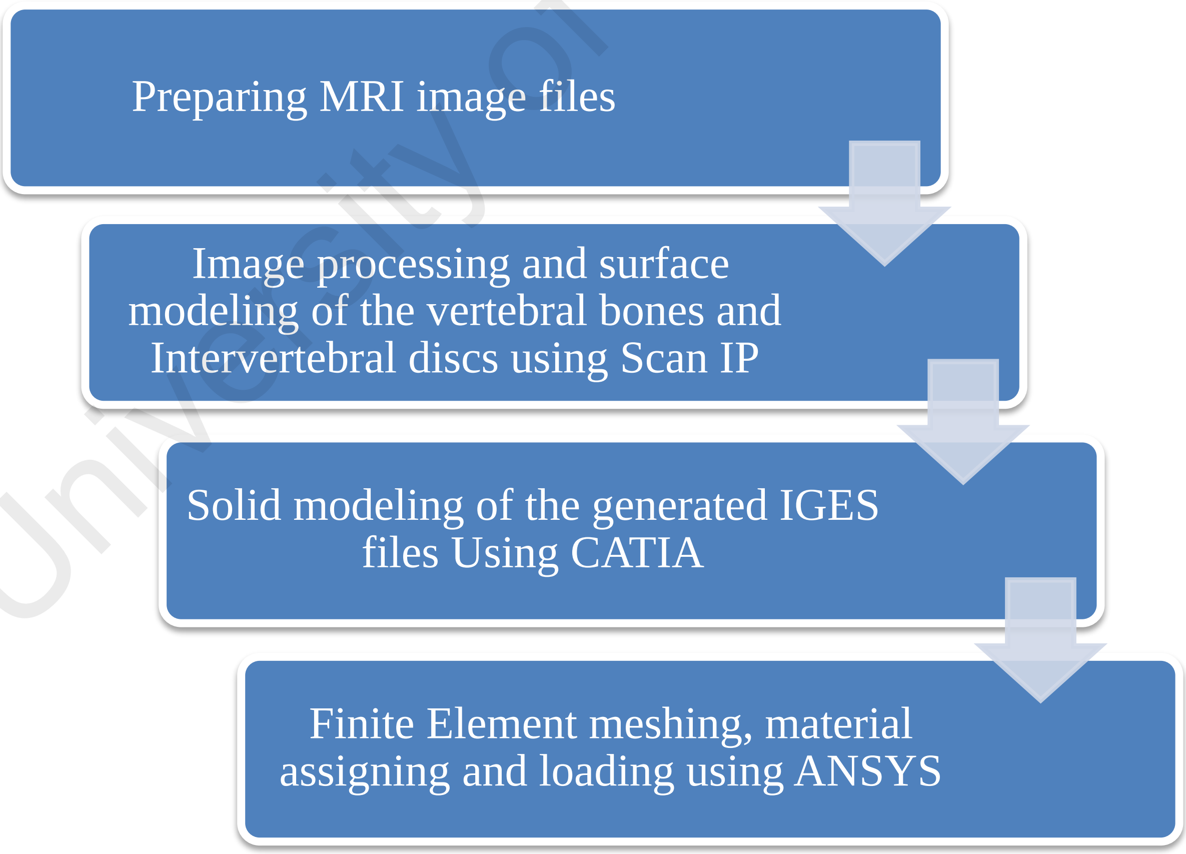

Figure 3-1. Methodology steps

The procedure of Magnetic resonance imaging (MRI) images preparation can be conducted the same as (Li and Wang, 2006) with some differences including the gender of subject which is an adolescent or adult lady of 22 to 26 with the positions of standing on three different high heeled shoes and flat support. Since there is no difference in bone material properties of young female bones compare to the male's therefore, no concern was taken into account in this regard. MRI collections were preceded in ScanIP software in order to generating a 3D model.

1.1 Geometrical validation and preparation of the model

3.1.1Scan IP

The objective of this section is to export both FEA and Non-Uniform Rational B-Spinal (NURBS) models of the Functional Spine Unit (FSU) which is capable of further CAD modifications with the means of associated CAD software, CATIA, as well as creating an importable mesh model for ANSYS (Finite Element software). Consequently initial modifications including model smoothening, contour detection and determination of mesh characteristics are being defined.

- Scan IP segmentation process is used to convert the MRI Scan files to 3D model
- Model is created based on HU units which are grey scale values associated with the material density and known as threshold which is predefined in software.
- Tissues, fat and air are distinguished by lower value of HU while the higher values can be referred to the hard tissue or bones.
- Since the area of interest is lumbar, thresholding of bone should be conducted as well as thresholding the bone tissue for getting more accurate geometery of lumbar. In order to fixing the desired boundary conditions, manual manipulation of boundaries is done by using crop mask.

- For selecting the region which is only represent the lumbar, whole of the spine is cropped and by applying the edit mask at the desired area favorite lumbers (L3-L5) are selected.

As it was briefly mentioned, at the process of components distinguishing, it should be noticed that the gray value of bone is larger than tissues. Therefore, it is necessary to use the manual process of segmentation to adjust the boundaries and the contour detectioning. It's because of discontinuity of boundaries which are caused by automated set up as well as creation of artificial isolated section due to lower cancellous gray value rather than cortical or other soft tissues.

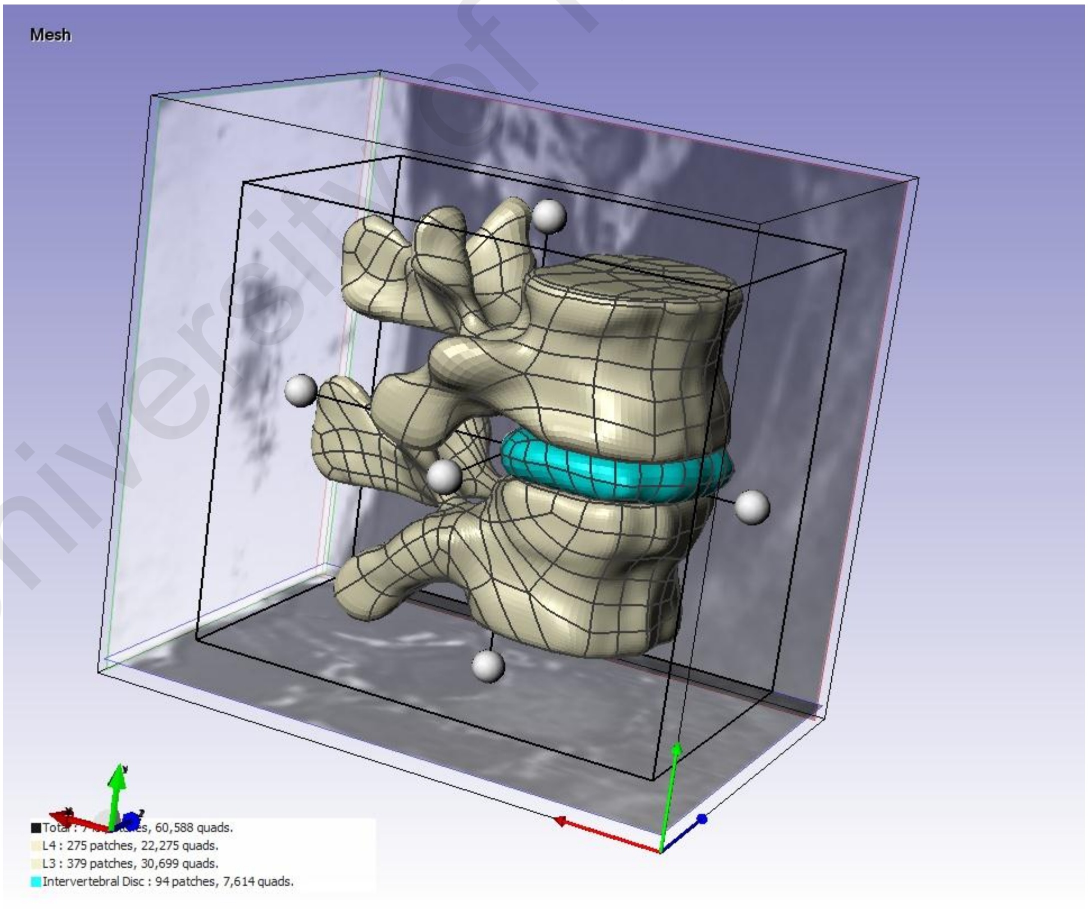

Figure 3-2. NURBES model of FUS based on contour detection method in ScanIP

3.1.2 Geometrical validation

The generated FE model was further compared with the reference dimensions suggested by Augustus A White and Panjabi, 1990; Sylvestreet al., 2007. To this end, anatomical landmarks of the vertebral were defined and geometrical verifications were accomplished by comparing the four important morphometrical parameters including size and angels of the pedicle cross-section width (PDW); pedicle axis inclination to the transverse plane (PDIt); Pedicle cross-section height (PDH); and pedicle axis inclination to the sagittal plane (PDls) (Augustus A White and Pnajabi, 1990). Figure 3-3 provides a graphical representation of the mentioned parameters.

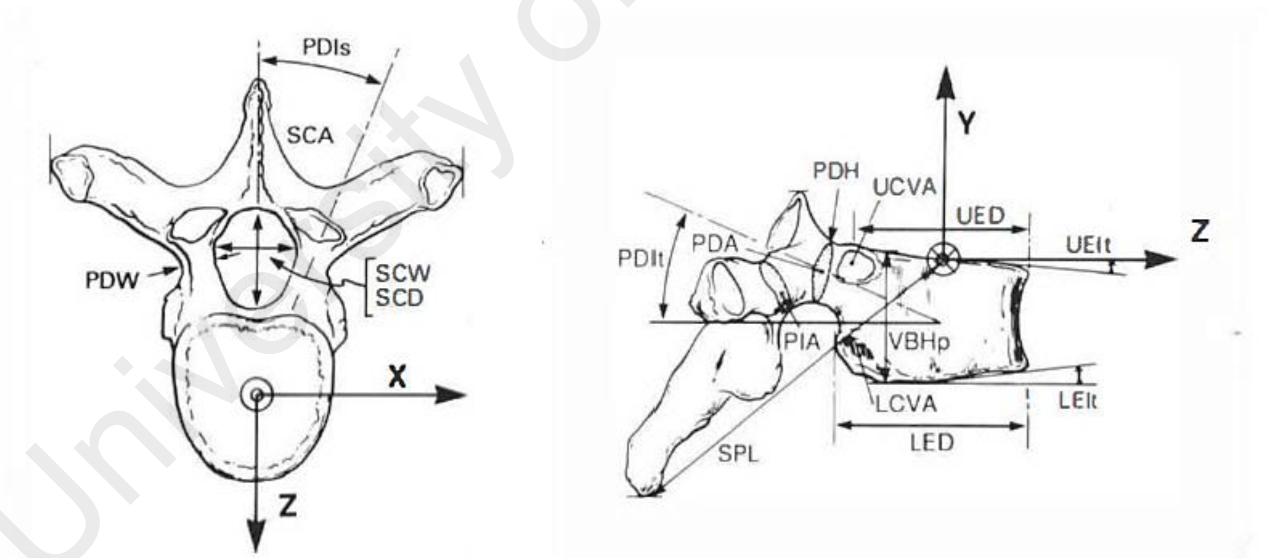

Figure 3-3. Geometrical dimensions and morphometrical parameters
(Augustus A White and Pnajabi, 1990)

Although mechanical properties of the vertebrae clinically do not differ from male to female but factors including shape, size and physical properties vary based on the age of individuals. Therefore, vertebras of different locations in the spine have approximately similar shape and design but parameters such as mass and size of the vertebra are increased from cervical to lumbar region. The reason for such a mechanical adaptation is to offer a sufficient support against compression loads which are increased progressively.

3.1.3 CATIA

The Next attempt is to modify the Initial Graphics Exchange Specification (IGES) model and defining the acting points of the Follower Load (FL), sketching the ligaments and further resizing the model components based on the suggested dimensions in reference studies. CATIA V5-R20 (Dassault Systemes) solid design software was used for solid modeling and modification of some parts such as cartilaginous endplates, tendons and ligaments which are not offered by primary model created by Scan IP. CATIA shape design module was used in order to joining the NURBS slices to provide a uniform surface and volume model of all components including vertebras, cartilages and intervertebral disc was assembled to complete a functional unite of spine (FSU) to investigate the distribution of stress specifically on intervertebral disc (IVD).

Jointed NURBES were transferred into the Mechanical part design environment for further solid modifications. Figure 3-4 indicates the process of CAD model preparation on CATIA shape design environment and steps taken toward generating a uniform jointed surface. More notices should be taken when coordinate system is transferred from one of the software to another. Accuracy of the final result which is mostly depends on the uniformity of distributed elements is mainly up to the smoothness of the surface that initially defined by Scan IP. Proper preparation of the initial three dimensional model in Scan IP can widely promotes the joining process and consequently procedure of thickness assignment to the slices.

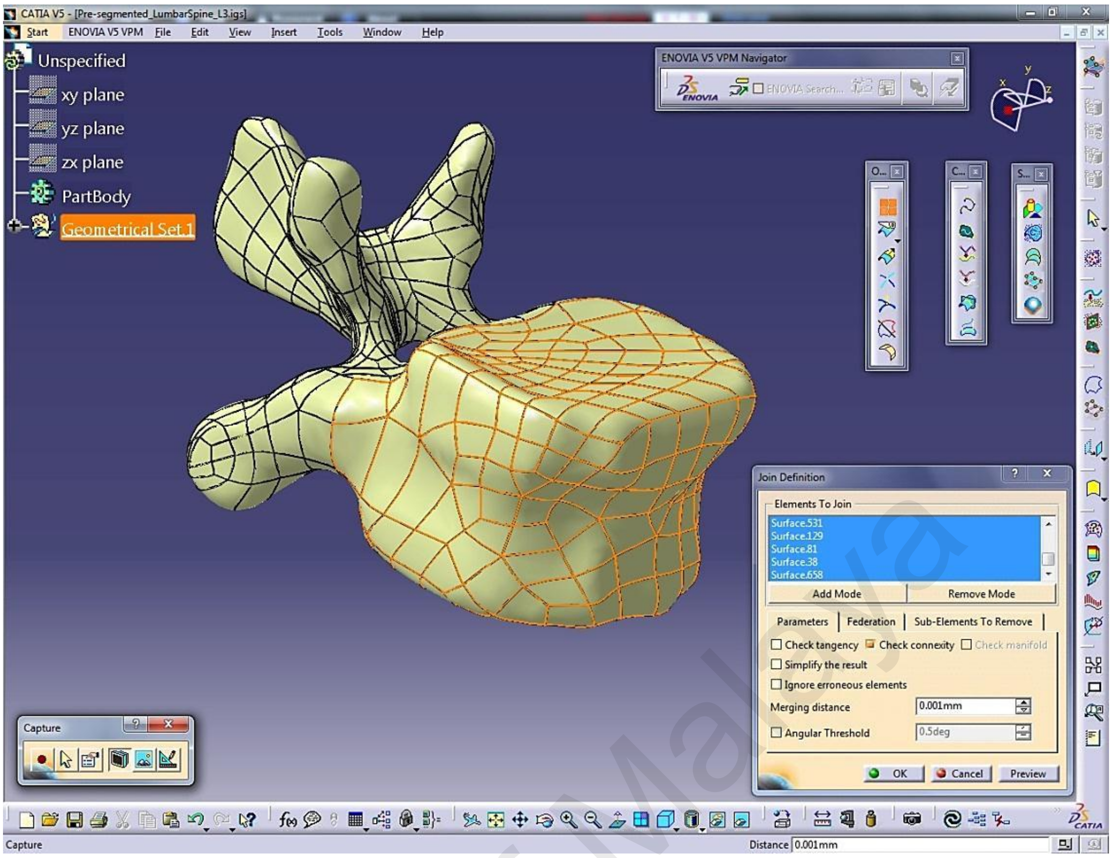

Figure 3-4. Preparation of solid model from NURBS slices

3.2 ANSYS (Finite Element Analysis)

3.2.1 Material properties and meshing

Finally assembled model is imported to the ANSYS 12.0 software for conducting the finite element analysis including pre-processing and post-processing which are consisted for defining; material properties of different parts of the model (Y.-X., 1999;Wong et al., 2003;Tanaka et al., 2004), operating load on lumbar, boundary conditions including constraints known as degree of freedom (DOF) (Y.-X., 1999), meshing, defining the general solution of the problem, interpolation function and invariant determination of the constitutive equations at each nodes. Following table briefly details the requirements for defining the element information such as type, and material properties which are necessary for modeling the several components of the L4–L5 motion segment (Li and Wang, 2006).

Table 3-1. Material properties of FSU componants

Component	Material	Element type	Invariant	
	Hyper elastic (Incompressible Material)			
Annulus matrix	Nonlinear Hyperelastic Mooney Rivilin	layered elements 3D-8 Nodes	C1,C2 D1,D2	8.00E005 4.5E006
Nucleus	Nonlinear Hyperelastic Neo-Hookean	Hyperelastic 3D - 8 Nodes	C10 D	0.16 0.024-1
Bone (Cortical and Cancellous)	Linear orthotropic	isoperimetric 3D - 8node	Ex,Ey,Ez PR (υ)	Polikeit[1]Sylvestre[2]
Growth plats (cartilaginous endplates)	Linear isotropic	Isotropic 8-Node solid hexahedral elements	E, PR	Polikeit[1]Sylvestre[2]
Ligaments	Linear isotropic	Bi-linear truss elements	E, PR	Polikeit[1]Sylvestre[2]

Reference to; (Goel et al., 1995;del Palomar et al., 2008; Sharma et al., 1995;Hibbit et al., 2004;Moramarco et al., 2010;Polikeit[1] et al., 2003;Sylvestre[2] et al., 2007;Kheng et al.,;Renner et al., 2007).

Intervertebral disc was considered as a hyperelastic disc covered by two cartilaginous endplates known as growth plate with the thickness of 2mm obtained from histological measurements (Esenyel et al., 2003;Linder and Saltzman, 1998;Mathews and Wooten, 1963). Annulus ground matrix was defined as a composite material made up of 3 lamina layers (Shirazi-Adl et al., 1986) with 12 embedded fiber bands (Polikeit et al., 2003). Based on the surface area ratio of the nucleus to the annulus ground matrix, intervertebral disc was divided into the two distinct parts including nucleus part with the surface area of 43% out of the total surface area of the whole disc (Smit et al., 1997; Polikeit et al., 2003) and annulus fiberus volume. Figures 3-5 represents both internal instruction of the annulus fiberus ground matrix and assembled model of IVD has been shown in figure 3-6.

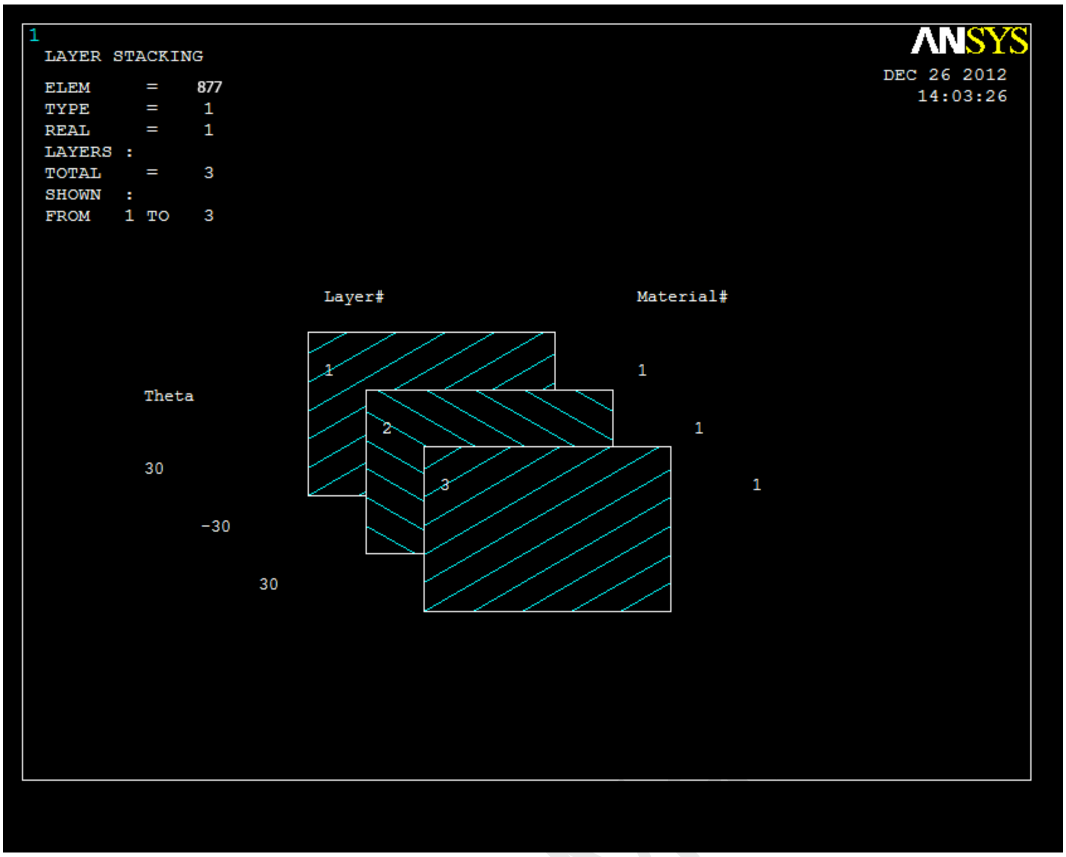

Figure 3-5. Annulus fiberus ground matrix instruction (fibers angle of ±30)

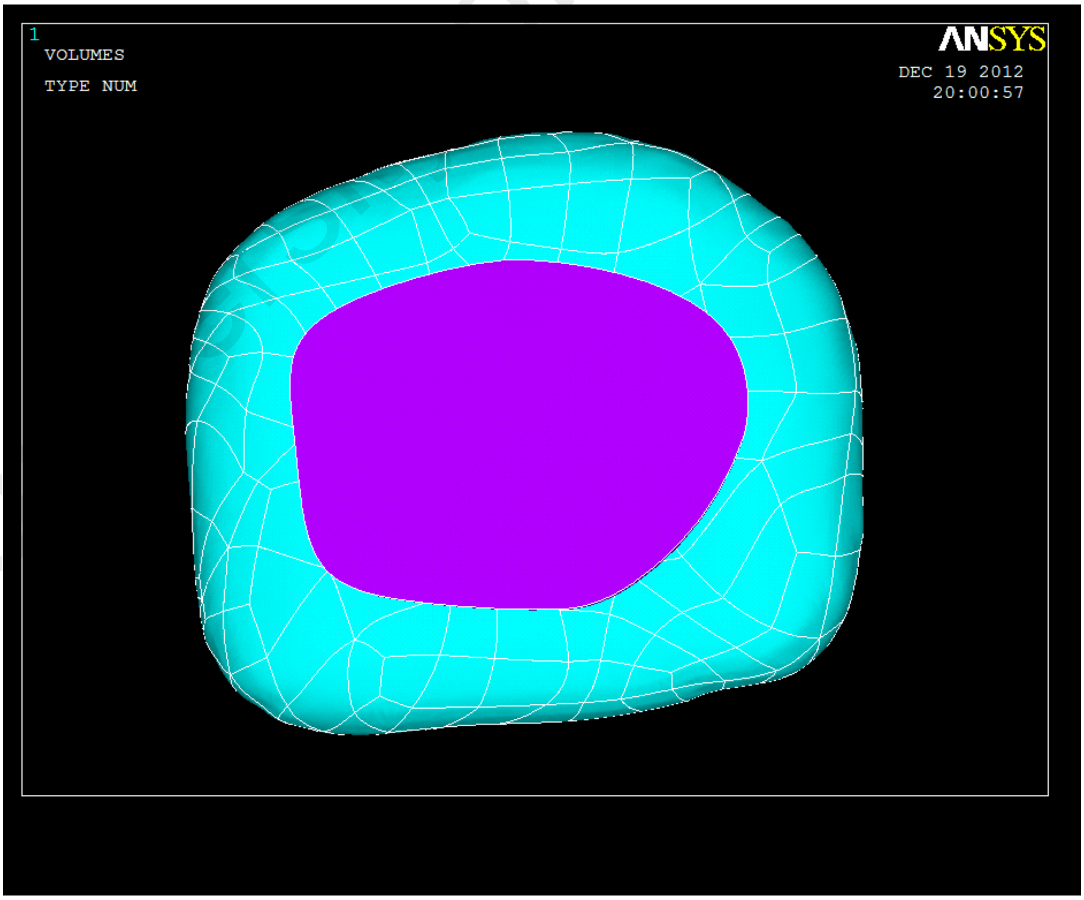

Figure 3-6. Intervertebral disc instruction in ANSYS (Purple region: Nuclus; Blue region: Annulus)

Contact surface between two articular processes of the consecutive vertebras, known as facet joint, was modeled as a cartilaginous gap of 5mm (Rohlmann et al., 2009a) and was solved as a frictionless contact problem (Chosa et al., 2004; Sylvestre et al., 2007). Locations and number of the element divisions (NED) for ligaments has been collected in following table (Sylvestre et al., 2007), figures 3-7 and 3-8.

Table 3-2.Locations and mesh details of the FSU ligament

Ligament	NED	Location
Anterior longitudinal elements (ALL)	10 Uniaxial Elements	IVD to the endplates anterior face
posterior longitudinal (PLL)	6 Uniaxial Elements	IVD to the endplates posterior face
flavum (LF)	3 Uniaxial Elements	Connected to the lamina surface
Capsular (JC)	8 Uniaxial Elements	Connects the periphery of adjacent facet joints
interspinous (ISL)	4 Uniaxial Elements	Attached to the adjacent spinous processes
supraspi- nous (SSL)	3 Uniaxial Elements	Attached to the adjacent spinous processes
Intertransvers (ITL)	2 Uniaxial Elements	Joining adjacent transverse processes

Final assembled and meshed model of all components including; vertebral discs of L4 and L5, intervertebral discs (annulus fibrus and nucleusporpuls) and all seven ligaments of ALL, PLL, LF, JC, ISL, SSL and ITL are shown in the figures 3-7 and 3-8. Ligaments were constructed in the ANSYS software using “modeling” tools in order to define straight lines. As an instance, to define the ISL ligament, three parallel lines were created by defining curve-tangent lines between lower curve of spinous process of L4 and upper curve of the spinous process of L5. Then this straight line meshed by two dimensional truss elements. Similar steps were taken to create rest of the ligaments; however more simple way would be to mesh the FSU first then create ligament lines using generated elements.

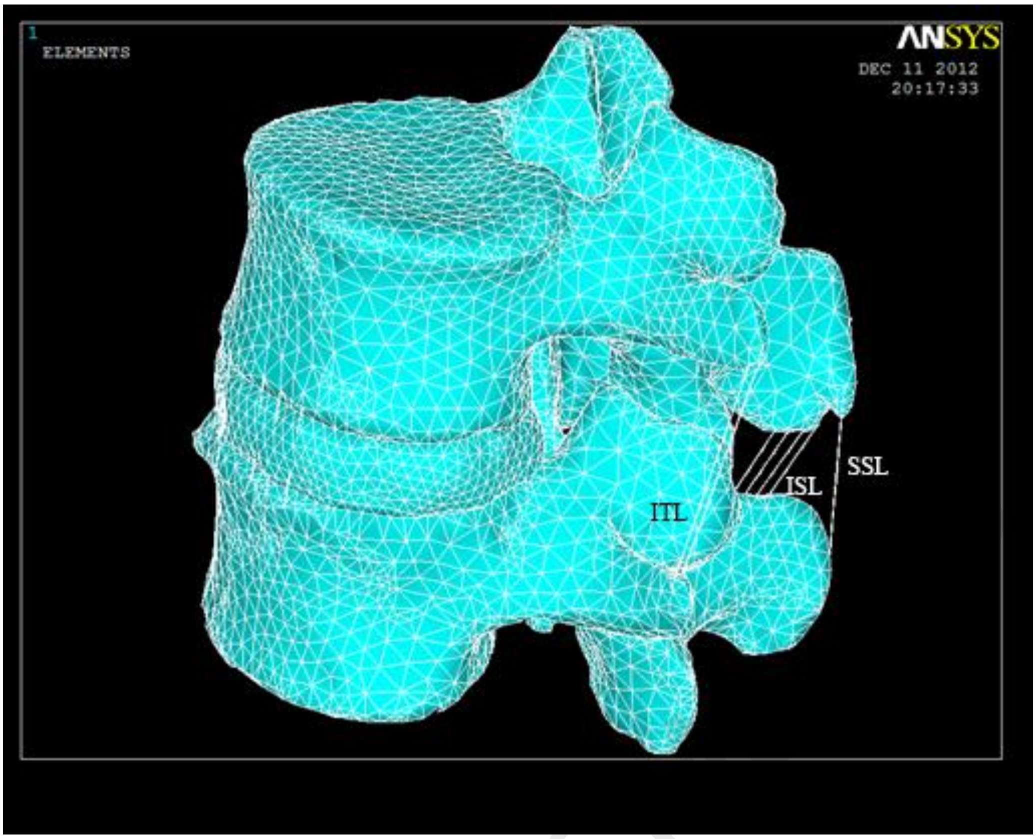

Figure 3-7. Locations of ligaments in sagittal plane (SSL, ISL, ITL)

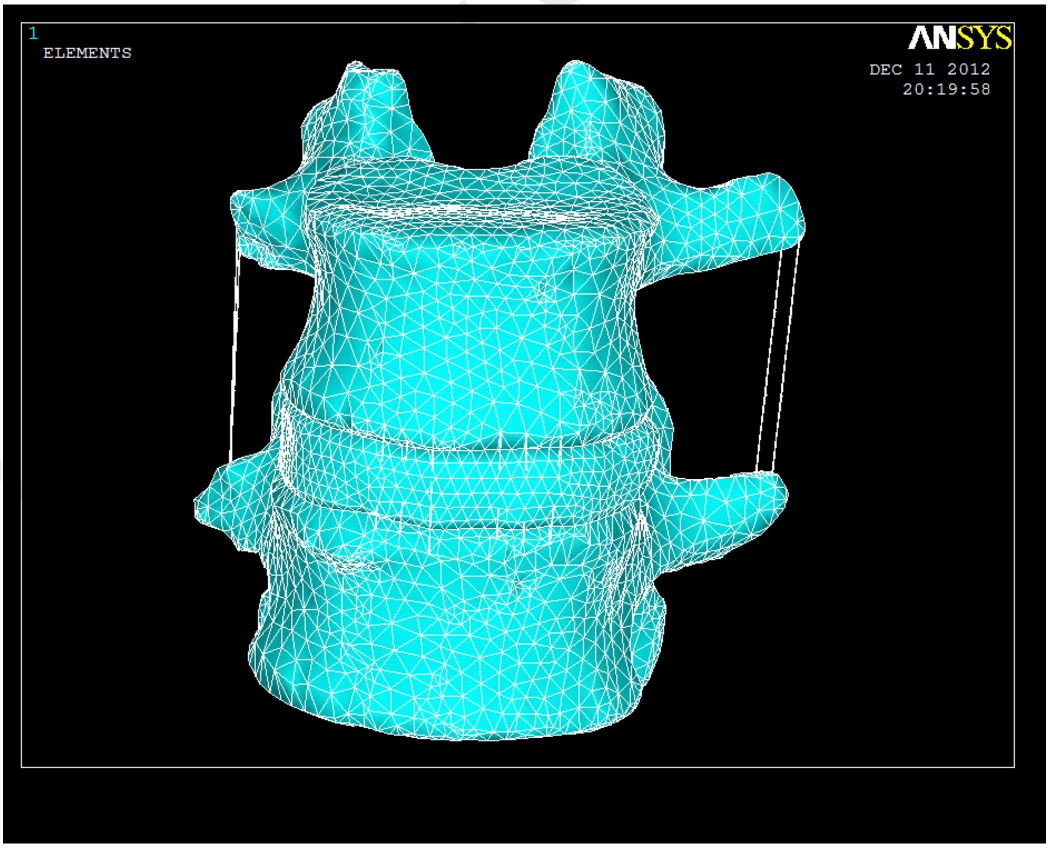

Figure 3-8. Locations of ligaments in frontal plane (ALL)

3.2.2 Load analysis and constraints determination

Load distribution on FSU when standing on high heels was defined as a combination of four loads including, FL which simulates the effect of local muscles (Rohlmannet al., 2006) and acts along the central line passed through the center of the vertebra bodies (Zanderet al., 2001; Rohlmannet al., 2006; Rohlmannet al., 2009a; Rohlmannet al., 2009b; Dreischarfet al., 2010) a vertical preload which simulates the upper body weight and acts at the anterior spine (Augustus A White and Panjabi, 1990; Rohlmannet al., 2006) and consequent bending moment which tends to flex the FSU. Variation of force values for erector spinae muscles were measured as EMG data for three different heights of 0, 4.5 and 8 cm during maximal voluntary contractions (MVC) similar to Lee et al., 2001; Ray and Guha, 1983 techniques and were normalized to yields the values in force unites. Following figure represents the free-body diagram of the functional spine unit including external loads and their acting points and their lever arms. Underneath endplate of the lower vertebra was constrained in all direction (Rohlmannet al. 2006).

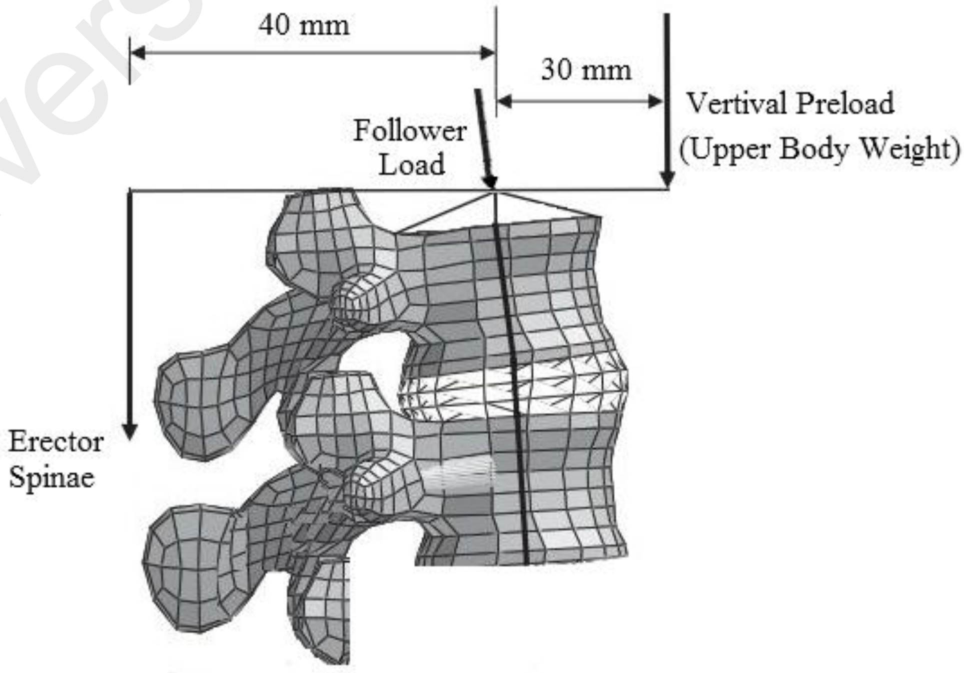

Figure 3-9. FSU free body diagram (force distribution diagram)

3.2.3 Analysis theory (defining the constitutive model)

Invariants of constitute model of incompressible material can be similarly defined using any of two terms equations of polynomial form and/or Mooney - Rivlin (MR) equation. polynominal form is an expanded version of the MR equation. The values of equation invariants are adopted from the literature for Annulus fiberus detailed in previous section and are substituted into the each of the following equations.

Polynomial form

$$W = \sum_{i=1}^{N} c_{ij}(\overline{I_1} - 3)^i (\overline{I_2} - 3)^j + \sum_{k=1}^{N} \frac{1}{d_k}(J_{el} - 1)^{2k}$$

Second term Mooney - Rivlin equation:

$$W = c_{10}(\overline{I_1} - 3) + c_{01}(\overline{I_2} - 3) + \frac{1}{d}(J_{el} - 1)^2$$

Invariants of Neo- Hookean model offered for nucleus pulposus were attained using following simple equations to convert the values suggested in the associated studies based on Yeoh model into what can be realized by Ansys as Neo-Hookean model. The similar relationship can be observed for the mentioned equation as Yeoh is the expanded version of Neo-Hookean.

Yeoh equation;

$$W = \sum_{i=1}^{N} c_{i0}(\overline{I_1} - 3)^i + \sum_{i=1}^{N} \frac{1}{d_i}(J_{el} - 1)^{2i}$$

For $i = 1$

Neo-Hookean equation;

$$W = \frac{\mu}{2}(\bar{I_1} - 3) + \frac{1}{d_i}(J_{el} - 1)^2$$

Therefore,

$$\frac{\mu}{2} = c_{10} \rightarrow \mu = 2c_{10}$$

All invariants values of C_1, C_2, D_1, D_2, and C_{10} was adopted from the previous studies which has been mentioned in the material properties table within the material and meshing section. Invariant μ is simply calculated based on the last equation which is originates from comparing the Yeoh equation with Neo-Hookean equation.

2. Result and discussion

Using a non-linear analysis based on the material properties defined within reviewed literature and consequent constitutive equations which was already discussed in the previous section stress-strain distribution was specifically analyzed over the intervertebral disc. The reason for focusing on the FSU including L4/L5 is the closest region to the lower back pain point complaint by the wearers (C.-M. Lee et al., 2001).Unlike bony structure, heyperelastic trait of the intervertebral disc besides many clinical reports, which associate the lower back pain to the in intervertebral disc damages when it undergoes different unbalanced loading patterns, motivated this study to more concentrate on the intervertebral disc behavior.

Von Mises yield criterion was considered (Zander et al., 2001) to define the stress thresholds around the annulus fiberus matrix of ground substances and nucleus pulpous. Respectively, three force unit EMG values adopted from conducted study by C.-M. Lee et al., 2001 were applied on the model as erector spinae force values. Consequently, corresponding values of maximum and minimum stresses obtained throughout running ANSYS simulation process. Graphical representation of the simulation results is shown in the following figures 4-1 and 4-2. Each of the results shown in the following figures was resulted from one time running a nonlinear finite element analysis for each of the three different force values of erector spinae. Rest of parameters including locations of the forces, constrains and values of FL and upper body weight and the consequent bending moment of it was maintained unchanged. Non-linear analysis parameters were set as, maximum substeps number of 1000 with automatic time stepping to make sure that convergence of the results is achieved beside assigning “large displacement static” required for analysis of hyperelastic the materials.

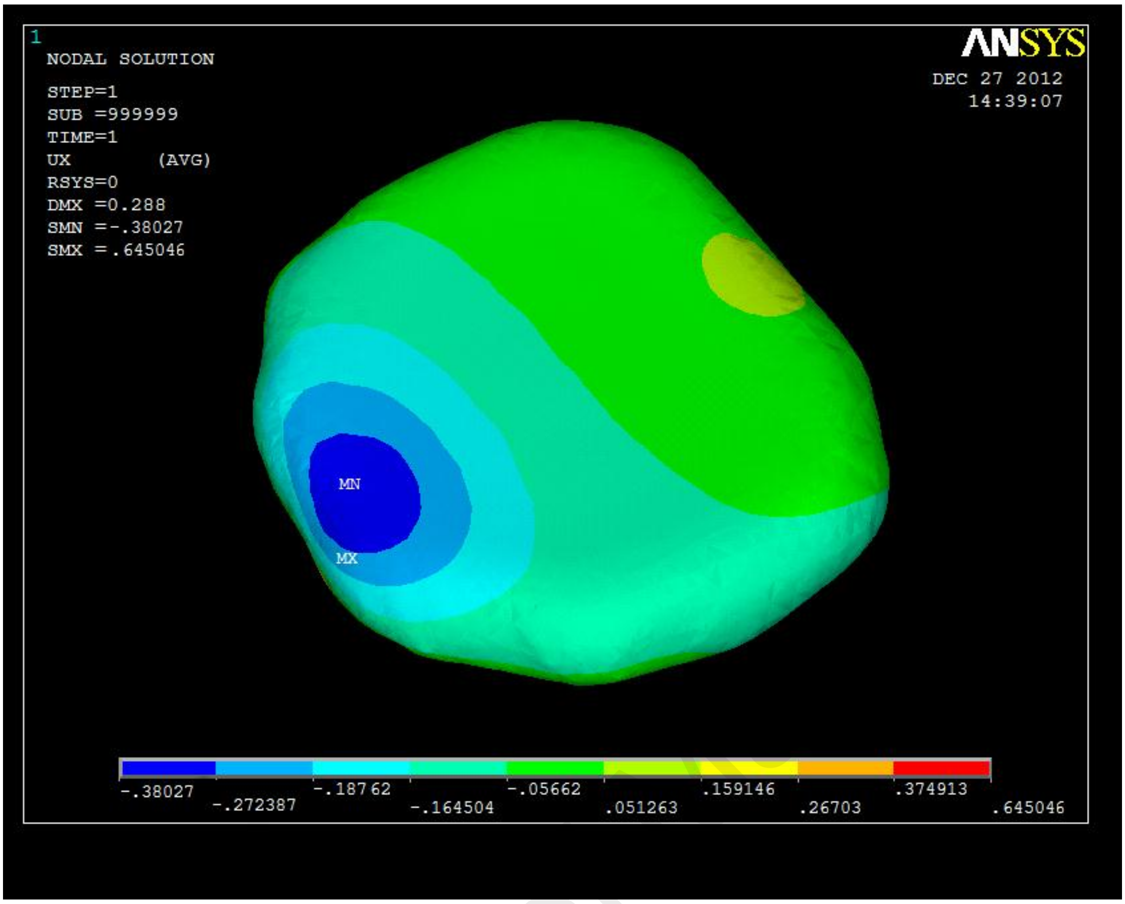

Figure 4-1.Stress distribution on IVD due to wearing 0 cm elevation of heel

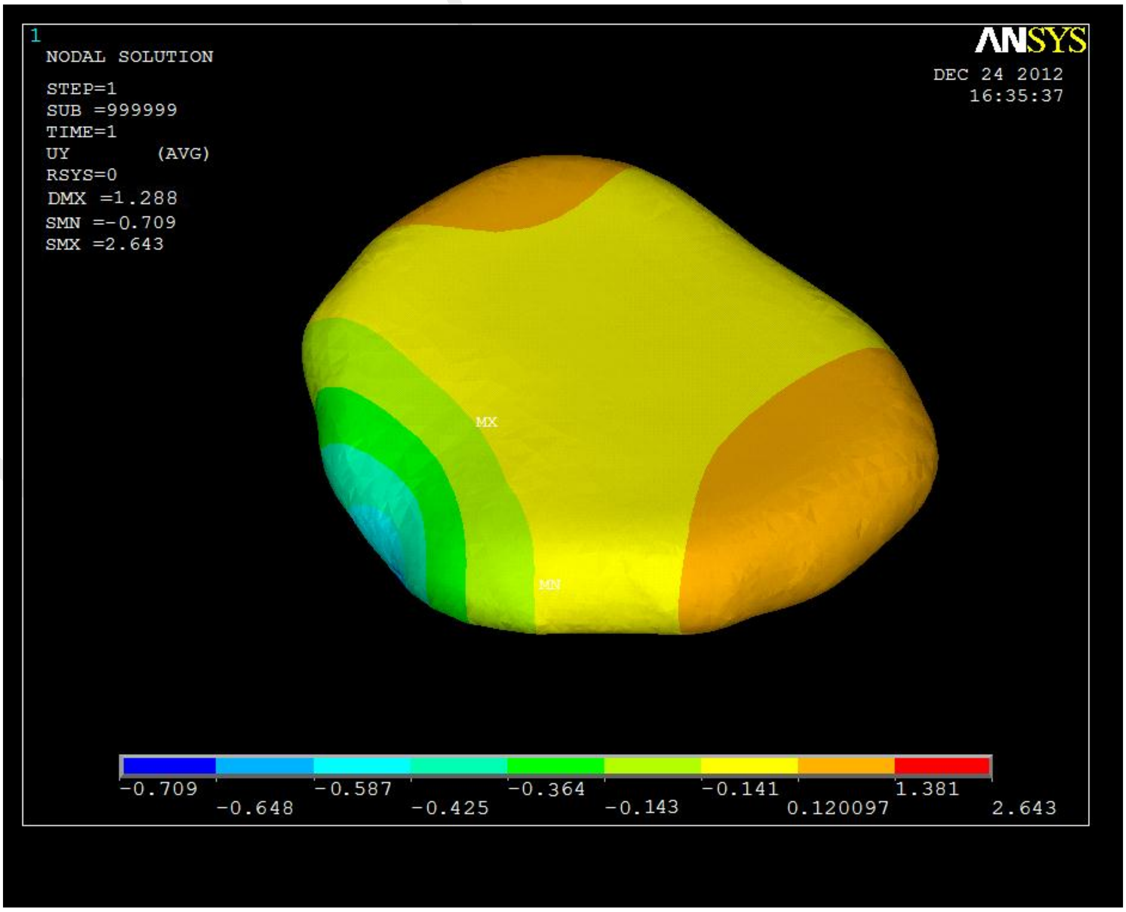

Figure 4-2.Stress distribution on IVD due to wearing 4.5 cm elevation of heel

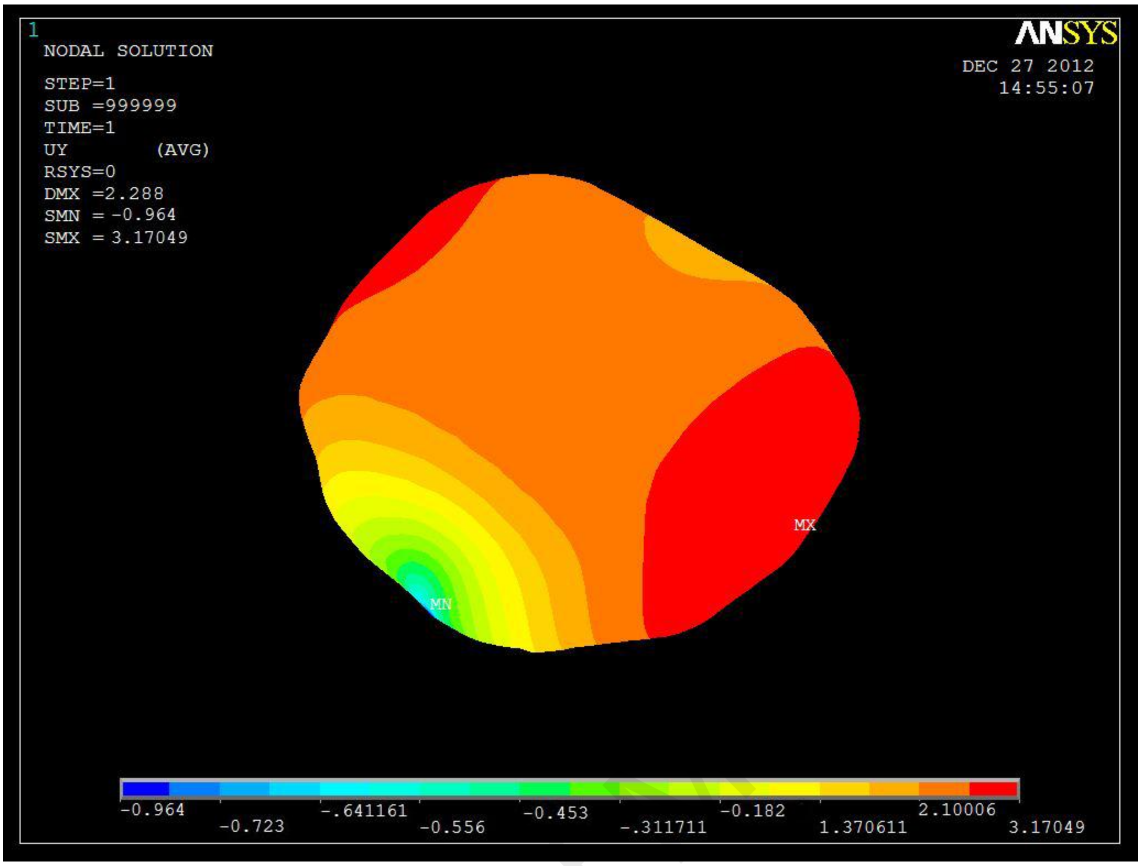

Figure 4-3.Stress distribution on IVD due to wearing 8 cm elevation of heel

The finite element simulation results of the current study are in satisfying agreement with the previous observations which predicates on taking proper assumptions and correct analysis method. As it is observed, increase in the height of the elevated shoes results in stress increment over the posterior and medial-lateral regions of the annulus fiberus ring. The reason for such stress distribution pattern can be referred to implementing an improper surface modification and also a weak smoothening process prior to the finite element analysis.

Indeed presence of any discontinuity on the surface due to improper surface preparation of the model results in stress concentration and consequently appearance of a sharp tension borders at the area where surfaces with different are joint. However results are theoretically representative for the stress distribution over the intervertebral disc as other parameters including endplates and constrain location can directly effect on pattern of stress distribution. Table 4-1 presents the maximum and minimum values of stress arrays corresponding to the three different heights of heel. The positive stress value indicates that the associated regions are subjected to the compression force and adversely, negative values indicate region which undergoes to the extension force. Compare to the previous studies good agreement is observed in overall for the stress values.

Table 4-1.Values of imposed stressed and displacement Correspondent to heels heights

Heels heights	Maximum stress	Minimum stress	Prevailing stress	Region of Max. stress	Region of Min. stress	(Y) Max displacement
Low (0 cm)	0.645 MP	-0.38 MP	-0.052 MP	Medial	Posterior	0.28 mm
Medium (4.5 cm)	2.643 MP	-0.70 MP	-0.141 MP	Medial Anterolateral	Posterior	1.28 mm
High (8 cm)	3.170 MP	-0.96 MP	2.1006 MP	Anterolateral	Posterior	2.288 mm

Stress concentration regions are determined based on the element locations provided by the ANSYS. Locations of stress concentrations can be conclusively justified regarding to the analysis primary data of erector spinae force values supplied by Lee et al., 2001. In consistence with lumbar flattening observed by Lee et al., 2001, locations of stress concentration obtained by this study can justify the spine lumbar flattening. However, recent study conducted by Russell, 2010 does not totally support the results of Lee et al., 2001. The issue is that the lumbar S curve would not be accurately represented based on lateral positioning of the land marks on the body trunk.

However, the hypothesis was further affirmed by another study conducted by Ebrahimian and Ghaffarinejad (2004).Location of the stress concentration which confirms occurrence of lumbar flattening can facilitate the process of pain region diagnosis of the disc injuries including disc bulging. As it is observed, intensity of the stress is increased from medial region toward anterior side of the intervertebral disc from stress value 0.645MP to 3.17 MP. Simultaneously posterior side undergoes to the extension stress from -0.38 MP to -0.96 MP to satisfy the equilibrium condition. Based on the simulation results, it is observed that the prevailing values of stresses are usually appeared on medial region of the intervertebral disc which is the nucleus pulposus area. The prevailing values appear on medial regions for both heights of 0 and 4.5 cm heights while high elevated heels impose the second high stress over the wide medial region. The prevailing values appeared mostly at the posterior ages of the intervertebral disc due to walking high elevated shoes.

Stress distribution bands shown at the bottom of each stress graphical representation provide more details about the regions and associated stress values of each particular area. As such, intensity of the stresses is represented by the color bands from blue color related to the minimum values of stresses up to the red which indicates the maximum regions of the stresses. Following chart represents the stress variation trends of all three heel categories for maximum, minimum and prevailing values of stresses.

Following bar chart facilitates the comparison of all biomechanical concerns discussed within the content. Interestingly the medial area related to the prevailing stress values undergoes extension for both low and medium elevated heels while this region is subjected to the compression due to wearing high elevated heels. There is an orderly progressive growth of 1 mm in displacement values along the vertical axis of Y corresponding to each class of the heights.

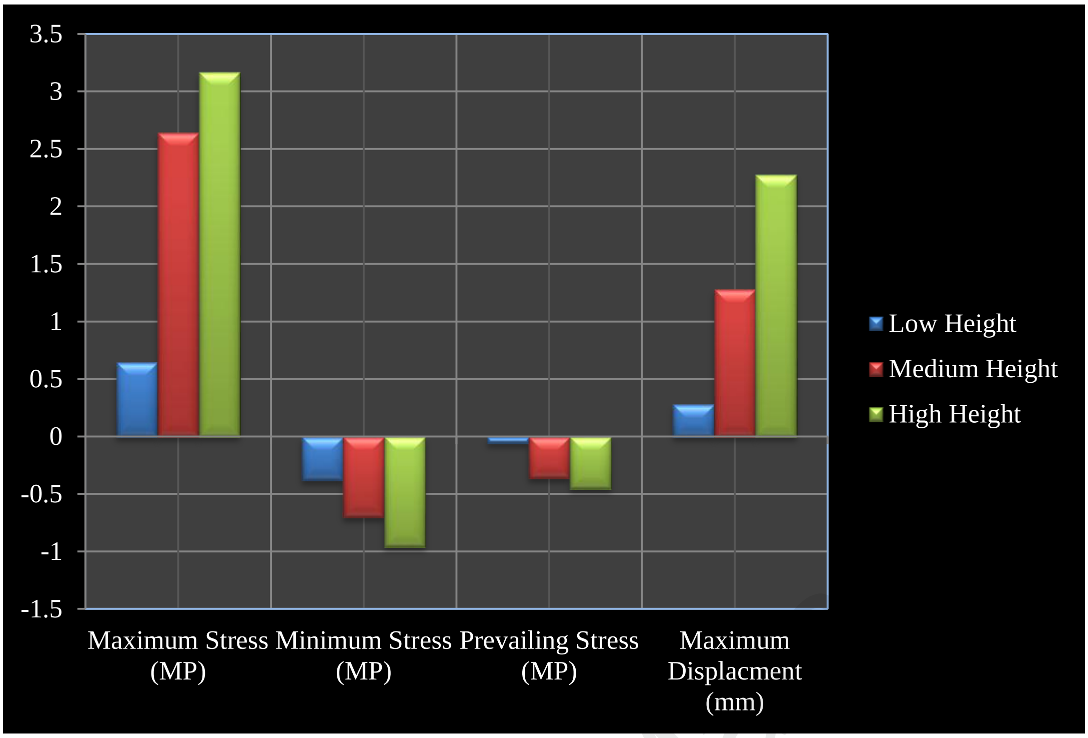

Figure 4-4. Chart representation of stress and displacement distribution

Stress growing pattern and displacement results observed in the chart clearly declare the issue of thinning disc injury. Therefore, it seems necessary to investigate the tissue stiffening problem in the regions with the higher concentration of the stress. There exist progressive trends of maximum stress and displacement values with height increment of the heels. The higher height of the heels the higher stress and displacement imposes to the intervertebral discs.

Vertical displacement values for both low and medium sizes of the high heels are approximately half of the correspondent values of stresses. It can be concluded that for any height of heels from flat to medium size of 4.5 cm displacements have half of the values of the imposed stresses. Compression stresses can be quickly distinguished based on the upward positive direction of the stress bars displayed in the chart while negative downward direction of the minimum stresses bar convey the distribution of extension stresses over the intervertebral disc.

Obviously there should be equal values of both compression and extension stresses due to wearing flat heels which has been truly obtained via simulation and represented in the displayed graphical bar chart. High elevated shoes exceptionally impose higher mechanical concerns to the intervertebral disc including stresses and vertical displacement based on the results of the current study compare to the flat and medium classes. Therefore, prolonged standing on the high elevated shoes should be avoided to prevent from disc degeneration which is caused by hydrostatic pressure (Handa et al. 1997).

Degenerative disc disease results in direct nerve compression as the spinal disc flattens. Disc height reduction (disc flattening) cusses the nerve canal between the bones to narrow leading to impingement. This can progress to create a sever debilitating symptoms such as; sharp pain, burning, numbness, tingling, and muscle weakness over time. This degenerative process can compromise the integrity of the nerve and ultimately distrusting the vital flow the information between brain and truck.

3. Conclusion

Theoretically, stress increment up to the yield stress threshold results in failure of the objects. Based on the stress values obtained by the simulations as a result of combining different loading patterns used by other researchers, prolonged standing on the elevated heels of 8 cm height may results in anterolateral stress concentration for almost 3MPa. This is about third of the imposed stress due to wearing heels of 0 cm height. Therefore, stress concentration on anterolateral regions of the intervertebral discs may consequences to the disc stiffening. Clinical concerns with disc stiffening can be addressed to further disc injuries including disc bulging and consequently disc herniation. However, Static analysis does not offer the realistic simulation for daily life compare to the dynamic analysis and in this regard fatigue analysis should yields better conclusion on disc stiffening problem.

One can suggest for conducting a dynamic analysis based on gait analysis information in order to performing more realistic simulation. However, the concern of dynamic loading can be summarized in fatigue failure while dynamic loading is a necessary factor for intervertebral disc healthiness as disc is an avascular tissue which needs fluid secretion and absorption in order to doing oxygenation.

Displacement variation trends of the intervertebral disc confirm forward bending of the FSU and consequently lumbar flattening which means reduction in the flexion angle of lumbar that already reported by Lee et al., 2001. Respectively, for those who suffer from slight posterior or posterolateral disc thinning problem it might be theoretically advisable to start wearing mid high heels in order to shifting the pressure from posterior regions to the anterior side. This can provides intervertebral disc more posterior free space to regain its fluid lost and recover its normal shape.

Defining a risk threshold of spine diseases according to the stress generated as a result of using certain range of high heels. More accurate primary estimation of the back pain region can be done for those who frequently wear high heels according to the stress concentration on different regions of the discs. Therefore, better disc injury diagnosis can be implemented based on the intensity and region of the imposed stresses.

Imposition of higher mechanical concerns to the intervertebral disc due to wearing high heels would provide middle age women benefit of subjecting the intervertebral disc to more activities which is a vital criterion for health of the intervertebral disc. As a matter of fact, tending to the sedentary lifestyle and having more rest are normal life features of the middle age persons that caused them to neglect about the health of the intervertebral disc. In this regard, more study including dynamic analysis coupled with the gait analysis information of high heels gait cycle can provide more justifiable advices.

Predictable bending in forward flexion of the lumbar based on the stress concentration regions determined by the simulation process causes the convex side of the annulus undergoes tensile stress while the concave side is subjected to the compressive stress. Consequently convex side is stretched and concave side is bulged (Jensen 1980). Bending is a mechanical concern which is most probable reason for causing spine trauma as it involves different types of stresses including shear, tensile and compression simultaneously. Mechanical characteristic of the bending is appearance of the stresses on the surface of the subject. Therefore, it can be concluded that the peripheral surface of the convex side of the intervertebral disc undergoes highest tensile stress values while peripheral surface of the concave side of the intervertebral disc is subjected to the highest compressive values of stresses.

Prolonged standing on high elevated shoes subjects the intervertebral disc under continues hydrostatic pressure that can causes the cartilaginous end plates to collapse. This hydrostatic pressure which compresses the intervertebral disc generates the radial stresses toward end plates as annulus fiberus is stretched. Nucleus fluid pressures pass the compressive stresses, which has been distributed on the annulus interiorlamina layers, to the upstream and downstream vertebras. While exterior lamina layers of annulus fiberus absorb the imposed tensile stresses based on the fiber thicknesses so that the intensity of the stress on thin-posterior fibers of the annulus fiberus is almost 5 times of the applied load.

5.1 Future work

This study only focused on the static simulation of normal standing on the high heels using only three different high heel heights and therefore cannot be generalized for the high heels with different heights. Additionally, more concern should be taken about dynamic analysis as it considers more important mechanical concerns like heel strike and consequently impact intensity which probably effects on fatigue property and relatively yield stress threshold.

Therefore, gait analysis information would be required in order to define the dynamic loading pattern. As a matter of fact, dynamic loading has complex patter which compromises different loading types including torsion that is essential for providing more realistic designs. Moreover, this study just considered the compression stress imposed by bending of the spine forward flexion and other stress arrays including shear generated by torsion phenomenon was not taken into account within the results. Torsion is resulted from movement of one vertebra over another one simultaneous to bending. Other out puts like annulus fiberus fatigue analysis and viscoelastic creep, which is a time dependent strain increment resulted by step constant stresses, can be carried out in further studies.

Respectively other possible results of simulation including interadiscal disc pressure, force distributed on the facet joints and contact force between the facet joints can be considered. Intersegmental rotation can provide more details about shear stresses that require preparing a full model of thoracolumbar region from L1 to L5 rather than only modeling a FSU.

5.2 Ethical principles of study

As the primarily data required for the modeling is supplied by the medical imaging techniques subject must be fully informed about the health risks associated with exposure beams. Scientific findings and reports regarding to concerns about using MRI techniques can be referenced and presented to the subject to make her fully understand about probable consequences of her participation. As a matter of fact, magnetic resonance imaging (MRI) technique or ultrasound Scanning has nothing to do with radiation exposure therefore is clinically preferred to costume compute tomography (CT) scan technique when one deals with the soft tissue like intervertebral discs. However, this study just used pre-collected data of the patients who clinically diagnosed to have no injuries on the thoracolumbar region in order to create a 3D-solid model. Therefore no more concern was taken into account regarding to radiation exposure.

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