NAZARBAYEV UNIVERSITY

School of Engineering and Digital Sciences

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BIOMECHANICS OF LOW BACK PAIN

by

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Declaration

We hereby declare that this report entitled “Biomechanics of low back pain” is the result of our own project work except for quotations and citations that have been duly acknowledged. We also declare that it has not been previously or concurrently submitted for any other degree at Nazarbayev University.

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Abstract

Low back pain (LBP) is a worldwide health issue with one of the highest ratings among patients as a result of mechanical stress and structural disruptions in the lumbar spine, particularly in the L4-L5 segments. It plays a significant role in the maintenance of a healthy life, and more importantly, it is still challenging to diagnose symptoms accurately due to the multifactorial nature of its causes. Specifically, the success rate of current treatment plans is limited by the inability to reliably link structural abnormalities with symptoms. To better understand the mechanical causes of LBP along with better guidance of clinical decision-making, this project aims to cover the necessity of sophisticated biomechanical modelling. This study is aimed to investigate the biomedical response of lumbar spine FEM (L1-L5 bones) under 3 static loading conditions in ANSYS software. The three distinct loading conditions include pure axial compression (force applied at 0° to the vertical axis on L1), forward bending (force applied at 45° on L1), and side bending (force applied at 20° on L1), while L5 being fixed. The simulations focused on total deformation, equivalent elastic strain, and equivalent (Von-Mises) stress as functions of force from 0 N to 1000 N with the step of 250 N. The results of simulations showed a noticeable increase in all above mentioned biomechanical parameters when exposed to bending conditions compared to pure axial compression in equivalent forces. The most increase is noticed at the 45° forward bending scenario, leading to a non-linear increase at higher force magnitudes. The novelty of this work is in the direct and quantitative comparison of 3 specified loading scenarios in a framework of consistent computational simulations. The comparison and results highlight the significant differences in mechanical demands because of these postures, giving insights on how load direction and different load scenarios can alter the stress and strain in simulation environments.

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1 Introduction and Problem Statement

1.1 What is Low Back Pain

Low back pain (LBP) is a major and ever-growing health problem worldwide, recognised as the single leading cause of disability and is vastly reducing the overall quality of life of people with it. In the year 2020, an estimated 619 million people worldwide suffered from LBP, and this is very likely to rise to 843 million by the year 2050. Such prevalence is mostly driven by the high population growth and ageing demographic [1]. While LBP can occur at any age without any exception, in general, it is more prevalent in elderly people, reaching a peak impact in terms of disability among the elderly (80-85 years), although the highest number of cases occurs around the ages of 50–55 years [2]. The vast majority of LBP cases, estimated at around 90%, are classified as non-specific, meaning no identifiable specific pathology or structural abnormality can fully account for the symptoms. Epidemiological data also indicate that women are more susceptible to this condition compared to men at most age groups [2]. Such widespread presence of the LBP translates into a significant challenge for public health with serious consequences for society as a whole. This condition highly restricts life and simple daily activities, not only physically affecting the person, but also putting high pressure on the mental health as well by hindering engagement in work, family, friends and any social life. The projection predicting an increase in the number of cases of people with LBP underscores the need for a better understanding of the problem and developing strategies to prevent and manage it. Since LBP can occur at any age, there is a need to understand the underlying drivers of LBP across the lifespan of a person. LBP during mid-life could be caused by acute injuries or the cumulative effects of constant occupational exposure, such as a sedentary job, lifting heavy objects, etc. LBP in later life is mostly caused by the degenerative changes in the intervertebral discs and reduced tissue resilience. Understanding how various mechanical loads contribute to spinal stress could therefore hold differential relevance for various age demographics.

In addition to the individual inconveniences and suffering, LBP also imposes high economic difficulties. It includes both direct expenditures for healthcare, like visits to doctors, imaging and diagnosing, rehabilitation and medication, and also includes indirect losses due to lost productivity and inability to do work at all or perform at a sufficiently high level [3]. Globally, LBP is a leading cause of job-related disability and accounts for millions of lost work days annually. Cost estimates vary by region and methodology but consistently point towards billions of dollars in annual expenditure, with chronic LBP cases driving the majority of these costs [4]. This enormous economic cost of the LBP shows the societal need to address the condition through research and simulations.

1.2 Biomechanics of LBP

The human spine is a very complex biomechanical structure that is engineered to support the body weight of a person, protect the spinal cord and most importantly enable movement. It comprises vertebral bodies, which primarily resist compressive forces, and intervertebral discs (IVDs), which act as flexible cushions allowing motion and distributing loads [5]. Each IVD consists of a soft, gel-like nucleus pulposus encased by the tough, layered annulus fibrosus. Under compression, the nucleus pressurises and exerts outward pressure, generating tensile "hoop" stresses within the annulus fibres, effectively functioning as a shock absorber [5]. Mechanical factors such as maintaining awkward postures for prolonged periods, engaging in tasks involving high spinal loading (e.g., heavy lifting), performing repetitive movements, and exposure to whole-body vibration are recognised risk factors [1]. Such factors lead to excessive stress in the spinal tissues and, over time, cause injuries or micro-damage in the structure of intervertebral discs or joints, causing such conditions as annular tears, disc herniation or disc degeneration. Interestingly, while LBP development is often linked to gradual, cumulative processes like fatigue damage or degeneration, clinical presentation frequently involves a sudden onset of symptoms associated with a seemingly innocuous activity or a moderate loading event, often involving a flexed (bent forward) posture. Therefore, investigating how different physiological postures and movements, such as axial compression, forward bending, and side bending, influence the distribution and magnitude of stress and strain within the spinal components is essential for understanding potential injury mechanisms and identifying postures or activities that pose a higher mechanical risk, especially in individuals with pre-existing spinal vulnerability.

This report focuses on the lumbar region of the spine, which is particularly vulnerable, particularly its lower segment, since it plays a vital role in carrying body weight. The mechanical elements include vertebrae, intervertebral disks, ligaments, and facet joints. It is subjected to mechanical forces that may lead to degenerative conditions, traumatic injuries, and pain from overuse injuries. The structural and functional significance of the intervertebral disks, particularly that of L4-L5, is related to high mechanical stresses [6]. This is also because both disks provide flexion, extension, and rotation, yet are most liable to bulge and herniation.

Generally speaking, a profound understanding of the biomechanical factors that provoke LBP and the non-standardised character of current diagnostic methodology are seen as the most pressing issues that require an individualised diagnostic and therapeutic approach. The aim of this paper is to, using 3D Slicer, construct a high-fidelity 3D model of the lumbar spine. This model will later be used in Finite Element Analysis (FEA) simulations in

ANSYS to replicate the lumbar spine's biomechanical environment and analyse stress distribution under various movement scenarios.

1.3 Aim and objectives

The primary objective of this capstone project is to employ ANSYS FEA to simulate and quantitatively compare the biomechanical response of a representative lumbar spine model subjected to three distinct static loading scenarios. These scenarios are designed to represent different physiological postures or movements:

1. Pure axial compression (simulated by applying force vertically downwards, at 0° to the spinal axis).
2. Forward bending (simulated by applying force at a 45° angle to the vertical axis in the sagittal plane).
3. Side bending (simulated by applying force at a 20° angle to the vertical axis in the coronal plane).

The analysis will specifically focus on comparing the resulting maximum total deformation, maximum equivalent elastic strain, and maximum equivalent (von Mises) stress across these three conditions as the magnitude of the applied force is incrementally increased. While numerous FEA studies have explored spinal biomechanics under various loading modes, including compression, flexion, extension, lateral bending, and torsion, the specific novelty of this investigation lies in the direct, quantitative comparison of the mechanical responses generated by applying forces at these three specific angles (0°, 45°, and 20°) within a single, consistent ANSYS FEA model. The findings are intended to contribute model-specific insights into how significantly the direction of applied load alters the resulting stress, strain, and deformation environment within the lumbar spine segment.

The two authors, Nagyn Seitkhan and Sanzhar Nugmanov, contributed to the literature review, methodology, and results of the capstone project, also being responsible for simulation analysis. The second author, Temirlan Ilyassov, contributed to the simulations of the 3D model in ANSYS software, also being involved in the literature review. The fourth author, Timur Zhursinow, contributed to the 3D model of our project, also involved in the literature review.

The remainder of this report is organised as follows: Chapter 2 reviews the relevant literature on lumbar spine biomechanics and LBP. Chapter 3 details the methodology for constructing the 3D model and conducting FEA simulations. Chapter 4 presents the results and discusses their implications.Chapter 5 provides analysis of results and discussion. Finally, Chapter 6 provides conclusions and recommendations for future research.

2 Literature Review

LBP is a widespread health concern affecting a significant portion of the population. While biomechanical factors undoubtedly play a crucial role in LBP, most of the research papers point out the limitations of a purely biomechanical approach to understanding this problem. Therefore, scientists argue for a more comprehensive and diverse framework that recognises the connection between biomechanical factors and other biopsychosocial elements that can cause the LBP. Biomechanics refers to the mechanics of the body, including neuromuscular control [5].

2.1 Diagnostic Approaches to LBP

Modern diagnostic strategies, such as MRI and CT, enable the determination of structural lesions, but most of them do not indicate a good relationship between the lesion and the symptoms. Non-association, however, has led to inconsistency in clinical decision-making and the inability to attribute a cause for LBP [6]. In addition, disparities in access to appropriate care, due to socioeconomic and ethnic considerations, serve as another aspect of challenges in LBP management [6].

2.2 Mechanical Stresses, Disc Degeneration and chronic LBP

These studies noted that most of the stresses are exposed to the intervertebral discs L4-L5, where mechanical stresses tend to accumulate due to high carrying load potential and involvement in movement. Because such physical and biomedical exposure exists, over time, these discs are more susceptible to different degenerative pathologies like herniation and bulging [5]. Therefore, this means that additional biomechanical modelling, which is more advanced in nature, needs to be done to explain the distribution of stress across the lumbar spine with respect to LBP.

In his research, Adams [8] demonstrated the role of structural and mechanical factors in the development of LBP. Mainly, he focused his attention on the fact that chronic pain often originates in the lumbar intervertebral discs, apophyseal joints, or sacroiliac joints. Factors attributed to mechanical loading (like bending and compression) eventually damage spinal tissues. Thus, this leads to degenerative changes like disc prolapse and endplate fractures. It is important to note that degeneration of the discs, in comparison with natural ageing, highly contributes to biomechanical changes. Moreover, degeneration results in uneven stress distribution in pain-sensitive regions. Environmental influences, such as repetitive strain and poor posture, worsen tissue damage. Accordingly, Adams [8] focuses on the importance of such biomechanical and environmental factors that develop stress concentration and lead to LBP, even without visible structural damage. These findings support a multifactorial view of LBP, integrating mechanical, genetic, and psychosocial considerations for effective diagnosis.

As it was mentioned before, the lumbar spine often experiences excessive stress, which usually contributes to the growth of LBP. This can also be explained by the fact that this region is a transitional region between the spine and legs, thus, there is a significant amount of motion along with a load transfer. Likewise, Cholewicki et al. [9] explored the interaction of mechanical, neuromuscular, and psychosocial factors in LBP and emphasised the need for an extended research that integrates biomechanical, psychological, and social factors. More specifically, the nonspecific effects of various therapies suggest the complexity of LBP and the high need for developed strategies. The authors argue for a systems strategy that incorporates biopsychosocial factors alongside biomechanical measures.

2.3 Lumbar Stability and LBP

The lumbosacral region is a very important region when it comes to spinal mobility and weight-bearing and this makes it important to the study of LBP. Appropriately, Nakipoglu [10] investigated the spine parts like sacral inclination, and its relationship with the LBP lumbar lordosis - as a result, no significant differences between acute and chronic LBP patients have been found. Nevertheless, this study found that the lumbar instability was significantly higher in very severe cases, 62.1%, in comparison with chronic cases, where it was 36.8%. Essentially, this indicates that reduced stability might be a contributing factor in the chronicity of LBP. Existing correlations between lumbar stability and angular measures, including, but not limited to lumbar lordosis and sacral inclination, further suggest the need for stability in LBP. Similarly to previous research, the findings suggest that further research into the characteristics of lumbar stability and angular parameters is necessary to explain how lower back pain progresses.

2.4 Advances in Biomechanical Modeling and Surgical Interventions

Nicholas et al. [11] developed a 3D finite element model of the L3-L4 vertebrae using several software like MIMICS, Geomagic, and ANSYS, which resulted in accurate simulations of lumbar anatomy and biomechanics. For our research, the model of the author gave a lot of detailed insights into stress distribution and range of motion under various movements, validated against in vivo experiments and clinical data. Neural networks have also predicted stress values at about 0.99666 for the regression score, while the fuzzy logic system categorised subjects into healthy bone mineral density, low bone mineral density, and osteoporosis. Essentially, this integrated approach gives a lot of insights and also improves biomechanical analysis and decision-making concerning the lumbar spine within research and treatment.

Ghista et al. [12] provided a comprehensive biomechanical analysis of a surgical procedure in the case of herniated lumbar discs. In this particular research, surgical treatment involves the reduction of pressure between the discs and compression on nerve roots/pain receptors by partial removal of the nucleus pulposus. By that, the study highlights that it is important to carefully regulate the amount of nucleus removed, meaning that the nucleus plays a critical role in maintaining disc mechanics which includes factors such as stress distribution or deformation. Using a biomechanical model, the above-mentioned author [12] presented the relationship between nucleotomic size, aspiration pressure, the volume of nucleus material removed, and the duration of the procedure. Moreover, Ghista et al. [12] developed a biomechanical model based on principles of fluid mechanics, specifically the Hagen-Poiseuille equation, to describe the flow of the nucleus pulposus through an aspirating tube.

2.5 Stimulating factors of LBP

The research done by Sany et al. [13] focuses on the prevalence of low back pain among medical students in Bangladesh - as a result, it revealed a high 12-month prevalence rate of 63.3%. Furthermore, another factor such as the impact of body mass index (BMI) on the prevalence of LBP has been found, with individuals who had a BMI over 25 kg/m^2 experiencing a higher prevalence (73.2%) compared to their lower BMI counterparts (56.7%). Accordingly, these findings suggest a common correlation between obesity and increased body weight contributing to LBP. Moreover, the study also finds a positive association with physical activity and further mitigation of LBP: participants with higher levels of physical activity in their daily lives demonstrated significantly lower rates of back pain. Similarly to that, prolonged sitting for more than six hours per day was associated with a higher prevalence of LBP. All these findings suggest that a sedentary life and prolonged sitting along with physical activity are key risk factors for LBP among students.

The same study by Sany et al. [13] also identified gender differences in LBP symptoms, with females having a higher prevalence (72.2%) compared to male counterparts (52.2%). This finding reveals another correlation that needs to be considered when proposing a preventive method for our project. Indeed, women are assumed to be more prone to LBP due to risk factors such as anatomical, physiological, and hormonal factors [13].

Another factor that is mostly said to be associated with LBP - the problem of prolonged sitting in daily life. As a part of the literature review, it was challenging to get more details on the different factors while sitting, rather than finding the existence of positive correlation itself. One of the studies that cover both mentioned aspects [14]. In this study, the author investigated the environments in which experiments were conducted to check if sitting stimulates LBP. More importantly, the author indicated all details on the experiments - time of sitting, type of experiment and its environments. As a result of the review towards 10 articles with over 1000 participation overall, it was found that sitting for total durations ranging from 1 hour to 6.96 h/d for 5 days is associated with immediate increases in LBP in people with and without a clinical history of LBP. More importantly, these results are taken from objectively measured experiments in both laboratory and field settings.

In terms of managing LBP, there are many ergonomic interventions such as dynamic sitting have been proposed worldwide to reduce discomfort. However, several articles have found that these propositions need to be checked thoroughly. It was found that while dynamic sitting can play a significant role in improving the back's state, the overall impact is not tremendous and does not propose a piece of clear evidence over static sitting practices. For instance, the systematic review done by O'Sullivan et al. [15] reported a lack of evidence that supports the effectiveness of dynamic sitting as the research suggests that despite several studies exploring dynamic seating mechanisms, the results mainly remained inconclusive.

2.6 FEA of the Lumbar region

Finite Element Analysis (FEA) has become a crucial part in spinal biomechanics research, offering a non-invasive means to estimate internal stresses, strains, and deformations under any needed conditions. Nowadays, a ton of studies have employed FEA to investigate the lumbar spine. and common methodologies involve creating 3D models from CT or MRI scans using software like Mimics, processing the geometry in tools like Geomagic or SpaceClaim, and performing simulations in platforms such as ANSYS or Abaqus. In these simulations, material properties are usually assigned based on literature values. Despite the extensive body of work, a specific gap exists concerning the direct, quantitative comparison of the biomechanical effects of applying forces at specific angles (0°, 45° forward, 20° sideways) within a single, consistent FEA model. While studies analyse compression, flexion, and lateral bending individually, often using pure moments to induce bending, the approach of applying an angled force inherently creates a combined loading state (compression plus shear/bending). This study aims to fill this gap by providing a focused, comparative assessment of these three specific loading scenarios simulated via angled forces in ANSYS, contributing model-specific data on how load direction, under this combined loading paradigm, alters the geometry of the lumbar spine.

The paper reviews a wide range of literature concerning the biomechanical risk factors in low back pain. Conducting a literature review gives a lot of opportunities to build on past experiences and research of other people, also gaining some insights for future work. It also helps to reveal any gaps that exist in the literature, which for this paper is the sphere of the biomechanics of low back pain and points to the way where additional research is needed.

The main focus was concentrated on the aspects of mechanical properties and stress distribution of the lumbar intervertebral discs, namely L1-L5. In addition, the article Bojairami et al. [16] provided important information about Young's modulus and Poisson's ratio, which was crucial in order to get an accurate model. The analyses that researched axial load and shear stress effects on the degeneration of the lumbar spine fostered the simulation methodologies utilized [16]. Furthermore, the impact of anatomical accuracy in the models of the lumbar spine on the clear explanation of the pattern of load distribution and areas of stress concentration was emphasized [16]. These studies gave a lot of insights into the biomechanics of LBP and also formed a good basis for the accurate modelling of 3-D models and biomechanical simulations.

3 Methodology

The structured methodology of this paper is designed to achieve the set objectives of biomechanical understanding of LBP through modelling and simulation using the following steps: literature review, 3D modelling, finite element analysis, and data analysis to derive actionable insights.

3.1 Literature review strategy

A comprehensive literature review was conducted to understand the biomechanical factors contributing to LBP, current diagnostic and modeling techniques, and identify gaps in existing research. Focus areas included mechanical properties of lumbar vertebrae and intervertebral discs (IVDs), stress distribution patterns under load, and FEA methodologies applied to the spine. Key studies informed the selection of material properties (e.g., Young's modulus, Poisson's ratio) [16] and simulation approaches, particularly regarding the effects of axial and shear loads [16] and the importance of anatomical accuracy [9]. The review highlighted the need for comparative analysis under different, clinically relevant loading directions, forming the basis for this project's objectives.

3.2 Modelling

The main aim of this work is the creation of a very accurate and anatomically detailed three-dimensional model of the lumbar spine. The model focuses on the region of interest of the lumbar area with the vertebrae of L1, L2, L3, L4, and L5. Such a detailed model is particularly necessary for the improvement of our understanding of the mechanical features of the region as well as the region's reaction to different bodily and external factors. For the realization of this aim, Slicer 3D software was determined as the main tool for three-dimensional modeling. It is an open-source software application that makes the development of complex three-dimensional models of different human anatomical structures such as bones, organs, and tissues from CT or MRI images possible. Its versatility and friendliness make the software very suitable for scientific purposes that need detailed anatomical models.

The level of complexity inherent within this model is crucial for enabling accurate biomechanical simulations as well as the derivation of substantial insights. All the modeling steps were performed precisely to ensure a true representation of the complex geometry as well as the material features of the lumbar spine. The process required the three-dimensional segmentation of the anatomical region as illustrated by Figure A1 of Appendix A. Segmentation entails the careful application of color coding on the individual parts of the digital model for the purpose of explaining the anatomical composition of the model. Manual segmentation was applied to each millimeter of the region of the lumbar spine to achieve maximum anatomical accuracy.

Considering the poor quality of CT scans that are available via the web, this work used a sample CT scan obtained from the complete Slicer 3D database. The sample scan of the entire lumbar area was taken originally from a liver scan. It should be noted that no patient-identifiable data was included with this sample and thus there was compliance with ethics. After the segmentation procedure was performed on the sample, every vertebra of the entire lumbar region—the L1 through L5 region—was modeled separately for the purpose of increased accuracy. The individual models were then combined into a full anatomical model that included the intervertebral discs that are required for the biomechanical functionality of the spine.

The resulting comprehensive 3D model is not only anatomically accurate but also tailored for advanced biomechanical simulations using ANSYS software. This capability ensures that the simulations yield reliable results, making this model an invaluable tool for studies in biomechanics.

3.3 Simulations

All finite element simulations were performed using the ANSYS Workbench 2024 R2 platform, specifically employing the Static Structural analysis system within the ANSYS Mechanical environment. ANSYS is a widely recognised and extensively validated commercial FEA software suite appropriate for complex structural mechanics problems, including those in biomechanics. Its established use in numerous spinal biomechanics studies confirms its relevance and suitability for simulating the mechanical behaviour of the lumbar spine under various loading conditions. The Static Structural system within ANSYS Mechanical provides the necessary tools for defining material properties, meshing the geometry, applying loads and boundary conditions, and solving for outputs like stress, strain, and deformation, making it appropriate for the objectives of this study.

ANSYS will be used for stress and strain analysis, which will provide insights into how axial compression, bending, and twisting affect the spine. Loading conditions will be incorporated into the simulations to replicate real-life scenarios, such as bending and lifting. These simulations aim to identify how repetitive or high-intensity activities can contribute to the degeneration of the lumbar spine and the overall development of LBP. By simulating different movements and forces, the study will provide a deeper understanding of how mechanical stress interacts with anatomical structures to produce biomechanical risks.

3.4 Calendar plan

The following figure shows the calendar plan for the Capstone Project.

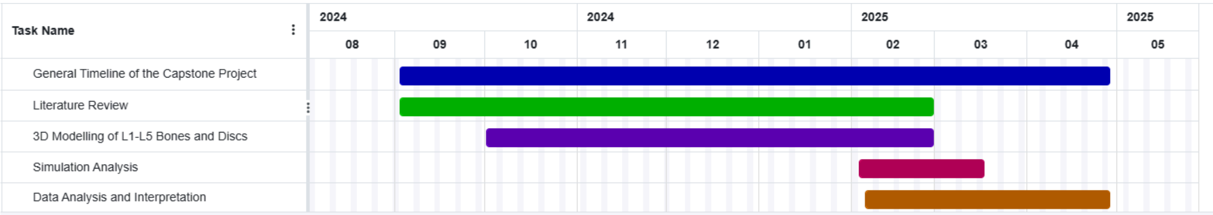


Figure 1. Calendar plan for the Capstone project.

As it can be seen from the calendar, the literature review took place from the start till the end of February, since the following capstone project needs extensive literature review on a lot of topics to:

- Shape the aim, problem statement, and find the essential materials regarding the problem of LBP;
- Find the essential references for 3D modelling of L1-L5 lumbar spine bones;
- Verify the material setup and force applied justifications for simulations.

3D Modelling also took most of the calendar plan due to its complexity and requirements for accuracy for models, starting from October till end of February. Simulations were set up simultaneously with the aim to reiterate the attempts with the goal for most accurate results. After simulation results were received, the data analysis and its interpretation were done, finalizing the Capstone Project.

3.5 Justification

The software and tools used in the following research paper were carefully chosen to ensure precision and accuracy, which are crucial for the project. Considering the fact that the 3D model should be as precise as possible since the simulations should be as close to real-life situations as possible, the 3D Slicer was chosen for its accuracy in anatomical modelling and versatility. The principle of work included scanning and working with real MRT scans of low back bones, so the resemblance is made as much as it is possible. The structure of the low back spine is not easy to replicate; it consists of a lot of curvatures, different surface types, and inconsistencies, thus the 3D Slicer, with the usage of scans, is a prime choice for replicating the L1-L5 discs as realistically as possible. Overall, it was chosen for its open-source nature, robust medical image segmentation tools, and ability to generate 3D models directly from DICOM data, ensuring anatomical fidelity based on the input scan.

For the simulations and biomechanical analysis that will follow, ANSYS software was chosen. Selected as an industry-standard, powerful FEA software with validated solvers for structural mechanics. Its capabilities in handling complex geometries, defining various material models, applying loads/constraints, and visualising results (stress, strain, deformation) are ideal for biomechanical simulations of the spine. The Static Structural system was appropriate for the project's focus on comparing responses under different static load directions. The choice for angles are not random, as 0° pure axial compression resembles the body weight of a person as a one scenario. The scenario with 20° side bending refers to side bending of the person or carrying some weight. The final scenario, 45°, resembles the forward bending of a person and carrying more weight. The 0° is self-explanatory, whilst 20° refers to the side bending as it affects the lumbar spine from the side, also resembling carrying some weight on one side (such as hanging the bag on one shoulder). The 45° forward bending, as it name says, refers to forward bending and also carrying some big weight, as carrying some weight needs bending forward, which this simulation shows.

4 Results

4.1 3D Model

Table 1. Comparison of the vertebral body anterior height of the model with existing research.

Vertebrae	Vertebral body height anterior (mm) [17]	Vertebral body height anterior (mm) of the model
L1	22.67-26.42	26.06
L2	23.36-28.13	28.96
L3	24.55-29.11	30.95
L4	24.73-28.98	31.63
L5	25.17-29.72	33.16

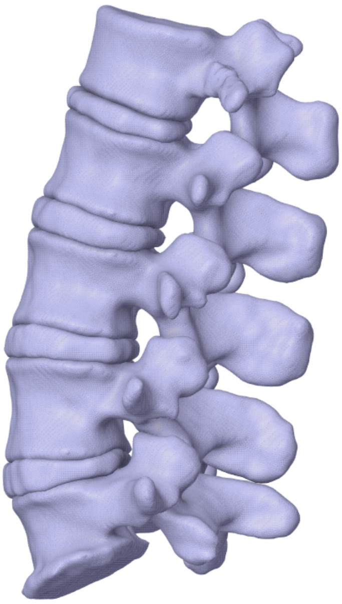


Figure 2. Assembled L1-L5 lumbar spine model used for FEA.

Table 1 depicts the comparison of anterior heights between the model and the values provided in the research by Bonczar et al. [17]. As can be seen from Table 1, the values of the anterior height of the model are very close to the real ones, so the model scale is 1:1. The L1-L5 vertebrae front views (with the anterior height dimensions) can be found in Appendix B, Figures B1-B5. The lumbar vertebrae in this project are made from the sample CT scan available in 3D Slicer. The final version of every model is precise to the real bone structure, as the model is taken from the CT scan. After all the models are done, they are assembled. The assembly includes L1, L2, L3, L4, L5 vertebrae and L1-L2, L2-L3, L3-L4, L4-L5 intervertebral discs, as can be seen from Figure 2.

4.2 Simulation setup details

Appropriate material properties were assigned to each distinct component of the 3D model to represent their mechanical behaviour. Based on common practices in spinal biomechanics modelling and data available in the literature, bony structures (cortical and cancellous bone) and cartilaginous endplates were typically modelled as linear elastic isotropic materials. The specific values for Young's modulus and Poisson's ratio for each material were assigned based on established literature values.

Table 2. Mechanical properties of 3D model [16].

Part	Young’s Modulus (MPa)	Poisson’s Ratio
Cortical bone	12000	0.3
Intervertebral disc	42.7	0.499

*(Note: IVDs were modelled as a single homogenous material for simplification)

Creating highly realistic spine models is challenging due to the intricate anatomy, the complex and often non-linear mechanical behaviour of biological tissues, consequently, simplifying assumptions are inevitably made in any FEA model. Therefore, results from FEA studies should always be interpreted within the context of the specific model developed, its inherent assumptions, and its limitations. Often, the greatest value of FEA in biomechanics lies not in predicting exact in vivo values but in conducting comparative analyses within a consistent model framework to understand the relative effects of different parameters, designs, or loading conditions.

As a next step, the geometric model was then divided into a finite number of elements and nodes through a process called meshing in ANSYS Mechanical. The model was meshed using tetrahedral elements, Moreover, a mesh sensitivity analysis was carefully considered to make sure that the results were not influenced by the element size. This helped us to aim for a better balance between computational accuracy and efficiency. A mesh sensitivity study was performed by comparing results (e.g., maximum von Mises stress under 1000N axial load) from meshes with average element sizes of approximately 2.5mm, 1.7mm, and 1.2mm. The results for maximum stress converged within 5% between the 1.7mm and 1.2mm meshes. Therefore, the 1.7mm average element size was chosen as a balance between accuracy and computational cost. The final mesh consisted of approximately *98538* elements and *474572* nodes.

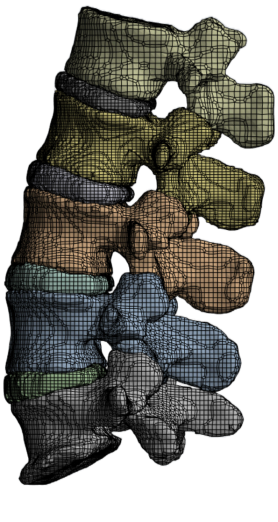
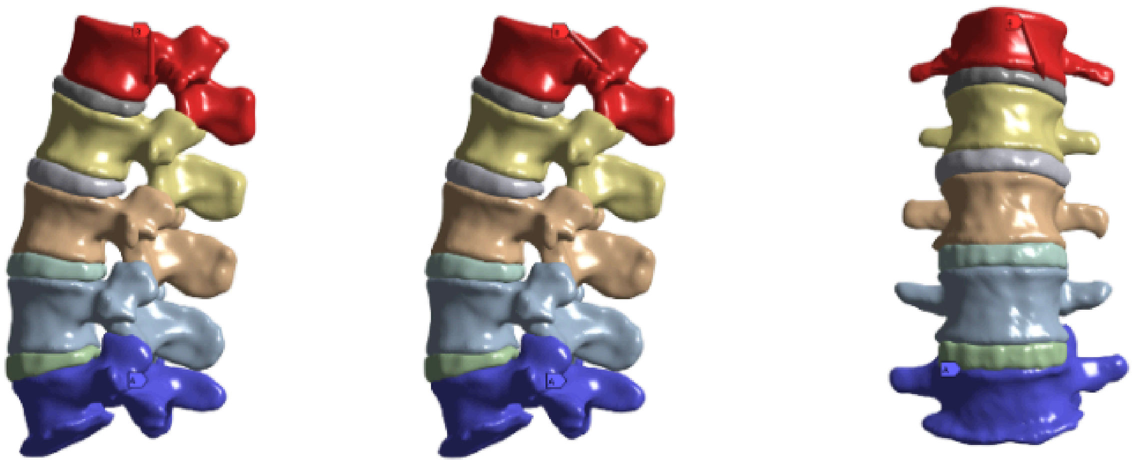


Figure 3. Image of the meshed finite element model

This figure shows the finite element mesh generated on the 3D geometric model. The discretisation of the vertebrae, intervertebral discs, and other components into a network of interconnected elements and nodes can be clearly seen.

It is important to note that to simulate the mechanical environment, appropriate boundary conditions (constraints) and loads were applied to the model. These conditions are consistent with standard practice in FEA studies of multi-segment lumbar spine models, the lowermost vertebral body, L5, in the model was selected as fixed support. As a result, three distinct static loading scenarios were simulated by applying force to the uppermost vertebral body, L1.



0 degrees | 45 degrees forward | 20 degrees sideways

Figure 4. Diagram illustrating the boundary condition (fixed L5 base) and the application of forces on L1 for the 0°, 45°, and 20° loading scenarios.

This schematic diagram illustrates the simulation setup for the L1-L5 model. It shows the fixed boundary condition applied to the inferior surface of the L5 vertebra. Arrows indicate the direction and point of application (superior surface of L1) of the force vector for each of the three simulated loading conditions: vertically downwards (0° Axial Compression), angled 45° downwards and forwards (Forward Bending), and angled 20° downwards and sideways (Side Bending).

The quantitative results from the simulations in ANSYS software are summarised in Table 3. The table includes maximum total deformation, maximum equivalent elastic strain, and maximum equivalent (von Mises) stress for the 3D model of L1-L5 bones within the range of 0 N to 1000 N with a step of 250 N. The data presented in the table is used to plot graphics and to compare the results between scenarios set up in ANSYS.

4.3 Simulation results

The quantitative results (maximum values within the L1-L5 model) for total deformation, equivalent elastic strain, and equivalent (von Mises) stress obtained from the ANSYS simulations for each load step and scenario are summarized in Table 3.

Table 3. Simulation Results

Angle (°)	Applied Force (N)	Total Deformation (mm)	Equivalent Elastic Strain (unitless)	Equivalent (von Mises) Stress (Pa)
0	0	0.00000	0.00000	0.00E+00
0	250	0.14929	0.030648	2.47E+07
0	500	0.21859	0.057296	4.44E+07
0	750	0.40788	0.085944	7.91E+07
0	1000	0.59718	0.12259	9.88E+07
45	0	0.00000	0.00000	0.00E+00
45	250	0.76426	0.17977	9.39E+07
45	500	1.32860	0.30956	1.80E+08
45	750	2.09290	0.50934	2.92E+08
45	1000	3.05710	0.71912	3.76E+08
20	0	0.00000	0.00000	0.00E+00

Table 3 (cont.)**.** Simulation Results

Angle (°)	**Applied Force (N)**	**Total Deformation (mm)**	**Equivalent Elastic Strain (unitless)**	**Equivalent (von Mises) Stress (Pa)**
20	250	0.22434	0.039366	3.23E+07
20	500	0.49868	0.098733	6.67E+07
20	750	0.67302	0.13810	9.30E+07
20	1000	0.89737	0.15747	1.29E+08

4.4 Graphical representation of results

The graphs from Table 3 were plotted in order to visualise the relationship between applied force and resulting biomechanical responses. Figure 5 shows the total deformation of all 3 distinct scenarios of the 3D model to the force applied from 0 N to 1000 N.

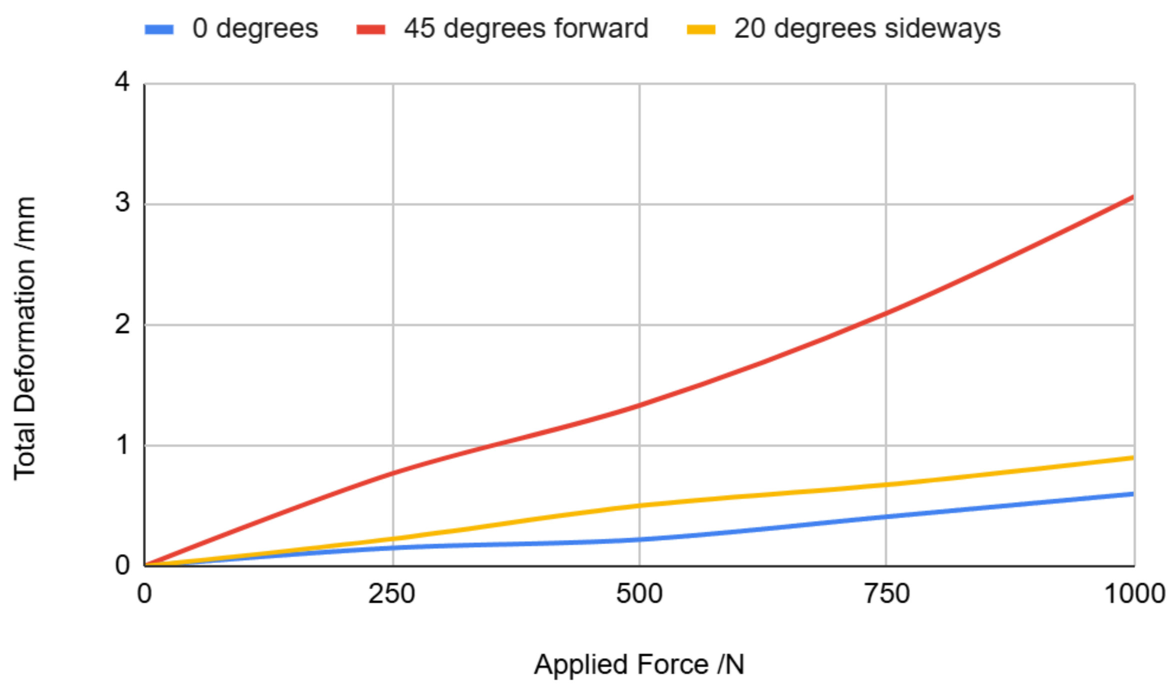


Figure 5. The relationship between total deformation and force applied for all scenarios

The figure above shows the distinct difference between total deformation for each scenario. The curve for 45° forward bending shows the steepest slope and the highest total deformation value, which indicates the greatest deformation under this scenario. The 20° side intermediate deformation between these 3 scenarios, while the 0° direct axial compression leads to the lowest deformation for the equal amount of force applied. The relationship appears to be approximately linear for 0° axial compression but shows increasing non-linearity, which can be seen in steeper slopes for higher forces, for both scenarios with bending, especially for 45° forward bending.

Figure 6 shows the relationship between the maximum equivalent elastic strain at the given force magnitude to applied force from 0 N to 1000 N.

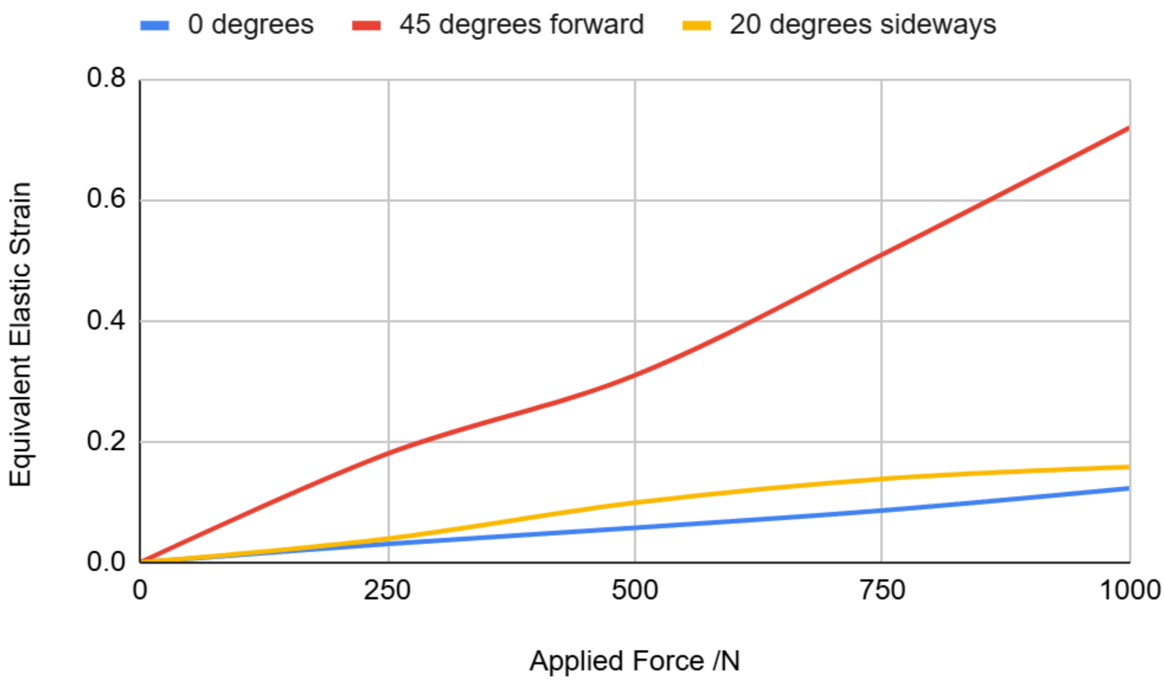


Figure 6. The relationship between equivalent elastic strain and force applied for all scenarios

This chart resembles the Figure 5 trend with increasing the equivalent elastic strain while applied force is increased gradually. The 45° forward bending shows the highest values for equivalent elastic strain followed by 20° side bending, last being the 0° axial compression. The slope of the curve for 45° is noticeably higher compared to other scenarios, especially for the forces beyond 500 N, which suggests the significant strain under compression-flexion load.

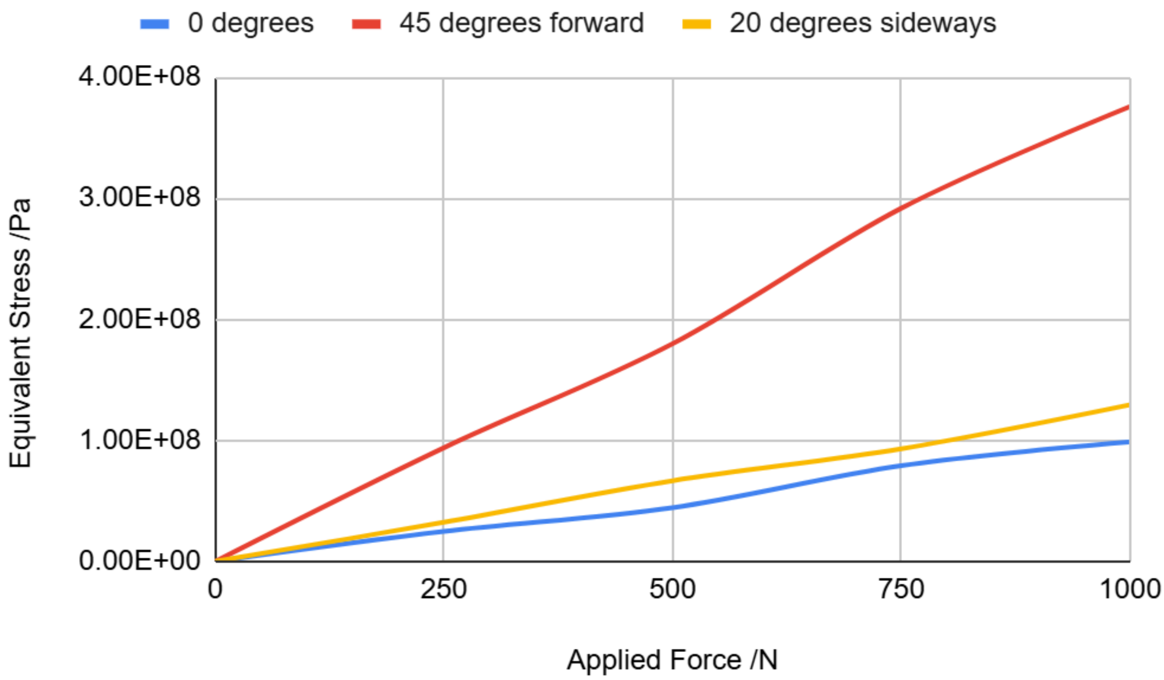


Figure 7. The relationship between equivalent (von Mises) stress and force applied for all scenarios

The trend is similar to both Figure 5 and Figure 6 for this simulation. The 45° forward bending simulation results are subsequently higher than the other two, which can be seen mostly on 750 N and 1000 N applied forces. Such stress increase can indicate a potentially critical stress concentration under forward bending scenario within this model.

5 Analysis and Discussion

5.1 Discussion

The results obtained from the simulations of L1-L5 spine bones provide a lot of insights on how the lumbar spine responds mechanically to different loading conditions.

Under direct axial compression at 0°, the force acts primarily on the longitudinal axis of the lumbar spine. This type of force can be compared to the daily routing, such as sitting straight or walking, and this type of loading is mainly resisted by vertebral bodies and discs. The structure of the lumbar spine is adapted for this function. The nucleus pulposus within the IVD pressurises and distributes the stress radially to annulus fibrosus and axially to the vertebral bones. The simulations in ANSYS showed that the above mentioned scenario has the lowest magnitudes of biomechanical measurements compared to other two scenarios with angle pressure at the same applied force (Table 3, Figures 5-7). This aligns with the lumbar spine's inherent capacity to withstand the compressive stress that mostly includes daily activities. The stress concentrations are mostly between the stiffer vertebrae and softer IVD, even potentially in the annulus fibrosus.

The forward bending scenario at 45° involves applying force at a significant angle in the sagittal plane. This force, compared to the direct axial compression, creates a complex loading state to the lumbar spine, leading to both substantial axial compression and large flexion-inducing moment. Discussing from a biomechanical point of view, flexion leads to anterior compression and posterior tension. This loading condition results in generating high compressive stresses and strains in the anterior parts of vertebral bodies and IVDs, consequently leading to high tensile stresses and strains in the posterior elements, including the posterior annulus, facet joint capsules, and posterior ligaments. The simulations in

ANSYS strongly reflect the above mentioned biomechanical response. The observed total deformation, equivalent strain, and equivalent stress are noticeably higher than direct axial loading and 20° side bending across all the applied forces (Table 3, Figures 5-7). The simulations are consistent and prove the point that forward bending imposes significant mechanical challenges on spinal structures and is often associated with mechanisms of disc injury and or LBP exacerbation.

In the side bending 20° scenario, the force was applied at an angle of 20° in the coronal plane, inducing both lateral bending moment and axial compression. The lateral bending occurring in this scenario leads to compression on the side where the spine bends (ipsilateral) and tension on the opposite side (contralateral). The results of this scenario showed the intermediate values between axial compression and 45° forward bending (Table 3, Figures 5-7). It suggests that while side bending is more influential to the lumbar spine than pure axial compression, the biomechanical response created is less severe than that in the 45° forward bending.

5.2 Comparative Analysis of Loading Conditions

The direct comparison of values of each scenario highlights the substantial differences in the biomechanical responses of a lumbar spine model. Analyzing the data at the maximum applied force of 1000 N provides a distinct illustration:

- **Total deformation:** The deformation in forward bending scenario (3.057 mm) was approximately 5.1 times greater than in pure axial compression (0.597 mm) and approximately 3.4 times greater than in side bending scenario (0.897 mm).
- **Equivalent Elastic Strain:** The equivalent elastic strain under forward bending (0.71912) was about 5.9 times higher than in pure axial compression scenario (0.12259) and roughly 4.6 times higher than in side bending case (0.15747).

- **Equivalent (von Mises) stress:** The equivalent stress also follows the same trend, leading to forward bending scenario (3.76E+08 Pa) being approximately 3.8 times greater than pure axial compression scenario (9.88E+07 Pa) and about 2.9 times higher than under side bending (1.29E+08 Pa).

These comparative analyses distinctly show the fact that the forward bending scenario imposes the most severe mechanical demands on the lumbar spine. Moreover, the graphs plotted on the base of these simulations (Figures 5-7) the difference in rate at which these mechanical responses increase with the applied force. Both curves for side bending and forward bending scenarios exhibit a more pronounced upward trend or non-linearity compared to the relatively linear trend in the pure axial compression case. It leads to the conclusion that sensitivity of the structure to load increases when bending force applied; each additional unit of force produces a progressively larger increment in total deformation, total equivalent strain, and total equivalent (von Mises) stress, which is distinctly seen in higher magnitudes of force. This non-linear trend underscores the potential for high tissue stresses and bone structure problems to develop while doing activities involving significant bending under load.

5.3 Novelty

The primary contribution of this study is the direct, quantitative comparison of the biomechanical responses (deformation, strain, stress) under these three specific loading conditions (0 degrees axial, 45 degrees forward angled force, 20 degrees side angled force) simulated within a single, consistent ANSYS FE model. While other studies analyse these loading modes separately or use different modelling approaches (e.g., pure moments), this work provides specific comparative data on how varying the angle of an applied force, creating combined compression and bending/shear, alters the mechanical environment in an L1-L5 segment model.

5.4 Limitations

It is important to note the limitations inherent to this study, which arise from the necessary simplifications made during the modelling and simulation process. Key limitations include:

- **Model Simplification:** The model represents the general structure of the lumbar spine, not including different pathologies or anatomical variations.
- **Material Assumptions:** Simplified material properties are assigned to the vertebrae and intervertebral discs, while the tissues have a much more complex structure and properties. In addition, the musculature around the spine, which enhances spine stability, was not included in this model. Both of these factors can lead to some imperfections or errors in the results.
- **Loading and Boundary Conditions:** The simulations assumed the static loads, even though most physiological activities are dynamic. Loads were applied as distributed forces on the L1 vertebrae, which is the closest to the force applied behavior yet simplifies how body weight and muscle forces act in vivo. The specific angles (0°, 20°, 45°) represent mostly discrete points as possible scenarios of lumbar spine movement and do not capture the full range of natural motion of low back bones. The fixed boundary condition at the base (L5) is idealization of the connection of L5 bone to the pelvis, which is a simplification made for the sake of simulations.
- **FEA Method:** Finite Element Analysis itself relies on numerical approximations and mathematical solutions, so results can be easily influenced by element quality and mesh density.

- **Scope:** This study is limited to three specific loading conditions to investigate, not including other scenarios like torsion or more complex combined movements. Moreover, this study does not include gender, age, nationality factors due to simplification and too wide scope for the investigation.

These limitations are mostly done for the simplicity of the simulation setup, as the setup for the scenarios needs complicated mathematical solutions with dynamic setups that are not available in ANSYS. Scope is limited for the sake of depicting the scenarios of interest, the age and other factors that can influence are not included in the research due to lack of information and dissimilarity of each bone for every individual. These limitations should be considered in the future work to obtain better and more accurate results.

5.5 Future Work

In our next steps, we plan to implement new loading simulations - for example, simulated walking or lifting simulations. This is done to ensure gathering the information about the true time-dependent behaviour of the lumbar spine under activities closer to reality. We also plan to use more major muscle groups along with the key ligaments in our finite-element assembly. Mainly, this is planned by modelling the erector spinae, multifidus, abdominal muscles, anterior longitudinal and ligamentum flavum - by that we can better interpret how active and passive tissues produce the necessary forces for reach loads and stabilisation functions.

Moreover, in order to personalise our predictions, we intend to build specific geometries based on CT or MRI scans that will consider a range of ages, sexes, and body mass indices of patients. Apparently, this will allow us to explore how different factors are affected by genetic/biological human factors. We'll also intend to simulate common pathologies - disc degeneration, facet arthropathy, or spinal stenosis - to understand more about how these risk conditions change the result of simulations. This requires more research to effectively collect the data about pathologies.

Finally, we will validate our computational findings against in vivo measurements such as motion-capture kinematics, intradiscal pressure recordings, and clinical outcome scores. By comparing simulated strains and stresses with experimental data, we can effectively review our material definitions and boundary conditions. Finally, this work will help us develop diagnostic preventive strategies as well as treatment guidelines that clinicians and ergonomists can apply in order to reduce the consequences of LBP.

6 Conclusion

To conclude, our study provides both clear and quantitative evidence of how posture position dramatically affects the mechanical loads on the lumbar spine. In the baseline case of our research, simulating the pure axial compression (meaning 0°) under a 1,000 N load, resulted in the deformation of by just 0.60 mm and generated a maximum elastic strain of 0.12 as well as reaching a von Mises stress of 99 MPa. Further in simulations, introducing a 20° side bend increased the deformation to 0.90 mm (which is 1.5 times axial), strain to 0.16 (1.3 times), and stress to 129 MPa (1.3 times). More importantly, adding more bend with a value of 45° as a final step caused deformation of 3.06 mm, which is over 5 times the axial case, whereas the strain rose to 0.72 (which is 6 times) and stress to 376 MPa (approximately 3.8 times).

Essentially, these numbers can be interpreted such that even the slightest shifts in load angle result in boosting the internal bone and disc forces by 80 %, which in turn can be interpreted as a significant contributing factor for injury or degeneration. By elaborating on these quantitative findings further, along with our original findings, where forward bending was found to be the most harming posture, we recommend both the precisely good and the practically feasible conclusion: even the small postural adjustments or implementation of lumbar support can cut peak spinal loads dramatically.

Overall, our approach confirms the critical role of ergonomic awareness (proper lifting techniques, supportive seating, task-specific back braces and so on) and also points toward a multidimensional strategy. We hope that combining our biomechanical insights with targeted workplace protocols and clinical guidance will result in a highly effective pathway that will definitely play a role in reducing the global burden of low back pain.

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Appendix A. Liver CT scan

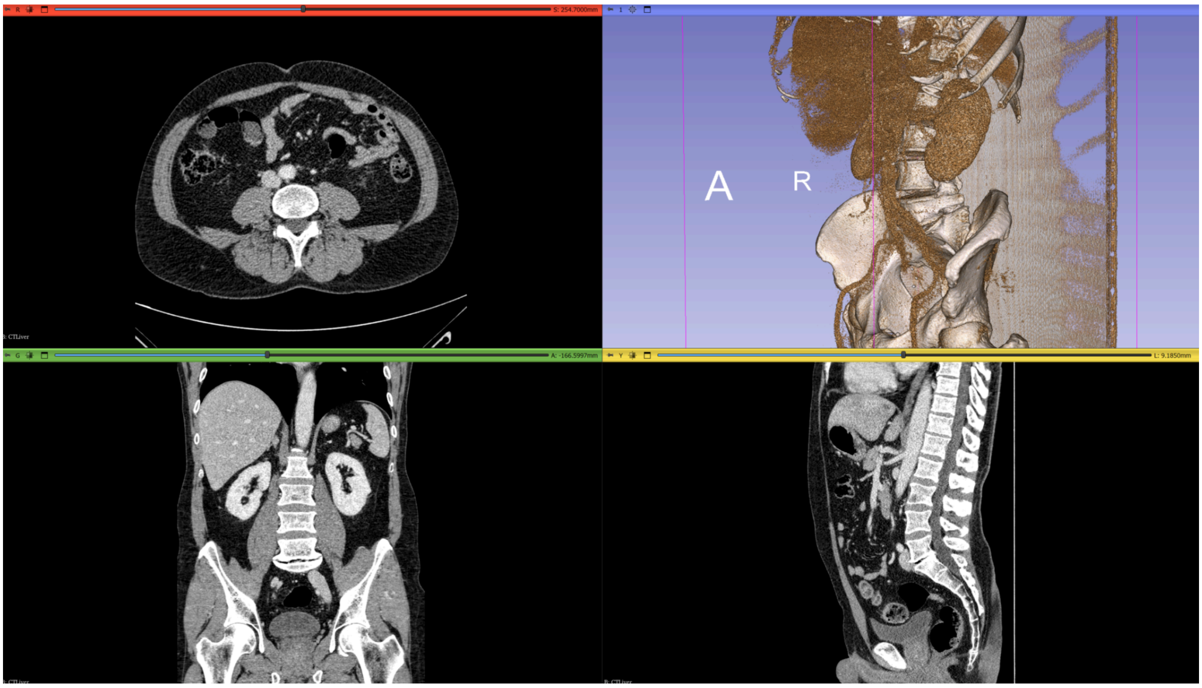

Figure A1. Liver CT scan in Slicer 3D

Appendix B. Vertebra Models

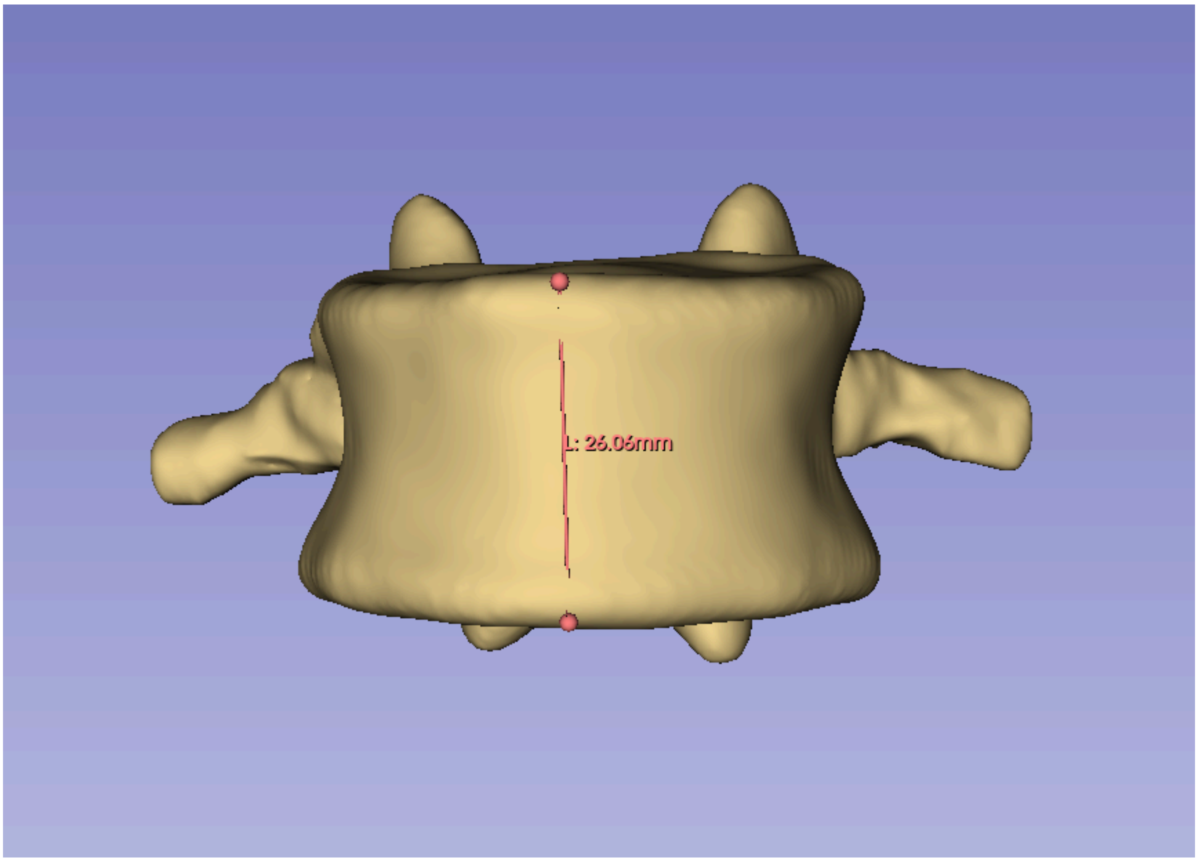

Figure B1. L1 Vertebra 3D Model

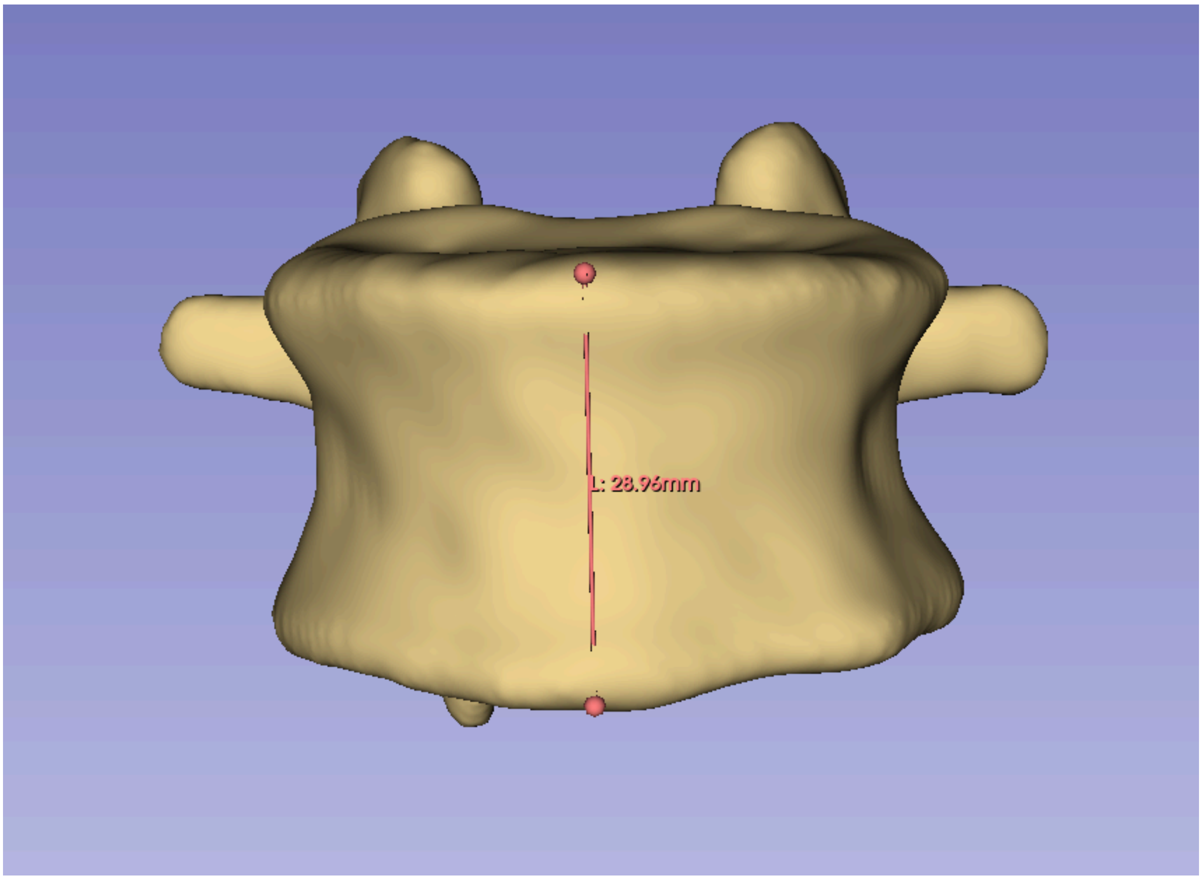

Figure B2. L2 Vertebra 3D Model

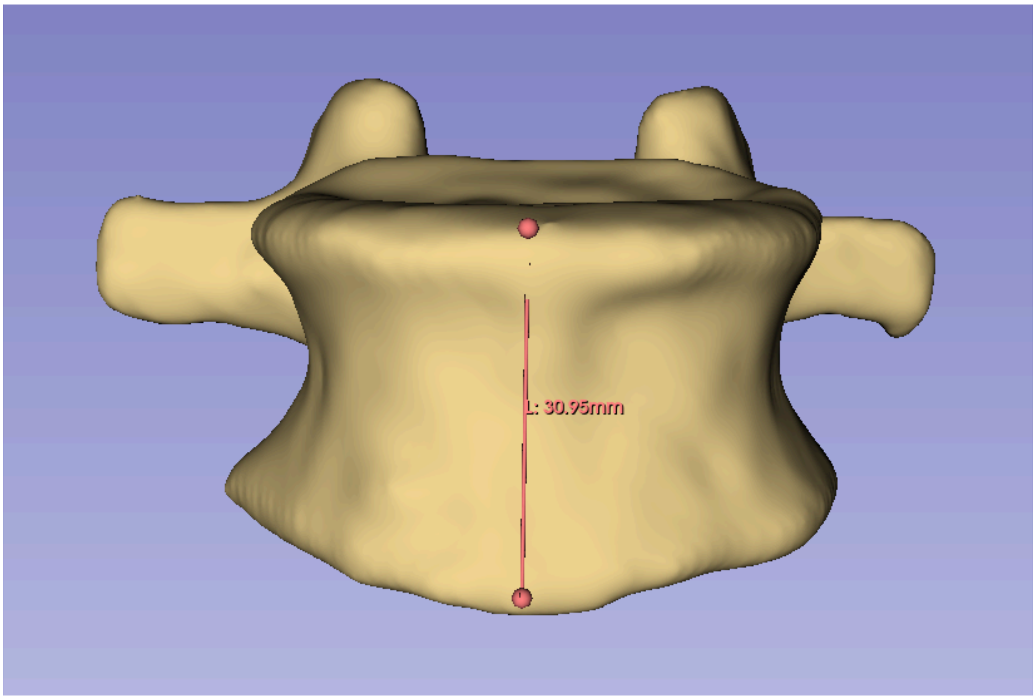

Figure B3. L3 Vertebra 3D Model

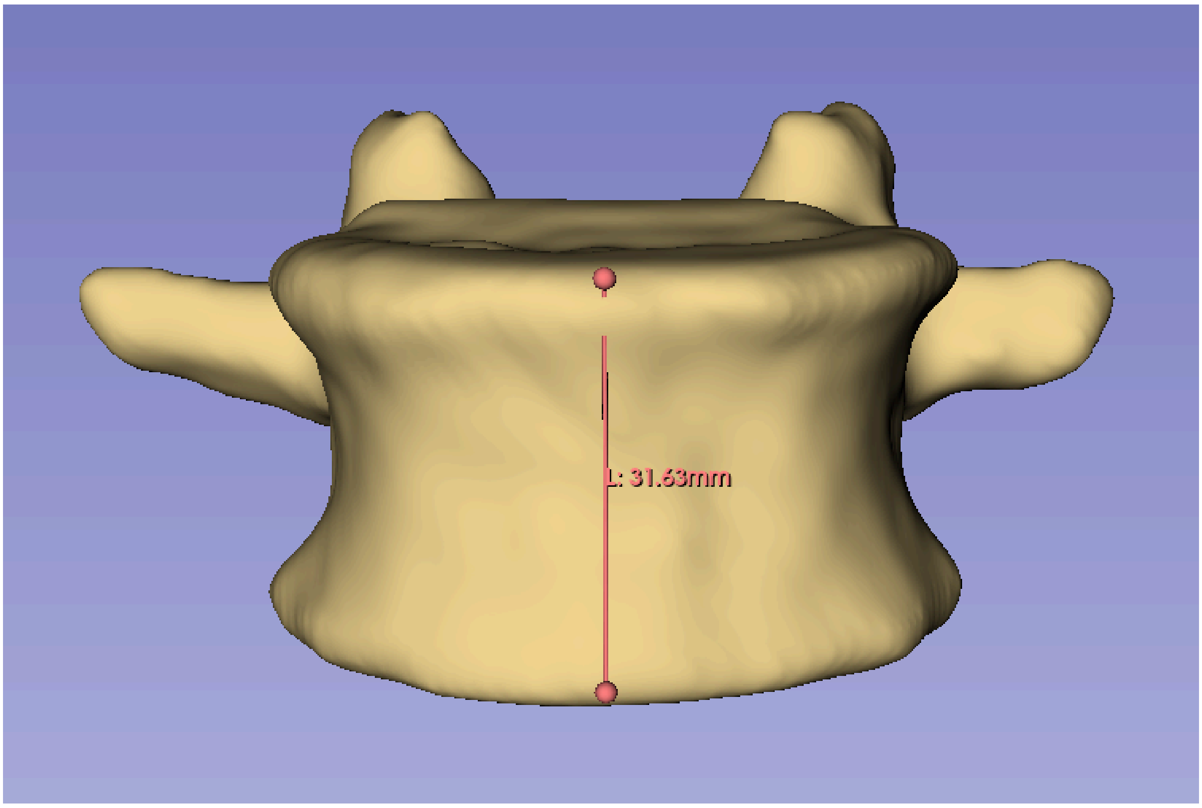

Figure B4. L4 Vertebra 3D Model

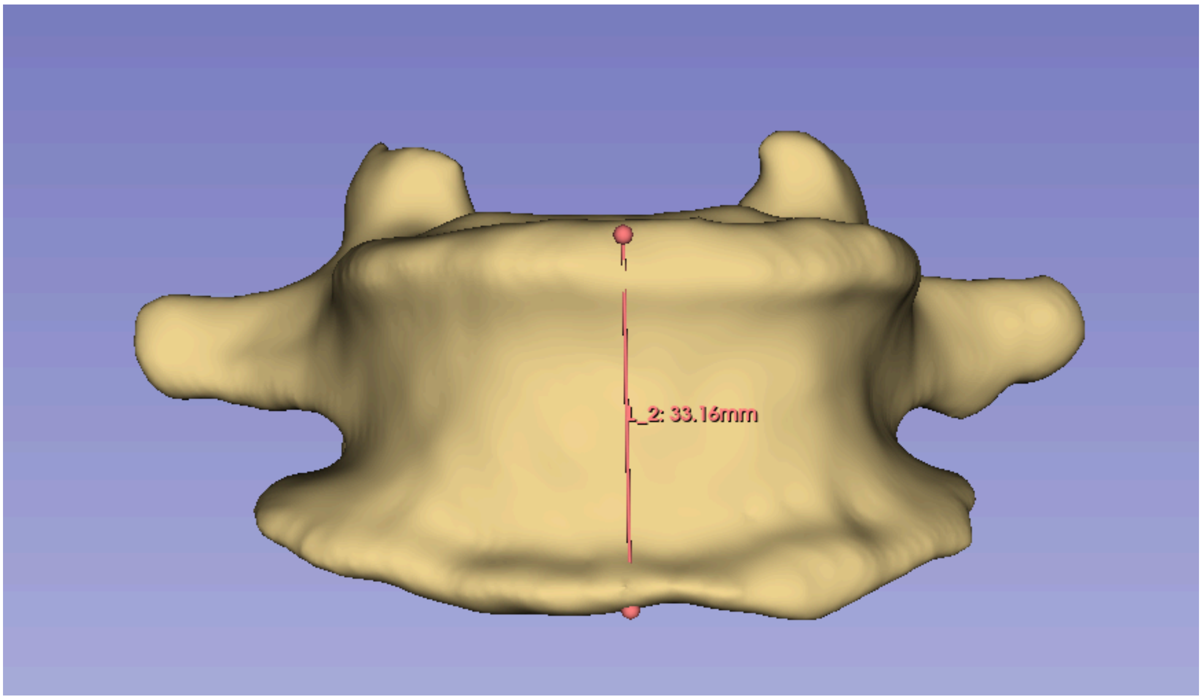

Figure B5. L5 Vertebra 3D Model

Appendix C. Simulation setup and results

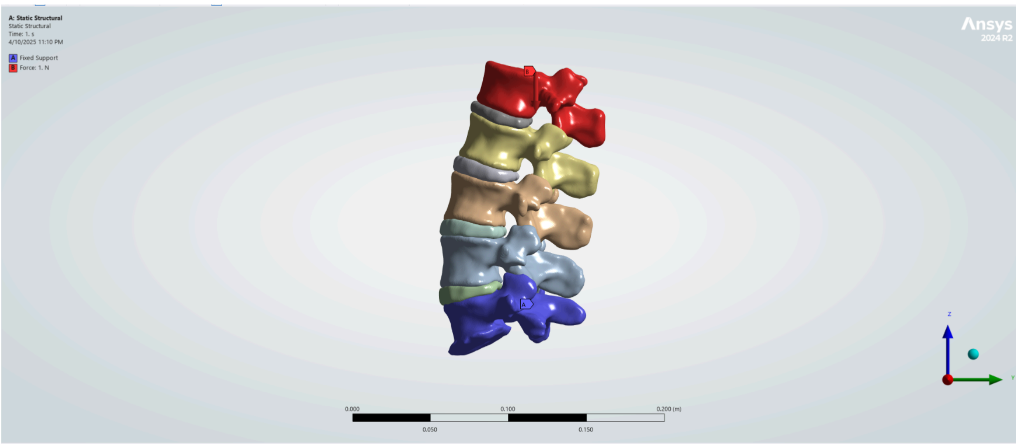

Figure C1. Model 0 degrees simulation setup

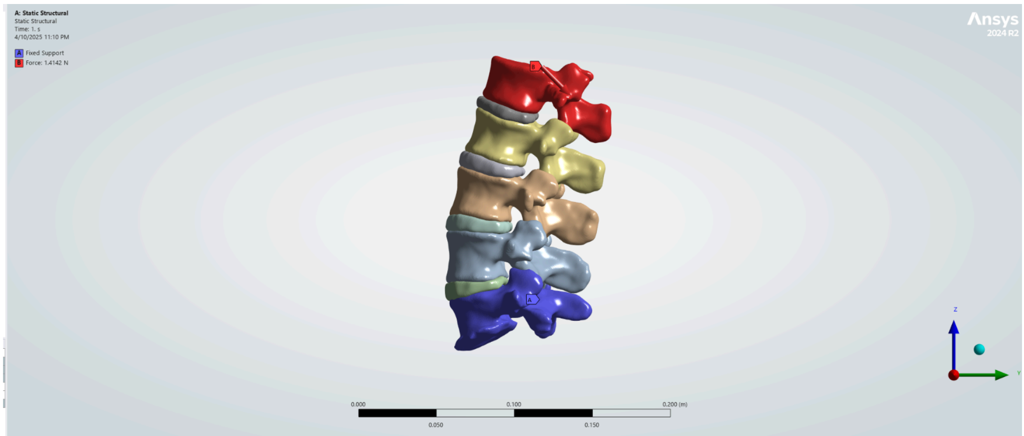

Figure C2. 45 degrees forward simulation setup

Figure C3. 20 degrees sideways simulation setup

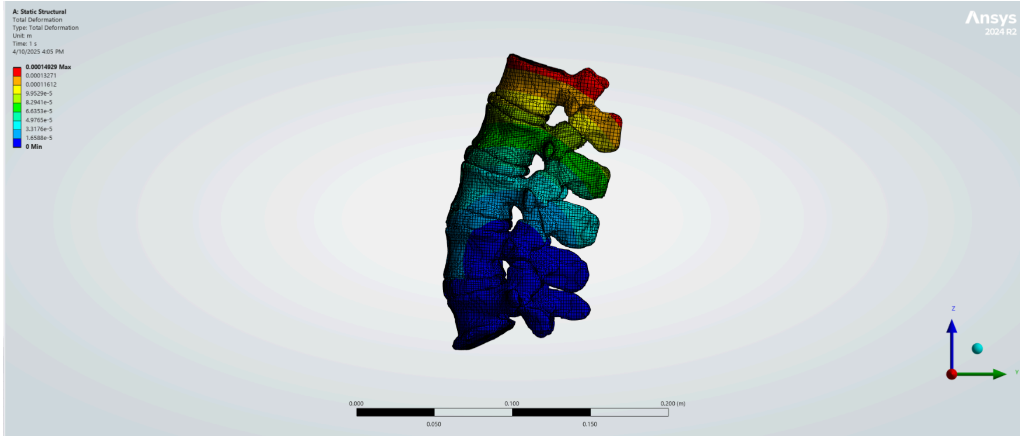

Figure C4. Total deformation for 250N, 0 degrees

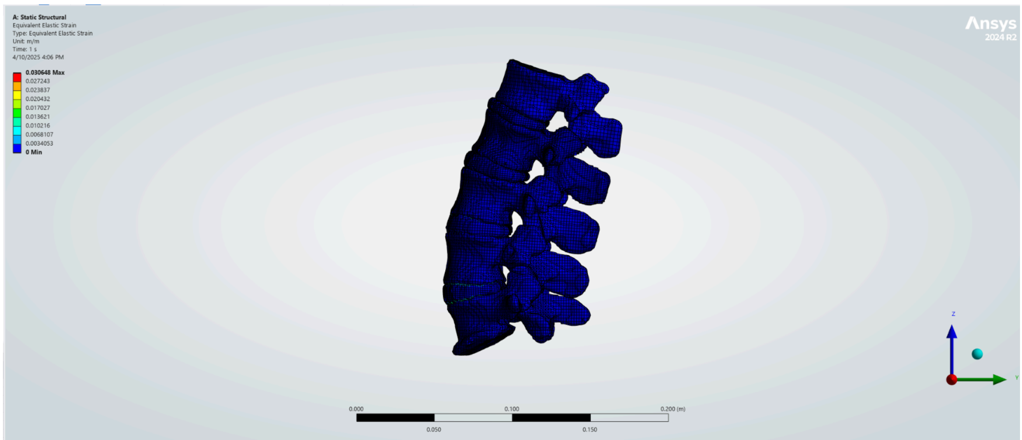

Figure C5. Equivalent elastic strain for 250N, 0 degrees

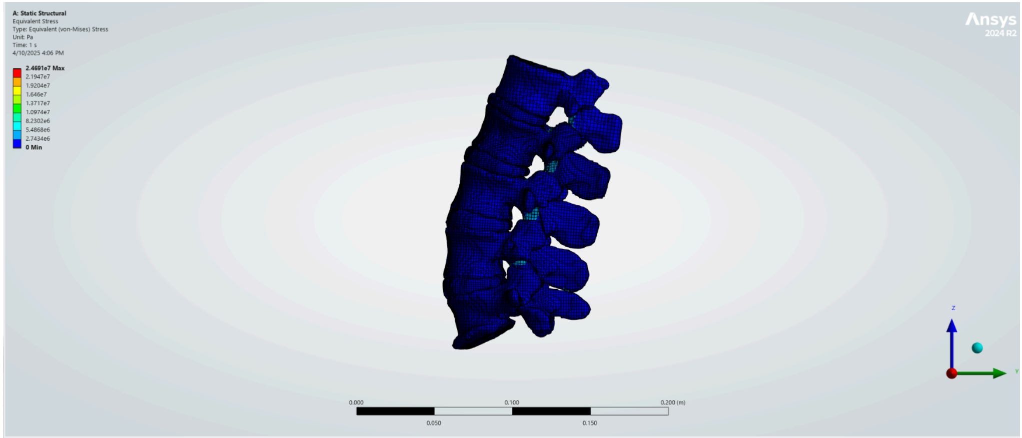

Figure C6. Equivalent stress for 250N, 0 degrees

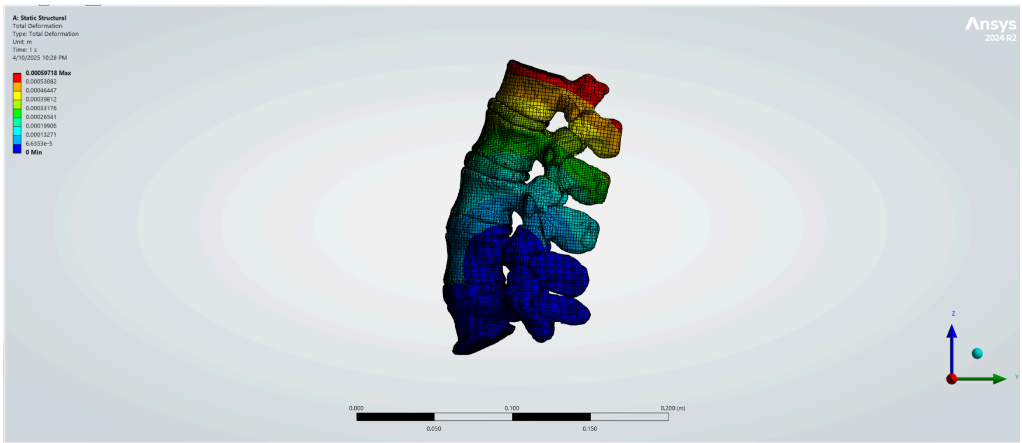

Figure C7. Total deformation for 1000N, 0 degrees

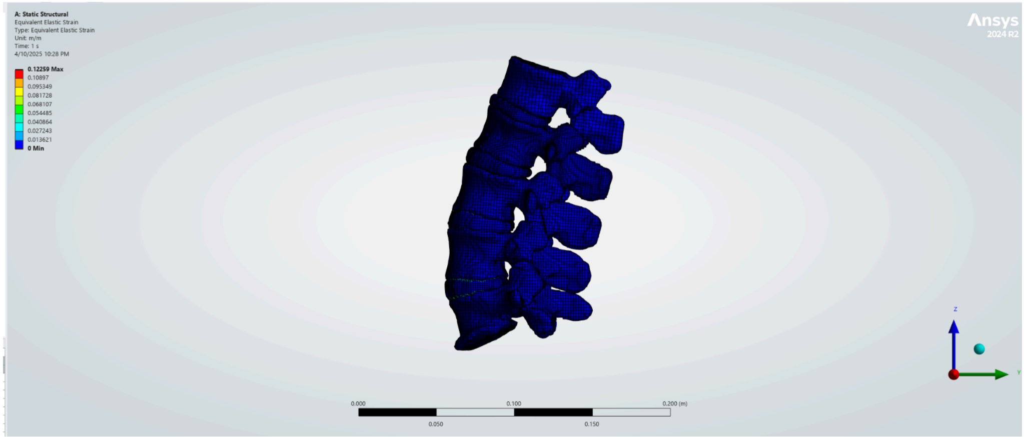

Figure C8. Equivalent elastic strain for 1000N, 0 degrees

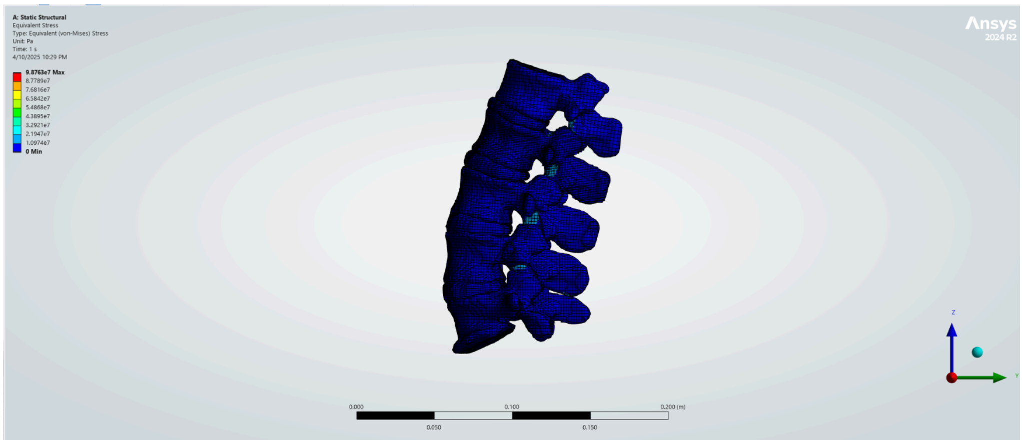

Figure C9. Equivalent stress for 1000N, 0 degrees

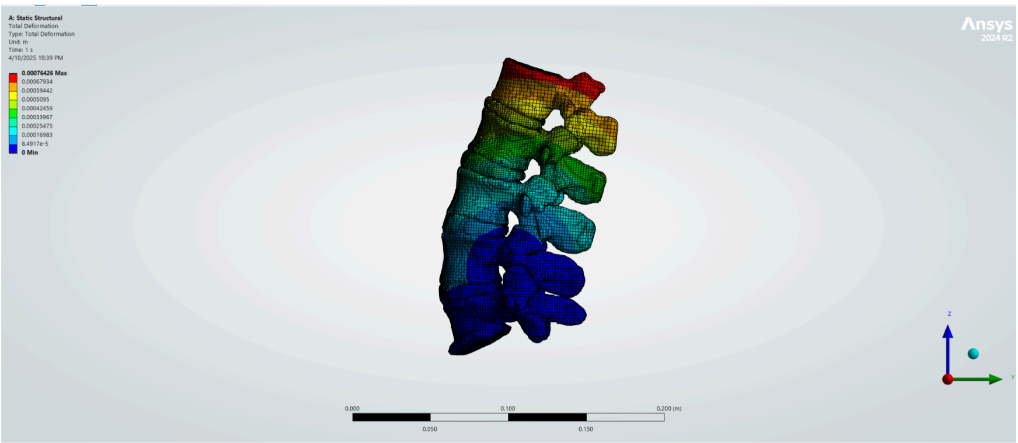

Figure C10. Total deformation for 250N, 45 degrees forward

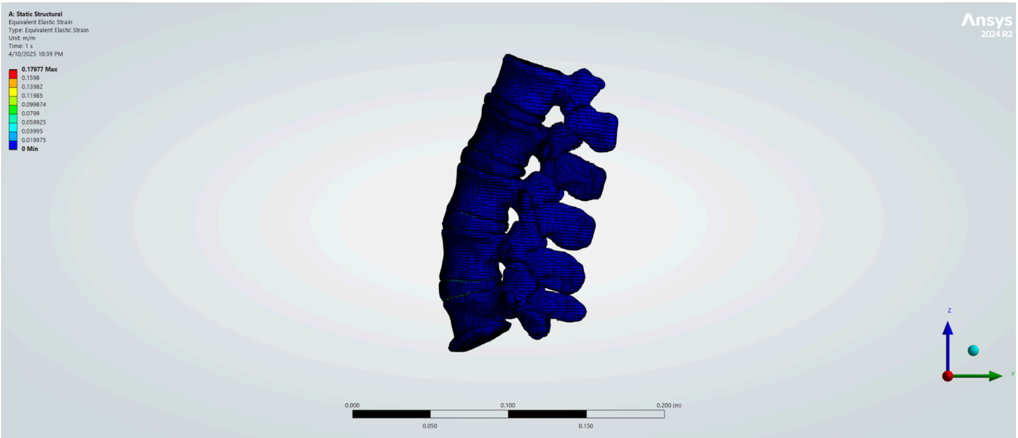

Figure C11. Equivalent elastic strain for 250N, 45 degrees forward

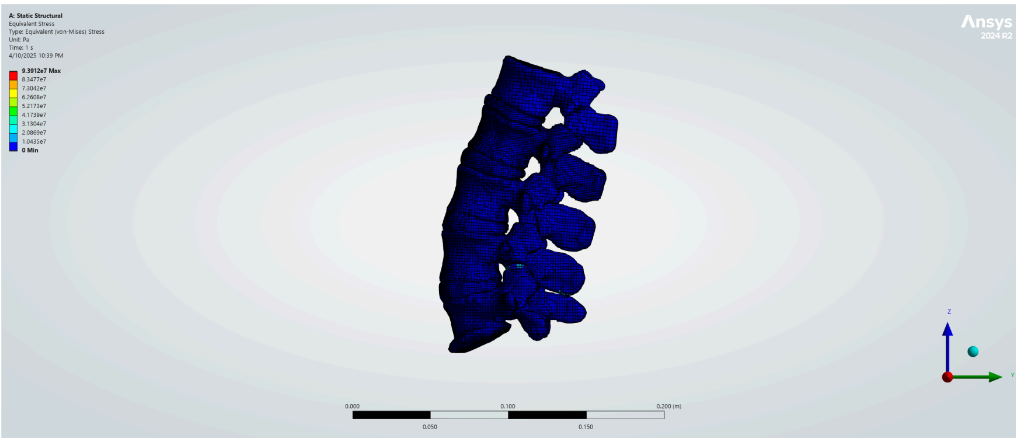

Figure C12. Equivalent stress for 250N, 45 degrees forward

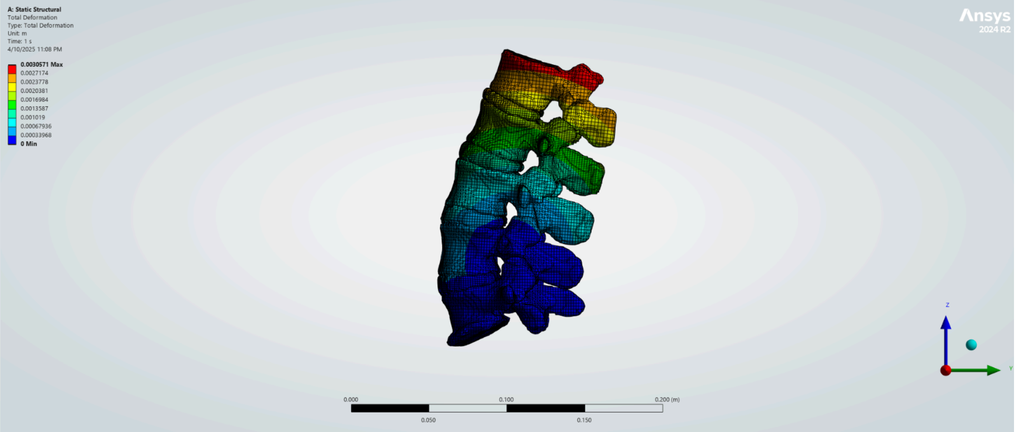

Figure C13. Total deformation for 1000N, 45 degrees forward

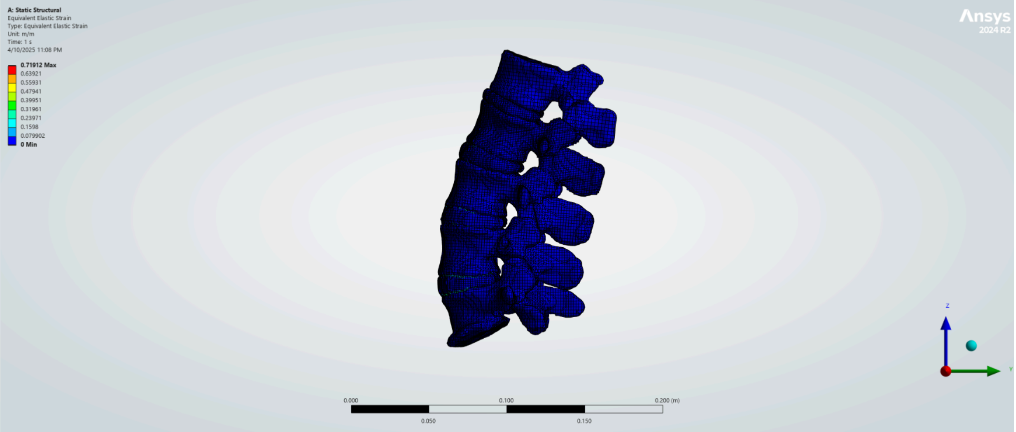

Figure C14. Equivalent elastic strain for 1000N, 45 degrees forward

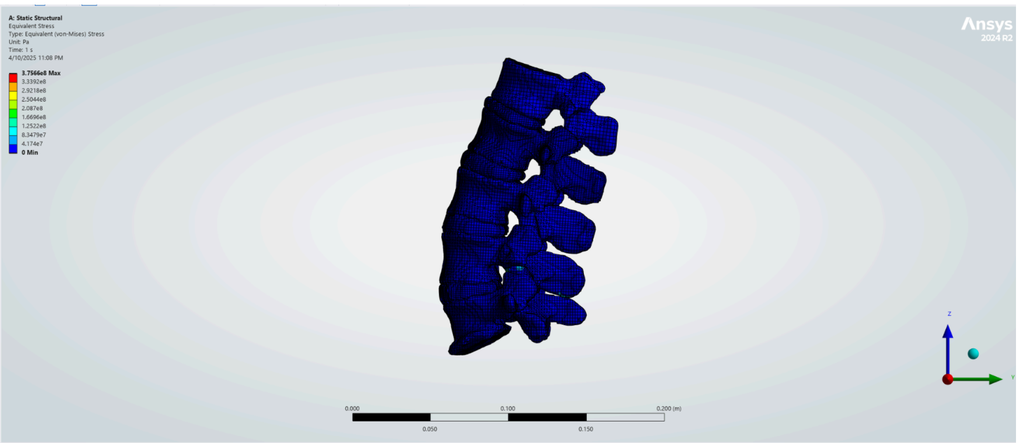

Figure C15. Equivalent stress for 1000N, 45 degrees forward

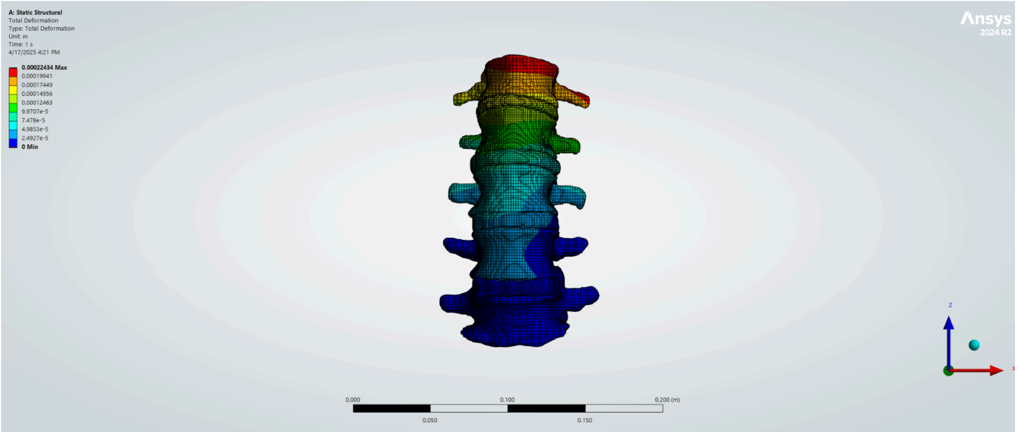

Figure C16. Total deformation for 250N, 20 degrees sideways

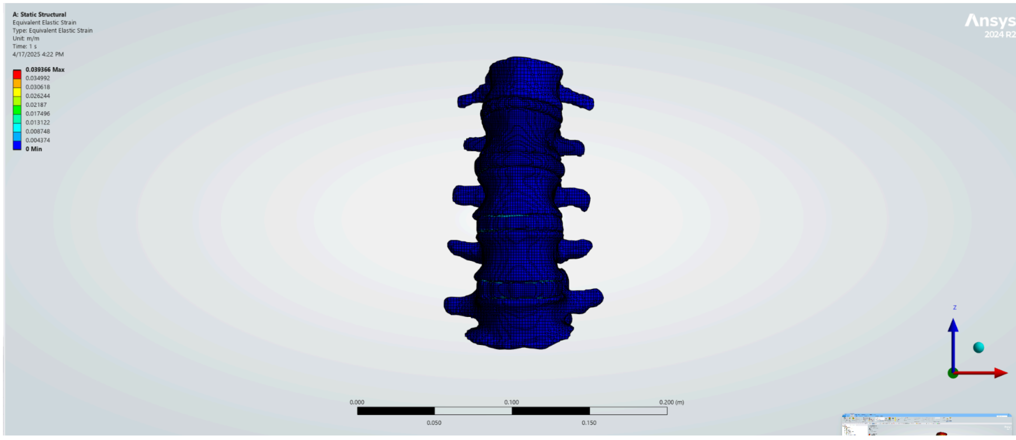

Figure C17. Equivalent elastic strain for 250N, 20 degrees sideways

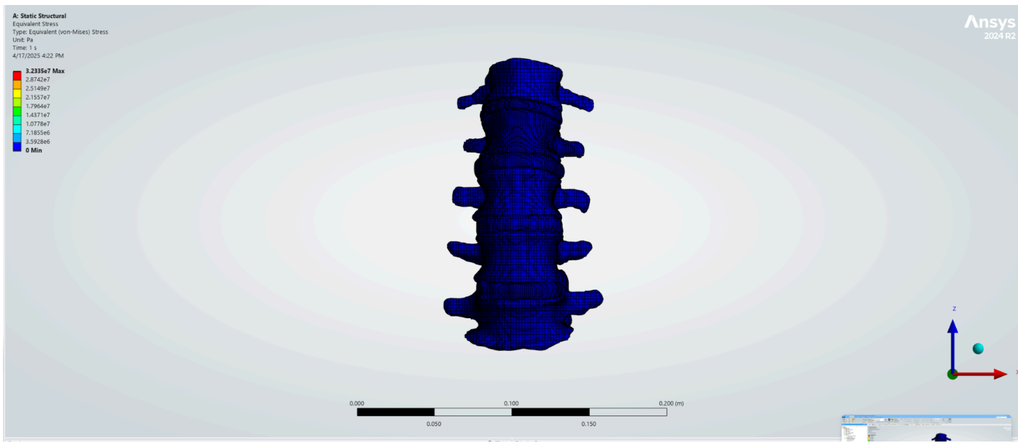

Figure C18. Equivalent stress for 250N, 20 degrees sideways

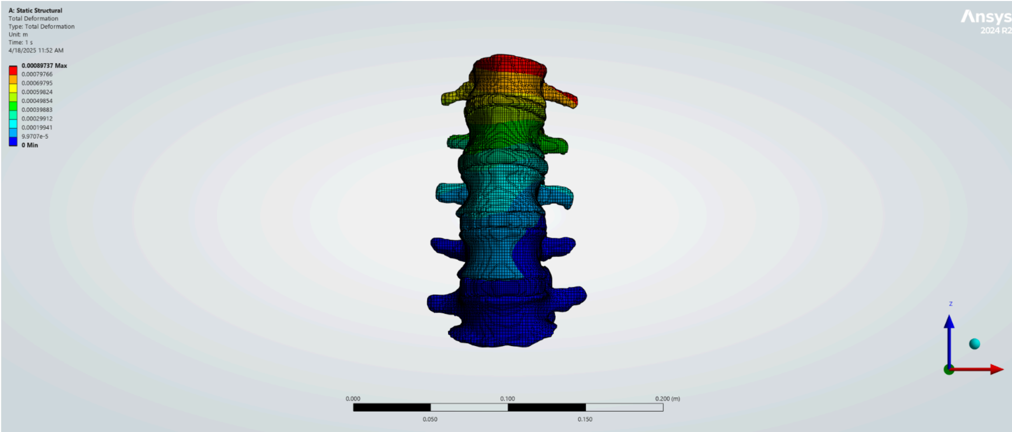

Figure C19. Total deformation for 1000N, 20 degrees sideways

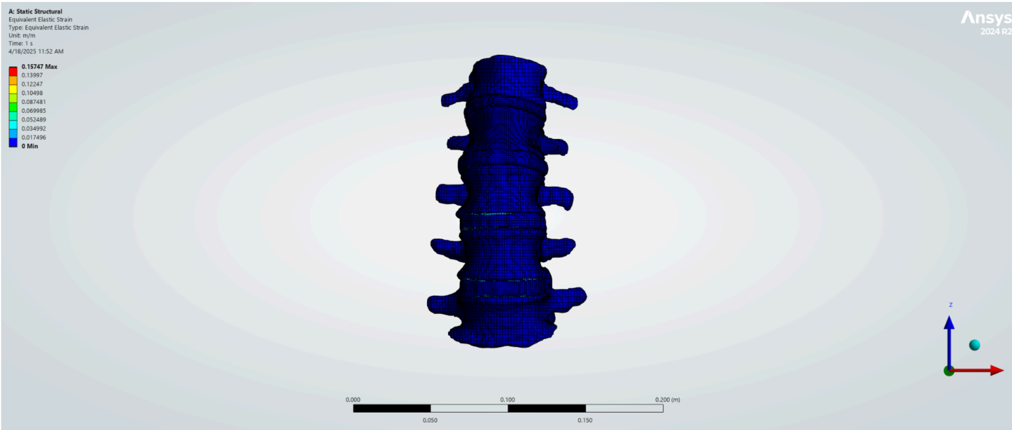

Figure C20. Equivalent elastic strain for 1000N, 20 degrees sideways

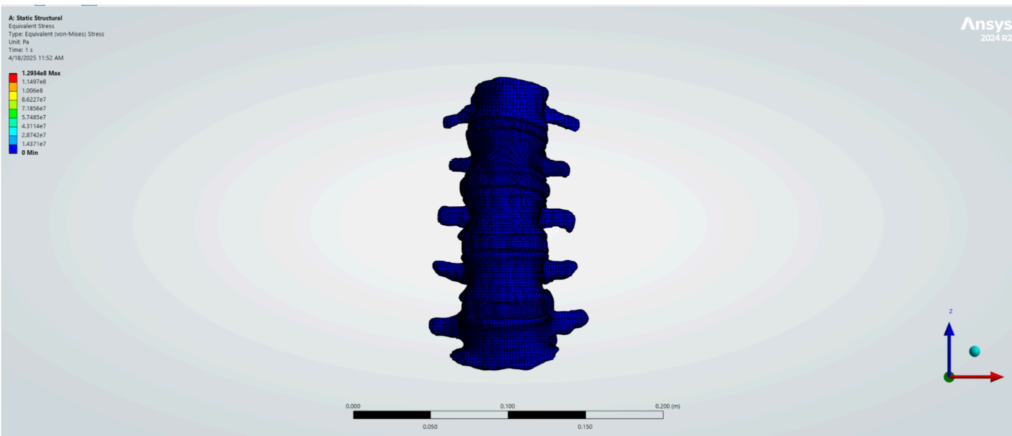

Figure C21. Equivalent stress for 1000N, 20 degrees sideways