Stress-strain distribution in intact L4-L5 vertebrae under the influence of physiological movements: A finite element (FE) investigation

Devismita Sanjay\textsuperscript{1}, Neeraj Kumar\textsuperscript{1} and Souptick Chanda\textsuperscript{1}

\textsuperscript{1}Department of Biosciences and Bioengineering, Indian Institute of Technology Guwahati, North Guwahati, 781 039, Assam, India

E-mail: devismitasanjay@iitg.ac.in

Abstract: This study is aimed at finding the stress and strain distribution in functional spinal unit of L4-L5 occurring due to physiological body movements under five loading conditions, namely compression, flexion, extension, lateral bending and torsion. To this purpose, 3D finite element (FE) model has been generated using 4-noded unstructured tetrahedral elements considered both for bones and intervertebral disc, and 1D tension-only spring elements for ligaments. The analyses were performed for a compression load of 500 N and for other load cases, a moment of 10 N-m along with a preload of 500 N was applied. The model was validated against in-vitro experimental data obtained from literature and FE analysis data for a range of motion (RoM) corresponding to various loading conditions. The highest stress was predicted in the case of torsion though the angular deformation was highest in case of flexion.

Keywords. Lumbar spine, finite element analysis, ligaments, stress development, strain, range of motion, functional spinal unit, intervertebral disc.

1. Introduction

The lumbar spine plays a vital role in transferring the weight to the pelvis. It also permits various types of physical movements during daily activities. The physiological movements may include the loading conditions of compression, flexion, extension, lateral bending and torsion. A major portion of the population, especially the elderly people, suffer from lower back pain and may need surgical help depending on severity of the ailment. It is, therefore, essential to gain insights into the biomechanics of lower vertebrae. In the past, many \textit{in-vitro} studies were performed to describe the mechanical behaviour of lumbar spine [1, 2, 3, 4]. Moreover, finite element (FE) based \textit{in-silico} analyses were also performed for different physiological movements for functional spinal unit (FSU) of L4-L5 [5, 6, 7]. During these experimental and FE analyses, the measurements or predictions, respectively, were carried out with regard to range of motions, displacements, intradiscal pressure and disc bulges. In the FE study by Tsouknidas et al [8], stress development in lumbar spine was investigated for two mobility scenarios of walking and running. In another FE study by Talukdar et al [9], the stress-strain distribution was investigated for the physiological movement load influences on L3-L4 FSU. In another FE studies, for physiological movements of compression and torsion, the stress-strain distribution was investigated on pediatric femur bone [10, 11]. There is, however,
hardly any study on the stress and strain distribution on the bones and intradiscal segments of L4-L5 FSU, occurring due to physiological loading conditions of body movement like compression, flexion, extension, lateral bending and torsion. In the present study, FE investigation was carried out in order to understand the load influence of daily movements for five loading conditions mentioned above, on L4-L5 vertebrae.

2. Materials and method:

The effects of physiological loading conditions were analysed using FE models of the functional spinal unit of intact L4-L5 vertebrae. The L4-L5 vertebrae was assessed for range of motions (RoM) and stress-strain distribution, and the validation of the in-silico scheme was conducted based on available literatures. The materials and method starting from preparation of virtual osteotomy till the biomechanical analyses have been described stepwise in the following subsections.

2.1 Development of model

Anatomically representative CAD models of L4 (model #3902) and L5 (model #3901) vertebra were procured from the manufacturer (Sawbones, Europe AB, Malmo, Sweden) and the FSU was anatomically placed in NURBS modelling platform of Rhinoceros v5.0 (Rhinoceros, Robert McNeel & Associates Seattle, USA). The cancellous and cortical bone were virtually segregated by performing Boolean operations in Rhinoceros. The average thickness of cortical bone was kept as 1 mm. The intervertebral disc (IVD) consisting of annulus fibrosus and nucleus pulposus was prepared thereafter. The nucleus pulposus occupies almost 43% of the entire disc volume [9]. A thickness of 0.5 mm was assigned to both bony end-plates and cartilaginous endplates. Facet cartilage was developed with an initial gap of 0.1 mm.

The 3D FE model for the FSU of L4-L5 was generated using Hypermesh v2019.1 (Altair Engineering Inc., Troy, Michigan, USA). The volumetric mesh for the model was generated using 4-noded unstructured tetrahedral elements (Figure 1). The average edge length of the elements was kept as 1 mm. Six CGAP elements (for ligaments)

Figure 1: FE model of intact L4-L5 with loading and boundary condition
types of ligaments were modelled using 1D tension-only spring elements, called CGAP elements. These elements were ideally suited to mimic soft tissue properties since they imparted stiffness only in case of tension whereas in compression the stiffness was zero.

2.2 Material properties

The material properties of the FE models were considered to be linear, elastic and isotropic and are presented in table 1. Low elastic modulus \(E = 0.1\) MPa was considered for nucleus pulposus in order to simulate the incompressible fluid like behaviour [13].

\[
\begin{array}{|c|c|c|}
\hline
\text{Components} & \text{Young's modulus (MPa)} & \text{Poisson’s ratio} \\
\hline
\text{Cortical bone} & 12000 & 0.3 \\
\text{Cancellous bone} & 100 & 0.3 \\
\text{Bony endplate} & 1200 & 0.29 \\
\text{Cartilaginous endplate} & 23.8 & 0.28 \\
\text{Nucleus pulposus} & 0.1 & 0.49 \\
\text{Annulus pulposus} & 9 & 0.4 \\
\hline
\end{array}
\]

The lengths of the ligaments were curated based on data from available literature [16, 17]. Stiffness values of the ligaments were calculated from the length, anatomical cross-section area and Young’s moduli of ligaments taken from the literature [6, 7, 13]. For calculating the stiffness, axial stiffness formula, \(K = AE/L\), was employed, where \(K\) is the stiffness, \(A\) is the cross-sectional area, \(E\) is the Young’s modulus and \(L\) is the length of the ligament. The calculated stiffness values were found to corroborate those from existing literature [16, 17]. The stiffness values of the ligaments are presented in table 2.

\[
\begin{array}{|c|c|}
\hline
\text{Ligaments} & \text{Stiffness (N/mm)} \\
\hline
\text{Anterior longitudinal} & 45.2 \\
\text{Posterior longitudinal} & 26.49 \\
\text{Flavum} & 43.71 \\
\text{Intertransverse} & 2.769 \\
\text{Interspinous} & 35.5 \\
\text{Supraspinous} & 12.72 \\
\hline
\end{array}
\]
2.3 Loading and Boundary Condition

The analysis was performed for five loading conditions namely compression, flexion, extension, lateral bending and torsion. A rigid body element (RBE3) was created, where a ‘dependent’ node was selected roughly at the centre of top surface of L4 cortical bone, and all surface nodes of L4 cortical bone were selected as ‘independent’ nodes. The loads were applied at central node of RBE3 which is at the top surface of L4 and calculated based on the weighted average of motions for all the surface nodes of L4 cortical (Figure 1). Thus, the loads were assumed to be acting through the center of gravity of L4. For the compression loading condition, a load of 500N was applied. For flexion, extension, lateral bending and torsion, 10 N-m moment was applied along with a preload of 500N [5]. All the bottom surface nodes of L5 cortical of FSU L4-L5 was constrained for all six degrees of freedom (Figure 1).

2.4 Contact Analysis

Surface-to-surface contact elements were employed (coefficient of friction, $\mu = 0.2$) [18], on both the left and right side at the facet cartilage interface of L4-L5. The contact problem was solved using Augmented Lagrangian method, where L4 cartilage was considered as slave and L5 cartilage as master body.

![Figure 2: Comparison of RoM in present study with existing literature [1, 5]: (a) axial compression, (b) flexion, (c) extension, (d) lateral bending and (e) torsion.](image)
2.5 Validation and analysis

The FE analysis was performed using ‘Optistruct’ solver of Hypermesh 2019.1. For the analysis, geometric nonlinearity was employed, wherein the stiffness is a function of displacement contrary to linear analysis where stiffness remains constant. The nonlinear analysis was implemented using piecewise linearity such that load was split into small increments and stiffness matrix was updated after each incremental load application. The iterations were performed to get equilibrium condition after every increment and Newton’s Raphson method was used to solve the equations [19]. The validation of the model was done considering the RoM under various loading conditions.

3. Result

In order to have an insight into the influence of different physiological movements on lower vertebra, stress-strain characteristics of the intact L4-L5 model were investigated in the present study. The results for range of motion (RoM) versus load were validated against the experimental data [1] and FE analysis data [5], as shown in figure 2. The contour plots of stress-strain distribution for different loading conditions were obtained and reported.

![Figure 3: Von Mises stress contours: (a) cortical bone and (b) cancellous bone](image-url)
3.1 Stress analysis

Von Mises stress contour plots for five loading condition are presented in figure 3. Both the cortical bone (Figure 3a) and cancellous bone (Figure 3b) were found to be following identical trend in stress distribution. The peak stress values in case of cortical and cancellous bone (Figure 3) were predicted to be ~50 MPa and ~1.6 MPa, respectively. The torsional movement predicted maximum stress followed by the extension. On the other hand, minimum stress was obtained in case of simple compression. In case of compression, the stress value for the cortical bone was found to be under 15 MPa (Figure 3a) while the value was less than 0.7 MPa for cancellous bone (Figure 3b). Under all loading condition, except lateral bending, the maximum stress region was found towards the posterior side. The lateral bending condition was found to induce the maximum stress towards the anterior side of the FSU.

![Von Mises stress contours](image)

**Figure 4:** Von Mises strain contours: (a) cortical bone and (b) cancellous bone

3.2 Strain analysis

Von Mises strain contour plot for the five loading conditions are shown in figure 4. The strain distribution was found to be following similar trend as stress distribution and also across the two types of bone – cortical and cancellous (Figure 4). The torsion was found to register peak strain for both cortical (~0.004) and the cancellous bone (~0.014) (Figure 4). The value of strain in case of simple compression for the cortical bone (Figure 4a) was found below 0.0015 and for the cancellous bone the predicted value was below 0.004.
4. Discussion

An intact L4-L5 model was assessed under different physiological movements to gain insights into biomechanical behaviour of lumber vertebra. This study is specifically helpful to know how the load transfer occurs during daily physiological activities. Previous studies suggested that proper loading and boundary condition plays a significant role for stress-strain analysis in lumbar FSU [9]. The validation of the model was done considering the RoM under various loading conditions. The loading conditions were obtained from the available literatures [1, 5]. Axial compression loading conditions considered as vertical force of 500N applied axially at the top surface of cortical of L4. This type of loading may arise while standing. In all the other four loading conditions, the axial force of 500N was considered along with the moment of 10 Nm. The magnitude of moment was kept same in all four cases of physiological movements though it differed in direction depending on the load case.

The FE results of RoM were found to be closely associated with the experimental results [1], as shown in figure 2. A maximum relative displacement of 0.38 mm was reported in the experimental study [1] whereas Xiao et al [5] predicted the value to be 0.88 mm under axial compression. In the present study, the predicted displacement was ~0.65 mm which was an average of the previous two findings. The maximum difference between predicted and measured angular deformation, however, was observed under torsion though the value was close to that reported by Xiao et al. [5]. In all other cases, a reasonable agreement with values reported in literature was found.

Earlier studies with implants (pedicle screw) [20-22] reported that the stress was found to be higher in neck region of pedicle for loading conditions such as flexion, extension, lateral bending and torsion. The stress distribution pattern in case of intact model of L3-L4 FSU by Talukdar et al. 2021 [9] and our FE result for L4-L5 FSU depicted a similar trend in the physiological movements. However, depending on the type of load, the peak stress was found to be in different locations, i.e. for extension it was highest towards posterior side whilst for lateral bending, it was in the lateral part of the vertebra. The von Misses strain distribution pattern for intact L4-L5 model was found to be similar in case of compression load of 500 N for both numerical value and the reported pattern [9]. Furthermore, the peak strain was found to follow a similar trend as that in the stress distribution.

5. Conclusion

An improved FE model of lumber vertebra was developed to understand the load transfer during physiological movements. From the study, it was evident that loading and boundary condition have significant influence on stress-strain distribution in L4-L5 vertebra. This study gave an insight into the stress-strain distribution corresponding to different loading conditions. The maximum stress obtained in case of torsion through our FE analysis was found near to the value reported by Biswas et al FE investigation performed on L3-L5. In their study, the maximum values of stress were reported but it did not display the contour plot of stress distribution. It is evident from previous study that the stress distribution pattern can help in better design of implants. From the present study it was found that, the highest stress was obtained in the
case of torsion but the angular deformation was highest in case of flexion. Therefore, it can be concluded that only range of motions (RoM) values may not give a wholesome picture with regards to estimation of the effect of loading. Rather, a stress and strain distribution pattern can help understand the effect of loading and failure criteria in a better way, and serve as a reference for improved design of implant.

Acknowledgements

The authors would like to acknowledge the computational facilities available at the Biomechanics laboratory of the Department of Biosciences and Bioengineering, Indian Institute of Technology Guwahati, which has helped to carry out this research study. The study has been partially supported by SERB, India (Grant no. SRG/2019/000235).

Reference:

[1] Markolf KL. Deformation of the thoracolumbar intervertebral joints in response to external loads: a biomechanical study using autopsy material. *J Bone Joint Surg Am.* 1972 Apr; 54(3):511-33. PMID: 5055150.

[2] Markolf KL, Morris JM. The structural components of the intervertebral disc. A study of their contributions to the ability of the disc to withstand compressive forces. *J Bone Joint Surg Am.* 1974 Jun; 56(4):675-87. PMID: 4835815.

[3] Virgin WJ. Experimental investigations into the physical properties of the intervertebral disc. *J Bone Joint Surg Br.* 1951 Nov; 33-B (4):607-11. doi: 10.1302/0301-620X.33B4.607. PMID: 14880588.

[4] Brown T, Hansen RJ, Yorra AJ. Some mechanical tests on the lumbosacral spine with particular reference to the intervertebral discs; a preliminary report. *J Bone Joint Surg Am.* 1957 Oct; 39-A (5):1135-64. PMID: 13475413.

[5] Xiao Z, Wang L, Gong H, Zhu D, Zhang X. A non-linear finite element model of human L4-L5 lumbar spinal segment with three-dimensional solid element ligaments. *Theoretical and Applied Mechanics Letters* 2011; Volume 1, Issue 6, 064001.

[6] Vena P, Franzoso G, Gastaldi D, Contro R, Dallolio V. A finite element model of the L4-L5 spinal motion segment: biomechanical compatibility of an interspinous device, *Computer Methods in Biomechanics and Biomedical Engineering*. 2005; 8:1,7-16, DOI: 10.1080/10255840500062914.

[7] Boccaccio A, Vena P, Gastaldi D, Franzoso G, Pietrabissa R, Pappalettere C. Finite element analysis of cancellous bone failure in the vertebral body of healthy and osteoporotic subjects. *Proc Inst Mech Eng H.* 2008 Oct; 222(7):1023-36. doi: 10.1243/09544119JEIM296. PMID: 19024151.

[8] Tsouknidas, A.; Savvakis, S.; Tsirelis, N.; Lontos, A. and Michailidis, N. (2013). A Non-linear Finite Element Model for Assessment of Lumbar Spinal Injury Due to Dynamic Loading. In *Proceedings of the*
International Conference on Bioinformatics Models, Methods and Algorithms - BIOINFORMATICS, (BIОСТЕК 2013) ISBN 978-989-8565-35-8; ISSN 2184-4305, pages 292-295. DOI: 10.5220/0004236902920295

[9] Talukdar RG, Mukhopadhyay KK, Dhara S, Gupta S. Numerical analysis of the mechanical behaviour of intact and implanted lumbar functional spinal units: Effects of loading and boundary conditions. Proc Inst Mech Eng H. 2021 Jul; 235(7):792-804. doi: 10.1177/0954419211008343. Epub 2021 Apr 9. PMID: 33832355.

[10] Kanu NJ, Patwardhan D, Gupta E, Vates UK, Singh GK. Finite element analysis of mechanical response of fracture fixation functionally graded bone plate at paediatric femur bone fracture site under compressive and torsional loadings. Materials today: Proceedings 2021, Volume 38, Part 5, https://doi.org/10.1016/j.matpr.2020.08.740

[11] Kanu NJ, Patwardhan D, Gupta E, Vates UK, Singh GK. Numerical investigations of stress-deformation responses in fractured paediatric bones with prosthetic bone plates. 2020 IOP Conf. Ser.: Mater. Sci. Eng. 814 012038

[12] www.sawbones.com

[13] Li J, Shang J, Zhou Y, Li C, Liu H. Finite Element Analysis of a New Pedicle Screw-Plate System for Minimally Invasive Transforaminal Lumbar Interbody Fusion. PLoS One. 2015 Dec 9; 10(12): e0144637. doi: 10.1371/journal.pone.0144637. PMID: 26649749; PMCID: PMC4674154.

[14] Goel VK, Kiapour A, Faizan A, Krishna M, Friesem T. Finite element study of matched paired posterior disc implant and dynamic stabilizer (360° motion preservation system). SAS J. 2007 Feb 1; 1(1):55-61. doi: 10.1016/SASJ-2006-0008-RR. PMID: 25802579; PMCID: PMC4365571.

[15] Wang W, Zhang H, Sadeghipour K, Baran G. Effect of posterolateral disc replacement on kinematics and stress distribution in the lumbar spine: a finite element study. Med Eng Phys 2013; 35:357–64. https://doi.org/10.1016/j.medengphy.2012.05.013.

[16] Yoganandan, N., Kumaresan, S., and Pintar, F. A. "Geometric and Mechanical Properties of Human Cervical Spine Ligaments." ASME. J Biomech Eng. 2000 December; 122(6): 623–629. https://doi.org/10.1115/1.1322034.

[17] www.wheelessonline.com/issls/lumbar-vertebrae-ligaments/

[18] Zhou Q, Zeng F, Tu J, Dong Z, Ding ZH. Influence of cement augmented pedicle screw instrumentation in an osteoporotic lumbosacral spine over the adjacent segments: a 3d finite element study. J Orthop Surg Res 2020; 15(1):132.

[19] Gokhale NS, Deshpande SS, Bedekar SV, Thite AN, Book- practical finite element analysis.,(2008)445.
[20] Kim K, Park WM, Kim YH, Lee S. Stress analysis in a pedicle screw fixation system with flexible rods in the lumbar spine. *Proc Inst Mech Eng H*. 2010; **224**(3):477-85. DOI: 10.1243/09544119JEIM611. PMID: 20408492.

[21] Biswas JK, Rana M, Roy S, Majumder S, Karmakar SK, Roychowdhury A. Effect of range of motion (ROM) for pedicle-screw fixation on lumbar spine with rigid and semi-rigid rod materials: A finite element study. *2018 IOP Conf. Ser.: Mater. Sci. Eng.* **402** 012146

[22] Kumbhalkar MA, Rambhad KS, Kanu NJ. An insight into biomechanical study for replacement of knee joint. *Materials today: Proceedings*, Volume 47, Part 11, 2021, Pages 2957-2965, https://doi.org/10.1016/j.matpr.2021.05.202