Verification of Human Lumbar Vertebrae and Intervertebral Disc Finite Element Models under Mechanical Forces

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Bone, being nonhomogeneous in nature need a complicated and time-consuming process to undergo computed simulation like finite element analysis. To overcome this hurdle, assuming a nonhomogeneous model as homogeneous could be a solution. The objective of this study is to focus on developing a homogeneous human lumbar finite element models and verify them under mechanical force by measuring disc stress, disc strain, disc deformation, total strain, and total deformation. Experimental and geometrical analysis were performed before verifying the lumbar model. To verify the models’ reliability, nonhomogeneous lumbar models were also developed. Five different static structural simulations were performed on four lumbar segments, and twenty parameters were measured. Numerically, out of twenty, eighteen parameters showed very less or no significant difference between homogeneous and nonhomogeneous models of the intervertebral discs and lumbar vertebrae. At the same time, proper caution to be provided while examining the results. With this validation procedure, researchers can process artifact images to get more information which enables them to contribute to the patient’s well-being.

Key words: homogeneous; validation; lumbar vertebra; IVD; FEA; mechanical

1. INTRODUCTION

A structure or a body made from the same material or element in all directions is termed as Homogeneous (H), for example, glass and wood. Materials that have the same properties in all the directions are termed Isotropic, for example, glass. Whereas, materials with different properties in different directions are termed as Anisotropic, for example, wood and bone. Generally, bone is considered as an anisotropic nonhomogeneous model. The shaft of a long bone like femur has longitudinal Young’s modulus of 9.15 ± 5.98 GPa in static and 11.05 ± 3.46 Gpa in dynamic. Whilst, in latitudinal, it is 3.05 ± 1.14 GPa in static and 8.3 ± 3.25 GPa in dynamic (Weerasooriya et al., 2016).

While running a Finite Element Analysis (FEA), the structure and material properties play a vital role. A patient-specific finite element model can be obtained through the CT scan. CT scan is a series of X-ray images taken at a regular interval like 0.2, 0.5, 1.0 mm. This series of images is called Digital Imaging and Communications in Medicine (DICOM) images (Pianykh, 2012). With a clear DICOM image, nonhomogeneous (NH) models can be easily developed. Whilst, when the DICOM images are filled with noises or artifacts caused because of metal implants, then it is very challenging to develop a nonhomogeneous model. In some surgical treatments like scoliosis correction, vertebrectomy, total replacement, a small portion or a part of the bone is removed, and it is substituted by a metal implant. When these patients with metal implants undergo a CT scan, X-rays from the CT scan gets refracted because of the metal implanted inside their body which results in noisy DICOM images.

Human bone is nonhomogeneous in nature. Both the bone and intervertebral disc (IVD) is made up of two layers. Bone
has an inner trabecular bone which is enclosed by an outer cortical bone. The outer thick white layer is cortical, and the inner is trabecular (refer to Figure 1(a)). Whereas, IVD has an inner fluid-like substance called nucleus pulposus which is covered by fibres called annular fibrosus (Erwin and Hood, 2014). Since post-operated patients have metal implants, the CT scan images have a lot of artifacts. Due to the artifacts, the exact layers of cortical and trabecular are unable to be distinguished (refer to Figure 1(b)). It is highly impractical or unachievable at all to develop a proper nonhomogeneous 3D model from the DICOM images which are affected by artifacts and noises. On the other hand, developing a homogeneous model from DICOM images that are affected by artifacts and noises is still achievable (refer to Figure 1(c)).

Figure 1. DICOM images of (a) normal, (b) post-operated with metal implants, (c) post-operated after image segmentation

Most of the prior bone FEA studies done had used an anisotropic nonhomogeneous model. At the same time, looking at the literature, isotropic homogeneous properties were used in several bone FEA studies, due to its convenience (Eswaran et al., 2007; Zulkifi et al., 2011; Shamnadh et al., 2017). But their results were not validated. Using homogeneous lumbar models and isotropic material property, range of motion (ROM) of the human lumbar segment was alone validated in our previous study (Palaniswamy et al., 2019). In this study, the same previously ROM validated human lumbar segments were used to validate the biomechanical properties like stress, strain, and deformation of the lumbar discs and lumbar vertebral segments. This type of validation procedure is convenient and helps the researchers to perform various research on the spine even if the images were affected with artifacts. These researches, in turn, help to develop a healthy society and better living as proposed in the global goals.

The literature review showed that several studies were done in the early days to find the mechanical strength of vertebra. Most of the studies used vertebrae from human cadavers and applied mechanical forces. Trexler et al., (Trexler et al., 2011) used a modified Split Hopkinson Pressure Bar (SHPB) to find shear loading in biological tissues. Bisschop et al., (Bisschop et al., 2013) found the torsional biomechanics of the spine which underwent lumbar laminectomy using Instron. Doulgeris et al., (Doulgeris et al., 2014) used Instron hydraulic apparatus to observe the effects of loading rate in axial rotation mechanics of the lumbar spine. Rahm et al., (Rahm et al., 2019) used MTS mini bionix hydraulic machine to determine the mechanical contribution of the intact rib cage during testing an instrumented specimen. All these studies mentioned above were conducted using human cadaver models.

Limitations to the cadaveric studies are numerous. IVDs in cadaver does not have the same physiological properties as in vivo due to lower fluid. It is also possible for the IVD to get damaged and show altered mechanics due to the presence of pressure sensors (Jones, 2013). Not only the IVD, but even vertebrae also face the same issues. Studies showed that approximately 2600 N is enough to cause failure or dislocation in the upper thoracic spine. However, a cadaver study reported that 613 N caused the failure (Oatis, 2016). Another study conducted by Hutton et al., on 58 lumbar samples between the age 17 to 65 years found that the load required to break lumbar vertebra ranges between 810 N to 15,559 N (Hutton et al., 1979). Adding to these varying results, the preservation and maintenance of cadavers need
sophisticated lab features and most importantly, the availability of human cadaver. To build and maintain a lab, researchers need more fund. Not all researchers can afford it. Particularly, the issues with human cadaver access. At least the cadaver of normal human can be obtained easily. But there are certain disease and musculoskeletal structural disorders like kyphosis, scoliosis, lordosis which affect only two to three percent of the population. When a researcher wants to research scoliosis, which affects the structure of the spine, obtaining scoliosis affected cadaver is very rare or highly impossible. Hence, there is a need to find an alternate way to solve these cadaver issues. It could be the subject-specific finite element analysis. Using the patients CT or MRI scan data, a three-dimensional model of the bone or an organ can be developed digitally and subjected to simulation. With appropriate material properties and simulation procedures, finite element analysis can yield accurate results like experimental studies.

To develop a patient-specific finite element model, researchers need clear DICOM images. Whereas, in certain scenarios, patients are implanted with metal implants. These metal implants refract X-rays and produce artifacts or noisy DICOM images. These artifacts can be bypassed by considering the bone as a homogeneous model (Eswaran et al., 2007; Zulkifi et al., 2011; Shamnad et al., 2017). This paper presents the framework of considering a normal human vertebral segment as homogeneous and validate it by subjecting it to mechanical forces.

II. MATERIALS AND METHOD

After obtaining the ethical committee approval from the Human Ethics Committee, Taylor’s University (HEC/2015/SOE/022), and data collection approval from the Dean of Madras Medical College (01108/MEI/2015), lumbar DICOM images of a normal young adult was obtained from the radiology department. The DICOM images were imported into Materialise version 20.0 (Materialise Inc., Belgium). Image segmentation was done and two separate masks were developed for lumbar vertebrae and IVD. It took nearly five hours to perform manual image segmentation. No muscles, ligaments, blood vessels, nerves, endplates, and metal implants were included in the modeling of lumbar segments. The developed masks were converted into parts. Parts were then smoothened and wrapped to cover tiny holes and sharp edges. An adaptive remesh was done with the triangle edge length of 1 mm to preserve surface contours. After performing a mesh independence study, a volume mesh was created using a four-node tetrahedral element with a maximum edge length of 2 mm. The 3D mesh models of four subject-specific homogeneous human lumbar segments L1-L2, L2-L3, L3-L4, and L4-L5 were developed and exported as .CDB files. The developed lumbar segments underwent geometrical analysis using Ansys Workbench version 17.2 (Ansys, Inc., U.S.A). The results of these analyses were found to be reliable. Very detailed explanations of the experimental analysis and geometrical analysis are presented in the previous study (Palaniswamy et al., 2019). The same four homogeneous human lumbar segments were used in this study as well.

| Table 1. Geometrical properties |
|---------------------------------|
| Lumbar Segments | Homogeneous | | Nonhomogeneous |
| Volume | Mass (Kg) | Nodes | Elements | Volume | Mass (Kg) | Nodes | Elements |
| L1-L2 | 85.62 cm³ | 0.6721 | 56125 | 320425 | L1-L2 | 85.62 cm³ | 0.6721 | 59235 | 341076 |
| L2-L3 | 95.93 cm³ | 0.753 | 62596 | 358597 | L2-L3 | 95.93 cm³ | 0.753 | 72233 | 388263 |
| L3-L4 | 100 cm³ | 0.785 | 65286 | 375760 | L3-L4 | 100 cm³ | 0.785 | 78661 | 427263 |
| L4-L5 | 114.50 cm³ | 0.8125 | 68262 | 394379 | L4-L5 | 114.50 cm³ | 0.8125 | 83261 | 498218 |

After performing the experimental analysis and geometrical analysis, validation analysis of homogeneous finite element models of lumbar vertebrae is performed. Linear homogeneous isotropic material properties were used for the validation of lumbar finite element models. This supposition was verified by developing nonhomogeneous lumbar segments for the same subject with separate material properties for trabecular bone, cortical bone, nucleus pulposus, and annular fibrous. Both the volume and mass of the models were the same in both the homogeneous and nonhomogeneous, but the number of nodes and elements were different. The geometrical properties of the homogeneous and nonhomogeneous models are presented in below Table 1.
acquired from the works of literature (Li and Wang, 2006; Li, 2011; Zheng et al., 2015; Wang et al., 2016) and provided in Table 2.

Table 2. Material properties of Homogeneous and Nonhomogeneous models

| Models          | Young’s Modulus (MPa) | Poisson’s Ratio | Reference       |
|-----------------|-----------------------|----------------|-----------------|
| Cortical        | 12,000                | 0.3            | (Li and Wang, 2006; Wang, 2011; Zheng et al., 2015) |
| Trabecular      | 100                   | 0.2            |                 |
| Nucleus pulposus| 1                     | 0.4999         |                 |
| Annular fibrous | 4.2                   | 0.45           |                 |
| Bone            | 200                   | 0.3            | (Li, 2011; Wang et al., 2016) |
| Disc            | 4                     | 0.4999         |                 |

All the connections were set to multi point constraint contact formulation. The solver type was set to direct. The large deflection was turned off and sub steps were added for gradual loading. By fixing the inferior surface of the vertebral body in the bottom part of the vertebra and applying a force of 1000 N on the superior surface of the vertebral body in the top vertebra, boundary conditions were added (refer to Figure 2). A force of 1000 N was selected because it was estimated that the lumbar spine receives a compressive load of 1000 N during standing and walking (Arjmand et al., 2015). Five different simulations were performed on each L4-L5, L2-L3, L3-L4, and L4-L5 lumbar segments.

Due to the application of force on the lumbar finite element model, changes like stress, strain, and deformation were developed. The normal stress is computed as:

\[ \{\sigma\} = [D]\{\varepsilon^{e}\} \]  

Where, \{\sigma\} is stress vector, [D] is elasticity or elastic stiffness or stress strain matrix, \{\varepsilon^{e}\} = \{\varepsilon\} - \{\varepsilon^{th}\} is elastic strain vector, \{\varepsilon\} is total strain vector, and \{\varepsilon^{th}\} is the thermal strain vector. While the strain is computed as:

\[ \{\varepsilon\} = [B]\{u\} \]  

Where, \{u\} is displacement vector, \{x\} is deformed, and \{X\} is undeformed. The overall stress of a FE model is calculated as equivalent or von-Mises stress. It is computed as:

\[ \varepsilon_{eq} = \frac{1}{2 + \nu'} \left( \frac{1}{2} \left( \varepsilon_{xx}^2 + \varepsilon_{yy}^2 + \varepsilon_{zz}^2 \right) + 2 \left( \varepsilon_{xy}^2 + \varepsilon_{yx}^2 + \varepsilon_{zx}^2 + \varepsilon_{xz}^2 \right) \right) \]  

Where, \nu’ is effective Poisson’s ratio. Whereas, the von-Mises strain of the FE model is computed as:

\[ \varepsilon_{eq} = \frac{1}{2(1+\nu')} \left( (\varepsilon_{xx} - \varepsilon_{y})^2 + (\varepsilon_{yy} - \varepsilon_{z})^2 + (\varepsilon_{zz} - \varepsilon_{x})^2 + 2(\gamma_{xy}^2 + \gamma_{yz}^2 + \gamma_{zx}^2) \right) \]  

Where, \varepsilon_{xx}, \varepsilon_{yy}, \varepsilon_{zz} is appropriate component strain values. The flexibility of a model in the finite element method is defined as:

\[ [D]^{-1} = \begin{bmatrix} 1/E_x & -v_{xy}/E_x & -v_{xz}/E_x & 0 & 0 & 0 \\ -v_{yx}/E_y & 1/E_y & -v_{yz}/E_y & 0 & 0 & 0 \\ -v_{zx}/E_z & -v_{yz}/E_z & 1/E_z & 0 & 0 & 0 \\ 0 & 0 & 0 & 1/G_{xy} & 0 & 0 \\ 0 & 0 & 0 & 0 & 1/G_{yz} & 0 \\ 0 & 0 & 0 & 0 & 0 & 1/G_{zz} \end{bmatrix} \]  

Where, \( E_x \) is Young’s modulus in the X direction, \( v_{xy} \) is major Poisson’s ratio, \( v_{yz} \) is minor Poisson’s ratio, and \( G_{xy} \) is shear modulus in XY plane. Whereas, isotropic materials without shear moduli \( G_{xy}, G_{yz}, \) and \( G_{xz} \) are computed as:

\[ G_{xy} = G_{yz} = G_{xz} = \frac{E_x}{2(1+\nu')} \]  

Figure 2. Lumbar finite element model. (a) Homogeneous model, (b) Nonhomogeneous model

\[ \{u\} = \{x\} - \{X\} \]  

Where, \{u\} is displacement vector, \{x\} is deformed, and \{X\} is undeformed. The overall stress of a FE model is calculated as equivalent or von-Mises stress. It is computed as:

\[ \varepsilon_{eq} = \frac{1}{2 + \nu'} \left( \frac{1}{2} \left( \varepsilon_{xx}^2 + \varepsilon_{yy}^2 + \varepsilon_{zz}^2 \right) + 2 \left( \varepsilon_{xy}^2 + \varepsilon_{yx}^2 + \varepsilon_{zx}^2 + \varepsilon_{xz}^2 \right) \right) \]  

Where, \nu’ is effective Poisson’s ratio.
Figure 3. Flowchart of this study
Total lumbar segment strain and deformation, IVD stress, strain, and deformation were measured. Total deformation and total strain tools were used to find the deformation and strain of the whole lumbar model. Probes like deformation, stress, and strain were used to measure the deformation, stress, and strain of the IVD. A deformation probe was applied to the circumference of the IVD. Stress and strain probes were applied to the whole IVD. In this analysis, deformation is measured in millimetre (mm), stress in megapascal (MPa), and strain in millimetre per millimetre (mm/mm). On the subject of stress and strain, von Mises was measured. As the IVD can get expanded in all the X, Y, and Z-axis, for deformation, instead of selecting a particular axis, the sum of all was selected. The flowchart of this whole study is presented in Figure 3.

III. RESULT AND DISCUSSION

It took nearly three hours to simulate the non-homogeneous models and lesser for homogeneous models in a computer with a 5th generation intel i7 processor and 16 GB RAM. Table 3 and 4 display the results of disc stress, strain, and deformation, total strain, and deformation in lumbar segments between nonhomogeneous and homogeneous models under 1000 N force. The difference between homogeneous and nonhomogeneous models ranges between 0.92 to 1.25 MPa in disc stress, 0.30 to 0.43 mm/mm in disc strain, 0.44 to 0.91 mm in disc deformation, 0.31 to 0.41 mm/mm in total strain, and 0.45 to 0.91 mm in total deformation. With the overall 20 parameters mentioned below in Table 3 and 4, only 2 parameters under disc stress in L2 – L3 and L4 – L5 segments show a difference of more than 1 MPa, but lesser than 1.3 MPa. The remaining 18 parameters showed a difference of less than 1 MPa, mm, and mm/mm. Overall, the percentage difference may seem high. But the direct difference based on the numerical value is less.

The lumbar and cervical curvature are generally ‘C’ shaped when viewed in the sagittal plane. When a single lumbar or cervical segment is viewed alone with an imaginary line running parallel to the superior surface of the top vertebra and inferior surface of the bottom vertebra, it looks like a wedge-shaped and the IVD typically looks like a wedge (refer to Figure 4). This happens because the anterior lumbar IVD height is always higher than the posterior lumbar IVD (Hong et al., 2010; Mirab et al., 2017). The contour plots of lumbar IVD in this study showed mild contour difference between homogeneous and nonhomogeneous models because of the difference in geometrical type and the assigned material properties.

Figure 5 represents the contour plots of von Mises stress, strain, and total deformation of the lumbar IVD under 1000 N of force. From these contour plots, it can be inferred that, while applying force, due to the wedge shape of IVD, ‘C’ shaped vertebral segment, and height of facet joints, the posterior or postro-lateral part of the IVD reacts more. The total deformation of homogeneous IVD in L1 – L2 is 0.52 mm, L2–L3 is 0.76 mm, L3–L4 is 0.55 mm, L4–L5 is 0.72 mm. Whilst, the total deformation of nonhomogeneous IVD in L1 – L2 is 1.22 mm, L2–L3 is 1.67 mm, L3–L4 is 1 mm, L4–L5 is 1.59 mm. Due to the layers of IVD, a noticeable difference was seen in the contour plot of IVD total strain. The total strain of homogeneous IVD in L1–L2 is 0.05 mm/mm, L2–L3 is 0.04 mm/mm, L3–L4 is 0.05 mm/mm, L4–L5 is 0.04 mm/mm. Whereas, the total strain of nonhomogeneous IVD in L1–L2 is 0.41 mm/mm, L2–L3 is 0.45 mm/mm, L3–L4 is 0.36 mm/mm, L4–L5 is 0.43 mm/mm. There were no visible or evident changes in the contour plots of homogeneous and nonhomogeneous IVD total stress.

In the present study, the IVD deformation of L3–L4 nonhomogeneous model was found to be 0.96 mm. Another
Figure 5. Contour images of IVD under total stress, strain, and deformation between Homogeneous and Nonhomogeneous model
finite element model study on L3–L4 lumbar vertebrae reported that the deformation on L3–L4 IVD under 1200 N of compression was to be 0.9 mm (Coogan *et al.*., 2016). This difference could be due to the finite element models used in the respective studies. Coogan *et al.*, used the CT images of cadavers with a mean age of 42.2 ± 13.7 years. Their finite element model included the cartilaginous endplate and cortical bone with shell elements. Vertebrae were meshed with four node tetrahedral elements, their IVD was meshed with eight node hexahedral elements, and used isotropic material properties for their model.

Another FEA study (Li and Wang, 2006) on lumbar disc biomechanical analysis which used an isotropic L1–L2 segment of a young adult reported that the total deformation of L3–L2 segment under 1000 N of the axial load was 0.8 mm. Whereas, in the present study, the total deformation of L4–L3 segment is 0.52 mm. The reported disc stress and deformation of L3–L2 segment cadaver experimental results under 1000 N of pressure in the age group of 22 to 77 years old was found to be 0.74 ± 0.15 MPa and 0.4 ± 0.2 mm (O’Connell *et al.*, 2007). The disc pressure of approximately 0.7 kg/cm² or 0.0686 Mpa, even when they are unloaded. When the vertical load is applied to the disc, the pressure in the nucleus is 50% higher than that applied externally. Despite the percentage difference between homogeneous and nonhomogeneous models, the results of IVD stress and IVD deformation from this study are within the range of data reported in the experimental studies.

There were a few limitations to this study. Eventhough the hexahedral elements provide more accuracy, this study used tetrahedral elements. The software which was used to develop 3D models of vertebrae and IVD has no option to export in hexahedral. There was no literature study supporting the idea of converting a tetrahedral into a hexahedral using additional software. Whereas, several studies used tetrahedral elements (Wang *et al.*, 2016; Meng *et al.*, 2013; Jaramillo *et al.*, 2015). The second limitation would be the usage of isotropic material properties. Bone is generally considered as nonhomogeneous and anisotropic in nature. This study used homogeneous models to validate, and there were no data found in the literature, using anisotropic material properties in a homogeneous model. It shall be experimented in the future research.

| Segment | Disc stress (MPa) | Disc strain (mm/mm) | Disc deformation (mm) |
|---------|------------------|---------------------|-----------------------|
|         | NH | H  | Difference | % Difference | NH | H  | Difference | % Difference | NH | H  | Difference | % Difference |
| L3–L4   | 1.03 | 0.09 | 0.94 | 91.26 | 0.34 | 0.02 | 0.32 | 94.11 | 0.76 | 0.32 | 0.44 | 57.89 |
| L2–L3   | 1.34 | 0.09 | 1.25 | 93.28 | 0.45 | 0.02 | 0.43 | 95.55 | 0.87 | 0.40 | 0.47 | 54.02 |
| L3–L4   | 1.07 | 0.15 | 0.92 | 85.98 | 0.33 | 0.03 | 0.30 | 90.90 | 0.96 | 0.44 | 0.52 | 54.16 |
| L4–L5   | 1.41 | 0.17 | 1.24 | 87.94 | 0.43 | 0.04 | 0.39 | 90.69 | 1.48 | 0.57 | 0.91 | 61.48 |

NH – Nonhomogeneous, H – Homogeneous

| Segment | Total strain (mm/mm) | Total deformation (mm) |
|---------|---------------------|-----------------------|
|         | NH | H  | Difference | % Difference | NH | H  | Difference | % Difference |
| L3–L4   | 0.41 | 0.05 | 0.36 | 87.80 | 1.22 | 0.52 | 0.70 | 57.37 |
| L2–L3   | 0.45 | 0.04 | 0.41 | 91.11 | 1.67 | 0.76 | 0.91 | 54.49 |
| L3–L4   | 0.36 | 0.05 | 0.31 | 86.11 | 1   | 0.55 | 0.45 | 45.00 |
| L4–L5   | 0.43 | 0.04 | 0.39 | 90.69 | 1.59 | 0.72 | 0.87 | 54.71 |
IV. CONCLUSION

The direct comparison between homogeneous and nonhomogeneous models on disc stress, strain, deformation, total strain, and deformation showed a difference of less than one in terms of numerical value in most of the data collected. But in terms of percentage, the difference is high. Hence, considering bone and disc which are nonhomogeneous in nature as a homogeneous does not produce a significant difference in the given scenario. The observation of IVD mechanical forces was consistent with those reported in the literature for IVD stress and IVD deformation. This assumption of nonhomogeneous as the homogeneous model allows the researchers to utilize the artifact affected images, which results in helping the patients to achieve a healthy lifestyle. Therefore, the homogeneous lumbar vertebral segment used in this study is reliable, validated, and will be used for further more analysis. Additional studies are required to analyse the validity of assuming human lumbar vertebrae as homogeneous with multiple segments.

V. ACKNOWLEDGEMENT

A heartfelt thanks to Dr. Nalli R Yuvaraj, orthopaedic spine surgeon from Rajiv Gandhi Government General Hospital, Chennai, for his help and constant support during the data collection. Cordial thanks to Dr. Chin Seong Lim, faculty of engineering, University of Nottingham Malaysia, for his help and support in performing structural simulations at the University of Nottingham Malaysia.

Declarations

Funding
Not applicable.

Conflict of Interest
On behalf of all authors, the corresponding author states that there is no conflict of interest.

Availability of data
Data sharing not applicable – no new data generated.

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