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Patient Specific Finite Element Modeling of Human Vertebra

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ABSTRACT

Many people suffer from low-back pain, a major reason of that pain is degenerative disc disease (DDD), as an alternative to intervertebral disc, an artificial disc serves to replace the damaged degenerated disc. To reduce later complications after implantation surgery it's important to use a matching vertebrae endplate geometry artificial implant. Vertebra bone material properties and geometry depend mainly on age, sex and effect of some bone diseases such as Osteoporosis which decrease bone density and even change its geometry. Finite element simulation is an effective way to analyze vertebra bone models in order to provide a custom made implant design to the patient as each patient's vertebra has unique endplates geometry and different bone material properties. This study targets to developing a methodology to create patient-specific finite element model with volume mesh generation, and material assignment using medical digital imaging in order to design a matching endplate patient-specific lumbar disc artificial implant and test its validation. A patient-specific finite element model of the lumbar spine vertebra L5 has been developed, CT data are used to create geometrical model and assign its materials. To test validation of the model, Finite element analysis of the created model is performed and stress, strain distribution and total deformation has been established.

Keywords: CT-scan, patient specific, lumbar spine, material assignment, finite element

INTRODUCTION

The lumbar vertebra model was considered one of the most simulated bones due to the effect of low back pain in quality of life and work performance. Lumbar spine, especially its L4 and L5 vertebrae, is subjected to greater loads than the rest of the spine as they carry almost 70% of body weight. Hence it continues to receive more attention clinically and experimentally [1]. Degenerative disc disease (DDD) cause the intermediate disc between vertebrae becomes thin and narrowing the space between the vertebrae. Also bulge or break off Pieces of the damaged disc could cause excitation to the nerves causing back pain [2]. Total disc replacement is rapidly becoming an option in treating patients with symptomatic DDD and has been suggested as an alternative to fusion [2-3]. Commercial lumbar disc prostheses are available in various sizes of endplates but consist of endplates that are designed relatively flat in comparison to the vertebrae endplate geometry which could cause later complications. Several studies have attended the importance of conformity of implant surface and endplate geometry [4] to reduce usual complications that are still observed in some patients after implantation such as anterior migration of the disc and partial dislocation of a joint [3]. One of the most prevalent reasons for such disc failures is incorrect positioning of the implant [4] as during implantation surgery, surgeons try to select the most suitable size of available commercial prostheses to match the vertebra endplate but every patient have unique size which could lead to implant subsidence [3-4], and they flatten the vertebrae's endplates and make a perpendicular cut in the vertebra or even use fasteners to fix the implant that could affect bone strength and lead to vertebral fracture [4].

Human bone has a non-uniform geometry and a dissimilar structure, there's difference from bone to bone and from person to person. In order to consider these essential differences [5], it was important to develop a modeling technique describing the geometric and material properties in order to design a matched geometry vertebrae's endplate and consider vertebrae's material properties which could be effected by age, sex or some bone diseases so that decrease Probability of implant immersion and other complications. The vertebral body consists of an outer shell of high strength cortical bone its thickness and density depend on sex and age [6], strengthened internally by the cancellous bone as a network of vertical and horizontal narrow bone supports [7]. In most studies the vertebra bone mechanical properties considered as a homogenous isotropic composite materials made of two different materials, spongious Cancellous bone (high porosity) and cortical bone (compact bone) [6], although bones are typically inhomogeneous anisotropic materials [8-9].Considering the model bone as isotropic has a major effect on the accuracy of bone behavior established by finite element models, which make the error in obtained numerical results from finite element solution could be as large as 50% [9].

Medical computed tomography (CT-imaging) became a useful tool for applying finite element modeling for human bone and determining its material properties by modifying the existing empirical relations of bone elasticity-density [9]. CT images are a pixel map of the linear X-ray attenuation coefficient of tissue. The pixel values are scaled (air= -1024, water= 0), this scale is called the Hounsfield scale (HU) after Godfrey Hounsfield. Using this scale, HU of fat is -110, HU of muscle is around 40, HU of trabecular bone is in the range of 100 to 300 and HU of cortical bone about 2000 [9,10]. HU obtained from clinical computed tomography scans that are made for diagnosis of bone diseases provide an alternative method for determining bone density without any additional cost to the patient [11]. Conversion of the HU to density of the bone could be obtained by Quantitative Computed Tomography (QCT) which considered a valid technique used in many studies, where solid phantom of known density used to calibrate the CT images during patient scanning [12]. Ultrasonic transmission technique is another method used to determine mechanical properties of bone by obtaining CT value from scanning bone in water [13]. Every type of CT scanner has obtained HU to density relation and density to Young's Modulus relation depending on scanner energy, pixel resolution, image resolution, CT slice thickness, and scanning technique [10].

The objective of this study was to propose a computational procedure could be used for each patient which helps in designing a suitable disc Implant considering patient's vertebra bone density with matched endplate geometry; a case study used to describe the procedure the following steps would be performed.

METHODOLOGY

CT scan on a healthy 74 years old male subject was taken. This scan was taken with x-ray energy of 120 kVp, mA settings of 320 mA, pixel resolution of 1 mm/pixel, image resolution of 512x512 pixels array and slice thickness of 1 mm. 1563 CT images of the legs and low back are exported in common medical file format DICOM (Digital Imaging and Communication in Medicine).

Creating 3D Model of L5 Vertebra

Processing CT images needed to evolve necessary geometric data in order to generate a geometric model of the vertebra. The program MIMICS (Materialise Software, MIMICS innovation suite (research and medical editions) version 16) used to process the CT images and derive the geometry for the model. First the project is cropped to separate the lumbar vertebra L5 images (fig. (1) shows axial(right), sagittal (bottom left), and coronal (top left) views of one CT scan slice of L5 vertebra) from the imported CT scan images, then 2D segmentation is performed by applying a threshold method in order to separate the vertebra bone from the surround mussels, tissues and nerves.

Fig. (2) Shows the variation of HU values across patient vertebra and surround tissues where the highest HU value represent cortical layer of vertebra (its HU is about 526) and its thickness is about 1.5 mm and cancellous shell HU ranges from 126 to 226, it's important to select suitable threshold value that cover vertebra cortical and cancellous (trabecular) shells, the lower threshold is 126 and upper threshold 1000 limits are used to create a mask which filters and highlights all areas on each of the slices of the CT scan that fall within these upper and lower boundaries and separate the intended bone in every slice.

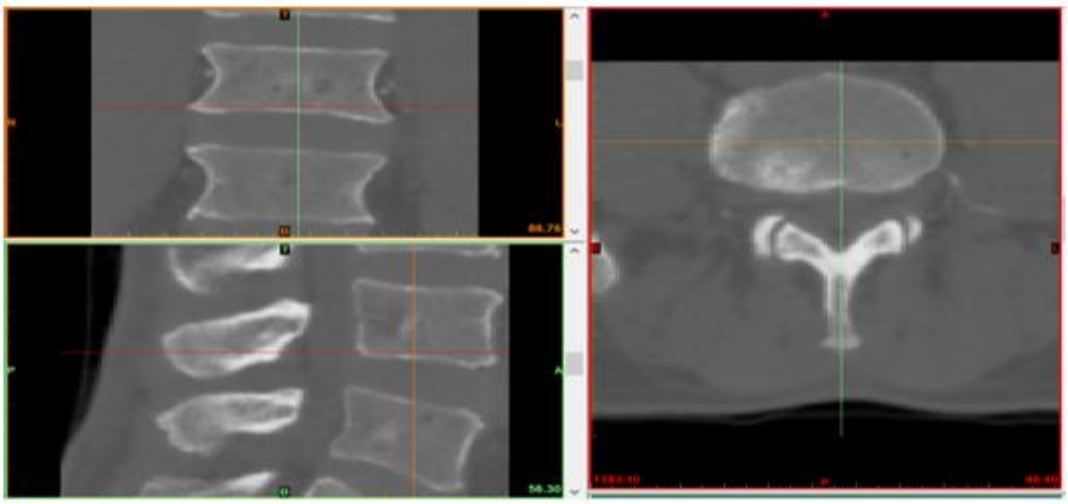

Fig. 1 Cropped vertebra project in MIMICS

Before calculating 3D model editing the mask created in the last step in order to remove the extra pixels that could have the same threshold limits and threshold missing pixels. The final mask after editing is shown in fig. (3). The next step 3D model will be calculated from the driven 2D mask by sticking together the thresholder slices and sufficient smoothing iterations which is an available function in MIMICS, will be processed to remove any noise in the model and improve its quality. Fig. (4) Shows the calculated 3D L5 vertebra model before and after smoothing.

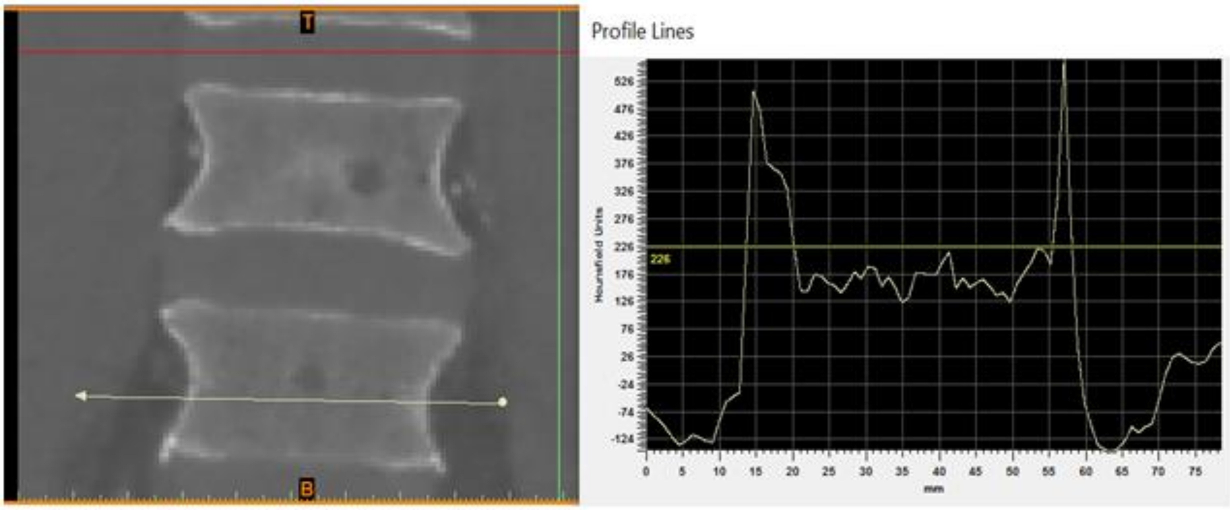

Fig. 2 HU values over a profile line across vertebra

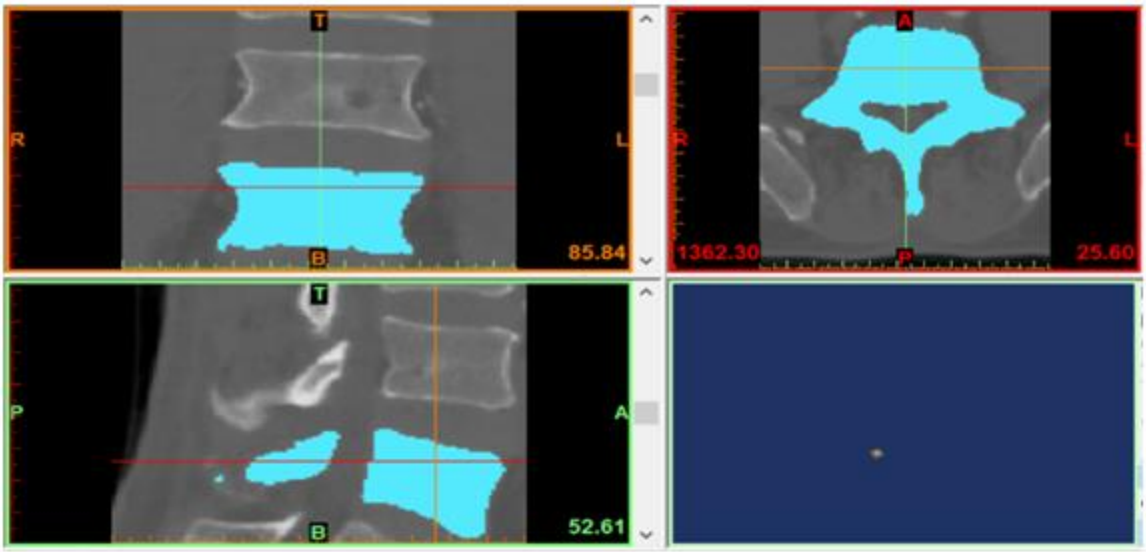

Fig. 3 2D model of vertebra in three planes

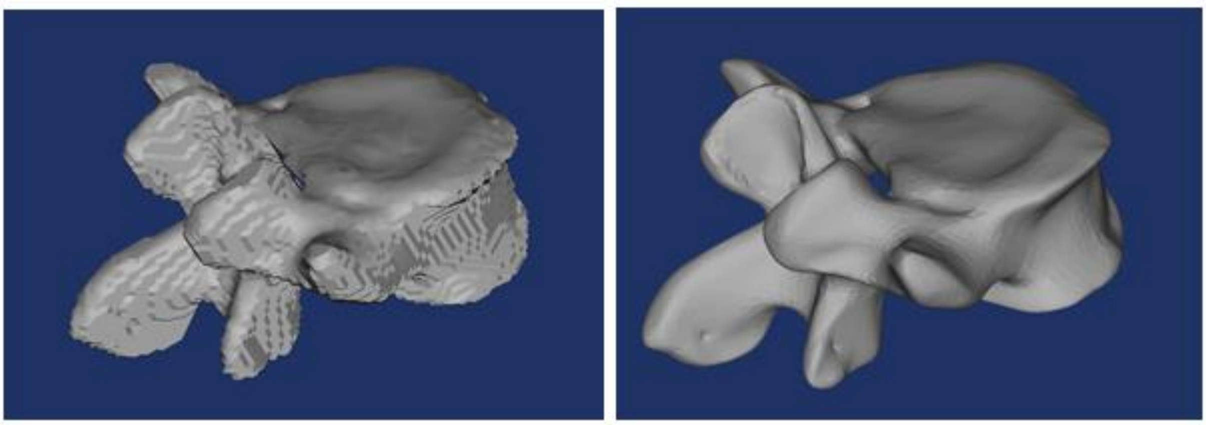

Fig. 4 3D model of vertebra before and after smoothing

Meshing

3-matic (Materialise Software 3-matic Research version 9) is used to create surface mesh to the vertebra then generating volume mesh with a tetrahedron element type with minimum edge size 2.05 µm. Nodes number of the model is 50845 and element number of the model is 262745. Fig. (5) shows the L5 finite element model after meshing.

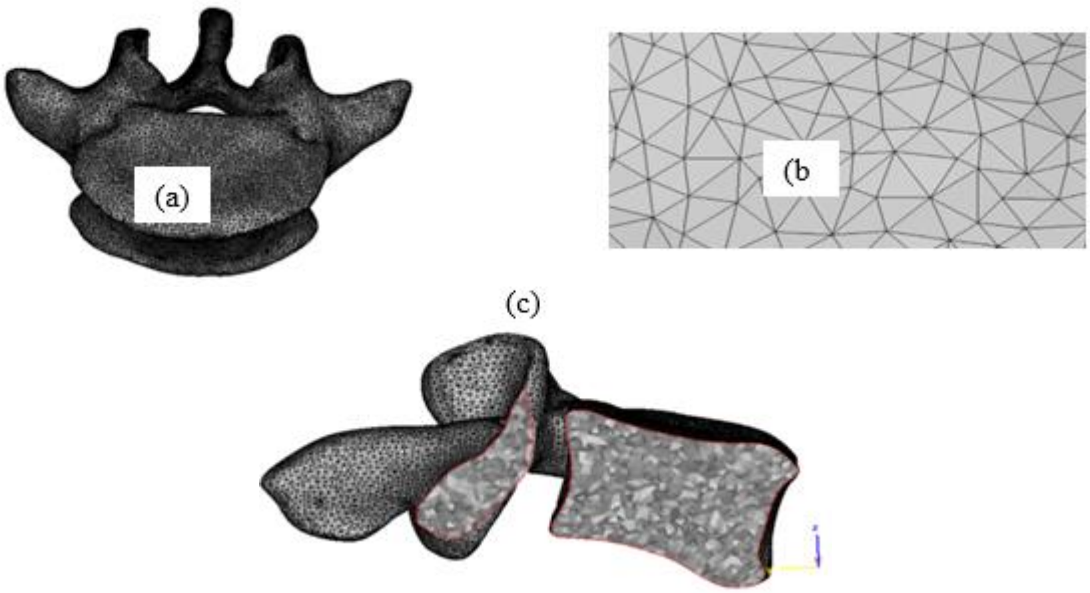

Fig. 5 (a) Meshed model of L5 vertebra, (b) Surface mesh and (c) Volume mesh

Material Assignment

Finite element model as a volume mesh is imported back to MIMICS (Materialise software) in order to assign its material using HU of the CT scan of L5 patient vertebra using expressions between lumbar spine region density ρ and HU of the CT scan, and the relation between density ρ and modulus of elasticity E (Young's modulus) for number of common used scanner types and there technique of measuring Young's modulus [10,14].

$$\rho = 1.122\, HU + 47 \quad (1)$$

$$E = 0.63 * \rho\hat{}1.35 \quad (2)$$

Equations (1), (2) are used with the model of this study as they are suitable for the scanner type specifications mentioned before and it's scanning technique (Ultrasound velocity measurement at 50 kHz) according to the study of Rho, Hobatho, Ashman, 1995 [13].

Fig. (6) shows the model after material assignment and distribution of 30 materials on the model depending on their HU in the left and the value of density and Young's Modulus for each material in the right.

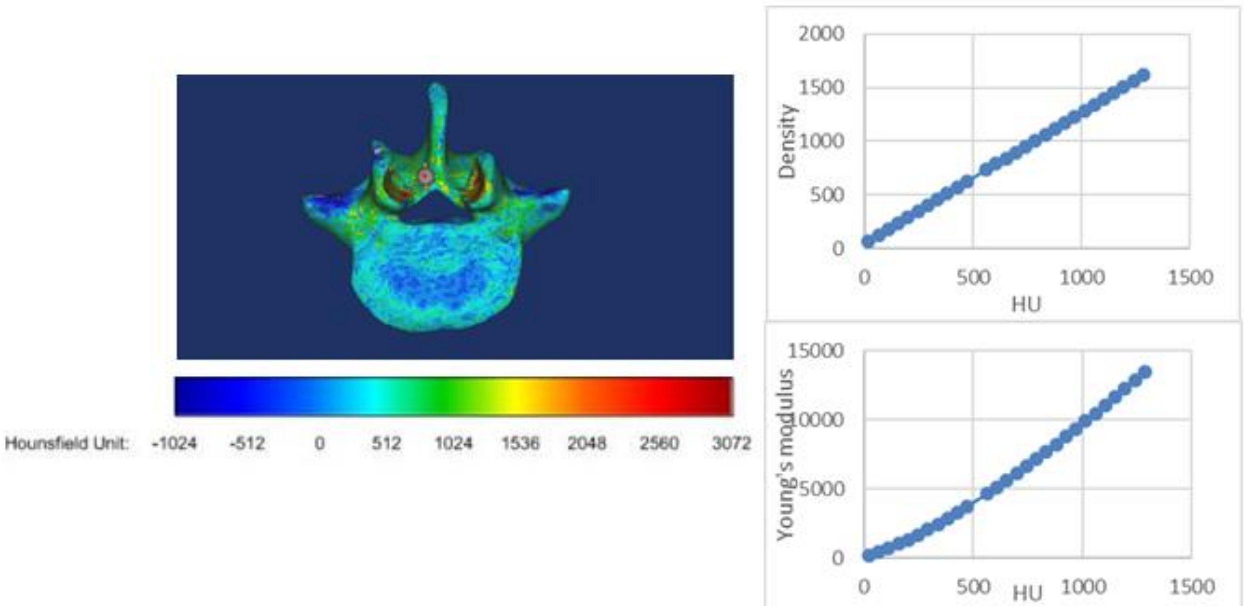

Fig. 6 Materials distribution over vertebra depending on their HU and histogram of the materials

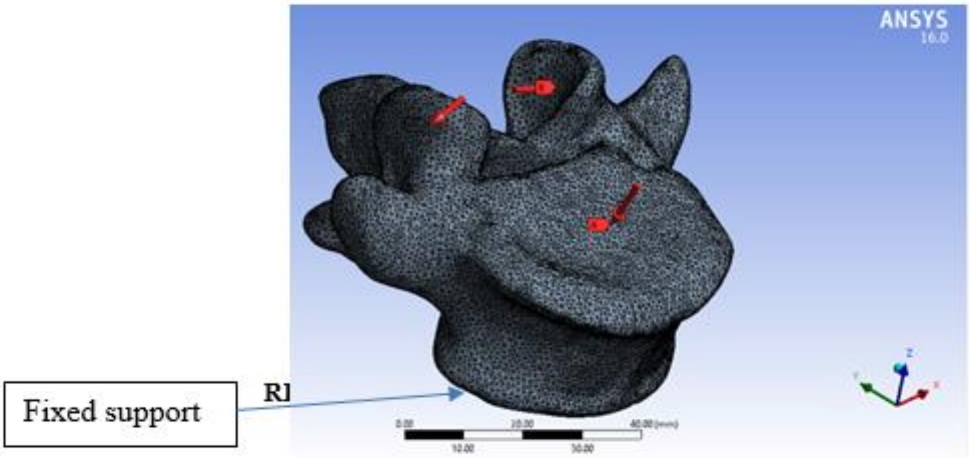

Fig. 7 Loading and boundary conditions of vertebra model

Loading and Boundary Conditions

Finite element analysis is performed to test validation of created model. The model is imported to ANSYS 16.0 Workbench in finite element modeller as (.cdb file) which exported from MIMICS and transfer the data to static structural engineering data to read assigned materials properties, and to model to read created geometry and mesh.

A fixed support is inserted in bottom end-plate of vertebra and pressure applied to upper end-plate of vertebra and on facet joints. The value of load depending on patient weigh tit's about 500N load when a healthy person weighted 70 kg stands straight in a relaxation state [15]. However, the number goes up to 2000N when he lifted 100N with two arms stretching straight [7].

The load sharing of articular facets ranges from 0 to 30% [16] ,0.37 MPa Pressure applied in upper end plate and 0.25 MPa in facet joint [17]. Fig. (7) shows loading and fixation of model. Before start solving project a command file (text file which exported from MIMICS) is inserted to project to define material of each element of the model.

RESULTS AND DISCUSSION

Fig. (8) illustrates total deformation distribution over model. Maximum deformation observed was 0.0026 mm at articular facets, and minimum deformation is in vertebral end-plate around fixed support. Fig. (9) illustrates equivalent (Von-Mises) stress distribution over model. Maximum stress was observed in end-plate of vertebra (fixed support) with value of 2.234 MPa, and minimum stresses is acting on spinous process of vertebra. Fig. (10) shows strain distribution over vertebra with maximum strain value of 0.00023 mm/mm.

Results are compared to studies [17] which used almost the same loading conditions of the current study and [12] which used different density and Young's modulus relations and its model was on a healthy middle-aged female subject, the finite element analysis outcomes are in the range of estimated values which indicates that the model mesh and its material assignment is accurate, table (1) summarize results of current study, study [17] and study [12].

Fig. (11) displays the effect of vertebra bone density reduction in deformation of the vertebra under the same loading condition which verify the importance of considering the patient's bone density in designing artificial disc implants as well as the vertebra geometry to reduce after implantation surgery complications.

Table - 1 Maximum equivalent stress, maximum equivalent strain and maximum deformation of current study and previous studies

Study	**Von-Mises stress (MPa)**	**Elastic strain (mm/mm)**	**Deformation (mm)**
Current study	2.234	0.00023	0.0026
Jovanović & Jovanović (2010) [12]	4.04	0.0052	0.092
Ho Quang,et all(2016) [17]	2.75	0.0002	0.0045

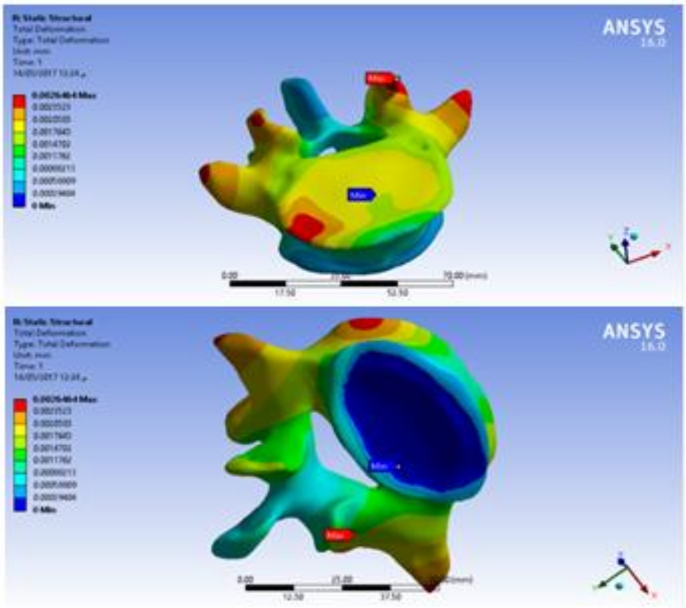

Fig. 8 Total deformation distribution over FE model of vertebra

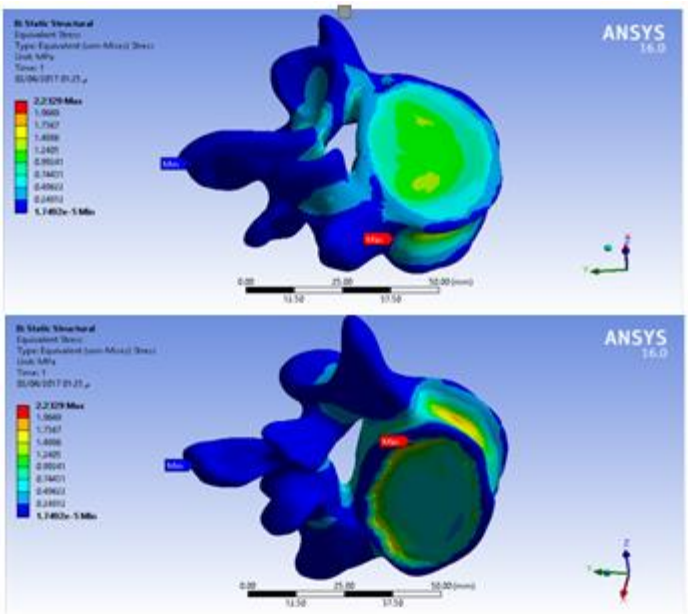

Fig. 9 Equivalent (Von-Mises) stress distribution over FE model of vertebra

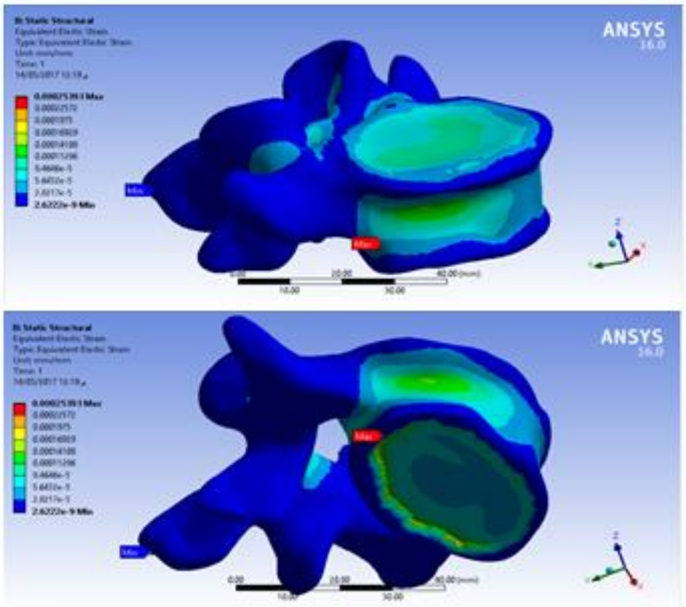

Fig. 10 Equivalent elastic strain distribution over FE model of vertebra

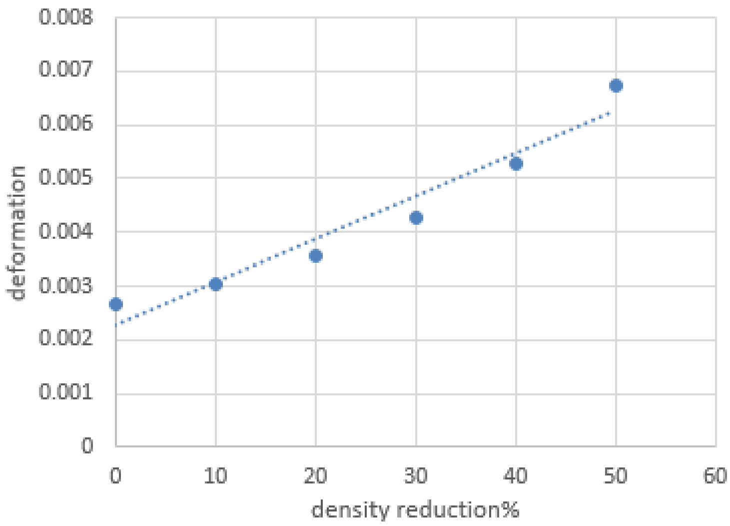

Fig. 11 effect of bone density reduction on vertebral deformation under the same loading condition

CONCLUSION

This study proposed a complete modeling procedure which includes geometric modelling, volume meshing and CT-based material assignment in order to create a patient specific finite element model for human lumbar vertebra L5 which helps on designing a patient specific artificial lumbar disc. Finite element analysis using a linear analysis (pure compression pressure) of the created model had been performed to test validation of the model. Compered resulted stress distribution, strain distribution and deformation of the model to previse studies revealed that the model was accurate. The study also illustrated effect of bone density reduction due to some diseases as osteoporosis in vertebral deformation under same loading condition which confirm the importance of considering bone material properties as well as vertebra geometry in designing artificial disc implants to reduce later complications.

REFERENCES

[1] CS Aparna, RP Tewari and AK Govil, Biomechanical Analysis of a Three Dimensional Finite Element Model, *International Journal of Mechanical Engineering and Robotics*, **2013**, 1(2), 149-152.

[2] Caroline Sköld, Hans Tropp, Svante Berg, Five-Year Follow-Up of Total Disc Replacement Compared to Fusion: A Randomized Controlled Trial, *European Spine Journal*, **2013,** 22(10), 2288-2295.
[3] Andre´ van Ooij, F Cumhur Oner and Ab J Verbout, Complications of Artificial Disc Replacement: A Report of 27 Patients with the SB Charite´ Disc, *Journal of Spinal Disorders & Techniques*, **2003**, 16 (4), 369–383.
[4] Neal de Beer, Andre´ van der Merwe, Patient-Specific Intervertebral Disc Implants Using Rapid Manufacturing Technology, *Rapid Prototyping Journal*, **2013**, 19 (2), 126-139.
[5] MC Ho Ba Tho, Patient Specific Bone and Joints Modeling and Tissue Characterization Derived from Medical Images, *WIT Transactions on State of the Art in Science and Engineering*, **2005**, 19, 63–79.
[6] I Leichter, A Weinreb, GC Robin and R Steinberg, The Effect of Age and Sex on Bone Density, Bone Mineral Content and Cortical Index, *Clinical Orthopaedics and Related Research*, **1981**, 156, 232–239.
[7] M Kurutz, Finite Element Modelling of Human Lumbar Spine, *Finite Element Analysis [David Moratal (Ed.)]*, **2010**, 1987, 210–236.
[8] C Hellmich, C Kober and B Erdmann, Micromechanics-Based Conversion of CT Data into Anisotropic Elasticity Tensors, Applied to FE Simulations of a Mandible, *Annals of Biomedical Engineering*, **2008**, 36 (1), 108–122.
[9] S Kazembakhshi and Y Luo, Constructing Anisotropic Finite Element Model of Bone from Computed Tomography (CT), *Bio-Medical Materials and Engineering*, **2014**, 24 (6), 2619–2626.
[10] Materialise Softwere Help.
[11] JJ Schreiber, PA Anderson, HG Rosas, AL Buchholz and AG Au, Hounsfield Units for Assessing Bone Mineral Density and Strength: A Tool for Osteoporosis Management, *J. Bone Joint Surg. Am. [The Journal Of Bone And Joint Surgery]*, **2011**, 93(11), 1057–1063.
[12] JD Jovanovic and LJ Miomir, Finite Element Modeling of the Vertebra with Geometry and Material Properties Retreived from CT-Scan Data, *Facta Universitatis, Mechanical Engineering*, **2010**, 8(1), 19–26.
[13] Rho JY, Hobatho MC, Ashman RB, Relations of Mechanical Properties to Density and CT Numbers In Human Bone, *Medical Engineering & Physics*, **1995**, 17(5), 347-355.
[14] JY Rho, MC Hobatho and RB Ashman, Relations of Mechanical Properties to Density and CT Numbers in Human Bone, *Medical Engineering & Physics*, **1995**, 17(5), 347–355.
[15] KS Han, A Rohlmann, T Zander and WR Taylor, Lumbar Spinal Loads Vary with Body Height and Weight, *Medical Engineering & Physics*, **2013**, 35(7), 969–977.
[16] C Lai, J Wang, G Chang and C Chung, The Load Sharing Contribution of Spinal Facet Joint During Impact Loading – A Porcine Biomechanical Model Cheng-Chuan, *IMECE*2003-42940, **2003**, 1, 3–4.
[17] HQ Nguyen, TT Dao, A Rassineux and MC Ho Ba Tho, Material-Driven Mesh of the Lumbar Spine Derived from CT Data, *Computer Methods in Biomechanics and Biomedical Engineering: Imaging & Visualization*, **2016**, 1163 (6), 1–9.