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Unravelling the Biomechanics of Spondylolysis: A Computational Approach to Optimising Conservative Treatment

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Abstract

Spondylolysis is a fracture in the pars interarticularis of the lumbar vertebrae and is one of the most common injuries in adolescent athletes. Early intervention and conservative treatment are a favourable approach to surgical techniques in preserving patient's quality of life. However, currently there is uncertainty regarding the timing of sports resumption, rehabilitation duration, and recommended rest periods, highlighting the need for clearer treatment guidelines. To address these uncertainties, biomechanical analysis of a finite element model has been completed, for the phases of spondylolysis fracture recovery, with the aim of improving our understanding of rehabilitation strategies and enhancing clinical outcomes.

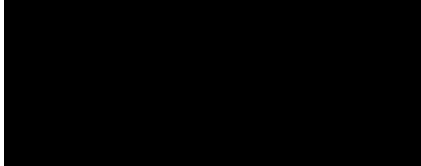
Two patient-specific three-dimensional finite element L4-S1 spine models were generated for intact and spondylolytic presentation, progressing through the stages of fracture healing from granulation to mature bone. The model included intervertebral disc geometries, facet joint models and ligament models. The model was analysed under activity-specific loading conditions of combined compressive preloads and moments of 6 degrees of freedom. A series of convergence tests, sensitivity analyses and validation tests were conducted to analyse the intact models.

Results showed that in the early healing stage of a pars fracture, stress increased to 26.94, 28.88 and 23.8MPa during flexion, ipsilateral lateral bending, and contralateral axial rotation, particularly in the cortical bone of the posterior lamina, right pedicle, and contralateral pars. As healing progresses, stress distribution becomes more balanced across both sides, with increased stress in the lysis region and decreased maximum cortical stress across the vertebra. Range of motion was found to be an ideal proxy metric for spondylolysis healing. This study highlights the importance of fracture specific rehabilitation in spondylolysis, particularly during the granulation phase. These findings emphasise the need for clearer, evidence-based rehabilitation guidelines and further research to refine treatment strategies and enhance clinical outcomes for young athletes.

Declarations

This work has not previously been accepted in substance for any degree and is not being concurrently submitted in candidature for any degree.

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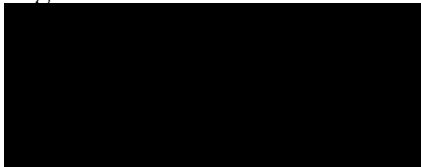


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This thesis is the result of my own investigations, except where otherwise stated. Other sources are acknowledged by footnotes giving explicit references. A bibliography is appended.

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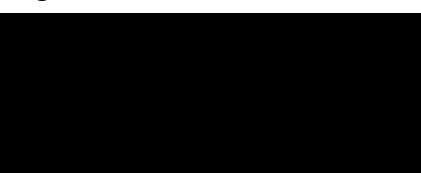


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The University's ethical procedures have been followed and, where appropriate, that ethical approval has been granted.

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List of Symbols and Abbreviations

Symbols	
N_{11}	The normal direction of the superior vertebrae's inferior face
N_{22}	The normal direction of the inferior vertebrae's superior face
$\mathbf{n}$	Average of the normal vectors
$\hat{\mathbf{n}}$	Normalised average of the normal vectors
h_{disc}	Height of the L4/L5 or L5/S1 intervertebral discs
m	Gradient of the intervertebral disc
f	Offset factor for intervertebral disc creation
P_{11}	Control point of the superior vertebrae's inferior face
P_{22}	Control point of the inferior vertebrae's superior face
p	Control points of the Bézier curve
$B(u)$	Bézier curve function
n_s	Number of Bezier curve steps
u	Curve progression parameter
P_s	Point spacing/ distance between mesh nodes
i, j	Index of the formulation
E	Young's modulus
A	Ligament cross-sectional area
L	Ligament Length per group
k_{total}	Total spring constant for a ligament group
$k_{individual}$	Spring constant for each individual spring
$n_{parallel}$	Number of parallel springs in the ligament
n_{series}	Number of springs in series in the ligament
r_{max}	Maximum node radial distance

$\Delta x, \Delta y, \Delta z$	Nodal moment arms
M_{total}	Total prescribed moment
M_{ij}	Moments for each specific node
F	Total force to achieve a total moment
σ_v	von Mises stress
$\sigma_{1,2,3}$	Three-dimensional principal stress
σ_{vw}	Volume weighted mean von Mises stress
V_{elem}	Volume of each element
V_{total}	Total volume of a spinal region
t_i	Time points
v_1, v_2	Vertebrae vectors
$\theta_{initial, final}$	Initial and final angles between the position vectors
Ex, Ey, Ez	Directional Young's Moduli
Gxy, Gyz, Gxz	Planar shear modulus
ν	Poisson's Ratio
$\nu_{xy}, \nu_{yz}, \nu_{xz}$	Direction dependent Poisson's Ratio
$c1, c2$	Mooney-Rivlin material parameters
Abbreviations	
AF	Annulus Fibrosus
ALL	Anterior Longitudinal Ligament
AR (LAR, RAR)	(Left or Right) Axial rotation
CT	Computed Tomography
DoF	Degrees of Freedom
EP	Endplate
FCL	Facet Capsular Ligament
FE	Finite Element
Flex-Ext	Flexion-Extension
FSU	Functional Spinal Unit
GP	Growth Plate
ISL	Interspinous Ligaments
ITL	Intertransverse Ligaments

IVBD	Inferior vertebral body depth
IVBW	Inferior vertebral body width
IVD	Intervertebral Discs
LB (LLB, RLB)	(Left or Right) Lateral Bending
LF	Ligamentum Flavum
LPHC	Lumbopelvic Hip Complex
LSO	Lumbo-Sacral Orthosis
Lysis	Spondylolysis
MR	Mooney-Rivlin
MRI	Magnetic Resonance Imaging
NH	Neo-Hookean
NP	Nucleus Pulposus
NR	Not recorded
NZ	Neutral Zone
P1	Patient 1
P2	Patient 2
PI	Pars interarticularis
PLL	Posterior Longitudinal Ligament
RoM	Range of Motion
SL	Supraspinous Ligament
STL	Stereolithography (File Type)
SVBD	Superior vertebral body depth
SVBW	Superior vertebral body width
TLSO	Thoracolumbosacral Orthosis
USS	Ultimate Shear Strength

Chapter 1: Introduction

1.1 Introduction to the study

Lumbar vertebral posterior arch fractures are one of the most common injuries in athletes with a prevalence of 47%(1) in adolescent athletes. These fractures, known as spondylolysis, pediculolysis or laminolysis, are believed to be fatigue fractures of the anatomically weak structures of the neural arch such as the pars interarticularis (PI) or pedicles. The cause of these fractures is linked to extreme spinal movements, especially during activities that combine rotation and hyperextension. To maintain stability and prevent degeneration of the spine and its surrounding structures, vertebral fractures are often treated surgically. These fractures, also referred to as lysis, are currently managed with fixation or fusion surgery, involving the implementation of metallic implants. However, this approach may not always be favourable for all patients, with cases of persistent lower back pain and adjacent segment degeneration. If spondylolysis is diagnosed in its early to progressive stages, alternative conservative treatment, involving rest and rehabilitation exercise or aids can prevent the need for surgical intervention. However, this process has prolonged recovery time and currently has a lack of biomechanical evidence for the duration or progression of rehabilitation and return to activities. Given the challenges in current conservative treatment approaches and the need for patient specific understanding of the biomechanics involved in both the fracture and the healing, there is a clear need for research in this area. Finite element (FE) modelling is an ideal method for precise simulation of spinal and spondylolysis biomechanics. FE analysis can produce accurate replications of patient-specific spinal structures and fractures, providing insights into specific mechanical responses of spondylolysis under various loading conditions. This approach could be used as a tool to assess fracture progression and the effectiveness of treatments by offering a detailed understanding of lumbar spine fractures and healing.

This thesis aims to develop and analyse multiple finite element computational models constructed from patient imaging of the lumbar spine, focusing on the spondylolysis region (L4-S1). In addition to the model development, the primary aim is to create a flexible and user-friendly framework, that analyses the spine for a range of activities, healing stages,

material properties, and other biomechanical factors that influence fracture recovery. With the objective to provide a biomechanical evidence-based approach to conservative treatment methods for spinal rehabilitation. As well as determining the relationship between resultant spinal range of motion and the magnitude and location of stress exerted on specific structures, to provide parameters to aid treating fractures and preventing the onset of lysis.

To achieve these objectives, this research project involved the development of a pipeline for quantifying conservative treatment, as well as an analysis of its quality and suitability. The first section of this pipeline included the segmentation of vertebrae from patient-specific CT imaging. The second section then focused on the construction of finite element L4-S1 spinal models, including structures such as vertebrae, intervertebral discs, ligaments, facet joints and fracture presentation. The third and final step involved a biomechanical analysis of the models under a range of loading conditions across all six degrees of freedom, with these loads modelled to represent activities of increasing intensity, typically recommended in physical therapy rehabilitation programmes. This pipeline was demonstrated for two patient specific cases, a and b, and illustrated in Figure 1.

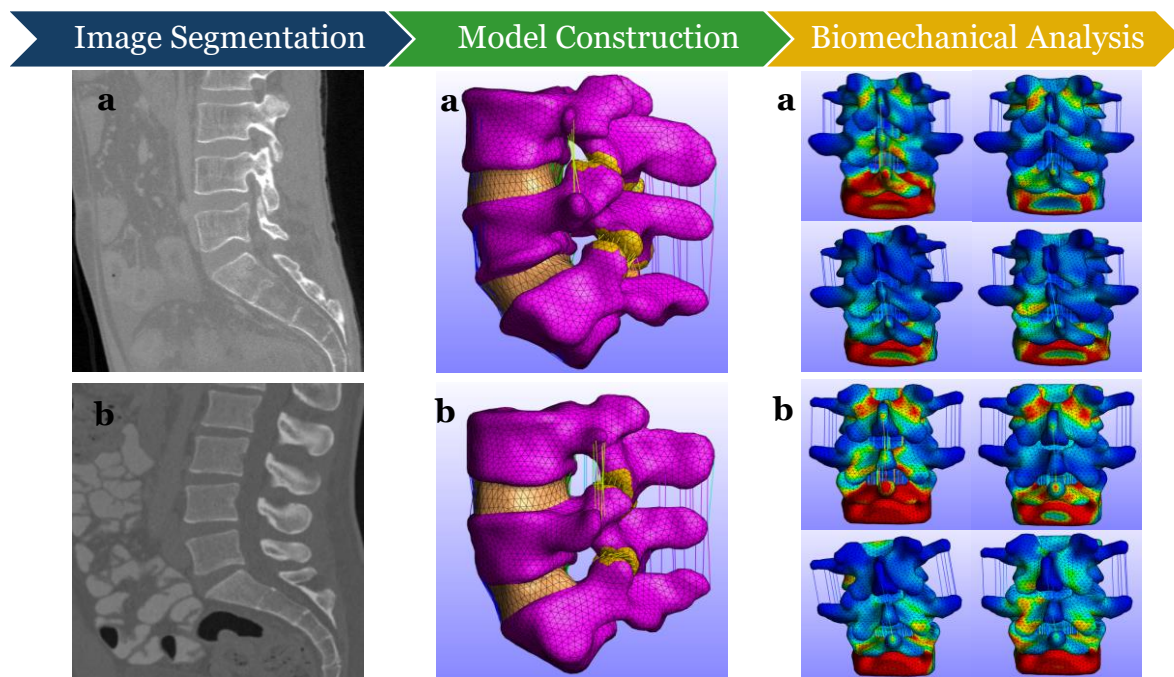


Figure 1: Three-Step Pipeline for Subject-Specific Finite Element Modelling of Spinal Structures – Parallel Pathways for Two Cases (a, b), Models and Four Example Loading Modes for Biomechanical Analysis: Flexion, Extension, Lateral Bending, and Axial Rotation.

1.2 Thesis Structure

This thesis is structured to comprehensively address the aim of developing and analysing patient-specific finite element spondylolysis spinal models, ensuring that both the models and their outputs are functionally robust and of suitable quality, whilst also ensuring that the framework is applicable for clinical use in conservative spondylolysis treatment. These models progressed from an initial analysis of intact spines to then analysing presentations of spondylolysis fracture and finally to phases of healing.

Chapter 1 provides a comprehensive overview of the research background, including the function of each lumbar spinal component and the spinal biomechanics to provide an understanding of why each parameter was considered for this model. Also included is a detailed introduction to spondylolysis and a preliminary literature review of conservative treatment, to present the current understanding of spondylolysis and limitations of treatment. Chapter 2 continues the preliminary research into the current finite element models available for spondylolysis, focusing on the causation and progression of fractures in the lumbar spine. This literature provided parameters for the construction of the spinal components, materials and loading conditions, as well as serving as models suitable for quality analysis and comparison.

Chapter 3 is split into two parts, with part 1 first focusing on the spinal framework construction including descriptions and visualisations of all spinal components and spondylolysis fracture presentations, boundary conditions and materials. This framework was then used to produce two patient-specific spinal models. Part two displays the post processing, quality analysis and application of the conservative treatment framework for a program of activities, analysed across the fracture's healing phases, in which key biomechanical outputs, are calculated and evaluated. Range of motion and vertebral stress were especially investigated using the three fundamental finite element analysis techniques, verification, validation and sensitivity analysis, which are required in producing a model of trusted quality.

The results and discussion are then discussed in chapter 4, which is also sectioned into two parts, focusing first on the results of the three finite element analysis techniques and then in the second half analysing the conservative treatment results across the range of activities.

Finally in chapter 5, this master's research project was summarised including evaluation of the framework for suitability for application, limitations and to highlight the next steps that should be taken to further enhance the framework for future use.

1.3 Introduction to the Spine

1.3.1 Anatomy and Function of Human Spine

The spine or vertebral column is a column of 33 bones and soft tissue connections that run from the skull to the pelvis. The primary function of the spine is to allow for range of motion of the body whilst protecting the spinal cord from injury, from external loads due to movement, environment or internal loads from the body weight of the head and torso (2,3).

The spine is split into five segments of vertebra columns, shown in Figure 2: starting from the base of the skull is the 7 cervical (C1-C7), 12 thoracic (T1-T12), 5 lumbar (L1-L5), 5 sacral (S1-S5) and 3-5 coccygeal vertebra. The cervical spine is a flexible structure that supports movement of the head and neck whilst protecting the spinal cord and blood supply to the brain. The blood vessels pass through the foramen to the base of the skull (4). The thoracic spine is the transition zone from the cervical to lumbar region with progressive increase in vertebral size in the caudal direction. The thoracic spine's function is to stabilise the rib articulations and provide support to the rib cage resulting in a more rigid structure than the cervical and lumbar spine(5). The lumbar spine is the bridge between the two rigid structures of the thoracic spine and pelvis. This region maintains the spine's upright posture, forms the foundational structure for the upper body, protects the abdominal organs(6), transfers weight to the lower extremities, and acts as a shock absorber due to its natural lordosis(7). The sacrum consists of fused rigid vertebrae and functions as the transverse centre of gravity(8) providing a stable and strong pelvis. The sacrum articulates with the coccyx through a fibrocartilaginous joint, functioning as a weight-bearing structure during seated positions(9).

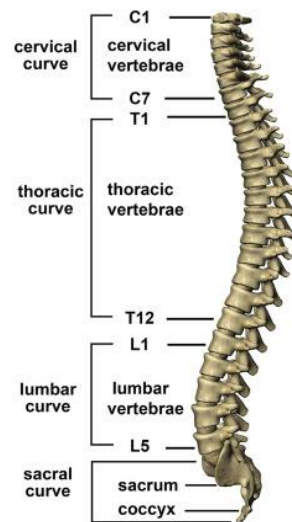


Figure 2: Anatomy of the spinal column (10)

1.3.2 Anatomy of the Lumbar Spine

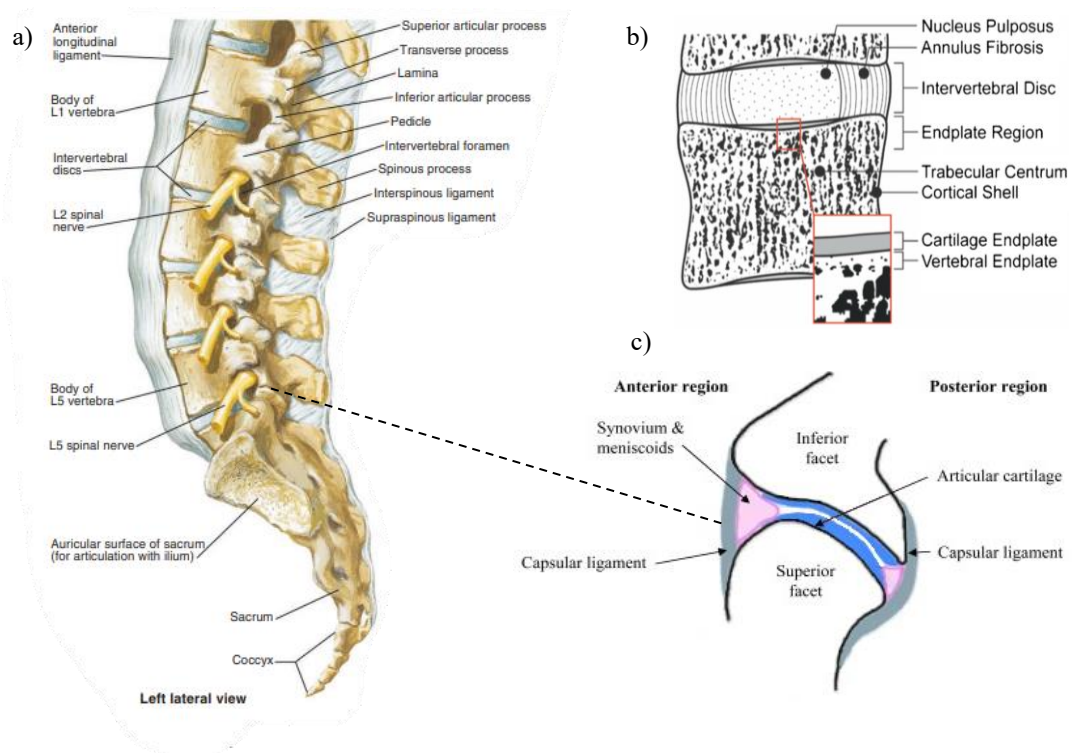


Figure 3: a) Anatomy of the ligamentous lumbar spine(11), b) vertebral structure(12) and c) facet joint components(13)

1.3.2.1 Vertebral Bodies

In the lumbar spine there are 5 vertebra which have unique anatomy compared to the other vertebral levels. These vertebrae are the largest in the spine and have larger vertebral bodies due to the increased proportion of weight distributed in the area. These vertebra also lack the transverse foramina which allow for the passage of blood vessels in the cervical spine and costal facets of the thoracic and cervical vertebrae whose role is to support the ribs attachment (14). These vertebrae also have shorter, thicker spinous processes which are perpendicular to the body and narrower transverse processes(15). The vertebrae also comprise a posterior arch, each possessing seven processes, as illustrated in Figure 4. These processes primarily serve for ligament and muscular attachment or articulation between adjacent superior and inferior vertebra (as seen in Figure 3a). These processes are joined to the vertebral arch, which is made up of the lamina and pedicles, forming the walls of the spinal canal. The vertebrae bone structure, outlined in Figure 3b, comprises of cortical and cancellous/trabecular bone and endplates (EP). The endplates are located at the superior and inferior vertebral body faces and have the function of load transfer between the intervertebral disc and vertebral body. The endplates also allow for the transfer of blood to the avascular disc(16), maintaining the structure and height of the disc through the diffusion of water after constant daily axial compression, causing fluid loss of the annulus fibrosus. The cortical shell of the vertebra encloses cancellous struts which are aligned vertically to follow the path of compression and principal stress(17), whilst not increasing vertebral weight(18).

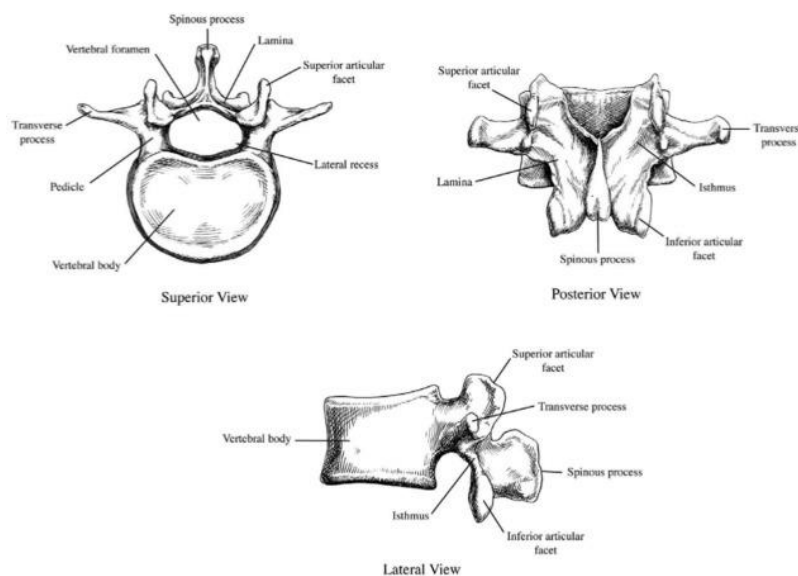


Figure 4: Anatomy of the lumbar vertebra displaying the three views of the processes. (15)

1.3.2.2 Intervertebral Discs

The intervertebral discs (IVD) are multi-tissued structures located between the vertebral bodies of the lumbar spine and consist of the nucleus pulposus (NP) and the annulus fibrosus (AF) as shown in Figure 3b. The nucleus pulposus constitutes 20-50% of the intervertebral disc and is a gelatinous material composed of 70-90% water, a matrix of proteoglycans, and type II collagen(19). The collagen fibres provide the structure with tensile strength and the proteoglycans allow for the binding of water to control the disc volume. The flow of fluid within the disc and its ability to undergo reversible deformation after loading, are regulated by the structural proteoglycan aggrecan, which binds to water molecules(20). Due to compressive loading of the spine, diurnal fluctuations in water content can reach up to 25%(21). During the 16-hour period of daytime loading, water diffuses out of the nucleus pulposus, resulting in a reduction in disc height. This is followed by an 8-hour period of rest, during which the disc rehydrates through the growth plate, restoring height and preventing degeneration(22). The annulus fibrosus contains the nucleus pulposus, with its structure of 15-25 concentric rings of lamellae(23). Each lamellae consists of collagen and elastin fibres which insert into the endplates and are arranged in an alternating pattern, with each sheet oriented obliquely at a 30-degree angle on average(23). The density of the disc increases with the radial distance from the nucleus pulposus with the outer layers providing stiffness to the disc to maintain the hoop stress(24) under loading, whilst the inner layers allow for increased diffusion with sparse layers. The primary functions of the discs are for shock absorption and as one of the three-joint complex components of the spine. When the spine experiences compression the incompressible nucleus deforms distributing pressure to the annulus fibrosus transferring load through the annulus fibrosus layers, producing tension to withstand the spinal load(25).

1.3.2.3 Facet Joint

The facet joints (zygapophyseal joints) are the second and third components in the three-joint complex of the spine, alongside the intervertebral discs. They play a biphasic role(13) in spinal motion, both permitting and restricting movement. This dual function is essential for maintaining spinal stability while allowing necessary flexibility.

The facet joints are located at the posterior aspect of the vertebral arch, between the inferior and superior articular processes of adjacent vertebrae. The joints, and their anatomical structures, vary depending on the individual and across different spinal regions and facet surfaces. In the lumbar region, the joints are more vertically oriented compared to the thoracic and cervical regions, with the superior facets having a convex shape and the inferior facets being concave. This orientation limits excessive motion in the lumbar region, protecting the spinal structures from high levels of mechanical stress. The facet surfaces in this region have an inclination angle of 82° to 86° in the sagittal plane and an angle of 15° to 70° (13) off the midline in the axial plane. These specific orientations control the degree of lateral bending and axial rotation, ensuring the integrity of the spine under varying load conditions. This specific structure permits movement of extension, flexion and lateral bending but restricts the movement of axial rotation(26).

As diarthrodial joints, these structures have a thin layer of hyaline cartilage(26), illustrated in Figure 3c. The cartilage layer can vary in thickness and shape, with it mostly covering the articular surfaces, leaving a region of exposed subchondral bone at the edges of the processes(13). The cartilage provides low-friction movement during tensile, compressive, and shear loads across the six degrees of spinal motion. The cartilage is surrounded by a synovial membrane, which produces synovial fluid to lubricate the joint(27). The joint is then encapsulated by a fibrous joint capsule, also known as the facet capsular ligament (FCL) as well as the ligamentum flavum. The FCL undergoes the most extensive deformations under motion out of the lumbar spinal ligaments, playing a crucial role in managing spinal stability and maintaining joint integrity during movement.

The multi-component facet joints play a key role in lumbar spinal deformations with multiple mechanisms to transfer load through the vertebra, transferring load through direct contact between the articular surfaces, ligaments and contact of the articular processes and neural arch. The facet joints transfer 20% of the axial force enacted on the spine during compression(28). The transmission of axial compressive loads to the lamina and pars interarticularis of the arch differ in specific loading conditions of the facets. Repetitive motions of extension or hyperextension can cause a pincer effect(29) on the pars interarticularis between adjacent inferior and superior articular processes, leading to stress fractures or conditions like spondylolysis.

1.3.2.4 Ligaments

The lumbar spine is supported and stabilised by seven main ligaments: the anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), ligamentum flavum (LF), interspinous ligaments (ISL), supraspinous ligament (SL), intertransverse ligaments (ITL) and facet capsular ligaments (FCL) as well as the iliolumbar ligament which connects and stabilises the lumbosacral spine and pelvis. These ligaments play a crucial role in maintaining stability and alignment of the spine during movement or weight-bearing activities. The behaviour and mechanical response of the ligaments are due to the structure, organisation, and interactions between its matrix of collagen fibres, elastin fibres and ground substance(30), where this matrix allows the ligaments to respond dynamically to mechanical loads. The collagen fibres are aligned in parallel bundles that twist at insertion sites between the bones and intervertebral discs of the spine(31). The parallel fibres have periodic “crimps” in which the usually straight fibres appear undulated(32). These crimps, which have a variation in size and angle, stretch and align with the direction of load during elongation or movement as tension increases, acting as a mechanical buffer(32) for the collagen fibres. During relaxation, these crimps regain their original shape acting as a recoil system(32) to preserve spinal stability. This biomechanical response of the fibres is the reason ligaments are primarily designed to resist and absorb tensile force. Due to the alignment of the fibres, the mechanical response to stress is non-linear, as the fibres deform from their wavy configuration to a parallel alignment (33). This process results in a load-deformation response that progresses from an initial toe region through to a linear phase, ultimately leading to plastic deformation and failure. Once this failure phase takes place in adult ligaments, full pre-injury structure and response is never recovered(34). Degeneration of ligaments and resultant loss of stability of the spine can lead to fatigue induced spinal conditions like spondylolysis or spondylolisthesis from vertebral slipping.

Each of the seven lumbar ligaments plays a crucial role in protecting the spine from injury, with each having a distinctive structure that supports this function. The positioning of these ligaments is illustrated in Figure 3a. First there are the ALL and PLL ligaments which run from the atlas to the sacrum. The ALL is a densely packed ligament composed of three distinct layers of collagen fibres, which attaches anteriorly to the vertebral margins with a

loose adherence to the intervertebral discs(35). The main functions of this ligament are to prevent hyperextension of the spine and absorb load in lateral bending. This structure is unique among all spinal ligaments as the sole structure responsible for restricting extension(36). The double-layered posterior longitudinal ligament, similar to the ALL, adheres to the vertebral margins on the posterior side of the vertebral body. However, in the disc region, its fibres fan out horizontally, covering the annulus fibrosus with a strong adherence to the disc in comparison to the vertebrae(36). The primary function of the PLL is to resist hyperflexion of the spine with a secondary function of preventing disc herniation. The remaining five ligaments (ITL, ISL, SL, LF, FCL) are positioned and join between adjacent vertebral arches. The LF attach superiorly and inferiorly between the anterior lamina of each vertebral level as well as medially to the ISL and the facet joint capsule(36). This ligament's primary function is to protect the spinal column during flexion and extension, preventing buckling supported by its rich composition of elastic fibres, allowing the spinal position to be restored after bending(37). The ISL are located between adjacent spinous processes and attach not only to the LF but also to the facet capsule. The alignment of the collagen fibres travels in a posterocranial direction originating from the LF, extending to merge with the SL, however, there are discrepancies in literature between the direction of the fibres with reported cases of horizontal, curved or vertical fibres(36,38). The main function of this ligament is to protect the spine during flexion by preventing separation of the spinous processes, however this ligament is weak in isolation(36). Also, it is important to note that orientation of these fibres effects the ligaments fibres response to flexion. The SL is composed of three layers of oriented collagen fibres and strong tendinous fibres which aids in limiting flexion. This ligament carries the greatest load of the posterior arch ligaments and has the greatest elongation(36). The ITL are thin membranous ligaments(39) that connect two adjacent transverse spinous processes. This ligament's function is to be loaded in flexion and lateral bending, but is the only ligament which does not act during axial rotation(36). The final ligament we are considering in this review and future models are the FCLs, which as previously discussed is connected between superior and inferior articular processes around the facet joints. The FCL releases synovial fluid to lubricate the cartilage and surfaces of the processes, but has an additional function to limit flexion, lateral bending, and axial rotation(27).

1.3.3 Lumbar biomechanics

The lumbar spine transmits axial compressive forces, originating from the weight of the upper body and muscle activity, through the vertebral bodies and intervertebral discs. Approximately 18% of these forces are carried by the facet joints(40), with the proportion of load in the posterior arch varying with bending. Increased bending not only alters the load distribution in the posterior arch but also introduces shear forces. These shear forces, often coupled with flexion, cause anterior-posterior movement of the vertebrae. However, this movement is restricted by the muscles, ligaments, and facet joints, which work together to protect the spinal canal and maintain stability under varying loads.

The spine moves through six degrees of freedom (DoF) during daily activities: flexion, extension (Flex-Ext), lateral bending, and axial rotation, as seen in Figure 5. Flexion and extension occur in the sagittal plane, rotating around the x axis(41). These movements involve spinal arching forwards (flexion) or backwards (extension). The lumbar spine's range of motion averages 55.4 degrees for flexion and 23.1 degrees for extension(42). Movements of flexion are controlled and stabilised by the ligaments of the posterior arch and the lumbopelvic musculature. Extension is restricted by the anterior longitudinal ligament, intervertebral discs and the facet joints to protect the spinal canal. Common activities of daily living, such as moving from sitting to standing positions, bending, walking, and inclining/declining stairs, involve flexion over other bending moments(41). Lateral bending (LB) occurs in the frontal plane with movements bending side to side, with rotation around the y axis. The lumbar range of motion from lateral bending ranges from 21.7-22.8 degrees of left to right bending(42). Lateral bending is limited and stabilised by the intertransverse and capsular ligaments. Axial rotation (AR) occurs in the transverse plane with rotational movements around the z axis. This motion is the most restrictive in the lumbar spine, with a left and right rotation range of motion of 13.1-14.1 degrees, due to limited flexibility given by the facet joints.

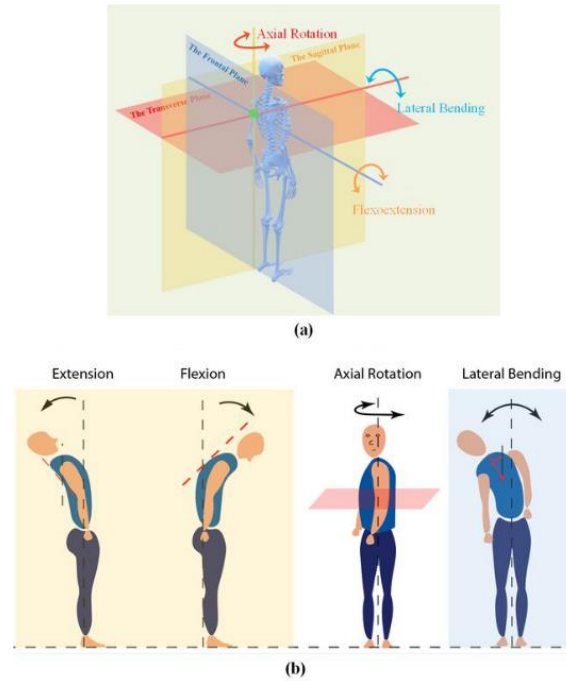


Figure 5: Planes of motion of the spine in the view of planes and axis of motion (a) and the six degrees of freedom (b)(43)

Movement patterns differ along the lumbar spine, where the inferior vertebra's movements are stiffer in extension, with the primary location of deformation occurring at L2-L3 (44). In contrast, flexion primarily occurs at the L5-S1 joint(44). These movement characteristics were highlighted in a finite element study conducted by Hsieh (2020)(45), who analysed lumbar spinal models. The study's findings, presented in Table 1, demonstrate the differences in the range of motion across various spinal segments. The stiffness across these spinal levels also influences the neutral zone (NZ), which is defined by Panjabi (1989)(46) as 'the range of intervertebral motion, measured from the neutral position, within which the spinal motion is produced with a minimal internal resistance.' An example of a neutral zone load curve is displayed in Figure 6, where loads outside of the ROM range will experience deformation that causes the segment to not return to its original position. The neutral zone is largest in flexion, extension, and lateral bending but for axial rotation the range of minimal resistance is much smaller(47).

Table 1: Maximum range of motion of lumbar segments in degrees:

	L1-L2	L2-L3	L3-L4	L4-L5	L1-L5
Flexion	3.93	3.94	3.89	5.33	17.09

Extension	2.80	2.39	2.25	2.21	9.65
Lateral Bending	3.30	3.17	2.98	2.88	12.3
Axial Rotation	1.95	2.21	2.59	3.63	10.38

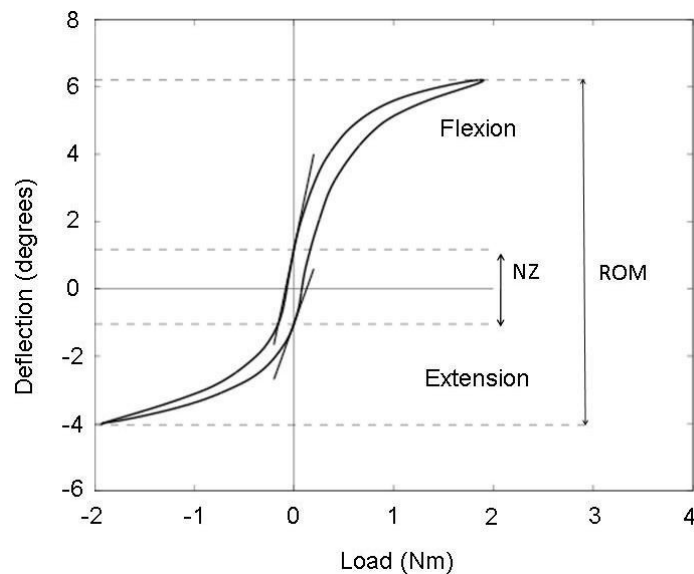


Figure 6: A load-deflection curve demonstrating the typical neutral zone (NZ) and range of motion presentation (48)

1.4 Introduction to spondylolysis

1.4.1 What is spondylolysis?

Spondylolysis is a lumbar spine condition presenting as defects in the vertebral arch, occurring in the pars interarticularis region. This region acts as a narrow bridge/isthmus between the inferior and superior articular processes, and the lamina(49). Spondylolysis can progress from unilateral to bilateral fractures on a single vertebra, to multi-level fractures. When bilateral fractures occur, there is a risk of progression to spondylolisthesis, where the fractured vertebra undergoes anterior slippage with respect to the inferior vertebra. Spondylolisthesis is most commonly found in adolescents during their growth spurt or in individuals between the age of 50-80(50,51). The prevalence of spondylolysis in the general population is approximately 8%(51), but it is notably a major cause of back pain in

adolescents(52), with rates of 4% by age 6 and 6% by age 14(53). Additionally, spondylolysis is present in 10% of athletes who engage in sports that involve frequent lumbar extension(53), with even higher prevalence reported in specific sports, such as professional football (38.1%) and baseball (44.1%)(51). Spondylolysis exhibits genetic patterns, with a higher incidence observed in males, showing a 2:1 male-to-female ratio(54). It is also more common among first-degree relatives of individuals with the condition and those with concurrent conditions such as spina bifida occulta, Marfan syndrome, osteogenesis imperfecta, and osteopetrosis(54).

1.4.2 Classification

Spondylolysis is classified using the Wiltse-Newman Classification(55) with the five main categories of dysplastic, isthmic, degenerative, traumatic and pathological spondylolysis as seen in Table 2.

1.4.2.1 Dysplastic

Dysplastic spondylolysis (type I) accounts for 14-21%(56) of lysis cases and is a congenital disorder characterised by a defect in the pars interarticularis, lumbar facets, intervertebral discs, or vertebral endplates(56). While the exact pathogenesis of spondylolysis remains unclear, there is evidence suggesting a genetic link, particularly associated with autosomal dominant inheritance(56,57). This genetic link is further supported by the higher incidence of spondylolysis and spondylolisthesis in family members of index patients, reported at 22%, compared to 8% in the general population(57).

1.4.2.2 Isthmic

Isthmic spondylolysis occurs in the pars interarticularis or isthmus and is characterised as a stress fracture. Type IIa is a stress fracture that occurs after repetitive stress and extreme motions from activities such as gymnastics, bowling, or rugby(53). Type IIb also results from repetitive stress fractures, but after repeated micro-traumas and healing, the pars interarticularis begins to elongate(52), which leads to increased stress or eventual degeneration of the surrounding lumbar structures. Type IIc is an acute fracture type which occurs from a traumatic high load event causing immediate fracture of the pars.

1.4.2.3 Degenerative

Degenerative spondylolysis (type III) occurs due to degeneration of the structures of the spine excluding the pars interarticularis including the intervertebral discs and the facet joints(58).

Degeneration of these structures leads to spinal instability and weakening of the ligaments and eventually progressing to spondylolisthesis(52).

1.4.2.4 Traumatic

Traumatic spondylolysis (type IV) presents as fractures other than the pars interarticularis in regions of the pedicles, facet joints or lamina(59). These fractures are often acute and can be coupled with isthmic fractures.

1.4.2.5 Pathological

Pathological spondylolysis (type V) occurs due to vertebral bone diseases or infections (metastatic or neoplastic processes, Paget's disease), or tumours which result in complications of the vertebra(58,59).

Table 2: Classification of spondylolysis or associated fractures of the vertebrae (55,58)

Type of spondylolysis	Title	Pathogenesis
Type I	Dysplastic	Congenital abnormalities
Type II a	Isthmic	Stress fractures in the pars interarticularis
Type II b		Elongated intact pars
Type II c		Acute fracture of the pars
Type III	Degenerative	Degeneration of structures not including the pars
Type IV	Traumatic	Acute fractures in areas other than the pars
Type V	Pathological	Bone diseases, tumours or infections

1.4.3 Spondylolysis progression

Spondylolysis pars interarticularis fractures are classified in stages by the severity of the stress injuries as presented on MRI images(60). Grade 0 is an intact pars interarticularis with no signal changes. Grade 1 demonstrates the oedema, or “very-early” lysis phase induced by stress but with no fracture lines. Grade 2 is also known as early-stage lysis, where signal abnormalities of the pars can be detected. In this stage, the pars interarticularis has visible irregularities presenting as thinning or fragmentation(60) or a hair-line defect(61). Grade 3 indicates either the progressive or terminal stages, with a complete unilateral or bilateral fracture in imaging with cortical disruption. Progressive lysis shows a wide defect at the pars with bone fragments(61). Where the terminal stage presents as a pseudoarthrosis(62) (a persistent gap or false joint) with visible sclerotic (dense bone formed during failure of bony union) changes of the bone seen in imaging(61). Finally, grade 4 represents a united bone fracture from a previous injury.

1.4.4 Symptoms, Impact on Quality of Life, and Associated Complications

Individuals with spondylolysis are usually asymptomatic (87%)(63) and can continue through daily activities of living, however, for progressive or acute fractures lower back pain can develop. As the condition develops, lower back pain radiates into the gluteal muscles, proximal lower extremities and posterior thighs(49,54). These symptoms worsen with activity, particularly in movements of extension(64) and standing in flexion for prolonged periods of time(52). The symptoms of this condition are progressive, and initially may only occur during strenuous or extended periods of activity. However, if the condition remains untreated the symptoms can appear during activities of daily living and interfere with quality of life due to the necessary periods of rest required to temporarily relieve the symptoms. Progression of the condition to spondylolisthesis can cause damage to the nerves, resulting in numbness, tingling and loss of function of the lower extremities and issues with bladder and bowel control(65).

1.4.5 Diagnosis

Diagnosis often starts with the patient reporting lower back pain as well as the presentation of tightness/contracture of the hamstring muscles(64). The patient also may present with lumbar hyperlordosis (increased concave curvature) and in late-stage cases lumbosacral kyphosis (increased convex curvature) detectable during palpation(52). However, confirmation of the diagnosis is done with X-ray or CT (Computed Tomography) imaging to identify the fracture and progression of the condition. Presentation of spondylolysis on oblique radiographs can be used to identify the “Scottie-dog” presentation(66) where the fracture across the pars interarticularis presents as a “collar” between the transverse process and articular processes as demonstrated in Figure 7. This presentation is of progressive to late-stage spondylolysis. The initial stage of lysis or bone bruising can be detected through Magnetic Resonance Imaging (MRI) using axial T2-weighted scans where the high signal contrasts are indicative of bone oedema. The pedicles are the most frequent sites of high signal contrasts, compared to the pars interarticularis which is harder to identify in imaging(67).

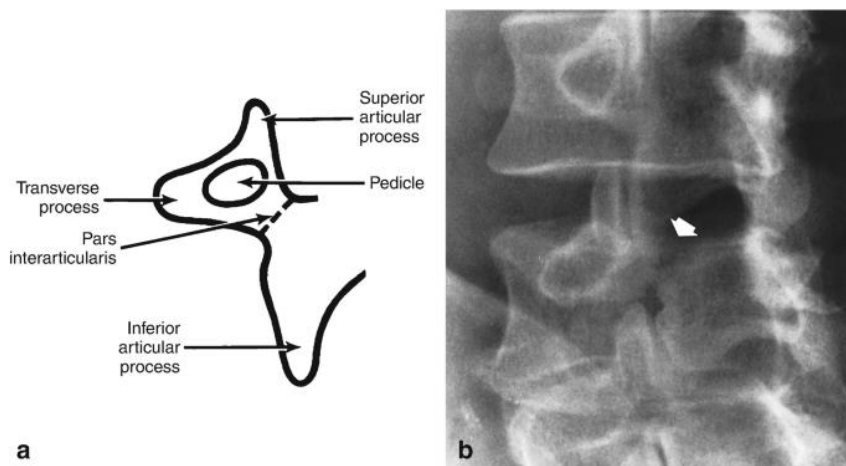


Figure 7: Spondylolysis fracture presentation identified by the Scottie dog sign(68)

1.4.6 Spondylolysis Recovery

The oedema and initial stages of spondylolysis offer the greatest potential for bone healing, while the likelihood of successful healing diminishes in the progressive stage and becomes negligible in the terminal stage(69,70). A study performed by Sakai (2017)(71), demonstrated that for a group of sixty patients that underwent conservative treatment of rest and activity

restriction, bony healing was achieved in 100% of very-early cases, 93.8% of early stages cases, 80% of progressive stage cases and 100% for all recurrent cases.

Successful bony healing in which the bone is returned to its pre-injured state requires a series of steps including hematoma formation, granulation tissue formation, bony callus formation and bone remodelling(72). Mechanobiological regulation plays a critical role throughout these stages, as mechanical forces applied to the healing bone influences cellular activity, tissue differentiation, and overall healing outcomes(73).

Haematoma formation occurs immediately after a fracture as a result of the disruption of blood vessels at the injury site. These vessels supply the damaged area with haematopoietic cells, leading to the formation of a clot that serves as the initial framework for the healing process(72). The next phase is granulation tissue development, occurring 3-7 days post-fracture(73), during which the haematoma transforms into granulation tissue, followed by fibrous connective tissue. This transformation is facilitated by the migration of mesenchymal stem cells(74), which differentiate into fibroblasts that produce the extracellular matrix. Additionally, hematopoietic stem cells give rise to mononuclear cells(75) that contribute to the immune response and support the healing process. Granulation tissue is highly deformable and can “elongate to twice its original length(75). Hence in this early phase, it is important to cease activity to allow for the fractures to heal. The next phase is cartilage formation, where the granulation tissue matures into collagen-rich cartilage through the activity of chondrocytes(76), forming a cartilaginous callus or bridge across the fracture gap. This process typically occurs 2-4 weeks after injury(73). Around 2-4 months after the injury, the callus transitions from a cartilaginous structure to an immature, hard bony structure that surrounds the fracture site(73). This process involves the resorption of cartilage and its replacement by bone, primarily through the activity of osteoblasts that are recruited to the site. Chondrocytes in the cartilaginous callus undergo hypertrophy and apoptosis, facilitating ossification(76). The final step is bone remodelling of the woven immature bone into organised mature bone(74). This process takes up 70% of the total fracture healing time and may last for 6-9 years(75).

Factors that can hinder the healing process and lead to delayed or non-union include insufficient stability or excessive fracture loading. Patient conditions also affect fracture recovery such as poor blood supply, hormonal imbalance, or underlying conditions such as

diabetes, hypothyroidism, or osteoporosis(77) or lifestyle choices such as smoking or drinking. To optimise healing and prevent complications, it is crucial to address and manage these factors through early intervention and tailored treatment strategies.

1.5 Conservative Precautions and Treatment Literature Analysis

The following section consists of a preliminary literature review on current conservative treatment strategies for spondylolysis, from twenty studies. This review aims to analyse the existing approaches, identify areas for potential improvement, and explore how advancements in biomechanical understanding could enhance treatment outcomes. Current conservative techniques include movement restriction, physical therapy, and the use of orthoses, while less frequently applied methods involve bone or electrical stimulators and kinesiology tape. Conservative treatment is effective in the early to progressive stages of spondylolytic lesions, with fractures healing after 6 months of treatment for unilateral cases and in 50% of bilateral cases(78).

1.5.1 Rehabilitation programs

1.5.1.1 Activity Restriction

Activity restriction is an important first step of conservative treatment with recommendations from nine studies in this review(64,79–86). This phase involves either complete rest, adjusting daily activities to limit spine movement or the elimination of specific aggravating activities such as sports. This period aims to minimise the abnormal spinal biomechanics due to spondylolysis or sporting activities to prevent the propagation of vertebral fractures(80,87). The time recommended for activity restriction varies significantly across studies. Three studies advise stopping all activities until symptoms improve and the patient is ready for physical therapy(64,82,85), while others recommend reducing painful activities such as sports for three to six weeks(79,83), or even up to 3 months of rest with allowance for daily living activities(80,81,86).

These guidelines for activity restriction are based on the time to complete bony healing, however, the duration of activity restriction is based on low-level evidence(64), highlighting the preliminary nature of these guidelines. This restriction is also not without its drawbacks, including potential risks such as muscle atrophy and disruptions to daily routines, which is an important consideration for adolescents who may miss school or athletes who risk a reduction in muscle conditioning(64). Initiating rehabilitation during fibrous formation presents a potential compromise, as the spondylolysis may be asymptomatic at this stage(82), allowing for a smoother transition to subsequent treatment phases.

1.5.1.2 Physical Therapy

Thirteen studies(64,79–86,88–91) highlight the necessity of a physical therapy treatment program for patient recovery, aiming to facilitate a return to daily activities or sports. These programs are either tailored to the individual's specific spondylolysis condition, adjusting the speed of progression and type of activity based on the patient's vertebral fracture progression, current physical level, or future needs(64,79–83,85,90). The second approach is a program of predetermined activities and durations, with outcomes assessed based on bony healing or symptom improvement(86,88,89,91). The debate around bony healing's necessity for rehabilitation success is highlighted by the variance in clinical outcomes. Standaert (2000)(80), notes that with a program combining activity restriction, bracing, and physical therapy, 78-96% of patients achieved good clinical results, with 25-37% showing bony healing through radiographic union. Conversely, studies(85,86) highlight the necessity of bony or fibrous healing to minimise recurrence risk or prevent progression to conditions like spondylolisthesis. Choi's (2022)(86), analysis supports this, showing that post-treatment 98% of individuals returned to sport, where 49.8% exhibited bony healing, but 23.9% developed spondylolisthesis. The success rates of these treatment programs also correlate with the initial grade of spondylolysis, with Standaert (2002)(80), finding healing rates of 73% in early defects, 38.5% in progressive defects, and 0% in terminal defects after 3-6 months of treatment with lumbar corsets and sports discontinuation.

Each treatment program aims to address the underlying cause of spondylolysis by addressing deficiencies in spinal strength, flexibility, and stability(64,89), or by rectifying sports techniques that place excessive strain on the vertebral structures. Muscle strength and coordination are important to distribute internal and external forces in the body. If a muscle is

fatigued, there will be a loss of synchronisation between muscle groups, with more force being redistributed to the spine, resulting in injury as well as a loss of spinal stability. Hence, to enhance stability, physical therapy stabilisation and endurance exercises should focus on addressing muscle deficiencies by targeting muscles interacting with vertebral fractures, including the transverse abdominis and multifidus. This approach encourages coactivation and coordination(64,79,83) to achieve a "biomechanically sound spinal function"(82). Flexibility and mobilisation of the hip and leg segments are also important, as stiffness of the mobility joints and extremities can lead to increased compensatory stress and forces on the stability joint or the lumbar spine(88). Therefore, physical therapy should target hamstring and quadricep tightness(79,88) and hip flexor flexibility(83) with gentle stretching and manipulations so as not to aggravate the fractured vertebra. The programs should be specifically designed to improve physical capabilities, ensuring they are customised to meet the individual needs of the patient. This approach is essential to prevent the recurrence of spondylolysis once the patient resumes their usual level of activity, with the aim of adequately preparing the patient for the biomechanical demands of their daily activities.

Example treatment programs are outlined in several studies of this review(64,81–83,85,88–91), with exercises that slowly increase in intensity as the patient's symptoms decrease and physical capabilities increase. The exercises start with low-level static muscle activation exercises in a neutral position, then progress to gentle mobilisation of the spine in flexion and bilateral bending but avoiding extension and rotation, then to building flexibility and muscle endurance through dynamic exercises, and finally, if necessary for the patient, to sport-specific exercises that correct technique and increase overall fitness. The physical therapy approaches for treating spondylolysis are informed by studies of similar spinal conditions and expert insights(64), yet an optimal treatment protocol has not been established. Although these programs are supported by biomechanical reasoning, further research is essential to gauge the impact on vertebral fractures, determine the most appropriate timing for initiation, and establish the duration for which the treatments should be continued.

1.5.1.3 Orthoses/Bracing

The use of orthoses during the initial treatment phase is advocated by ten studies(79,81,83–85,89,91–94). Spinal braces are recommended to ensure proper posture and immobilisation of the spine during the rest period after a spondylolysis diagnosis or once more strenuous

activities are resumed to prevent excessive rotation or hyperextension. The recommended duration for wearing a brace varies: one study suggests using braces only while symptoms persist due to patient burden(79), while others propose a usage period ranging from three to six weeks(83,91). Additionally, four studies recommend wearing the brace for three to six months(80,84–86) while the patients commence rehabilitation programs, to protect the spine as activity levels increase. The duration of orthotic usage and the effectiveness of the orthoses also rely on the advancement of the lysis fracture. Studies(91,92), employing signal-change MRI techniques, revealed that early spondylolysis exhibited 87-94% bony healing within 3-4.3 months, whereas progressive fractures showed 60-64% healing after 5-6 months, and terminal defects exhibited no bony union. Orthoses are still beneficial for terminal stage spondylolysis as the area of the defect is stabilised through restriction of lumbar motion and therefore reduces the patient's pain; however, the extent of lumbar motion reduction is not quantified(94). The goal of bracing in spondylolysis treatment is to restrict motion in the fractured area, providing improved stabilisation by reducing total vertebral column movement. The studies examined various orthotic devices, including Thoracolumbosacral Orthosis (TLSO) and Lumbo-Sacral Orthosis (LSO), which come in rigid, semi-rigid, or hybrid forms, as well as braces designed for sports re-engagement. The specific type of brace has a significant impact on the range of motion of the spine. For instance, while rigid or semi-rigid LSO or TLSO braces might limit all six lumbar spinal movements, hybrid braces, which may combine soft front panels with hard back panels, allow for greater flexion. Meanwhile, braces intended for sports activities generally impose fewer restrictions on rotation. There is debate among the studies as to whether spinal flexion is a risk factor for spondylolysis. Research from Nakashima (2021)(91), indicates that biomechanical evidence supports spinal flexion without adverse effects, and Ibiebele (2021)(92) confirms that the main purpose of LSOs is to target mechanical factors contributing to lysis caused by extension and rotation. Therefore, these studies advocate for LSOs that offer greater mobility, such as the hybrid brace. In contrast, Fujimoto (2020)(94), recommends the use of rigid LSOs for patients in the progressive stages of spondylolysis to enhance the likelihood of bony healing and prevent further complications.

While these studies provide promising biomechanical reasoning for the usage of braces in spondylolysis treatment, studies(64,80,82,92) advise against their usage. The restrictions

imposed by TLSO, and rigid LSOs are restrictive to daily life and often result in low patient compliance due to the “solidity, tightness, heat, humidity, and general discomfort”(94) that comes with wearing the brace. Furthermore, these studies point out the minimal evidence supporting the effectiveness of braces in treatment, often noting that braces are tested alongside other treatments to not compromise patient care. The literature also suggests a lack of biomechanical analysis on the impact of braces on patients with spondylolysis, indicating braces primarily restrict overall body movement without significantly affecting intervertebral mobility(81). Watkins (2010)(82), also states that bracing serves as an effective temporary pain relief method, but removal often leads to increased tenderness due to muscle atrophy from compensatory support from the brace. To further evaluate the effectiveness of bracing, future research could include conducting a randomised controlled trial focusing solely on bracing as a treatment or employing computational biomechanical analysis to understand the internal forces and motions in patients with and without a brace.

1.5.2 Conservative Management Methodological Limitations

The primary limitation of conservative treatment seems to be the significant variation in the timing of the conservative treatment’s initiation, duration and progression. The time of each phase of rehabilitation from periods of rest, exercise or application of orthoses depends on expert opinion and monitoring of patient-reported pain. As a result, the healing of spondylolysis may be impacted by the lack of standardised guidelines or patient-specific tools to track the recovery process.

Further limitations of the studies are that only short-term effects of conservative treatment were studied, there were little analyses of spondylolysis recurrence and therefore it is hard to predict the long-term success of these treatments. Biomechanical reasoning was also given in several of the studies, but no direct biomechanical testing was done from either experimental or computational testing on spondylolysis-specific cases.

Additionally, there were study limitations across the conservative treatment subject groups, with the most prevalent issue being a lack of evidence from mostly descriptive-based studies. The studies that had an experimental methodology often did not have any control groups, as to not compromise patient care, for comparative analysis meaning that the understanding of

which rehabilitation technique is still not developed. The patient groups in these studies also had debatable compliance for the activity restriction, physical therapy and orthoses usage where the studies had to rely on a patient's word on how rigorously they performed these activities.

To address these limitations of current conservative treatment strategies, the development of a computational model to monitor patient healing progress could provide a framework for tailored treatment programs. By analysing each patient's specific biomechanical and kinematic data related to their spondylolysis, a model could offer personalised recommendations for treatment duration and intervention based on the individual's healing progress. Providing an upfront treatment plan could also aid with patient compliance to improve the likelihood of success.

Chapter 2: Literature Review

Biomechanical Analysis of Spondylolysis: Causes, Progression, and Management for Fracture Prevention

2.1 Introduction

This chapter of the review aims to build understanding of the current biomechanical knowledge of spondylolysis and associated lumbar spinal fractures. This review will be conducted from a unique perspective through the analysis and comparison of fracture biomechanics through experimental and computational methods. Combining the biomechanical knowledge from experimental analysis of a patient's unique morphology and loading conditions, as well as computational modelling of their spine's stress and force patterns, would advance personalised treatment plans. In addition to these two primary objectives, the secondary objective is to understand the current quality and complexity of vertebral fracture models, providing insight into future advancements necessary for incorporating biomechanics into patient treatment.

This review will analyse computational methodologies utilised for assessing the biomechanics and presentation of lysis in the lumbar spine, complemented with an

experimental analysis section. When considering experimental techniques, *in vitro* experimental testing obtains measurements by evaluating spinal manipulations under mechanical load, however, these approaches encounter challenges in precisely analysing the intricate micro biomechanics. Moreover, such techniques are constrained by the accessibility and diversity of the specimens utilised. *In vivo* testing, on the other hand, is more desirable due to its capacity to provide insights into the spine's dynamics in its natural environment and interactions between biological systems. The application of *in vivo* methods is restricted, primarily due to the necessity for minimal invasiveness to avoid harming the subjects involved in the study. Computational methodologies, or *in silico* techniques, have been developed to overcome these limitations. Computational approaches utilise Finite Element (FE) analysis, a sophisticated computational tool that models complex structures, like the spine, into smaller, more manageable elements. This method allows for a detailed examination of the biomechanical behaviour such as stress and strain patterns within the spinal structures under various physiological conditions, offering information which is more complex to measure through experimental methods. Combining these methods will enhance the understanding of lysis aetiology, thereby guiding effective treatment, rehabilitation, and prevention strategies.

2.2 Methods

2.2.1 Search strategy and criteria

This review was conducted according to the Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA)(95). The review protocol was established prior to the study at the International Prospective Register of Systematic Reviews (PROSPERO) (CRD42024512484). Three online databases (ScienceDirect, PubMed and Scopus) were searched to extract articles up to March 25th, 2024, with a subsequent snowball search and rerun, ending on April 5th, 2024. The search strategy consisted of the following keywords: Lysis, spondylolysis, laminolysis, pediculolysis, vertebrae, fracture, biomechanics, pathomechanics with Boolean operators (AND, OR, NOT). Article abstracts were screened and extracted by one author (P.K) and assessed independently by two reviewers (vL.R, A.S)

to compare against the eligibility criteria. Any disagreements were resolved through discussion. To conduct this review, exclusion and inclusion criteria was selected to retrieve relevant publications, outlined in Table 3.

Table 3. Article eligibility criteria

Section	Inclusion criteria	Exclusion criteria
Experimental & Computational methods	Biomechanical analysis of lumbar vertebral fractures: spondylolysis, pediculolysis or laminolysis	Absence of biomechanical analysis of vertebral fractures
	Pre- & post-injury biomechanics	Non-fracture spinal conditions
	Experimental analysis	Surgical treatments of fractures
	FE models	Non-English articles
	Human and animal studies	Repeats of articles
		Full text or PDF unavailable

2.2.1 Data extraction

Included articles obtained were grouped by methodology into experimental and computational techniques, outlined in Table 4 and 5. Global variables extracted: author surname, publication year, study purpose, experimental design, subject model (target population, sample number, region and condition of interest, anatomical structures), loading modes and data collection, main findings, and limitations. Extracted information was sectioned into secondary groups of fracture causation, location, morphology, and progression and will include variables such as spinal structures RoM, stress distributions, and spinal loading. Data was recorded in Microsoft Word and Excel.

2.2.3 Quality & bias assessment

Methodological quality and bias assessments were completed by one author (P.K) and required a series of tools. Computational articles were assessed using Methodological Quality assessment of Subject-Specific Finite Element analysis for computational orthopaedics

(MQSSFE)(96). Experimental articles were analysed using the JBI critical appraisal checklist for quasi-experimental studies(97).

2.3 Results

2.3.1 Search results

The search results are outlined in Figure 8 where a total of 732 papers were extracted from three databases. This number was reduced to 419 papers after excluding “fixation” as a keyword. Vertebral fusion/ fixation is a prevalent topic in lysis biomechanical research but was not considered within the scope of this study, as the aim was to analyse the original, unaltered biomechanics of lysis. During the title and abstract screening process any duplicate studies were removed. Studies discussing only the topic of spondylolisthesis were also excluded if there was no mention of lysis or vertebral posterior arch fractures. Further articles were removed if there wasn't sufficient biomechanical analysis of lysis, such as epidemiological studies, pain management techniques or case reports. As a result of this screening, 389 articles were excluded. After a final full-text article assessment, the final review included 20 studies from three databases: PubMed (n= 9), ScienceDirect (n= 10), and Scopus (n= 1). The studies were then separated into 2 categories: Experimental Analysis (n=10) and Computational Analysis (n= 10).

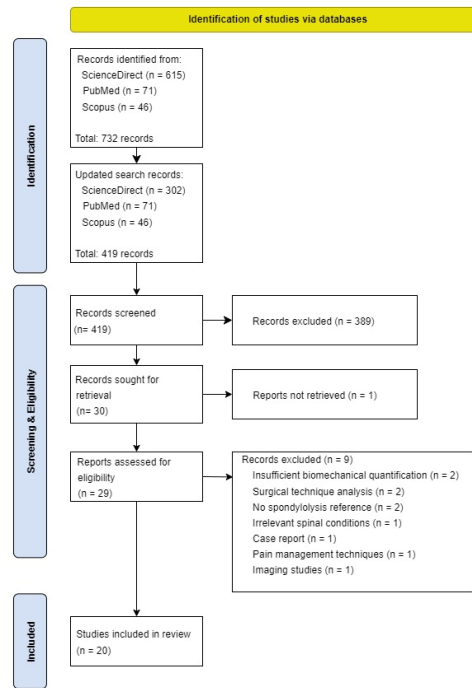


Figure 8. A flow diagram of the study selection process outlining each screening section using the PRISMA 2020 flow diagram guidelines for reporting a systematic review.

2.3.2 Experimental method analysis

The 10 studies in this section(98–107) analysing lysis biomechanics through experimental methods are outlined in Table 4. The studies used subjects with no pathology (n=6), existing (n=2), or induced vertebral posterior arch defects (n=6) to analyse lysis. The region of interest included the lumbar spine (n=8) or lumbopelvic hip complex (LPHC) (n=2), with model complexity ranging from in situ living subjects (n=3), to isolated human (n=2) or animal spinal segments (n=6).

Table 4: Methodological analysis of experimental studies

Author, Year	Purpose	Experimental design							Main findings	Main weaknesses
		Research method	Target population	Sample number	Area of interest	SP presentation	Measured parameters	Loading technique and modes	Data collection	

Nagamoto, 2021	Analysing the relationship between sports, fracture presentation and hip mobility.	ROM analysis	Human Athletes in rotational and non-rotational sports Age: 15.3 ± 1.7 years Gender: 44m/10f	54	LPHC PI	Unilateral SP: 20 Bilateral SP: 34 L3-L5 defects.	Fracture angles. Hip internal rotation RoM	NA	CT imaging	Rotational sports group had greater percentage of rotation-type fractures and larger fracture angles, but smaller hip internal rotation range of motion compared to non-rotational sports group	Selection bias due to unknown SP prevalence in each sport Exclusion of asymptomatic cases Hip rotation measurements vary by method Patient activity variation
Colloca, 2012	Analysing SP progression & movement (6 mo., no intervention)	In vivo biomechanical testing	Live animals Merino sheep Age: 4-24 months	12	Lumbar spine	Induced bilateral L5 terminal defects (4mm)	Spinal stiffness Spinal displacement	Dynamic mechanical testing apparatus with indenter rod. PA forces at chiropractic spinal manipulation levels (2.0, 6.0, & 12 Hz; -13 to -48 N for stiffness and 80N for displacement)	Data acquisition system: Force and displacement data recorded at 2500 Hz Histology CT imaging	Increase in dynamic spinal stiffness, as well as reductions in vertebral displacements occurring in response to spinal manipulation	Anesthetised specimens may have differed mechanical responses PI is wider in sheep than in humans.
Chamoli, 2014	The effect of bilateral SP defects on spinal RoM.	In vitro biomechanical testing	Cadaver kangaroo specimen	14	L1-L6	Intact specimens Bilateral DM: L4 induced defects (2mm)	Global and segmental RoM	Six-degrees-of-freedom kinematic spine simulator Bending moments of ±5 Nm: Flex-Ext, LB, AR testing modes	Motion capture: Pedicles digitised and tracked with infrared markers. Interpedicular kinematic metrics recording changes in motion.	L4-L5 motion and interpedicular displacement significantly increased after defect creation, other translations remained unchanged.	Only single and not multi-motion analysis.
Bisschop, 2014	Effect of laminectomy on segmental motion and stiffness for predicting resultant SP.	In vitro biomechanical testing	Cadaver human specimen Age: 50-90 years	12	L1-L5	Intact specimens L2 or L4 laminectomy.	Global and segmental RoM and stiffness.	Hydraulic materials testing Machine. Preload compression: 250N Bending moments of ±4 Nm: Flex-Ext, LB and AR testing modes	Motion capture: Force and displacement data digitised at 100 Hz	Laminectomy increased RoM at the level of surgery but only affected LB of adjacent segments. Stiffness was not affected.	Limited sample size forced segment variation. Excluded faulty data and used lower force to protect potentially weak spines (unusual for target population).

Gallagher, 2012	A review of lumbar spine shear tolerance and recommendations of pre-fracture shear force limits.	Literature review and statistical analysis	Cadaver human specimen Cadaver porcine specimen	NA	Lumbar spine	Intact spines, SP & Spondylolisthesis	Shear tolerance Fatigue failure	NA	Review Weibull analysis	Workers can withstand shearing forces up to 1000N. But for repetitive tasks a lower limit of 700N is recommended for most people's safety.	Excluding existing damage, complex loading, and shear-compression interaction. Human fatigue data is scarce with poor female representation.
Howarth, 2013	Quantifying shear force to fatigue life for SP prediction.	In vitro biomechanical testing	Cadaver porcine specimen	32	Lumbar spinal units Using 16 C3–C4 and 16 C5–C6 porcine vertebrae	Intact specimens	Cumulative shear force Number of cycles to failure Fracture Patterns	Servohydraulic material testing system. Compression (300N) and Flex-Ext. Cyclic shear (20%, 40%, 60%, 80% of failure tolerance, 1 Hz) Until failure or 21,600 cycles	Xray imaging Shear displacement linear potentiometer Shear force: 2 uniaxial load cells Ultimate shear failure force: vertebral measurement calculations	Higher forces led to fewer cycles before failure. Injury potential increases at 20-40% cumulative shear but comparable at 60-80% Greater load magnitudes led to more variable fracture patterns.	For safety testing stopped early pig before total vertebral fracture. Porcine specimens have weaker PI and had differing results to human lumbar spinal data.
Bright, 2018	Investigation of the asymmetry of facet joints and stress induced SP.	In vitro biomechanical testing	Cadaver porcine specimen Age: 6 months	21	Lumbar vertebrae L3, L4 & L5	Intact specimens Induced tropism	Fatigue cycles, strain, creep, secant modulus and hysteresis	Compression testing apparatus, 2 methods: With gauges: compressive cyclic load (2 Hz), amplitude ($\pm 50\%$), set to target average strain of $1500 \mu\epsilon$ Without gauges: -400 ± 200 N load. Asymmetrical load offsets (0° , 18° , 28°) mimicking tropism	6 3-element stacked rosette strain gauges	Facet angle variations cause uneven stress on facet joints, leading to higher strain on one side. Increased strain leads to fracture, observed through higher creep rate and increased toughening mechanisms.	Several vertebrae damaged in butchering process and had to be excluded Testing limited to compression forces. Narrow range of angles explored.
Beadon, 2008	Producing a biomechanical model of spondylolysis and spondylolisthesis	In vitro biomechanical testing	Cadaver porcine specimen Age: 6-8 months	5	Lumbar spinal units L6-S1	Induced SP: notches in the PI, facet joints and anterior AF	Conditions to initiate SP and spondylolisthesis and injury	Actuator paired with a material testing device Cyclic: 2 Hz, 300-600 N, 7200 cycles	Optical markers for 3D motion. Visual inspection for vertebral or disc changes.	Pig spines fractured bilaterally at the PI, EP and facet joints remained intact.	Spine movement and crack visualisation limitations may have led to overestimating the number of cycles before failure

							presentati on: Cycles to failure, impulse, compressi on & cyclic shear loads	Cyclic + Impulse: 2 Hz, 1500 N impulse/1200 cycles Elevated Cyclic: 2 Hz, 500-800 N		Model mimicked a moderate human spinal slippage (Grade II spondylolisthesis) with an average slippage of 33%.	
Konz, 2001	Identifying the pathomechanisms of SP and spondylolisthesis	In vitro biomechanical testing	Cadaver immature Chacma baboon specimen Age: 2-3 years Gender: 4m/3f	12	Lumbar spinal units L5/L6-S1	Induced SP: Bilateral Pars defects on the superior vertebrae Intact & incised IVD groups	Load- displacement curves Failure load and location Ultimate displacement	Uniaxial MTS machine Vertebral bodies tested under A-P shear force (25.0 mm/min) to failure.	Radiographic scans Histological Analysis	Fractures occurred in the growth plate, regardless of disc integrity. Steeper facet angles increased the load the growth plate carried, but weaker PI sections didn't significantly impact overall load distribution.	Limited availability of the specimens Baboon model has limitations to facet representation
Alway, 2021	Analysing the relationship between LBSI and Cricket bowling	RoM analysis	Human Athletes in rotational and non- rotational sports Age: 18.9 ± 1.9 years	58	LPHC	Participants with and without LBSI (39 with/ 11 without)	LPFC angles, orientation, flexion and rotation Kinetic Parameters: force, loading rate, impulse, velocity	Bowlers performed maximal velocity deliveries	18-camera Vicon Motion Analysis System (300Hz) 47 14mm reflective markers Kistler force plate (1500Hz) Radiographic scans	Bowlers with LBSI displayed different back, hip, and spine postures compared to healthy bowlers at different time points in a single throw.	Discrete rather than continuous data measurements.

2.3.3 Computational model and method analysis

The computational biomechanics section contains 10 studies(67,108–116) which are outlined in Table 5. These studies analysed models with no lysis pathology (n= 9), artificially created unilateral defects (n=4), or bilateral defects (n=2). All studies (n=9) investigated the models under loading conditions of extension, flexion, LB, AR, or axial compression. An additional study was included(113), to aid with quality analysis of the included computational models.

One article(106) included experimental and computational methods and therefore will be discussed in both sections.

2.3.3.1 Spinal & Bony Structure

All 10 models considered the lumbar spine, but there was a wide variation in the complexity of model structures. 8/10 models included the L5 vertebra in their selection, 4 models included 2-3 vertebra level spinal units, 5 models included the sacrum in their models, 1 model analysed a single vertebra. 9/10 models considered both the cortical and cancellous bone types considering the structures as linear orthotropic hexahedral elements, where the cortical thickness is 3mm. The material properties of the bone structures were either applied to the whole geometry or a third material property was considered for the posterior elements of the arch.

2.3.3.2 Intervertebral Discs

Nine of the lysis models detailed intervertebral disc creation in which all models included the nucleus pulposus and annulus fibrosus with only three models including the endplates. Disc model geometries were constructed either through extraction using 3D imaging segmentation software from CT imaging, or through reconstruction using the vertebral bodies or endplates as templates. From the studies that detailed the nucleus pulposus' construction, the geometry was constructed with hexahedral elements and given the material behaviours: incompressible or quasi-incompressible and low stiffness. However, all models did not go into detail about the material models and laws used for the incompressible models. The annulus fibrosus was most commonly modelled as a composite material with fibres, cables or link elements positioned at 30-degree angles embedded in ground substance, with ground substance modelled as solid elements. The fibres material behaviours were modelled as either linear, bilinear or non-linear and tension resistant, however apart from this designation no detail was given about the laws or equations used. The number of layers of the annulus fibrosus was discussed by three studies and ranged from 2-14. Endplates were only briefly discussed in two studies and were modelled as hexahedral elements with no further detail.

2.3.3.3 Ligaments

Six lysis models included all seven spinal lumbar ligaments (ALL PLL ITL LF FCL SL ISL) with only one model that included the ALL, PLL, ITL, LF and CL. The ligaments were modelled as either cable elements or 3D truss elements (also described as link elements depending on the software) and were arranged and attached to anatomically accurate structures. The ligament's primary function is to resist tension in the direction of the fibre orientation to ensure stability of the spine. Tension resistance was accounted for in the choice of either cable or truss elements. Choice of the type of element was not elaborated on but were likely influenced due to software constraints. Cable elements are ideal to model tension resistance, while truss elements can simulate cables using tension-only non-linear properties.

Ligament geometry description was limited with only two models specifying the number of elements per ligament. The number of ligaments is a parameter that influences the range of motion and flexibility of the spine as well as effecting model accuracy and convergence. Five models also describe the ligament area, an important parameter for defining the mechanical properties.

All ligament model's material behaviours were described as non-linear/bilinear and elastic/hypoelastic. However, there was a lack of detailed information on the specific linearity laws applied, which affect the accuracy of the ligament mechanics. The information on material behaviours and properties were provided in tables, derived from literature rather than experimental evaluation. Additionally, material properties were sourced from multiple references, which affected the level of detail provided for each ligament. Models included ligaments with values that varied depending on the spinal level, while others described whole ligaments only in terms of non-linear elasticity. This variability can significantly impact the quality and reliability of patient-specific models, as ligament properties can differ based on individual patient characteristics.

2.3.3.4 Facet Joints

Lysis models in this review that included the facet joints in their spinal models, defined the structures as 3D contact problems (n=8) using either gap (n=5) or surface-to-surface (n=2) contact elements. Gap elements simulate contact when the two surfaces become within a set distance, where during contact the stiffness reaches a set value and when the surfaces move

away the stiffness tends towards zero(117). Gap elements are often used for bilinear or friction problems, however in this review none of the model's considered friction between the surfaces. The frictionless properties simulate the two articular surfaces lined with cartilaginous layers to enable motion between the spinal joints(13). No models included cartilaginous tissue, meniscoid or synovial fold structures that are found in the facet joints, but rather replicated their behaviour through defining specific material behaviours, including the terms "frictionless" and "non-linear" behaviour. Four studies in this review stated that their facet joint models were non-linear but did not elaborate further on the type of non-linear models or material laws used. Material values such as Poisson's ratio and Young's modulus were also not unique to the facet joint structures and consisted of either the properties of the cortical bone or the posterior arch region. All models in this review did not elaborate on the facet joint construction either due to using previously defined models or software defined features.

2.3.3.5 Spondylolysis

The spondylolysis defects were modelled as either unilateral (n=5) or bilateral (n=2) in the models. The methods for creating the defects included selecting a group of elements around the pars interarticularis, also referred to as an 'oblique direction through the isthmus,' or by defining a specific line and width (0.1 mm) to represent the fracture. These elements were either deleted to simulate a complete fracture, or used to simulate a contact problem between the surfaces. All models were assumed to represent a segmental fracture, this meant spondylolysis was only studied during the terminal phase assuming no material between the surfaces. Other fracture stages such as oedema, early or progressive lysis were not analysed.

Table 5: Methodological analysis of computational studies

Author, Year	Computational model purpose	Complexity of model(s)						Model focus	Loading modes	Main findings	Main weaknesses
		Full model	Origin	Bony structure	IVD	Ligaments	Facet joints				

Haj-Ali, 2019	Generating models to predict subsequent lumbar fractures after unilateral SP	L1-S1 ligamentous spine Nodal tie constraints between disc and vertebra 275,570 elements 101,558 nodes	CT scans of a 53-year-old-female spine with no pathology.	6 vertebrae Cortical & cancellous bone as hexahedral & tetrahedral elements	5 IVDs Hexahedral and tetrahedral solid elements	Anterior longitudinal ligament (ALL) Posterior longitudinal ligament (PLL) Intertransverse ligament (ITL) Ligamentum flavum (LF) Capsular ligament (CL) Interspinous ligament (ISL) Supraspinous ligament (SSL) Non-Linear Manually added	Nonlinear, frictionless 3D contact problem	Intact Model: No pathology Unilateral defect model: L5 left PI defect, free edge. Bilateral defect model: L5 defect in both L5 PI	10Nm moments applied to superior surface of L1: Flex Ext RLB LLB RAR LAR Inferior surface of S1 fixed in all directions.	Unilateral DM: increased stress on contralateral L5 PI during flexion, left LB, right AR. Bilateral DM: increased RoM in lumbar spine levels, not just next to defect. The ligaments have a strong influence on RoM/ flexibility of the spine.	Material properties iteratively selected from one study to match experimental results and reproduce RoM loading modes.
Chosa, 2004	Evaluating stress distribution in the PI under lumbar loading	L4-L5 ligamentous spine 11194 elements	CT scans of a 29-year-old male with no pathology	2 vertebrae Hexahedral solid elements for cortical and cancellous bone	1 IVD Composite material with fibres in ground substance. AF: nonlinear, 3D cable elements in tension, 8 fibre layers at 30° angles to the horizontal (criss-cross) make up 20% of the disc. EP: hexahedral solid elements.	ALL, PLL, ITL, LF, CL, ISL, SSL Non-linear cable elements	3D contact problem, nonlinear, frictionless 180 gap elements on L4 inferior and 144 contact elements on L5 superior surface	Intact Model: No pathology	Compression: Axial force 0-500N on central L4 superior surface over 15 increments. Combined loading with compression over 15 increments: Flex/Ext: anterior/posterior axial forces on L4 up to 15 Nm. LB/AR: Paired forces on lateral ends of L4 up to 15 Nm.	Highest stress at thinnest point of superior PI (dorsal) and similar location (ventral). Highest stress under compression + extension > rotation > flexion > LB > compression alone.	Effects of bone remodelling not considered. Muscle forces not considered. Viscoelastic properties of disc material not considered. Only single and not multi-motion analysis.
El-Rich, 2009	Analysing the biomechanics of lysis induction in a	L5-Sacrum	CT scans of a cadaver	L5 and Sacrum	1 IVD Manually generated	ALL, PLL, ITL, LF, CL	3D-contact problem, non-linear, frictionless,	Listhesis Model with no SP.	3 conditions: 1) 500 N axial force	In an altered spondylolisthesis geometry, stress concentrations	Disc neglects time dependent response: intradiscal

	spondylolisthes is model		ic intact specim en combin ed with scans of Xrays of a 14-year-old male with grade-1 isthmic spondyl olisthes is	3mm thick EP and cortical bone. Cortical and cancellous bone: linear, orthotropic material	volume from EP geometry. NP and AF: 3D hexahedral isotopic solid elements. AF: fibres AS 3D bilinear cable elements with tension resistance, 2 layered at 30° angles to the horizontal (criss-cross) NP: quasi-incompressible, elastic	Bi-linear cable elements Tension resistance Isotropic	Surface-to-surface contact elements Closed gap or 2 mm inter-facet-surface distance used.		2) 500 N axial force combined with 15 Nm Flex moment 3) 500 N axial force combined with 15 Nm Ext moment Axial forces applied to superior L5 EP. Flex-Ext moments applied to anterior and posterior L5 end plate face. Sacrum fixed at femoral head region	form in the pedicle, PI and AF, and are potentially responsible for SP induction.	pressure, porosity and viscoelasticity. Relaxation and daily cycle responses of the spine not included. Spinal muscles forces not included. Effects of lumbar or thoracolumbar integrity not included. Personalised patient properties, with respect to age and spinal tissue properties excluded. Long term loading responses excluded
Jia-can, 2009	Exploring mechanical behaviour of the lumbar spine to provide a base knowledge for fracture classification.	L4-Sacrum 121 239 nodes, 112 491 units	CT scans of a 39-year-old female no pathology.	NS	NS	NA	NS	Intact Model: No pathology	Impact load of 0-8000N over 40minutes: Rising time:10 minutes & Falling time: 30 minutes Ext, Flex, LB: Axial compressive stress on superior/inferior surfaces of L4 and sacrum.	Stress transmits from the articular processes to the vertebral body under axial compression	Did not analyse effects of ligaments or muscles. Could not obtain model characteristics stated in papers where pdf is not available
İnceoglu, 2012	Evaluating stress	Single vertebra L3	CT scans	Cortical and	NA	NA	NA	Intact Model:	Extension: Uniform	Unilateral DM:	Vertebrae material

	distribution in the pedicles before and after PI fractures using a multi-scale model.	24895 tetrahedral elements for vertebra 24560 elements for internal pedicle model.	of a human spine for outer bony structures and micro-CT internal structures.	cancellous bone of pedicle modelled in 5 layers. Rigid body constraints. Hexahedral elements used for bones. Internal and external structures connected with tie constraints.				No pathology Unilateral Defect Model: PI fracture created by removing 310 elements.	distributed loads applied to superior and inferior facet joints (1N/mm ²) and neural arch (2N/mm ²) to represent non-homogenous pressure	Under extension increased stress in contralateral PI and pedicle, especially around outer boundaries.	properties not from a single patient (age, gender, and tissue variations). Soft tissue effects excluded. Global structure effects not captured.
Teraï, 2010	Evaluating the site of origin of spondylolytic fractures.	L3-S1 ligamentous spine	CT scans of a cadaveric intact specimen	6 vertebrae Cortical and cancellous bone as hexahedral elements.	3 IVD Composite material with fibres in ground substance. AF: fibres embedded at 30° angles to the horizontal in concentric rings. NP: hexagonal elements, low stiffness and near incompressibility, pseudofluid.	ALL, PLL, ITL, LF, CL, ISL, SSL Elastic, 3D 2-node truss elements, non-linear properties, hypoelastic material	3D gap contact elements to represent cartilaginous articular surface	Intact Model: No pathology	Axial compression of 400N combined with 10.6Nm moments: Flex Ext LB AR	Highest stress distribution in the caudal-ventral PI for all movements except flexion.	Cadaveric spine used in intact model which has very different biological properties to analysis group
Wang, 2006	Evaluating the biomechanical changes in the lumbar bony and soft tissue structures in an intact, unilateral and bilateral SP model.	L2-L4 lumbar ligamentous spine 4804 elements and 3911 nodes	CT scans of a 19-year-old male with no pathology.	3 vertebrae Cortical & cancellous modelled with hexahedral solid elements.	2 IVD AF: fibres embedded in ground substance, two-node link elements, resistance to tension at 30° angles to the horizontal.	ALL, PLL, ITL, LF, CL, ISL, SSL Homogenous and isotropic, two-node link elements with tension resistance.	3D nonlinear contact problem Surface contact elements	Intact model: No pathology Unilateral defect model: L3 right posterior PI fracture Elements: 4816	10Nm moments applied to superior surface of L2: Flex Ext RLB LLB RAR LAR	Unilateral DM: Maintains stability, but increased stress on contralateral PI. Bilateral DM: Increased rotation angle, vertebral displacement, disc stress, and endplate stress under extension/torsion.	Only single and not multi-motion analysis. Multiple sources of material properties for one spine. The effects of muscle force,

					EP: hexahedral solid elements.			Nodes: 3906 Bilateral defect model: L3 posterior PI fractures on both left and right sides. Elements: 4830 Nodes: 3919 Fractures: Contact problem with node-to-surface contact elements.			orthotropic behaviour of the vertebral body, nonlinear properties of ligaments, and viscoelastic properties of the disc were not considered in this FE model.
Sairyo, 2006	Evaluating the pathomechanics of stress in the pedicles as an SP indicator	L3-L5 ligamentous spine	CT scans of a human cadaveric specimen	3 vertebrae Cortical and cancellous bone as 3D hexahedral elements.	2 IVD Composite material with fibres in ground substance. AF: fibres embedded at 30° angles to the horizontal in concentric rings. NP: hexagonal elements, low stiffness and near incompressibility, pseudofluid.	ALL, PLL, ITL, LF, CL, ISL, SSL Elastic, 3D 2-node truss elements, non-linear properties, hypoelastic material	3D gap contact elements to represent cartilaginous articular surface.	Intact Model: No pathology	Axial compression of 400N combined with 10.6Nm moments: Flex Ext LB AR	Under loading highest stress in PI, followed by pedicle under all loading conditions.	Cadaveric spine used in intact model which has very different biological properties to analysis group
Sairyo, 2005	Modelling biomechanical rationale for SP and the pathomechanisms.	L3-S1 ligamentous spine	CT scans of a human cadaveric specimen	4 vertebrae Cortical and cancellous bone as 3D hexagonal elements.	3 IVD Composite material with fibres in ground substance. AF: fibres embedded at	ALL, PLL, ITL, LF, CL, ISL, SSL Elastic, 3D 2-node truss elements, non-linear properties,	3D gap contact elements to represent cartilaginous articular surface.	Intact Model: No pathology Unilateral defect model: L5 left PI defect,	Axial compression of 400N combined with 10.6Nm moments: Flex Ext LB AR	Unilateral DM: Increased stress in contralateral PI (up to 12.6x compared to intact spine).	The facet joint elements were able to transmit only compression. Cadaveric spine used in intact model which has very

					30° angles to the horizontal in concentric rings. NP: as hexagonal elements, low stiffness and near incompressibility, pseudofluid.	hypoelastic material		0.1mm crack.			different biological properties to analysis group
Konz, 2001	Analysing the effects of facet orientation and PI thickness on SP and spondylolisthes is	L4-L5 immature ligamentous spine	CT scans of a cadaveric ligamentous immature baboon specimen (T12-S1)	2 vertebrae Cortical and cancellous bone as 3D isoparametric hexahedral elements, divided by element boundaries GP: 3D isoparametric 8-nodal elements	1 IVD Composite material with fibres in ground substance. AF: fibre reinforced concrete elements, collagen fibres in 8 axial & 6 radial layers, make up 20% of the disc. Only allows for tension. NP: incompressible	Hypoelastic, 3D cable elements, Only allows for tension.	3D interface/gap elements Only allows for normal compression	Intact Model Thin PI model: surface elements for one PI halved, weakened GP with reduced GP-AF nodes Unilateral defect model: elements removed at PI	50N and 100N A-P shear force	A reduced thickness of the PI increases the likelihood of spondylolytic fractures. More sagittal oriented facets lead to listhesis progression.	Lack of human material properties for modelling for IVDs, ligaments and GP. Baboon facet orientation does not give suitable representation for GP analysis.

2.2.4 Quality & bias study

2.2.4.1 Computational

For the computational studies, quality and bias analysis was performed using the MQSSF tool(96) and Jones' (2008) framework(113) for assessing study design, verification, and validation. The results are outlined in Table 6 with an average quality score of 60%.

Complete evaluation of certain FE models(112) proved impossible due to the inaccessibility of foundational papers outlining the model's construction.

The quality assessment parameters include verification, validation, and comprehensive sensitivity analysis. All studies met the verification criteria by using commercial FE software, ensuring numerical precision in meshing, processing of geometries and parameters, and mathematical solutions. Sensitivity analysis criteria are satisfied through the incorporation of real-world experimental and patient data into the models. The studies construct spinal units or vertebrae using CT images from patients or cadavers, with subsequent processing in the software to meet study objectives. Models are adjusted to include soft tissue structures not captured by CT, such as ligaments, or use combinations of micro-CT and CT for differentiating cortical and cancellous bone geometries. Model material data in each study is obtained through literature search and previous FE spinal studies. This is appropriate for general population biomechanics but poses limitations for patient-specific studies. Validation of these studies necessitates experimental techniques to obtain accurate material properties. Validation is particularly challenging for complex models, as seen in studies(109,110,115,116), involving multi-layered spinal units with intricate interactions and parameters. These models require meticulous mesh construction and appropriate material allocation, either on an element-by-element or structural basis, a process that is both time-consuming and computationally demanding. Simplification is often necessary for feasibility and cost-effectiveness, without compromising quality. The majority of studies focus on models with 2-3 vertebral layers(67,108,109,112,116), with one study(111) examining a single vertebra. Sensitivity analysis in these models is less emphasised, as parameters are frequently derived from prior studies. However, some studies(67,114,115) reinforce their findings with clinical data, enhancing the reliability of their models.

Table 6: MQSSFE tool: Computational quality and bias analysis(96), criteria outlined in original article

+ Sufficient evidence, – Insufficient evidence, / not relevant

REF.	HAIJ-ALI, 2019	CHOSA, 2004	EL-RICH, 2009	JIA-CAN, 2009	İNCEOĞLU, 2012	TERAI, 2010	WANG, 2006	SAIRYO, 2006	SAIRYO, 2005	KONZ, 2001	TOTAL SCORE (%)
STUDY DESIGN											
1	+	+	+	+	+	+	+	+	+	+	100
2	+	+	+	+	+	+	+	+	+	+	100
3	+	+	+	+	+	+	+	+	+	+	100

4	+	+	+	+	+	+	+	+	+	+	100
5	-	+	-	+	-	-	-	-	-	-	20
6	+	+	+	+	+	+	+	+	+	+	100
7	+	+	+	+	+	+	+	+	+	+	100
8	/	/	-	/	-	-	/	/	+	/	25
SUBJECT RECRUITMENT											
9	+	+	+	/	+	+	+	+	+	+	100
10	-	-	+	-	-	-	-	-	-	+	20
11	-	+	+	+	-	+	-	+	-	-	50
12	+	/	+	/	+	+	+	+	+	+	100
MODEL CONSTRUCTION & CONFIGURATION											
13	+	+	+	/	+	+	+	+	+	+	100
14	+	+	+	/	+	+	-	+	+	+	88.9
15	-	-	+	/	-	-	-	-	-	-	11.1
16	+	+	+	/	+	+	+	+	+	+	100
17	+	+	+	/	+	+	+	+	+	+	100
18	+	+	+	/	+	+	+	+	+	+	100
19	+	+	+	/	+	-	+	-	-	-	55.6
20	+	-	+	/	+	-	-	-	-	+	44.4
BOUNDARY AND LOADING CONDITIONS											
21	+	+	+	+	+	+	+	+	+	+	100
22	+	+	+	+	+	+	+	+	+	+	100
23	-	-	-	-	-	-	-	-	-	-	0
24	-	+	+	-	-	-	-	-	-	-	20
25	-	-	-	-	+	-	-	-	-	-	10
26	+	+	+	+	+	+	+	+	+	-	90
MODEL VERIFICATION AND VALIDATION											
27	+	-	+	/	-	-	-	-	-	-	22.2
28	+	-	+	/	-	-	-	-	-	-	22.2
29	+	-	+	/	+	-	+	+	+	+	77.8
30	+	+	+	/	+	-	/	-	-	-	50
31	+	+	+	/	+	-	+	+	-	+	77.8

MODEL ASSUMPTION AND VALIDITY											
32	+	+	+	/	+	-	+	-	-	+	66.7
33	-	+	+	/	-	-	+	-	+	-	44.4
34	+	-	+	/	-	-	+	-	-	-	33.3
35	-	-	+	/	+	-	-	-	-	+	33.3
36	+	-	+	/	-	-	-	-	-	-	22.2
37	+	+	+	+	+	+	-	+	+	+	90
Total (%)	75	68.6	89.2	75	67.6	48.7	60	55.6	54	61.1	<u>60</u>

2.2.4.2 Experimental

Quality and bias analysis for the experimental studies were analysed in Table 7 with an average score of 80.8%. The JBI critical appraisal checklist(97) provided a broad framework to cover all types of experimental study, however the analysis was less detailed than the computational tool.

Table 7: JBI Critical Appraisal Checklist: Experimental quality and bias analysis, Ref. criteria outlined in original article(97)

+ Sufficient evidence, – Insufficient evidence, / not relevant

REF.	NAGA MOTO , 2021	COLLOC A, 2012	CHAMOL I, 2014	BISSCHOP , 2014	GALLAGHE R, 2012	HOWART H, 2013	BRIGHT , 2018	BEADON , 2008	KONZ, 2001	ALWAY, 2021	TOTAL SCORE
1	+	+	+	+	/	+	+	+	+	+	100
2	+	+	+	+	/	+	+	+	+	+	100
3	/	+	+	+	/	+	+	+	+	/	100
4	-	+	+	+	/	-	+	-	+	+	66.7
5	/	-	+	+	/	/	+	-	-	+	66.7
6	/	+	/	/	/	/	/	/	/	/	100
7	+	+	+	+	/	+	+	+	+	+	100
8	+	+	+	+	/	+	+	+	+	+	100
9	+	+	+	+	/	+	+	+	-	+	88.9
Total (%)	83.3	88.9	100	100	/	85.7	100	75	75	100	<u>80.8</u>

2.4 Discussion

2.4.1 Experimental analysis of lumbar vertebral fractures

2.4.1.1 Causation

Analysing forces and loading conditions aids understanding of the underlying mechanisms for lysis. Studies(104–106) suggest that excessive anterior-posterior shear loading is the primary mechanism for lysis initiation, due to repeated exposure or exceeding a bone's ultimate tensile strength. During extension, shear forces act perpendicular to the spine upon the vertebral arch, ligaments, and intervertebral discs (IVD)(104,106). In combination with AR, additional stress is imposed upon the facet joints and its ligaments(104).

Studies(104,105) state that the limits of vertebral ultimate shear strength (USS) are 1.7kN for females and 1.9kN for males(104). Additionally, 1.325kN at 60% shear failure tolerance constitutes risk for spinal fracture(105). However, injury can develop as low as 20-40% of USS(105), potentially occurring at 30-40% USS after 1000 cycles and 40-60% USS after 100 cycles(104).

Repetitive shear loading and extension are common in rotational sports like baseball, tennis, and volleyball(98,107). These high-intensity activities require repetitive motions of Flex-Ext, LB, and AR of the LPHC. To prevent bone damage, a balanced path of kinetic energy from upper to lower extremity is required for each motion. For example, when bowling, in baseball, excessive rear hip flexion increases the likelihood of stress fractures by a factor of 4, increasing to 11 when paired with extension(98). The compromised hip motion could be a result of insufficient LPHC flexibility, requiring increased trunk rotation and extension exerting stress on the facet joints(107). An unbalanced strain on the facet joints leads to microfractures if the bone's threshold is exceeded.

2.4.1.2 Morphology

Lumbar spinal morphology affects the transmission of forces throughout the bony and soft tissue structures and hence can influence the risk of developing lysis. For studies analysing lysis biomechanics, it is important to choose a model that represents human morphology. In this review, 6 studies(99,101–103,105,106) utilised animal models where porcine specimens are the most widely used due to morphological similarities. Porcine spines have comparable

facet angles of 45° (105) but narrower PIs. The spines are also adapted to a quadrupedal loading posture and will have differing loading responses, such as increased shear failure tolerance(105).

Humans have inherent variations in their vertebrae, such as facet joint orientation. Facet joints should have symmetry ensuring equal force transmission between the PI during loading. Asymmetry (tropism) of the facets, where one is more coronally oriented, alter the loading environment of the spine, resulting in an increased strain on the less coronally oriented side(101). For a congenitally obtained defect, daily-living loading conditions may not lead to immediate fracture due to the body's capabilities to adapt to abnormal loading environments(101). However, under increased compression, fractures may develop. Increased asymmetry also exacerbates the chance of lysis, where an 18° and 28° facet angle asymmetry results in increased steady-state creep loads by 23% and 67% respectively(101), indicating that the bone is deforming at a faster rate.

Acute spinal injuries, such as cortical notching or surgeries like laminectomy and discectomy, can significantly increase the risk of lysis(105). After a laminectomy, the RoM capable by the affected vertebrae, increases by 7-12% across all 6 spinal motions(100). An approach representative of a discectomy was tested by Beadon (2008)(99) where 25% of the anterior IVD was removed. After the functional spinal unit (FSU) was subjected to complex loading, lysis was obtained in 60 minutes(99).

2.4.1.3 Location

Lysis is most commonly present at the L5 (59%(107), 33%(98)) and L4 vertebrae (33%(107), 35%(98)). All studies discussing lysis stated that the PI is the most common location for fractures followed by the pedicles. The number of fractures and exact presentation, depend on movement, morphology, and force to initiate fracture. Fracture location due to sports can be classified into rotational and non-rotational, depending on the angular velocity of rotation(107). Rotational sports produce sagittal-oriented fractures(107) with larger angles between the posterior cortex and defect ($>20^{\circ}$), whereas non-rotational fractures, with angles $<20^{\circ}$, are located closer to the pedicles. Rotational fractures are more commonly found on the non-dominant side(98,107), likely due to increased stress on the facet joints due to rotation, which is further increased with a lack of hip RoM(107). Magnitude of force can affect the predictability of fractures. Howarth's (2006)(105) analysis of shear loading to ultimate

anterior failure tolerance, found that higher percentage loads produced varying location results. For their FSU specimens, a shear load of 40-60% produced bilateral cranial PI fractures, whereas 80% shear loads, produced compound fractures with defects in the cranial and caudal PI and endplate avulsions(105).

2.4.1.4 Propagation

The presence of lysis defects has a significant increase in segmental motion during Flex-Ext, LB and AR. This was shown by Chamoli (2014)(102) who found, for a L4 bilateral lysis kangaroo spinal model, there was an increase in motion of 188.3%, 40.9% and 363.4% respectively. In the same model, global differences in motion were found in the superior vertebral segments for Flex-Ext and AR of -23.3% and +94.2%(102). An additional enlargement of the neutral zone at L4-L5 suggested spinal instability(102).

Lysis defects resulting in abnormal RoM, increase the risk of secondary defect locations. The reduced stability in the lumbar spine causes an individual to adopt compensatory motions from poor LPHC control(98), altering the loading conditions across the neural arch and producing a positive feedback loop of stress-related injury, potentially progressing the condition to bilateral fractures or spondylolisthesis.

With increased RoM of the spine and insufficient stabilisation, the chance of successful bony union decreases. Consequently, the spine employs adaptive mechanisms to stabilise the change in mechanical environment, such as stiffening. One stiffening mechanism is ankylosis of the facet joints(103). Colloca, (2012)(103) observed ankylosed stiffening of 4.48-6.28N/mm and 12.29-18.12N/mm in 6-month post-defect in vivo bilateral lysis models, after excitation of L5 vertebrae at 2Hz and 12Hz. Ankylosed structures cause long-term reductions in spinal displacement. To avoid this side effect, lysis patients should reduce excessive spinal movement or loads to encourage the body's optimum healing process and prevent fracture propagation.

2.4.2 Computational analysis of lumbar vertebral fractures

2.4.2.1 Causation

Computational studies(67,108,115) analysed the stress distribution across an intact vertebra through 6 spinal motions of Flex-Ext, LB and AR, and concluded that stresses in the PI and pedicles were highest in extension. Extension combined with compression, resulted in the

highest stress. The stress levels then decrease in order for the following; compression-AR, compression-flexion, compression-LB and compression in isolation(108). Hyperextension motions in activities of daily living are rare compared to flexion, hence the spines natural curvature favours flexion(67). Repetitive extensive loads result in fatigue fractures over time, initially starting with a bone bruise or oedema(67). Incidence of lysis is highest in athletes who experience repeated extension and rotation of their lumbar spine.

2.4.2.2 Morphology

The PI has been highlighted as the primary fracture location due to the narrowness of the isthmus. Individuals may have variations in the thickness of this bone and therefore have increased susceptibility to lysis. FE immature spinal models of an intact and artificially narrowed PI were created by Konz, (2001)(106). For the intact model, shear load (100N) was balanced between the facet joints (78%) and the growth plate (GP) (21%). For a narrowed PI, load enacted on the facet joints increased to 85% and decreased to 14% for the GPs(106). Von Mises stress in the PI also increased (2.26-3.18MPa(106)). This was the only test in geometry variation in the computational studies and therefore additional research of a wider range of vertebral morphologies would be ideal to improve population representation, aiding diagnosis, and treatment for a broad range of patients.

Conditions such as spondylolisthesis alter the alignment of the lumbar spinal column due to a slip of a vertebra. Changes in spine morphology can lead to the development of lysis causing an altered biomechanical response of the spine(109). El-Rich's (2009)(109) analysis found that after a minor L5 slip, von Mises stress in the L5 vertebrae under single axial compression and combined with flexion or extension was 9.9, 49.4 and 21.0 MPa for the PI, and 13.7, 64.1, and 32.6 MPa for the pedicle(109). The right and left facet joints experienced differing stresses of 1.2-1.6 MPa, 0.6-0.1 MPa, and 7.3-6.9 MPa respectively, due to the asymmetry of the spinal model constructed from patient data(109). The motion of flexion produced the greatest stresses likely due to the anterior positioning of the vertebrae unlike typically oriented spines, proving the importance of understanding a range of morphological models.

2.4.2.3 Location

Unlike experimental methods, computational FE methods can analyse the location of stress and strain distribution across the vertebral structures and therefore can be an effective tool for lysis prediction. Under compression, stress transfers from the facet joints to the articular

processes and then the vertebral arch(112). Analysis of the six spinal motions shows there is variation between motions and areas of highest stress. The PI experiences the highest principal stress of 2.5, 59.5, 10.7, 5.0, 14.5 and 33.8 MPa at the ventral wall and 9.6, 26.7, 10.1, 1.5, 2.2 and 28.2 MPa at the dorsal wall for flexion, extension, ipsilateral bending, contralateral bending, ipsilateral rotation, and contralateral rotation, respectively(115). These locations of stress align with CT imaging of early stress fractures where defects occur at the caudal-ventral wall(115). During extension, the ventral wall of the PI can develop microfractures due to tension during maximal elongation(115), as well as increased stress on the superior facets from contact(108). The pedicle wall also experiences elevated principal stress during the 6 spinal motions, but the spinous process, lamina and transverse process stresses are minimal(67). There is debate between four studies in this review(67,108,111,115) about the location of the highest stress during extension, however this could be attributed to the complexity of each FE spinal model, regarding number of vertebrae and complexity of soft tissue structures.

2.4.2.4 Propagation

Unlike experimental models, FE techniques can analyse stress distributions across the vertebral structures during motion. Studies compared the effects of unilateral and bilateral fractures on stress distribution(110,111,114,116) and motion analysis(110) across 6 spinal motions. After unilateral PI fracture, stress decreases in the fractured area and increases in the contralateral edges of the PI and pedicle cortical walls(111). Maximum von Mises stress is concentrated at the dorsal wall for LB and AR, and there's no significant difference for the ventral and dorsal walls for Flex-Ext(114). Unilateral models demonstrated increased contralateral stress of 15.6% for flexion, 79.3% and 105% for contralateral and ipsilateral AR, and 102% and 104% for contralateral and ipsilateral LB, respectively(116). More complex models suggest that there are greater stresses in the contralateral pars of L5 during flexion, ipsilateral LB and contralateral AR due to the lack of support structures to provide load absorption in these motions(110,114). Bilateral models show a percentage decrease in PI stress for flexion, extension, AR, and LB of 74.02%, 69.60%, 40.81%, 68.36%(116), and an increase in IVD and endplate stress(110,116). Increase in intradiscal pressure suggests that bilateral lysis will progress to IVD degeneration(110). Motion analysis(110) shows that unilateral fractures have local effects with an increase in motion only at the affected vertebrae

whereas bilateral fractures have more global effects. A L5 unilateral fracture's motion increases in the L5-S1 region during extension (20%), ipsilateral LB (18%), and contralateral AR(110) (NR%). For L5 bilateral fractures, motion is comparable to intact models until inferior to the L3-L4 region during extension, ipsilateral LB, and contralateral AR where there is a 91%, 48% and 172% increase(110). For ipsilateral AR between L2-S1 there is a maximum difference of 218% motion with an increased stiffening superior of L3-L4 and an increased flexibility inferior of L3-L4(110). A similar pattern is found for contralateral LB with a decrease in motion of 10% superior to L3-L4 and an increase in motion of 56% inferior to L3-L4(110). This change in motion and stress concentration provides pathomechanical explanation for the progression of lysis due to the significant changes in loading patterns. If the vertebral structures are unable to withstand this load during the healing process, the condition may progress to bilateral PI fractures or lysis in additional vertebral components(116).

2.5 Literature Conclusion

From the biomechanical literature that is outlined in this review, it suggests that there is an in-depth analysis of RoM and stress distribution of intact vertebrae and unilateral and bilateral lysis defects, measured extensively by computational and experimental models for loading conditions of compression, Flex-Ext, LB, and AR. Experimental models provided analysis for a series of complex testing environments from isolated mechanical testing models to patient and sport kinematic studies providing a comprehensive overview of spondylolysis. However, the analysis of animal models will provide limited representation of human biomechanics and cannot fully capture the progression of spondylolysis across the different stages of initiation. The experimental models also lacked sufficient consideration of individual factors such as age, gender, or existing spinal conditions, limiting their ability to generalise findings across varied patient populations. The computational models provided detailed biomechanical analysis of spondylolysis before and after fracture initiation, as well as the response to loading conditions in both unilateral and bilateral spondylolysis. However, the complexity of some of the spinal models was difficult to analyse due to the little detail provided for some of the structures due to software specific element types and material settings.

Despite this valuable analysis, this field has experienced a recent decrease in research. The minimal biomechanical analysis of population factors in both methodologies underscores the

need for continued research in this field. Factors like age, gender, race, body composition, lifestyle choices, existing medical conditions or resolved spinal defects could enhance our understanding of vertebral fracture pathomechanics. Another limitation was the simplicity of the fracture presentation in both methodologies, where artificial defects are used to represent lysis. Therefore, future computational analysis is needed on accurate fracture geometries across all stages of lysis alongside an increased range of fracture locations.

This systematic review provides a comprehensive analysis of the biomechanics underlying spondylolysis from a unique combined perspective of experimental and computational approaches. Combining the analysis from both techniques provides a well-rounded understanding of the conditions from multiple perspectives. The quality and bias assessment ensured included studies met acceptable standards with experimental models scoring between 75-100% and computational models between 48.7-89.2%. There was the potential for selection bias in this review due to only analysing studies in English with the specific time frames that were included.

2.6 Key Takeaways: Literature Review Model Methods

Models from this literature review directly influenced the choices made in the construction of the finite element spinal models used in this study. Listed below are the key parameters that were influenced:

2.6.1 Structure & Materials

Overall, most finite element models in this review included the components: vertebra, intervertebral discs, ligaments and facet joints to construct a model suitable for spondylolysis biomechanical analysis. Therefore, the model in this study has also included these components and likewise excluded additional lumbar spinal components such as the muscles or spinal cord. The vertebral structures, cortical and cancellous bone, and intervertebral disc components, nucleus pulposus and annular ground substance, as well as their material properties and governing laws, were directly modelled after El-Rich (2006)(109) and Haj-Ali (2019)(110) due to their high quality analysis scores and detailed spinal modelling. For the ligaments, disc fibres, facet joints, and lysis, external literature sources on non-spondylolysis-related spinal models or alternative construction methods were required due to the insufficient detail provided for modelling these structures. For instance, the models for the

ligaments did not specify the equations governing the non-linear material response to applied force, or similarly for the facet joints the type of contact model was not specified.

Spondylolysis models in this review presented as a terminal line fracture or a region of deleted elements at the pars interarticularis, and therefore did not list material properties or laws for this region. This study aimed to analyse the early-progressive phase of lysis with a variation in material parameters representing the five stages of healing, and therefore did not replicate this approach, and hence had to procure material data from external literature of bone healing, that was not available in the spondylolysis models reviewed.

2.6.2 Model outputs

The biomechanical outputs that were analysed most frequently across the models were vertebral effective stress or von Mises stress, and range of motion of the spinal segments. Effective stress is a measure of potential fracture or yielding of the bone and therefore is a useful tool for measuring the progression of a fracture or identifying secondary locations of initiation. Segmental range of motion directly relates to locations of stress where specific motions can lead to consistent presentations of stress across the posterior vertebral arch, as seen in flexion, contralateral axial rotation and ipsilateral bending. Range of motion is also a clinically measurable parameter used in spinal rehabilitation. Therefore, these parameters were chosen for analysis of the spinal models and conservative treatment framework for spondylolysis progression during healing. Additionally, comparing the identical biomechanical output variables with those from the literature review allowed for direct validation against established computational models for assessing the framework's quality.

Chapter 3: Methods

Chapter 3: Part 1, Model Construction

After consideration of the spinal components, spondylolysis and conservative treatment techniques from anatomical and biomechanical perspective, the spondylolysis conservative treatment framework could begin development. This methods chapter is divided into two parts: the steps for model construction and the steps for model analysis, including validation and the development of the conservative treatment framework.

3.1 Introduction: Finite Element Spinal Model

Two patient lumbar finite element models were constructed to identify biomechanical trends, providing evidence for a framework aimed at developing patient-specific rehabilitation protocols. The primary aim of this model was to analyse the biomechanics of the different stages of spondylolysis healing in the lumbar spine, specifically focusing on the L5 vertebra. This model is designed to simulate the condition's development, beginning with a biomechanical assessment of a healthy, intact spine, and progressing to a fractured pars interarticularis, with subsequent stages of healing including granulation, fibrous tissue, cartilage, immature bone, and mature bone. By simulating these stages, we aim to identify the physiological ranges and stress thresholds of the lumbar vertebra during different loading conditions.

To create an anatomically accurate representation of the L4-S1 vertebral segment, the following key structures were included that were previously discussed in the background section:

- Detailed vertebral geometries obtained from patient imaging
- Intervertebral disc structures
- Facet joint articulations
- Major spinal ligaments

To analyse this condition, a wide range of considerations for model complexity were made, as shown in Figure 35A, to ensure the model is suitable for assessing this condition. These considerations ensure that the model is capable of accurately assessing the biomechanics of spondylolysis while balancing computational efficiency and physiological relevance. The Figure 35A outlines the wide range of potential anatomical structures, material properties and parameters that affect boundary conditions. After consideration, the model was chosen to be an intact L4-S1 spinal model without existing pathology. This model includes the structures of the vertebral bony and intervertebral disc, facet joint contact pairs, and the seven lumbar spinal ligaments: ALL, PLL, FL, ISL, ITL, SL, FCL. The level of spinal unit of L4-S1 was chosen in order to capture the pincer effect of the superior L4 and inferior S1 facet joints on the L5 pars interarticularis during the 6 spinal motions(29). To model spondylolysis, an

artificial defect was added, with material properties adjusted to reflect distinct stages of healing.

3.2 Software

A finite element spinal model was constructed using MATLAB (MATLAB Version: 9.13.0.2080170 (R2022b) Update 1) and the open-source toolbox GIBBON (The Geometry and Image-Based Bioengineering add-On)(118). Processing tools TetGen and FEBio are included in addition to the toolbox, for tetrahedral meshing and finite element analysis.

The spinal model was originally constructed from two MATLAB scripts provided by the GIBBON toolkit repository(118) including the benchmarks

DEMO_febio_0060_vertebrae_disc_01 and DEMO_febio_0074_vertebrae_compression_01.

The DEMO_febio_0060_vertebrae_disc_01 model demonstrates a two vertebra, one-disc spinal unit which is simulated under compressive loading to the superior vertebral body. This model provides a foundation for constructing a spinal segment providing developmental steps for constructing meshes, assigning material properties, defining boundary conditions, and running FEBio simulation. The second benchmark

DEMO_febio_0074_vertebrae_compression_01, was used to build upon the foundation for mesh-based loading and boundary conditions. While the original code has been extensively modified to meet the criteria of this research, the foundational structure provided by GIBBON was a valuable template.

The models included in the demos provided the template for the construction of the spinal model with adjustments to meet the requirements of a validated spinal and lysis model. Such adjustments include new vertebral, disc and ligament geometries and contact models between the facet joints. Additional modifications and additions were made to the code to improve the flexibility and adaptability of the code for different vertebral geometries, material properties and boundary conditions. This was crucial for this study of lumbar spinal mechanics and spondylolysis fracture models.

3.3 Geometry Acquisition

The lumbar vertebral structures L4, L5 and S1 for patients 1 and 2 were acquired from patient CT imaging from the VerSe2020 database (sub-verse533 and sub-verse820)(119–121). The vertebral geometries were isolated and sectioned using the program 3D Slicer(122) (Image

Computing Platform for the Quantitative Imaging Network) and TotalSegmentator v2(123), an extension to initially isolate the three vertebrae. Post-processing 3D Slicer functions included filling holes up to 5mm in diameter and applying Gaussian smoothing to the geometries with a 1mm kernel size to improve mesh quality and continuation of the segmented regions. Care was also taken to correct any excessive selection during segmentation, with particular attention to the facet joints and articular processes to allow for smooth movement of the arch during deformation and facet contact. Replacing the original GIBBON benchmark vertebra was crucial to ensure accurate lumbar spinal geometry for lysis analysis. However, the origins and patient characteristics of the lumbar spines used in these models were unknown. This limitation constrains the patient specificity of the boundary conditions and materials of this finite element model. All other structures included in this model, such as the intervertebral discs and ligaments were generated in-code as described in the following sections, because the available scans do not provide enough clarity to accurately distinguish these structures. Once the vertebrae were obtained, they were imported into the finite element model code for spinal unit construction.

3.4 Vertebrae and Intervertebral Disc Modelling

Importing and processing of the vertebra required a flexible method and framework capable of handling multiple vertebra geometries. The process of importing the vertebrae began by applying a rotation in the XY plane to achieve correct anatomical positioning. This was done by aligning the centroids of the vertebral bodies and using a rotation matrix to ensure that the inferior and superior faces of the vertebral bodies of L4, L5, and S1 were parallel. The vertebrae were aligned in the z axis with a spacing for disc height and to allow for spacing between the articular processes. Disc heights were chosen for the two patients to be 11.6mm and 11mm for the L4L5 disc and 10.7mm and 10mm for the L5S1 disc. These heights were selected from literature(124) paired with inspection of the source CT scan data. The S1 vertebra required an additional x-axis offset to maintain the lordotic shape with a distance (2-4.5mm) which was selected from measurements from the CT scans. The importing process was optimised for model flexibility and speed in which the vertebral centroids were selected using the coordinates of the external vertebral anterior, posterior, superior, and inferior walls. This process ensured that multiple vertebral models could be imported and saved for simulation.

The vertebrae were remeshed to provide a surface mesh with the Geogram remesh GIBBON plugin using a defined variable named point spacing (P_s) to control the distance between the vertices (mm) in the remeshed surface. The meshing program produced triangles of approximate uniformity whilst maintaining detail of the surface detail of the vertebrae. The point spacing variable was used to refine and coarsen the mesh for the h-method convergence studies.

Intervertebral discs geometries L4L5 and L5S1 were manually constructed based on vertebral geometry, due to a lack of clarity of the scans to give acceptable geometry extraction. The intervertebral disc model formation was maintained from the GIBBON benchmark [DEMO_febio_0060_vertebrae_disc_01]. The process of creating the disc involved a series of steps involving Bézier curve generation and surface sweeping techniques.

Each disc was formed from the boundary curves of the superior and inferior vertebral bodies neighbouring the disc. Vertebral body faces were selected using angular and distance thresholds specific to each vertebra to assign logical masks to the vertebral body face groups. The distance threshold was selected to replicate the superior width and depth (SVBD, SVBW) and inferior width and depth (IVBW, IVBD) of the vertebral body faces. The angular thresholds restricted the selection area to account for fluctuations in the geometries and ensure precise area selection. The dimensions of the inferior and superior faces used to construct the discs are outlined in Table 8. The distance and angular thresholds were selected in order to align with disc dimensions from literature(124). The disc face selections were remeshed to form a smooth disc perimeter and ensure connectivity between the disc and vertebra.

Table 8: Disc dimension parameters for the two lumbar discs.

Dimension (mm)	Patient 1		Patient 2	
	L4/L5	L5/S1	L4/L5	L5/S1
IVBD	32.9 (L5)	27.646 (S1)	26.79 (L5)	20.23 (S1)
IVBW	49.07 (L5)	48.26 (S1)	43.43 (L5)	42.4 (S1)
SVBD	34.14 (L4)	31 (L5)	28.98 (L4)	27.49 (L5)
SVBW	47.82 (L4)	46.73 (L5)	47.53 (L4)	44.60 (L5)
Disc height	11.6	10.7	11	10

To establish orientation of the disc, the normalised average ($\hat{\mathbf{n}}$) of the normal direction vectors of the inferior face of the superior vertebral body (N11) and the superior face of the inferior vertebral body (N22) was calculated using Equation 3.4.1-3.4.2. To construct and guide the disc shape, the central axis of each disc was defined for a Bézier curve, using control points. The control points included the superior (P11) and inferior points (P22) of the disc and two intermediate points, outlined in Equation 3.4.4. The intermediate control points were positioned along the average normal vector offset by a factor (f) that incorporates the disc height (h_{disc}) and a slope (m) gradient (Equation 3.4.3). The Bézier curve (B(u) where “u” represents curve progression) was generated from the parametric Equation 3.4.5, where each point on the curve is computed as a weighted sum of the control points. The curve was determined by the number of steps of the curve (n_s), a value determined by the height of the disc and the distance between vertices and the intermediate points (p) (Equation 3.4.5). The disc volume was created using sweeping operations formed from the boundary curves and the Bézier curve path. The volume created was converted from a quadrilateral mesh to a triangular mesh. To improve the mesh quality, smoothing operations were applied to remove irregularities in the conversion process.

$$\mathbf{n} = \frac{\mathbf{N11} + \mathbf{N22}}{2} \quad (3.4.1)$$

$$\hat{\mathbf{n}} = \frac{\mathbf{n}}{\|\mathbf{n}\|} \quad (3.4.2)$$

$$f = \frac{h_{\text{disc}}}{m} \quad (3.4.3)$$

$$p = [P11; P11 + f \hat{\mathbf{n}}; P22 - f \hat{\mathbf{n}}; P22] \quad (3.4.4)$$

$$B(u) = \sum_{i=0}^{n_s-1} \binom{n_s-1}{i} (1-u)^{n_s-1-i} u^i p_i \quad (3.4.5)$$

$$n_s = \frac{h_{\text{disc}}}{P_s}$$

$$u = \left(\frac{i}{n_s - 1} \right)$$

Once the surface meshes for both the intervertebral discs and vertebrae were obtained, the nodes at the corresponding surfaces of the disc and vertebrae were tied together. This was accomplished by merging the node sets of the structures, ensuring that the nodes on the disc share the same degrees of freedom as the corresponding nodes on the vertebrae. This approach assumes a perfect bond between the components, removing the need for contact definitions and improving the model's computational efficiency. The approach of joining the node sets between the disc and the vertebrae mirrors the physiological connection responsible for transmitting forces between the discs, endplates and the vertebrae(16), ensuring a stable connection between the segments.

After joining the discs and vertebrae, the model's surface mesh was converted into a volumetric mesh using the TetGen processing tool included in the GIBBON package. The surface mesh was transformed into tetrahedral elements with quality controls to ensure the mesh was suitable for finite element analysis. These controls included a minimum radius-edge ratio of 1.2 for the tetrahedra, region markers to distinguish between different spinal structures, and constraints to preserve the original geometry. The result for the volumetric meshing is demonstrated in Figure 9, where 9a shows the original spinal geometry from the benchmark DEMO_febio_0060_vertebrae_disc_01, with 9b and 9c demonstrating the progression from a L4-L5 spinal unit to a L4-S1 lumbar segment.

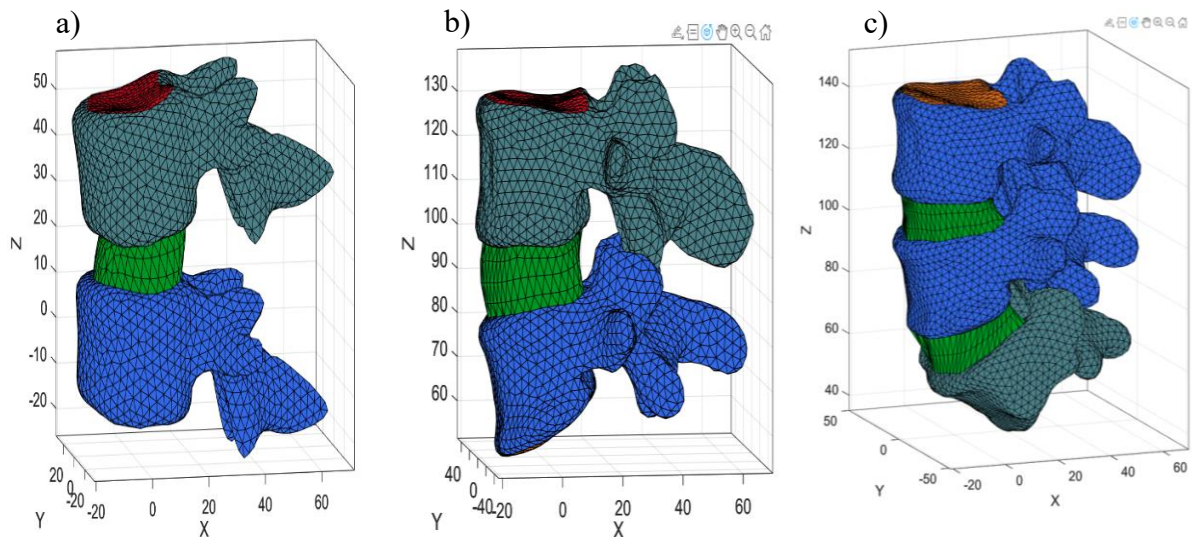


Figure 9: Progression of the vertebrae and disc model construction from the benchmark spinal unit mode (a) to the L4-L5 segment (b) to L4-S1 patient one meshed model.

After a volumetric mesh was produced the vertebra and disc geometries could be improved to include specific structural components. As the aim of the study was to analyse the vertebral structures mechanical responses, it was important to capture the different regions of bone types. Cortical and cancellous regions were defined throughout the vertebral body and posterior arch of L4, L5 and S1, as shown in Figure 10. Cortical thickness was defined as 1.80mm, 1.88mm and 1.36mm for the three vertebrae respectively(125,126). These regions of elements were assigned specific material properties as shown in Table 12, with orthotropic elastic material properties obtained from literature(109). Cortical and cancellous bone were included throughout the vertebral body and posterior arch to maintain the material distribution and mechanical behaviour, however, to reduce computational complexity specific anatomical replication of the cancellous bone was not included.

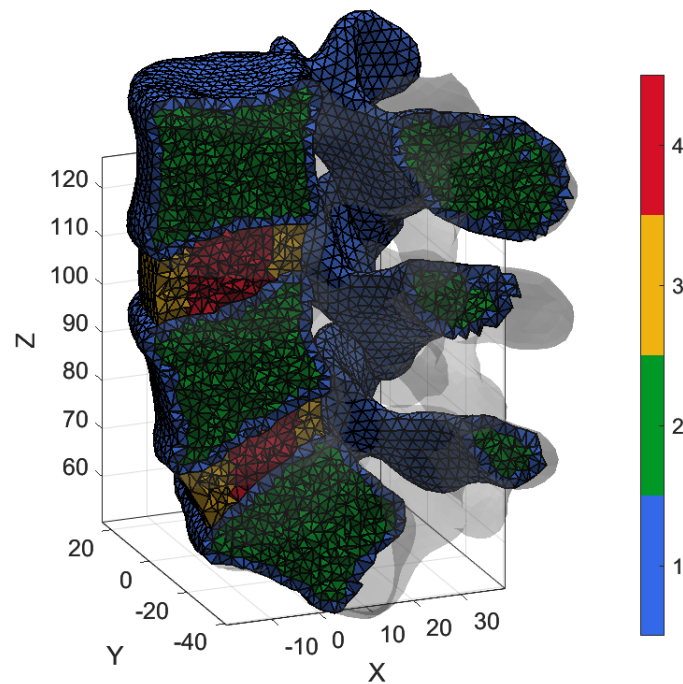


Figure 10: Cross section of vertebra and disc model regions showing the tetrahedral meshes: 4: Nucleus Pulposus, 3: Annulus Fibrosus, 2: Cancellous bone, 1: Cortical bone

The chosen intervertebral disc components in this study were the annulus fibrosus and nucleus pulposus. Elements groups were assigned so that the nucleus pulposus was to occupy 40% of the disc's cross-sectional area(127) for both the L4/L5 and L5/S1 discs. Material

properties were assigned to these areas as shown in Table 12, with neo-Hookean material behaviour obtained from literature(110). As the primary focus of this study was to analyse the vertebral and spondylolysis mechanics, the intervertebral discs were modelled with a simplified method compared to those in the literature review which modelled annulus fibrosus fibre structure. This approach allowed for a more efficient analysis of the vertebra, while still accounting for the structural differences between the annulus fibrosus and nucleus pulposus.

3.5 Facet joints

The facet joints play a crucial role in spinal biomechanics particularly in the context of spondylolysis. As previously discussed, the joints are integral to the distribution of compressive and shear load in the spine as well as acting as structural restraints to maintain spinal stability by limiting excessive motion. As demonstrated in multiple studies in the review, degenerated or abnormal facet joints can alter the load bearing and stress distribution throughout the vertebra, with an increase in this pattern in the presence of spondylolysis. Therefore, to accurately analyse the lumbar spine in both intact and fractured models the facet joints were an important structure that needed careful construction in this study.

The facet joint structures were carefully extracted during the sectioning process with manual tracing of the contours of the structure after total segmentation of the vertebra models. This approach ensured accurate selection of the morphology of the facet joints and maintained a uniform gap between the articulating facets.

To accurately represent the contact in a facet joint, a 3D contact model was established between specifically defined elements, with a single element layer to simulate a cartilage layer. The facets were constructed as specific regions selected on the concave and convex surfaces of the articular processes. To construct the surfaces, a bounding box was defined for each articular process through prescribing unique thresholds along the x, y, and z axes. Normal angle thresholds were also implemented defining specific angular ranges for each facet face to capture the specific curvature of the lumbar facet faces. These logical masks were then assigned to corresponding face groups on the spinal model. The facet selection

results can be seen in Table 19, demonstrating the number of elements included, as well as Figure 11 which demonstrates the anatomical structure.

The facet joints were modelled as a contact model, with an approach inspired from the literature models. After reviewing recent spinal models, including insights from the study “Biomechanical Modelling of the Facet Joints: A Review of Methods and Validation Processes in Finite Element Analysis” by Mengoni (2021)(128), a cartilage layer was incorporated to improve the model's accuracy. Cartilage was modelled as a one element thick layer, embedded in the articular processes. This approach was chosen over defining a specific cartilage geometry due to the complexity of the geometry of the hyaline cartilage. As previously discussed, the characteristics of facet joint cartilage varies significantly depending on its location within the spine and is influenced by patient specific factors such as age, weight, and degree of spinal degeneration. Cartilage is also challenging to identify and extract from CT imaging and therefore could not be extracted during sectioning of this model. Consequently, the material properties of cartilage used in this study were sourced from existing literature, given that "the material properties of the individual tissues of the facet joint have not been completely defined"(13). Additionally, the anonymity of the patient 1 and 2 spine sources further limits the ability to tailor material properties to specific patient characteristics.

The contact model was defined as a frictionless, non-linear, sliding-elastic facet-on-facet contact model to replicate the sliding of cartilage layers between the articular processes during movement. A sliding-elastic contact model in FEBio was chosen because it is a recognised approach for simulating simple contact problems in facet joint models and other diarthrodial joint contact mechanics cases(129). This model allows the superior and inferior surfaces to slide over each other tangentially while enforcing contact in the normal direction, thereby preventing separation during contact. To achieve this, the Augmented Lagrangian method was applied to manage both compression and tension in the contact surfaces, ensuring the surfaces remain connected without excessive penetration or separation. This method combines Lagrange multipliers, which apply corrective forces to prevent penetration, with the automatically calculated penalty stiffness provided by the auto penalty feature. The auto penalty feature allows FEBio to calculate the penalty stiffness required for contact enforcement, ensuring that contact forces are appropriately scaled according to the model's

material properties and geometry. However, since this is a soft-tissue contact problem, where cartilage allows for some compliance under compression, a gap tolerance of -1.25 mm was prescribed. This setting permits a small amount of penetration of the facet surfaces before the contact enforcement force is applied, reflecting the natural behaviour of cartilage under load. To further simulate the cartilage, frictionless behaviour was prescribed as the surfaces are lubricated by synovial fluid which result in a coefficient of friction of 0.01 or less(130) and therefore was assumed to be zero.

Additional parameters for this contact problem included disabling symmetric stiffness which, while increasing computational complexity, improves the accuracy of contact interactions and resultant load distribution. This adjustment was crucial, as the study is analysing the biomechanical effects of lysis on the vertebrae, which are significantly influenced by the behaviour of the facet joints. To enhance computational efficiency and manage problem-solving time, a tolerance value of 0.2 was set, ensuring a convergence criterion where the relative error of the current solution compared to the previous iteration is less than 20%, thereby reducing computational time. The Augmented Lagrangian process was limited to a minimum of one and a maximum of 10 iterations to balance accuracy with computational efficiency. A two-pass algorithm was implemented to ensure contact interactions were accurately detected on both surfaces of the facet joints. Finally, a detection area for potential contact was defined using the model's vertices, setting a vertex search radius of 1% of the model's overall size, with a search tolerance of 0.01, meaning surfaces would be considered for contact if they were within 0.01mm from each other.

As previously mentioned, the contact group of elements was also used to define the cartilage layer. This layer was assigned neo-Hookean material properties, as specified in Table 12. The material properties were selected from a comprehensive review of cartilage material models(128) in the context of facet joint finite element models. The selected cartilage model was chosen due to its higher Young's modulus. This choice is due to this spinal model's construction with an embedded cartilage layer, that merges with both cortical and subchondral bone, and therefore requiring the use of a material with a higher stiffness to accurately represent the integrated structure.

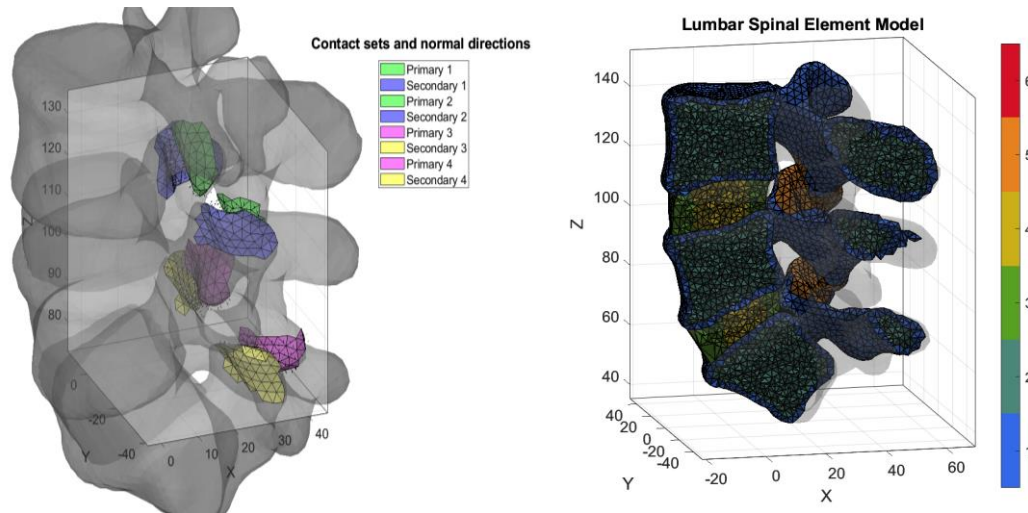


Figure 11: Contact problem faces and vertices, and the cross section of the spinal component regions: 5: Facet joint surface selection, 4: Nucleus Pulposus, 3: Annulus Fibrosus, 2: Cancellous bone, 1: Cortical bone.

3.6 Ligaments

The seven lumbar spinal ligaments are essential to include in this study's spinal model to ensure the structural stability and proper alignment of the vertebrae under deformation. Accurately representing the ligaments' function in resisting deformation is crucial in the context of spondylolysis, as it allows for a realistic distribution of load across the vertebrae during realistic ranges of spinal motion. While the effects of ligament degeneration and spondylolysis biomechanics were not extensively discussed in the review, the absence or buckling of ligaments would result in exaggerated spinal movement, impacting load distribution in the vertebrae. Therefore, it is crucial to include intact ligaments in the model to accurately capture and influence the results.

Finite element models of ligaments are typically represented using either one-dimensional spring elements or three-dimensional continuum elements(131). Continuum models offer a more precise approach for simulating the interactions between ligaments and bony structures, providing a detailed analysis of the regional biomechanical response both within the ligaments and in their interactions with surrounding tissues. However, this method is computationally intensive and requires a significant modelling effort and increased computational resources to accurately capture the complex multicomponent structure of ligament fibres and matrices. A comparative study of spring and continuum models found that spring models can reduce computational time by approximately 30%(131). Therefore, this

study adopts the approach used by seven of the models analysed in the literature review, which employed the one-dimensional spring element technique. In this study, the springs were modelled with linear behaviour to simplify the simulations and reduce computational demands. Given the need to simulate multiple fracture scenarios and analyse the resultant mechanical and kinematic responses of the spine, the linear spring model was chosen to optimise computational efficiency, providing a practical balance between accuracy and resource constraints. Linear models do not capture the non-linear characteristics of ligament behaviour; however, this study's primary focus is to understand the resultant effects of spondylolysis on vertebral effective stress magnitude location and lumbar range of motion, and therefore this model is suitable.

Ligament groups of springs were constructed for the ALL, PLL, ITL, ISL, SL, LF, and FCL, with subgroups designated for ligaments that occur in pairs, such as the left and right ITL, as well as for ligaments that connect between each vertebra. For the vertebra-vertebra ligaments, including the ITL, ISL, SL, and LF spring groups, the following method was employed. The anatomical areas for node selection were carefully defined based on the natural attachment points of each ligament. For instance, the ITL ligament nodes were selected from areas on the transverse process, specifically from the superior edge of one vertebra and the inferior edge of the adjacent vertebra. This ensured that the selected nodes accurately represented the actual ligament attachment sites. Vertebral nodes within these defined areas were isolated using bounding boxes and logical masks, with the selected area corresponding to anatomical ligament connection locations. Springs, modelled by connecting pairs of nodes located on adjacent vertebrae, simulated the natural position of ligaments, with the number of springs (outlined in Table 9) depending on the width of a ligament. For the specified number of springs, the closest pairs of nodes were identified by calculating the Euclidean distance between their x and y coordinates in the xy-plane. Once a node pair was selected, it was removed from the pool of potential spring connections to avoid duplicates. This iterative method of selecting node pairs within the defined area ensured the formation of straight ligament fibres. Once the defined number of spring node pairs was formed, they were assigned to a ligament group to be later processed by FEBio.

The formation method for the ALL and PLL included additional steps due to these ligaments attaching to multiple points on all three vertebrae (L4, L5, S1) and the intervertebral discs.

The anatomical areas for node selection were defined to capture the complex attachment patterns of these ligaments. The ALL and PLL attach to the inferior and superior vertebral margins and adhere to the surface of the annulus fibrosus; therefore, multiple springs arranged in series were needed to model each ligament fibre and capture these attachment points. Similar to the previous ligaments, logical masks were defined for the regions of the vertebral margins and discs. An iterative process was then performed in two stages to find the nodes in the most suitable alignment between the superior face of L4 and the inferior face of L5, and between the inferior face of L5 and the inferior face of S1. As in the previous process, these two loops identified the best groups of nodes that were in closest alignment in the xy-plane, storing the selected nodes and then removing them from the potential pool of spring connections. The segments were connected by using the endpoint of the L4-L5 segment as the start of the L5-S1 segment; initially separating these segments was crucial for reducing processing time. Once the defined number of spring chains was selected, they were assigned to the ALL or PLL groups. The ligament groups ALL PLL, ISL, SL, ITL and LF were stored in a cache specific to each spinal model and tailored to the mesh refinement selected for analysis in the mesh convergence study for simulation efficiency. The four FCL groups included in the model required a unique construction due to their relationship with the facet joint model. First, the facet joint surface elements were extracted to identify the boundary nodes where the springs would connect. Each superior and inferior facet joint surface boundary node was then paired. To ensure accurate pairing, the node sets were reordered starting from the closest points between the two surfaces. From this starting point, the spring nodes were paired using a step size function to account for differences in the number of nodes between the surfaces, thereby maintaining equally spaced springs. The paired spring nodes were then stored in the corresponding FCL group for each of the four facet joints. The final number of springs for each of the ligament groups can be seen in Table 9 and the total ligament model is displayed in Figure 12. This process of ligament creation was flexible between models, allowing for the quick formation of ligaments between multiple geometries and meshes, which was crucial for this study.

For each ligament group, specific linear material properties were assigned, including Young's modulus (E) and ligament cross-sectional area (A). The spring constant for each individual spring, $k_{\text{individual}}$, was calculated using Equations 3.6.1 and 3.6.2. To determine the total

spring constant k_{total} of the ligaments, the length of each ligament (L) was calculated by measuring the distances between node pairs for all springs comprising a ligament, and then averaging these lengths. The calculated lengths for each ligament are displayed in Table 9. The total spring constant for each ligament was then divided by the number of parallel springs in the ligament (n_{parallel}) to assign the individual spring constant to each spring ($k_{\text{individual}}$). For the PLL and ALL ligaments, the number of springs in series (n_{series}) had to be considered. To obtain the individual spring constants for these ligaments, Equation 3.6.2 was used. All material properties, obtained from the literature, are summarised in Table 12.

$$k_{\text{total}} = \frac{EA}{L} \quad (3.6.1)$$

$$k_{\text{individual}} = \frac{n_{\text{series}} k_{\text{total}}}{n_{\text{parallel}}} \quad (3.6.2)$$

Table 9: Ligament material properties obtained from literature and in model ligament measurements. Material properties obtained from Fan et al. (2019).

Ligament/ Spring group	Young's Modulus (MPa)	Area (mm²)	Length (mm)	Number of Springs
ITL Left	10	1.8	25.94	5
ITL Right	10	1.8	28.64	5
ALL	7.8	63.7	123.67	13 of 8 springs in series
PLL	10	20	102.85	10 of 8 springs in series
ISL L4/L5	11.6	40	10.33	8
ISL L5/S1	11.6	40	17.82	8
SL L4/L5	8	30	25.13	1
SL L5/S1	8	30	30.35	1
LF L4/L5	15	40	35.00	10
LF L5/S1	15	40	30.64	10

FCL SAP Left	7.5	30	6.09	19
FCL SAP Right	7.5	30	6.12	20
FCL IAP Left	7.5	30	9.25	21
FCL IAP Right	7.5	30	7.99	21

14 Ligament Model: ALL, PLL, ITL (2), SL (2), LF (2), ISL (2), FCL (4) 14 Ligament Model: ALL, PLL, ITL (2), SL (2), LF (2), ISL (2), FCL (4)

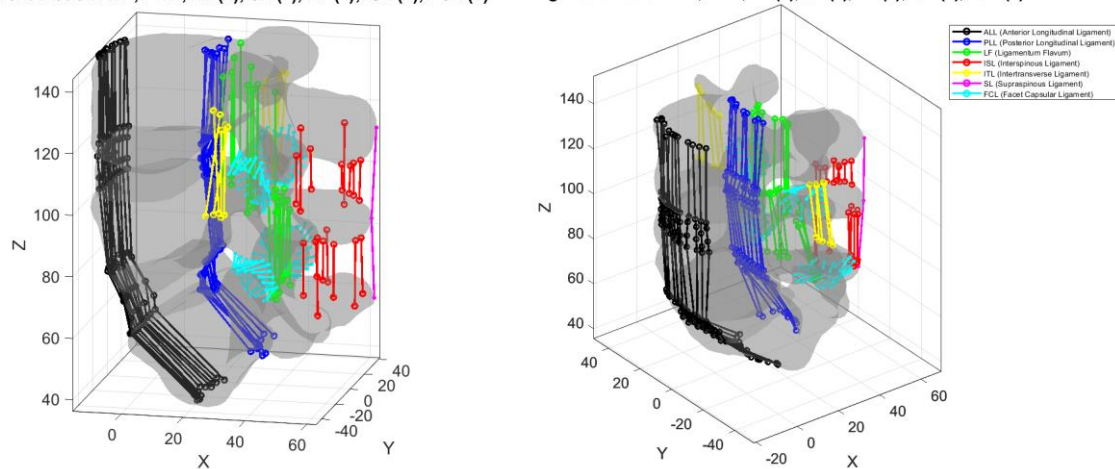


Figure 12: Ligament model of the spine where each ligament is represented as a group of discrete springs.

3.7 Spondylolysis

Based on the results of the literature review, it appears that finite element models of spondylolysis primarily concentrate on the mechanical factors leading to fractures or on the effects of the condition in its terminal phase. These models did not consider fractures from the perspective of grade 0 to early grade 3 fractures and the progression of fracture healing, through the phases of granulation, fibrous tissue, cartilage or immature and remodelled mature bone. Complex finite element models have analysed fracture healing of bone through either mechanobioregulatory or bioregulatory healing models(132). Mechanobioregulatory models incorporate mechanical stimuli to analyse tissue proliferation and differentiation within the fractured or callused region. Bioregulatory models focus on the cellular processes

and extracellular matrix interactions to analyse tissue repair(133). These models are highly complex and would be computationally intensive to apply to this multi-component spinal model. Previous fracture healing models(132,133) have focused on small sections of bone, and applying a similar method to this study's model of the L4-S1 spine would not be practical for detailed analysis of the healing process across different grades of spondylolysis. Given the complexity involved in capturing the interactions between the fracture healing stages, a static modelling approach is used. This approach analyses each healing stage, granulation, fibrous tissue formation, cartilage development, and the maturation of bone, by assigning specific material properties to the pars interarticularis area. This snapshot method simplifies the analysis while still allowing for the assessment of mechanical changes and their effects on the spine during conservative treatment.

As bony healing can only be achieved in early to progressive fractures this model assumed we have not reached terminal lysis. The unilateral spondylolysis area was defined in a similar approach to the literature review models, in which a group of elements are defined in the left pars interarticularis region. Instead of deleting the elements to simulate terminal lysis, these elements were assigned specific neo-Hookean material properties from Ghiasi's (2019)(73), finite element fracture model, for each of the healing stages, as shown in Table 12.

The region was defined using the following approach: A plane was created and translated to pass obliquely through the pars interarticularis to define the area of lysis. The lysis thickness was defined to select elements within a specified distance on both sides of the plane in a direction perpendicular to it. The cortical and cancellous bone elements within this specified distance were grouped into the lysis element group. The total number of elements in the lysis group are displayed in Table 19. The fracture was considered to be homogeneous across both the cortical and cancellous bone even in the mature healing phase as the degree and pattern of bone remodelling depends on fracture severity, treatment type and a patient's specific healing response. Cortical and cancellous bone healing also differs in speed due to their surface area and blood supply(134) so to simplify the fracture region all healing phases material properties and geometry was assumed to be homogeneous. Another assumption was that the fracture geometry was consistent throughout the healing phases, with no callus formation or volume expansion due to swelling pressure(73) during the initial inflammatory granulation stage. However, callus size is hard to quantify and depends on fracture size, interfragmentary

motion(73), and number of days post fracture(135). Therefore, fracture geometry was kept consistent to the pars interarticularis geometry. The final selection of elements is displayed in Figures 13 demonstrating the model of the left unilateral fracture.

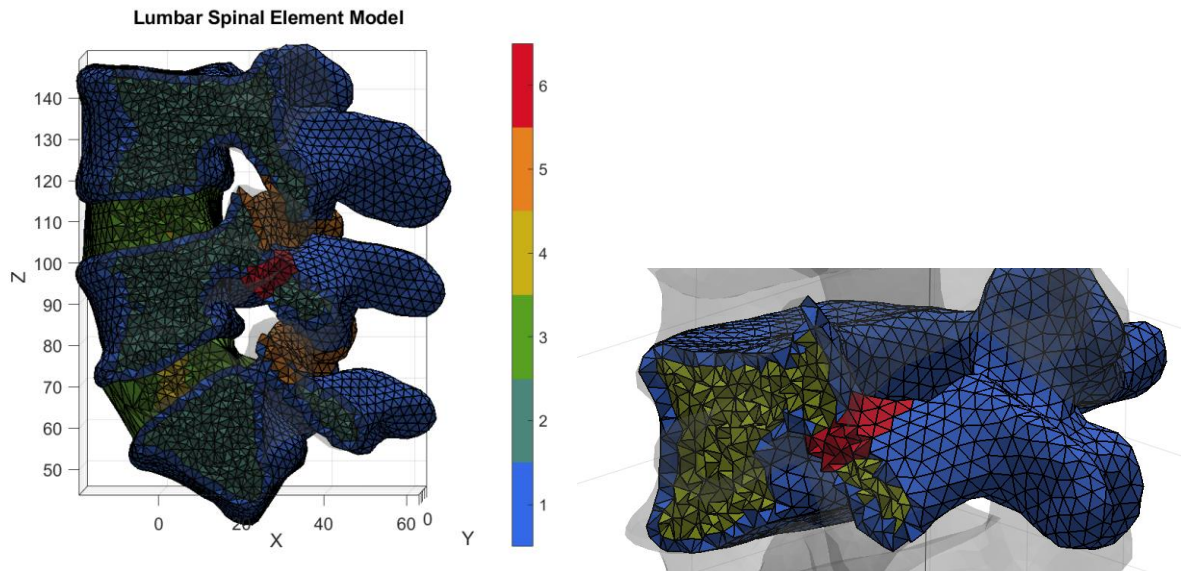


Figure 13: Selection of elements to represent a unilateral spondylolysis fracture taken from two angles. Lysis highlighted in red.

3.8 Boundary Conditions and Loads

The inferior surface nodes of the S1 vertebra were constrained in all translational degrees of freedom (x, y, and z axes) to simulate a rigid articulation with the pelvis and ensure model stability. The nodes of the superior face of L4 were grouped to prescribe the range of loads and moments for model analysis, these nodal groups are visualised in Figure 14. Loads were applied to the spine in two steps. First, a preload was applied to the spine as a compressive force distributed evenly between the nodes of the superior face of the L4 vertebra. Pre-loading mimics the natural compressive load of the upper body during upright standing(136) and helps to provide stability to the overall spinal model by increasing load resistance and hence reducing excessive motion and rotation(137). The preload was established for the first 10 time increments to reach stabilisation before the moments were applied. After the preload, an additional moment was applied to the prescribed nodes specific to flexion, extension, lateral bending and axial rotation. As L4 was not a rigid body, moments had to be applied as a distributed load across the surface.

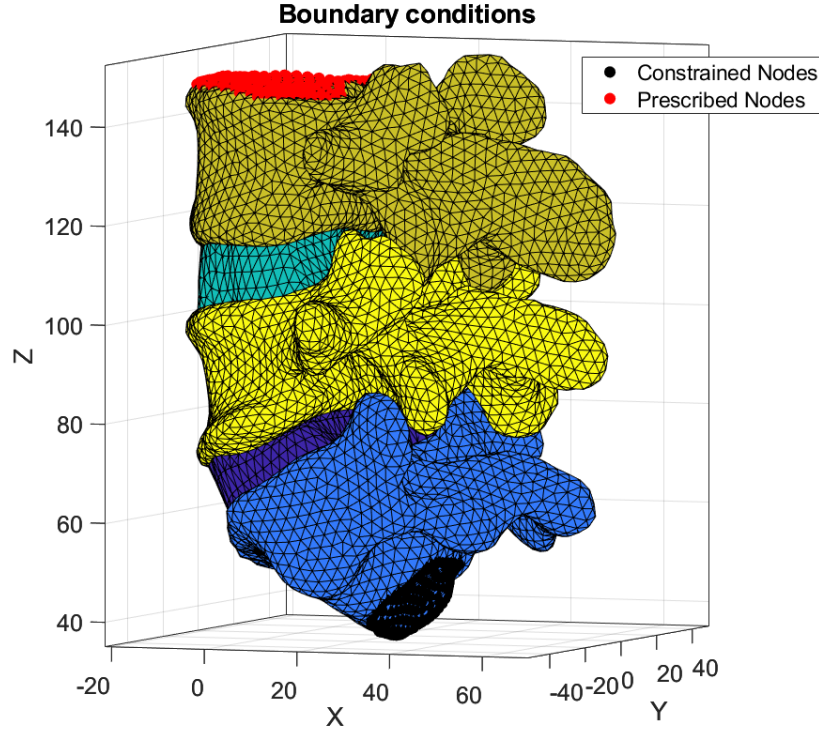


Figure 14: Boundary conditions of the spine. With constrained nodes on the inferior face of S1 and nodes for loads to be prescribed to on the superior face of L4.

The moments were constructed through a systematic process, including determining the force distribution across the nodes of the L4 superior surface and the lever arm that corresponded to each node to provide the resulting moment around the specified axis. After defining the type and magnitude of the moment loading, the code identified the appropriate axis of rotation for each specific moment:

- Y-axis for flexion and extension, with forces applied in the X-Z plane
- X-axis for lateral bending, with forces applied in the Y-Z plane
- Z-axis for axial rotation, with forces applied in the X-Y plane

The direction and moments are illustrated in Figure 15. For each loading mode the central rotation axes were determined for each of the moments. From these axes the corresponding maximum radial distance (r_{max}) were determined from the moment arms Δx , Δy or Δz for the axis of bending to find the total force (F) needed to prescribe a total moment (M_{total}). Each moment arm was calculated as the distance from the node coordinate to the centre with equation 3.8.1. The force calculations are represented in Equations 3.8.2-3.8.3. Force

distribution across nodes was proportional to their distance from the axis of rotation, with the maximum force applied at the greatest distance. The forces were broken down into their plane components and the force was distributed between the directions, this step also adjusted the direction of the components for the opposing bending moments. The bending moment of the individual components was calculated using the cross product, to determine the moments for each specific node (M_{ij}), generated by the force relative to the axis, shown in Equation 3.8.3. The cross product was used to determine the twisting effect created by the forces, and calculates how much each node contributes to the overall bending moment. The total moment of the x-axis was obtained by summing these individual moment contributions across all nodes. Once the moments were calculated any discrepancies were solved with a scaling factor to ensure the desired moment was achieved. This iterative process ensured that the resulting moment closely matched the target, thereby achieving the desired bending effect. A range of bending moments and preloads were prescribed to analyse the mechanical response of the intact and spondylolysis models.

$$r_{max} = \sqrt{(\Delta(x, y, z))^2 + (\Delta(x, y, z))^2} \quad (3.8.1)$$

$$F = \frac{M_{total}}{r_{max}} \quad (3.8.2)$$

$$M_{ij} = (\Delta(x, y, z)_i \times F_{(x,y,z)i}) - (\Delta(x, y, z)_j \times F_{(x,y,z)j}) \quad (3.8.3)$$

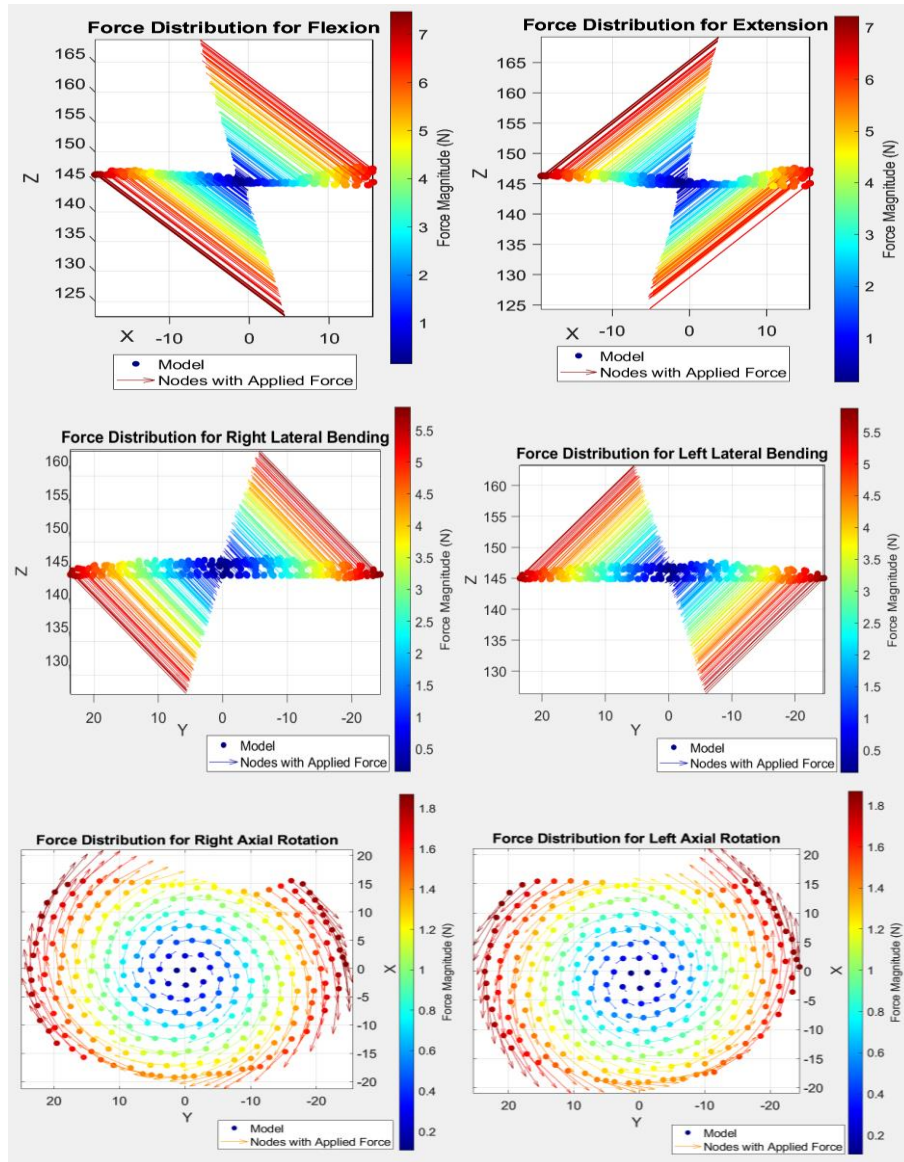


Figure 15: Example of the bending moments applied as distributed force to the superior face of L4, with arrows showing the direction of applied moments.

Chapter 3: Part 2, Post Processing & Evaluation

3.9 Data Analysis

After curating the model elements, defining the material properties, and applying the boundary conditions, the FEBio input structures were established for the simulation. The model was then run using specified parameters, including loading modes, model types, lysis material types, and mesh convergence settings such as point spacing, moment values, and body loads. Outputs analysed for this model include range of motion, von Mises stress of the

cortical, cancellous and lysis regions of the bone, mean volume-weighted von Mises stress, segment displacement and stress patterns across the L5 vertebra.

Stress patterns were analysed using FEBio studio for analysis of the model over the timesteps of compression and bending moments. Range of motion, stress and displacement were all imported and analysed in MATLAB after external simulation in FEBio.

For the stress outputs, principal stress data (σ_1 , σ_2 , and σ_3) was imported for the cortical, cancellous and lysis regions of the L5 vertebra over all time steps. The von Mises stress (σ_v) was calculated for each region over all time steps from the principal stresses using Equation 3.9.1. Both the mean and maximum von Mises stress were also calculated across all elements for each region and for all timesteps. Volume-weighted mean von Mises stress (σ_{vw}) was also calculated using the Equation 3.9.2 where V_{elem} is the volume of each element and V_{total} is the total volume of the region. Where maximum σ_{vw} was also found across all time steps for each region.

Displacement data was imported for the entire spinal structure from FEBio and then separated into lysis, cortical and cancellous nodal regions. The displacement magnitude for each node was calculated using the Euclidean norm of the displacement vector. The mean displacement was computed for each region across all time points, from which the maximum displacement per region was then derived for the entire simulation period.

The angular range of motion was calculated using two vectors (v_1 , v_2) created from landmarks on L4 and S1. The landmarks were identified as two elements on each vertebra which were the centroids of the inferior and superior faces of the vertebral bodies. The position arrays of the coordinates for the landmarks were extracted after the FEBio simulation. The initial and final vectors were compared at the start and end of the bending moment time regions and created by subtracting the landmark positions at the initial and final time points (t_i). The initial and final angles between the position vectors were calculated based on these vectors. The angles were measured within the relevant anatomical planes and to measure the angle in two components of the vectors —sagittal, coronal, and transverse— during flexion-extension, lateral bending, and axial rotation, respectively. To find the initial ($\theta_{initial}$) and final angles (θ_{final}) between the two vectors, the four-quadrant inverse tangent function (arctan) was used, as demonstrated in Equations 3.9.3. The dot product input

provides the cosine of the angle which determines the alignment of the two vectors and calculates the magnitude of the angle between them. The cross product is used to determine the direction of rotation, allowing for a complete 360° analysis of the angular range of motion. The initial angles are subtracted from the final angles to produce range of motion of the L4 to S1 spinal unit.

$$\sigma_v = \sqrt{0.5[(\sigma_1 - \sigma_2)^2 + (\sigma_2 - \sigma_3)^2 + (\sigma_3 - \sigma_1)^2]} \quad (3.9.1)$$

$$\sigma_{vw} = \frac{\sum_i(\sigma_{vi}V_{elem,i})}{V_{total}} \quad (3.9.2)$$

$$\begin{aligned} \theta_{initial,final}(t_i) &= \arctan\left(\frac{\|v_1(t_i) \times v_2(t_i)\|}{v_1(t_i) \cdot v_2(t_i)}\right) \\ v_1 \cdot v_2 &= \|v_1\|\|v_2\|\cos\theta \\ \|v_1 \times v_2\| &= \|v_1\|\|v_2\|\sin\theta \end{aligned} \quad (3.9.3)$$

These specific outputs were chosen for either intact model analysis or for the spondylolysis mechanical behaviour study. The outputs maximum displacement of each region, maximum von Mises stress, maximum volume-weighted von Mises stress and range of motion were analysed in the data verification as a representation of local and global metrics of the model.

For the spondylolysis study, von Mises stress across the L5 vertebra was analysed to identify areas of high stress and at risk of fracture. Understanding stress distribution under the different loading conditions is crucial for rehabilitation to prevent progression of spondylolysis to secondary fracture locations or preventing union of the bone. By identifying movements that cause high stress in vulnerable areas of the spine, conservative treatment can be designed to avoid these movements, reducing the risk of re-injury. Therefore, the von Mises stress locations and magnitudes have been analysed to investigate mechanical limits of each healing phase of spondylolysis. Von-Mises stress analysis can be used as a parameter for rehabilitation intensity progression with fracture healing. Range of motion is analysed to see the instability of the spine with the fracture over the healing process to tailor rehabilitation treatment plans to accommodate for this instability. The instability can be used as a parameters to analyse the healing process and progress rehabilitation.

3.10 Data verification

A mesh convergence study was conducted on the intact L4-S1 spinal model including the facet joint and ligament additions. A h-refinement study was used with a reduction in element sizes controlled by a variable called “Point Spacing” which controlled the mesh size by altering the distance (mm) between vertices. A range of 3.05mm to 2.75mm was chosen to alter the linear tetrahedral elements of the vertebral and disc structures. The study analysed, as previously mentioned, for displacement, maximum von-Mises and volume-weighted von Mises stress and range of motion to analyse local changes of displacement and maximum von Mises, and global changes of range of motion and volume-weighted von Mises stress. Volume-weighted von Mises stress was analysed to address discrepancies in element size, particularly at remeshed boundaries such as the intervertebral disc-vertebra interface, which were resized after the initial element definition based on point spacing. Mesh convergence percentage error was calculated using Equation 3.10.1, where the smallest point spacing was taken as the “ideal point spacing” to compare changes in values between the meshes. The ideal point spacing was chosen to be the model with the smallest percentage errors across all output parameters and for the value that finds a stable stress distribution across the vertebrae.

$$Percentage\ Error\ (\%) = \frac{Ps - Ideal\ Ps}{Ideal\ Ps} \times 100 \quad (3.10.1)$$

3.11 Sensitivity Analysis

This section outlines the sensitivity analysis conducted to identify the components of the model that had the greatest influence on the results and to assess the model's robustness. Analyses were completed for intact model material property sensitivity and fracture model characteristics for the components of the lumbar spinal model. These tests evaluated the input parameters, as displayed in Table 10, in relation to the outputs of von Mises stress and range of motion. To perform the analysis a base line loading condition was established. All models were tested with an axial compressive body load of 530N and a 7.5Nm load for the prescribed moments, except for the intervertebral disc analysis, which followed the same method used in the range of motion validation study.

All models were analysed for the loading modes flexion, extension, lateral bending left, and axial rotation right with exception for the fracture location study and ligament stiffness analysis where all 6 DoF were analysed. The sensitivity analysis was conducted based on two key factors, (1) adjustment of component material properties and geometry and (2) comparison between the patient models. The ligaments were analysed through comparison of the minimum spring stiffness that would avoid simulation error and the maximum stated values for Young's modulus in literature(138), as this study did not adopt a non-linear approach to vary the modulus with strain. The material laws of the vertebral component, cortical and cancellous bones, intervertebral disc components, annulus fibrosus and nucleus pulposus, were tested for two different material laws. Vertebral bones were analysed with isotropic and orthotropic elastic material properties. The cortical and cancellous bone are stiffer along their longitudinal axis due to the direction of the bone structure therefore these material properties were analysed. The two material models were obtained from literature from Xu (2016)(139) and El-Rich (2009)(109) finite element models. Two hyperelastic material models for the intervertebral disc were analysed of Neo-Hookean and Mooney-Rivlin materials. Both material models are suitable for simulating soft tissue deformations and are frequently used in disc finite element models. Neo-Hookean models are computationally simple and efficient(140) and therefore are suitable for high run time or batch simulation applications. However, analysis of intervertebral disc tissue modelling has suggested that Mooney-Rivlin models fit experimental data of disc deformation with improved quality compared to Neo Hookean models(141). The two material models were obtained from existing literature, Xu (2016)(139) and Haj-Ali (2019)(110) finite element models. Fracture geometry analysis was conducted by varying the fracture widths from 1-4mm at the left pars interarticularis to evaluate their impact on the magnitude and distribution of von Mises stress within the vertebra. Additionally, the effect of fracture location was examined in both patient models by analysing unilateral fractures in either the left or right pars interarticularis.

The second consideration involved comparing vertebral models obtained from two different patients to analyse how variations in vertebral geometry influence the model outputs. These models were constructed identically apart from slight changes in disc height and width and facet joint surface area to adjust to the patient's anatomy.

The results analysis for the sensitivity tests included linear regression of the ligament stiffness trials. A Python script was developed to perform cumulative linear regression sensitivity analysis for L4-S1 range of motion and von Mises stress in the cortical and cancellous bone of L5, with increasing ligament stiffness. The Young's modulus of all ligaments was systematically increased for each trial. For every trial, the script computed a new regression coefficient, or slope, that quantified the correlation between changes in ligament stiffness and variations in stress or range of motion. The results from each trial were progressively incorporated into the regression calculation until all four trials were analysed. The slopes or "sensitivity" were then plotted against the trials to visualise the results. A constant sensitivity indicates small changes to stiffness, while negative or positive slopes show inverse or direct relationships. The y-axis represents the change in range of motion or stress per unit of overall ligament stiffness change, where larger magnitudes represent a greater change to stiffness.

Intervertebral disc analysis was performed with analysis against Jaramillo's (2016)(142) range of motion results for comparison to the validity study to analyse the effect on the resultant Pearson's coefficient. Vertebrae material analysis was done by comparing colour plots of the von Mises stress maps of the orthotropic and isotropic data to analyse stress distribution patterns. For the fracture studies, graphs were generated to visualise the von Mises stress in the cortical region, allowing a clear comparison of stress values between different fracture scenarios.

Table 10: Sensitivity Analysis Parameters for Analysis for intact and fracture models

Category	Parameter of Interest	Original property		Altered properties	
Ligament material properties	Stiffness	E (MPa)		3 increasing increments up to:	
		ALL	13.9	E (MPa)	
		PLL	15	ALL	20
		LF	17.25	PLL	20
		ITL	34.35	LF	19.5
		FCL	20.2	ITL	58.7
		ISL	10.8	FCL	32.9
		SL	11.5	ISL	11.6
				SL	15
		Cortical	Cancellous	Cortical	Cancellous

Vertebrae material properties	Material Law of cortical and cancellous bone: Orthotropic and Isotropic	Ex = 8000 MPa	Ex = 767 MPa	E = 12000 MPa	E = 100 MPa
		vxy = 0.3	vxy = 0.12		
		Ey = 8000 MPa	Ey = 40 MPa	v = 0.3	v = 0.2
		vxz = 0.3	vxz = 0.12		
		Ez = 14000 MPa	Ez = 1157 MPa		
		vyz = 0.3	vyz = 0.12		
		Gxy = 3280 MPa	Gxy = 20.4 MPa		
		Gyz = 3280 MPa	Gyz = 24 MPa		
		Gxz = 3280 MPa	Gxz = 24 MPa		
		v = 0.3	v = 0.12		
Intervertebral disc material properties	Material Law of annulus fibrosus and nucleus pulposus	NP	AF	NP	AF
		E = 1 MPa	E = 8.4 MPa	c1 = 0.12, c2 = 0.09, v = 0.4999	c1 = 0.56, c2 = 0.14, v = 0.45
		v = 0.4999	v = 0.45		
Fracture model design	Neo- Hookean and Mooney-Rivlin				
	Fracture width	Fracture covering pars area (4mm width)		3mm, 2mm, 1mm	
	Fracture location	Left unilateral fracture		Right unilateral fracture	

3.12 Data Validation

Indirect validation against experimental data and validation against existing computational models, were conducted to compare the two-patient intact and fracture models with existing spinal models and the spondylolysis models referenced in the literature review. Range of motion analysis has been analysed against Jaramillo's (2016)(142) finite element model which analysed segmental range of motion of L4-L5 and L5-S1. To validate the fractured models, the granulation healing phase was compared with the unilateral terminal defect models from Sairyo's (2005)(114) study. Validation of the models involved analysing the trends of output parameters between the data sets, including range of motion and the magnitude and locations of von Mises stress, under various loading conditions, as outlined in

Table 11. Since Jaramillo's model provides a complete dataset for range of motion analysis, the Pearson correlation coefficient was used to evaluate and validate the range of motion data statistically. The Pearson correlation technique is suitable for analysing the linear relationship between the two models, allowing validation of changes in range of motion with bending moment across similar finite element models(143).

Range of motion has been analysed for pure moments of 2 to 8Nm with 0N axial compressive body load, for the 6 degrees of freedom. Von Mises stress for lysis models were validated against Sairyo's (2005)(114) model for a combined axial load of 400N and bending moments of 10.6Nm for all 6 DoF.

Table 11: Validation studies: Loading conditions applied to the L4 superior face for the 6 DoF of the lumbar spine.

Test Name	Compressive force/ Body load (N)	Moment (Nm)
Flexion	0	2, 4, 6, 8
	400	10.6
Extension	0	2, 4, 6, 8
	400	10.6
Lateral Bending (Left/Right)	0	2, 4, 6, 8
	100	7.5
Axial Rotation (Left/Right)	0	2, 4, 6, 8
	400	10.6

3.13 Model Summary & Lysis Evaluation

Material properties for the overall spinal models are displayed in Table 12, where the material properties have been obtained from the literature review models or from an extensive review of material properties of lumbar spinal models(136). The final models derived from the spinal imaging of the two patients are shown in Figures 16a and 16b.

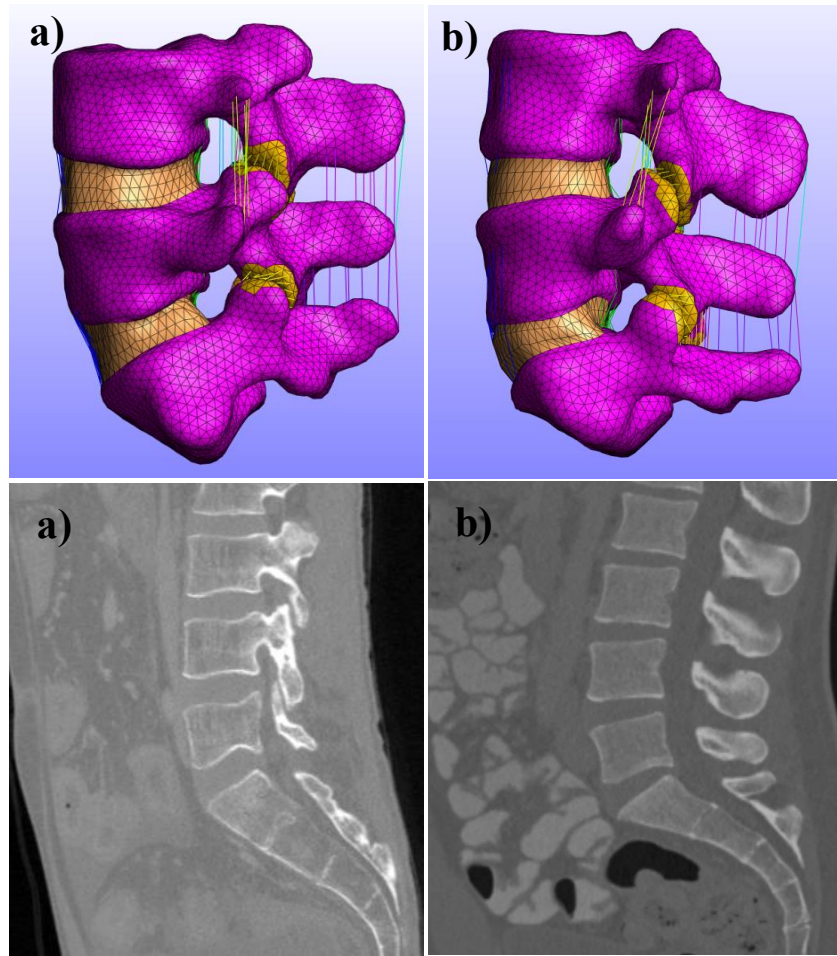


Figure 16: Comparison between patient-specific medical imaging and finite element models of the lumbar spine, a) Patient 1 L4-S1 Model, b) Patient 2 L4-S1 Model.

The two spinal models were evaluated in their intact state for a series of data validation, verification and sensitivity analysis. After analysis of the intact models, spondylolysis unilateral fractures were introduced for comparison of the intact L5 vertebra to the 5 cases of healing. To provide a framework for a conservative treatment tool one patient specific model was analysed for a series of activity specific loading conditions as displayed in Table 13. The axial loads and bending moments represented a progression of low to high activity, with low activity assuming the patient is stationary, medium activity demonstrating standing and walking and high activity representing moderate running. These example activities show how a progression of movement can be analysed during spondylolysis recovery phases, providing a scalable approach to assess the mechanical demands on the fractured area and inform targeted conservative interventions based on patient-specific limitations and recovery milestones. To calculate the activity loads, normalised compression values for various

activities were derived from Schäfer's (2023)(144) mechanical loading of the spine analysis. Axial compression calculations were based on the average body weight of an adult (60 kg), multiplied by gravitational force and adjusted for normalised compression values corresponding to different activities: 0.9 for standing at 0°, 1.4 for walking, and 1.8 for moderate running. Coupled with compression, a range of moments were tested for flexion-extension, lateral bending, and axial rotation to simulate various activity intensities. The methods and analysis from these spondylolysis models offer physiological ranges and stress thresholds to guide a conservative rehabilitation framework.

Table 12: Material properties of the structures of the model L4-S1 spine obtained from literature.

Anatomical Structure		Material Law	Elastic Modulus (MPa)			Shear modulus (MPa)			Poisson's Ratio	Reference
			Ex	Ey	Ez	Gxy	Gyz	Gxz	v	
Vertebral	Cortical bone	orthotropic elastic	8000	8000	14000	3280	3280	3280	0.3	El-Rich, 2006
	Cancellous bone	orthotropic elastic	767	40	1157	20.4	24	24	0.12	El-Rich, 2006
Intervertebral Disc	Annulus fibrosus	neo-Hookean	8.4						0.45	Haj-Ali, 2019
	Nucleus pulposus	neo-Hookean	1						0.499	Haj-Ali, 2019

Facet Joints	Inferior and superior articular processes L4, L5, S1	neo-Hookean	3500						0.25	Mengoni, 2020 Tang, Meng, 2011
Ligaments	ALL	“Linear spring”	13.9							Fan, 2019
	PLL		15							
	ISL		10.8							
	ITL		44.09							
	SL		11.5							
	LF		17.25							
	FCL		25.28							
Lysis stages	Granulation	neo-Hookean	0.001						0.17	Ghiassi, 2019
	Fibrous tissue		2						0.17	
	Cartilage		10						0.17	
	Immature bone		1000						0.3	
	Mature bone		6000						0.3	

Table 13: Lysis analysis loading conditions for activities with increasing intensity.

Activity	Loading Mode	Compression Force (N)	Moment (Nm)
Low intensity: Stationary	Compression	100N	0.1

	Compression + Bending (Flex-Ext/LB/AR)	100N	2,4,6,8
Medium intensity: Standing	Compression	530N	0.1
	Compression + Bending (Flex-Ext/LB/AR)	530N	2,4,6,8
Medium intensity: Walking	Compression	824N	0.1
	Compression + Bending (Flex-Ext/LB/AR)	824N	2,4,6,8
High intensity: Running	Compression	1060N	0.1
	Compression + Bending (Flex-Ext/LB/AR)	1060N	2,4,6,8

Chapter 4: Results & Discussion

Chapter 4: Part 1, Model Validation

4.1 Mesh Convergence

4.1.1 Results

The following subsections present the output results for the four parameters measured for the mesh convergence study. For this mesh convergence study, the moments of flexion and extension were analysed for a 7.5Nm moment coupled with a 500N body load. A summary of the total number of elements for each mesh is in presented in Table 19. The results for the four parameters, range of motion, maximum displacement, maximum von Mises stress and, maximum volume-weighted von Mises stress are displayed in Figures 17 and 18. Maximum displacement, von Mises stress and, maximum volume-weighted von Mises stress were analysed for the cortical, cancellous and lysis regions for the L5 vertebra for a mature bone lysis healing stage. The mesh convergence percentage error tolerance should be less than 5% due to the model and computational complexity.

4.1.1.1 Range of motion

The results of the mesh convergence analysis of range of motion are presented in Figure 17. Error percentages were calculated and are presented in Table 14 and Figure 18. The average values for range of motion for flexion and extension were 3.98° and 3.86°. Range of motion for the L4-S1 segment converged towards the ideal point spacing of 2.75mm with minimal changes (<2%) in error and magnitude between the 2.85mm and 2.75 mm values. As RoM is a global value and therefore does not fluctuate greatly with mesh size, the tolerance limit for a converged result was >5% and therefore a mesh size of less than 3.0 point spacing is acceptable.

Table 14: Range of motion convergence test error results for convergence analysis of point spacings of 3.05-2.75mm

Point Spacing (mm)	Flexion % Error	Extension % Error
3.05	2.24	11.44
3.00	0.75	3.23
2.95	0.75	3.73
2.85	1.00	1.99
2.75	0.00	0.00

4.1.1.2 Maximum displacement

The results for maximum displacement magnitudes and percentage error are presented in Figures 17 and 18 and Table 15. Considering the magnitudes of the displacement of the three L5 vertebra components there is minimal variation with fluctuations seen between 2.5-3mm for flexion and 1.25-1.50mm for extension. The error values show a general decrease from 3.05mm to 2.75mm. For both flexion and extension, the average percentage error across all three regions is 3.98% when the point spacing is 2.85mm. Cortical and cancellous bone regions reached the ~5% criteria however the lysis displacement did have higher percentage error results. Therefore, to minimise the effect of mesh size the smallest point spacing is the most suitable based on displacement results (2.75 mm).

Table 15: Maximum displacement convergence test error results for convergence analysis of point spacings of 3.05-2.75mm

Point Spacing (mm)	Cortical Flexion % Error	Cortical Extension % Error	Cancellous Flexion % Error	Cancellous Extension % Error	Lysis Flexion % Error	Lysis Extension % Error
3.05	9.40	0.00	9.28	0.69	4.33	10.48
3.00	5.37	0.70	4.81	0.69	7.94	2.42
2.95	5.03	1.41	4.81	1.39	5.05	3.23
2.85	5.03	3.52	4.47	3.47	5.78	1.61
2.75	0.00	0.00	0.00	0.00	0.00	0.00

4.1.1.3 Von Mises Stress

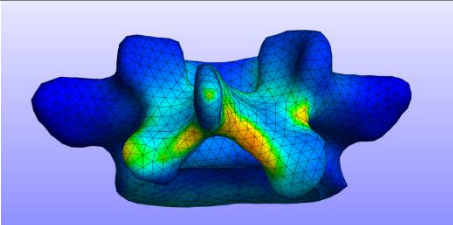
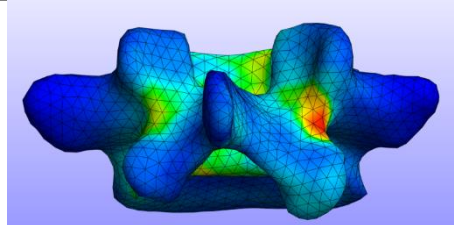
The maximum values and local distribution of von Mises stress were analysed for the three L5 vertebral regions. Von Mises stress magnitude results are displayed in Figure 17, and percentage error results are presented in Table 16 and Figure 18. The magnitude of von Mises stress for the cortical, cancellous and lysis regions are for flexion between 4.9-5.8MPa, 1.2-1.7Mpa and 2.4-3.8Mpa, and for extension between 3-3.9Mpa, 0.7-1.1Mpa and 1.5-2.7Mpa. Cortical and cancellous bone for both flexion and extension show a steady reduction in percentage error for the finer mesh values indicating convergence of the results moving towards the 2.75mm point spacing value. However, the results for the lysis section do not seem to converge with a finer mesh. As demonstrated in Table 19 the selection of elements varies between mesh study, which could explain the large fluctuations and resistance to convergence. For stress distribution, results exported from the simulations are displayed in Table 17. The flexion colour maps are analysed between the ranges of 0 and 4Mpa and from the results it can be seen that the colour gradient locations appear more uniform between the element sizes. Von Mises stress localised around the superior and inferior pars interarticularis regions and in the posterior pedicle regions. The extension colour maps produce varying results as seen in the varied percentage error analysis. Von Mises stress locations for extension were analysed within the range of 0-2 MPa. A reduced stress analysis was selected

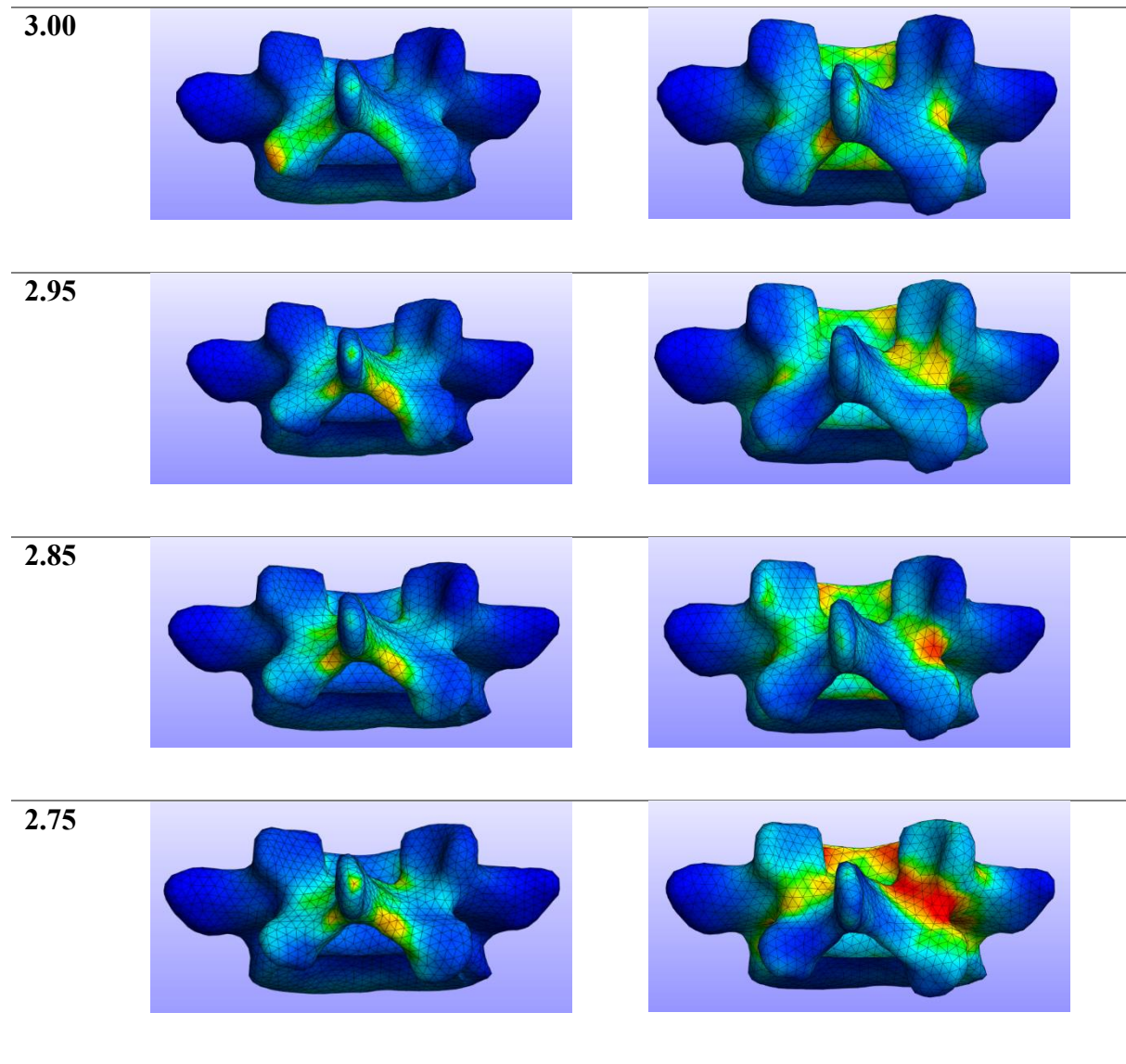
because the lower intensity force applied in this portion of the study resulted in only minor variations in von Mises stress. Von Mises stress locations varied from localised stress in the right inferior pars interarticularis region moving to occupying the total pars region. There were also distributions of stress in the region of lysis ranging from localisations in the inferior pars to both the inferior and superior pars regions.

Table 16: Maximum von Mises stress convergence test error results for convergence analysis of point spacings of 3.05-2.75mm

Point Spacing (mm)	Cortical Flexion % Error	Cortical Extension % Error	Cancellous Flexion % Error	Cancellous Extension % Error	Lysis Flexion % Error	Lysis Extension % Error
3.05	10.79	23.21	19.86	19.28	18.24	13.30
3.00	10.97	21.17	17.12	27.71	8.45	36.70
2.95	5.30	19.13	2.05	18.07	3.04	20.74
2.85	3.11	12.76	7.53	6.02	27.70	42.02
2.75	0.00	0.00	0.00	0.00	0.00	0.00

Table 17: Colour maps of von Mises stress distribution analysis for convergence analysis of point spacings of 3.05-2.75mm

Point Spacing	Flexion von Mises stress (4MPa)	Extension von Mises stress (2MPa)
3.05		



4.1.1.4 Maximum Volume-Weighted von Mises Stress

Max Volume-Weighted von Mises Stress results are presented in Table 18 and Figures 17 and 18. The magnitude of volume-weighted von Mises stress for the cortical, cancellous and lysis regions are for flexion between 1-1.21Mpa, 0.2Mpa and 0.7-0.9Mpa, and for extension between 0.6-0.7Mpa, 0.16Mpa and 0.5-0.7Mpa. Error percentage decreases with mesh refinement for both cortical and cancellous bone and during extension for lysis. Similar to maximum von Mises stress analysis it is likely that lysis element selection influenced the variation in results for these local L5 parameters.

Table 18: Maximum volume-weighted von Mises stress convergence test error results for convergence analysis of point spacings of 3.05-2.75mm

Point Spacing (mm)	Cortical Flexion % Error	Cortical Extension % Error	Cancellous Flexion % Error	Cancellous Extension % Error	Lysis Flexion % Error	Lysis Extension % Error
3.05	7.02	4.69	5.00	0.00	1.20	22.54
3.00	14.91	3.13	0.00	0.00	19.28	30.99
2.95	2.63	1.56	0.00	0.00	4.82	22.54
2.85	6.14	3.13	5.00	0.00	12.05	16.90
2.75	0.00	0.00	0.00	0.00	0.00	0.00

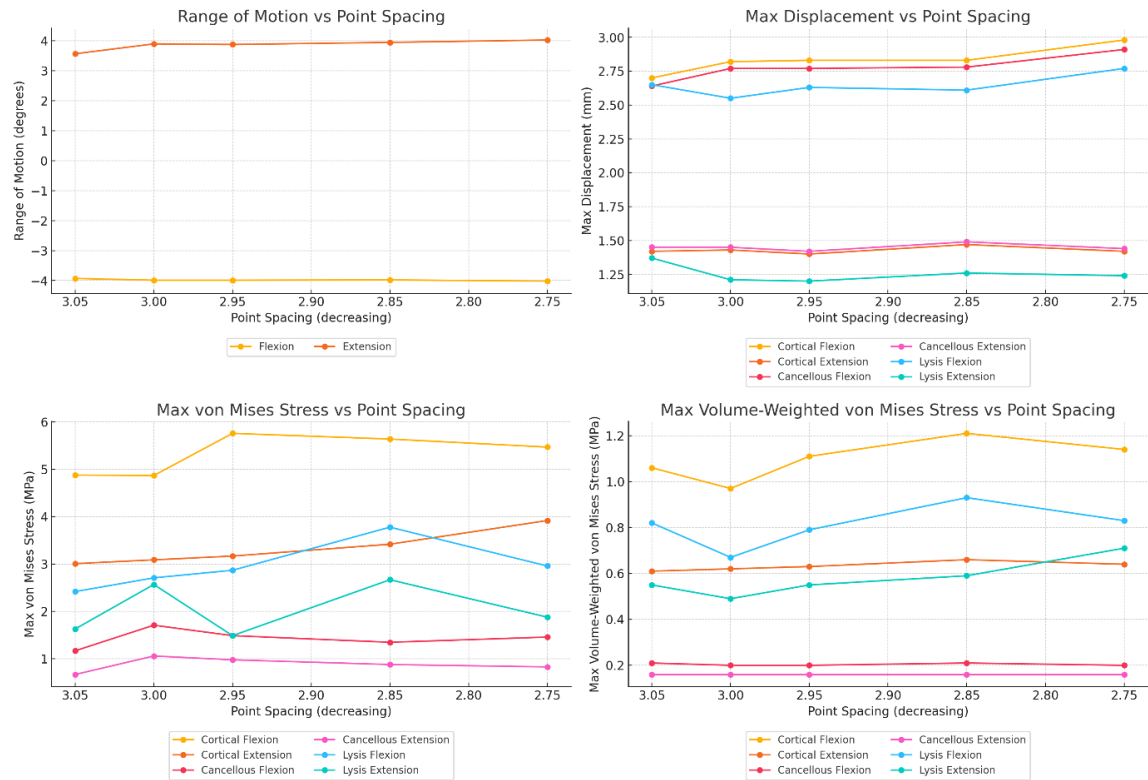


Figure 17: Mesh convergence analysis of mesh refinement from 3.1-2.75mm node distances (Point spacing)

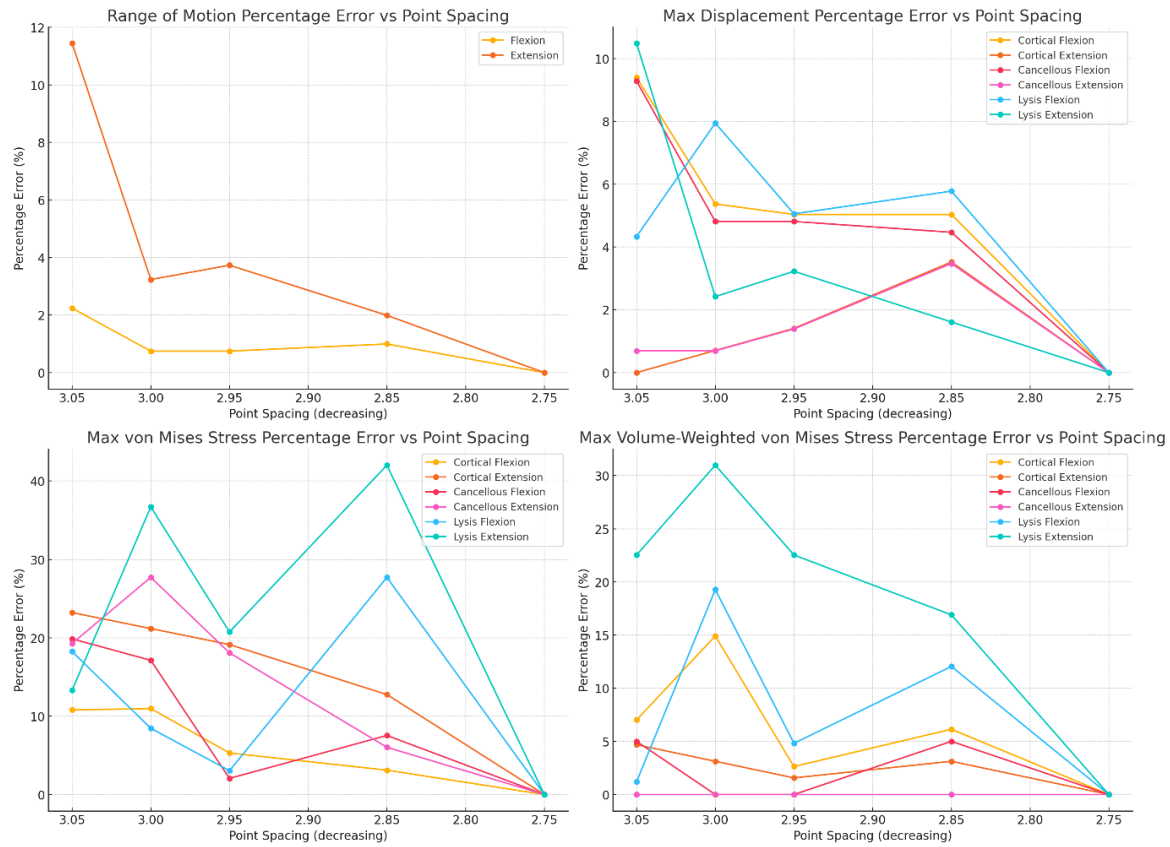


Figure 18: Mesh convergence analysis of percentage error for mesh refinement from 3.1-2.75mm node distances (Point spacing)

Table 19: Resultant total number of elements of each spinal convergence for convergence analysis of point spacings of 3.05-2.75mm

Anatomical Structure		Element type	Element number				
			Point spacing (mm)				
			3.05	3.00	2.95	2.85	2.75
Vertebrae	L4 Cortical bone	4-node tetrahedral	13334	13829	14345	15577	17000
	L4 Cancellous bone	element	26979	28476	30550	34528	38685
	L5 Cortical bone		12660	13072	13691	15122	17388

	L5		27308	28771	30434	34276	37588
	Cancellous bone						
	S1 Cortical bone		10105	10491	10796	11675	12577
	S1 Cancellous bone		21060	22289	23646	26916	30282
Intervertebral Discs	L4/L5	4-node	4273	4279	4619	5563	6256
	Annulus fibrosus	tetrahedral element					
	L4/L5 Nucleus pulposus		4154	4504	5018	5572	6206
	L5/S1 Annulus fibrosus		3307	3606	3511	4141	4496
	L5/S1 Nucleus pulposus		3092	3431	3562	4094	4596
Facet Joints	L4 IAP 1	4-node	213	226	232	253	259
	L4 IAP 2	tetrahedral	206	200	218	237	280
	L5 SAP 1	element	252	248	270	296	335
	L5 SAP 2		318	330	329	326	382
	L5 IAP 1		203	213	229	253	275
	L5 IAP 2		256	279	277	237	333
	S1 SAP 1		189	224	256	296	269
	S1 SAP 2		250	260	298	369	358
Ligaments	ALL		13 of 8 springs	-	-	-	-

		2-node	in				
		Spring	series				
	PLL	Elements	10 of 8	-	-	-	-
			springs				
			in				
			series				
	ISL		16	-	-	-	-
	ITL		10	-	-	-	-
	SL		2	-	-	-	-
	LF		20	-	-	-	-
	FCL	SAP1	19	19	21	21	21
		SAP2	20	18	18	19	23
		IAP1	19	19	19	21	21
		IAP2	19	22	19	23	22
Lysis		4-node	617	668	654	724	788
		tetrahedral					
		element					

4.1.2 Discussion

The mesh refinement process involved prescribing a series of point spacing parameters. These parameters were used to define the mesh of the spinal model at each iteration of the simulation. However, this process posed challenges for structures whose responses were dependent on the location of their attachment nodes. The ligaments had to be recreated for each iteration of the mesh convergence to redefine the attachment for the springs. The connection area and number of springs remained consistent, but the exact attachment points varied based on the most appropriate connection, determined by the Euclidean distance. Structures which also depended on specific defined thresholds for creation, were the facet joint surfaces, facet capsular ligaments and the disc face and lysis area. The area for these structures was consistent through the iterations, however, the number of elements changed depending on the element inclusion criteria, as displayed in Table 19. Small changes in attachment points and spacing between the ligaments and geometries of the facet joint

surfaces, disc and lysis regions had observable effects on the outputs analysed in this mesh convergence study. These factors could explain the divergence in the results, particularly in the lysis region outputs. The specific element size (point spacing) was chosen to be values of 3.05 mm, 3.00 mm, 2.95 mm, 2.85 mm, and 2.75 mm. The minimum element size was constrained by the volumetric meshing program TetGen, which encountered intersection errors when generating models with excessively small elements. The initial mesh size was 3.00 mm. To analyse mesh convergence, one coarser mesh and three refined mesh sizes were tested.

Further limitations were identified with the prescribed uniform mesh density of the vertebrae and discs. For each analysis a consistent element size was defined throughout, alternatively, a more complex yet superior approach would have been to utilise a non-uniform mesh. This method would allow for more complex meshing in detailed regions, such as the facet joints or areas of interest like the pars interarticularis. This approach would have provided a more detailed and reliable presentation of stress across the vertebra allowing for improved analysis. This approach could also have led to improved convergence by decreasing the size of elements in specific locations of the vertebra, without having to decrease the overall element size, and would reduce the computational complexity.

The convergence criteria for reaching a suitable mesh refinement were to achieve less than 5% error for the output parameters and to achieve a localised uniformity between the von Mises stress distribution across the L5 vertebra. A mesh tolerance of 5% was chosen to judge the model's reliability and sensitivity to changes in element size and organisation. Due to the complex surface geometry of the spinal components, the model would be sensitive to mesh size and therefore there is an expected level of error and results change between mesh refinement steps.

Von Mises stress outputs were the most sensitive to changes in mesh with varying percentage error trends, especially for the lysis group of elements which showed resistance to convergence. For both volume-weighted and local von Mises stress the results did not meet the ideal convergence tolerance. These results are attributed to the previously discussed limitations with the construction between convergence tests and small changes in the lysis geometry due to element selection.

Although, the stress results did not meet the ideal convergence tolerance, the fluctuations in stress were still in a physiologically relevant range for the L5 vertebra in a near intact stage. Comparing the von Mises stress distribution analysis of the L5 vertebra under extension to a similar study by Inceoğlu (2012)(111), the locations of maximum stress across the pars interarticularis are comparable, with stress magnitudes of 3Nm during extension for a low-intensity load distributed on the spinal unit.

The consistency of stress magnitude and localisation between finite element models reinforce the reliability of the model's structural response under physiological loading conditions. When considering the models application as a tool for clinicians to use in rehabilitation programs, quantitative stress results are less directly applicable to treatment. Each patient will have unique characteristics that alter the loading conditions, spinal anatomy and response. The model's ability to predict the overall response to loading conditions and the distribution of stress due to an individual's specific anatomy is of greater clinical relevance than precise numerical values. Therefore, from this convergence analysis the model can be deemed to be suitable for analysing biomechanical response patterns, and load distribution, as well as identifying physiological stress ranges that indicate 'safe' versus 'potentially harmful' loading conditions, for a patient-specific case.

Range of motion and vertebral displacement did meet the tolerance criteria for error for the point spacing of 2.85mm with only an exception for displacement for the lysis elements in flexion. Range of motion was the least sensitive to mesh refinement out of all the parameters because, as a global parameter of the L4-S1 model, it represents the integrated response of all the spinal structures. This effect means that at a coarser mesh, local discrepancies due to element size result in a reduced influence on the structure resulting in convergence. In contrast, von Mises stress is a local parameter which was analysed at just L5, in which peak stresses are captured in detailed regions. Therefore, smaller changes in mesh can significantly affect the distribution and magnitude of stress, increasing the complexity of convergence without the use of a fine mesh.

Even though range of motion converged at a coarser mesh of 2.95mm, a smaller mesh size was chosen, with elements of 2.75mm, to reduce error across all outputs. The related element numbers corresponding to this mesh size, for all components, are displayed in Table 19. This level of refinement increased the computational time from 174 seconds for a 3.05 mm mesh

to 255 seconds for a 2.75 mm mesh per simulation. While this will cumulatively extend the total run time across multiple simulations, reducing significant errors in all four outputs was essential for medical applications. Reproducing accurate range of motion values is crucial when developing a rehabilitation tool, as this parameter can be easily translated into clinical practice. By understanding and recommending appropriate motions recommended by this tool, clinicians can guide and inform patients of exact movements that support recovery without risking further harm to the healing spondylolysis. Therefore, achieving low error values in these two outputs is crucial as displacement will affect the range of motion. With overall errors of 3.98% for displacement and 1.5% for the range of motion, the outputs for the 2.75 mm mesh model were considered accurate.

Therefore, the mesh size for the two spinal models was determined to be set with the 2.75mm point spacing parameter. While finer meshes could have produced a further reduction in errors, limitations imposed by TetGen and the increase in computational time made 2.75mm the most practical choice for this study.

4.2 Validation

4.2.1 Results

4.2.1.1 Intact Model Validation

The patient 1 (P1) and 2 (P2) intact spinal models were subjected to pure bending moments of 2-8Nm with 0N of axial compressive force to replicate Jaramillo's(142) experimental analysis of L4-S1 segmental ranges of motion for 4 DoF. The results of these simulations are presented in Table 20. In comparison to Jaramillo's results on average the difference in range of motion was 2.79 and 2.43 degrees for segments L4-L5 and L5-S1 for patient 1 and 3.60 and 2.57 degrees for patient 2, with an overall decrease in range of motion for all loading modes and bending moments. The interquartile range method was used to identify outliers in the dataset to identify range of motion that deviated significantly. For both patient data sets, outliers were discovered in axial rotation loading modes, at 8Nm with a value of 3.48° and at moments 4, 6 and 8Nm with values of 5.51°, 8.73°, and 12.22° for patients 1 and 2 respectively. Analysis from the Pearson's coefficient correlation method for the total spinal segment (L4-S1) is outlined in Figure 19, for all data values (including outliers). The resultant Pearson correlation for patients 1 and 2 were 0.1355 and -0.1668 producing a weak positive and weak negative correlation result. When excluding the outliers from both data sets

the correlation increases to 0.6279 and 0.9327 providing a strong positive correlation between the data sets. As shown in Figure 20, the Jaramillo data for range of motion across the three loading modes exhibited a typical neutral zone curve, characterised by a rapid initial increase in range of motion in the low-resistance zone, followed by increasing resistance as the neutral zone ended. In contrast, the data from Patient 1 and Patient 2 displayed a more linear increase in range of motion with increasing load for both spinal segments, with an overall stiffer response to load.

Table 20: Comparison of finite element model data of intact models of patients 1 and 2 to experimental data (142). Loading mode of a pure moment of 2-8Nm for 4 bending modes.

Segment	Loading Mode		Range of motion (Degrees)		
		Bending moment (Nm)	L4-S1 Patient 1	L4-S1 Patient 2	Jaramillo (2016)
L4-L5	Flexion	2	0.33	0.57	3.7
		4	0.66	1.15	5.3
		6	0.99	1.74	5.8
		8	1.32	2.33	6.7
	Extension	2	0.33	0.59	2.3
		4	0.69	1.21	3.2
		6	1.07	1.87	3.5
		8	1.47	2.58	3.9
	Left Lateral Bending	2	0.47	0.7	3.5
		4	0.96	1.42	4.5
		6	1.46	1.95	5.4
		8	1.8	1.88	5.9
	Right Axial Rotation	2	1.03	2.6	1.2
		4	1.98	5.51	1.7
		6	2.82	8.73	2.1
		8	3.48	12.22	2.2
L5-S1	Flexion	2	0.54	0.5	4.4
		4	1.08	1.01	5.4

		6	1.62	1.55	6.4
		8	2.17	2.1	7.0
	Extension	2	0.51	0.49	2.9
		4	1.04	1	3.6
		6	1.6	1.54	3.7
		8	2.18	2.1	3.9
	Left Lateral Bending	2	0.58	0.7	2.1
		4	1.17	1.42	2.8
		6	1.78	1.95	3.3
		8	2.41	1.88	3.7
	Right Axial Rotation	2	0.22	0.01	1.3
		4	0.43	0.01	2.0
		6	0.61	0.05	2.4
		8	0.76	0.12	2.7

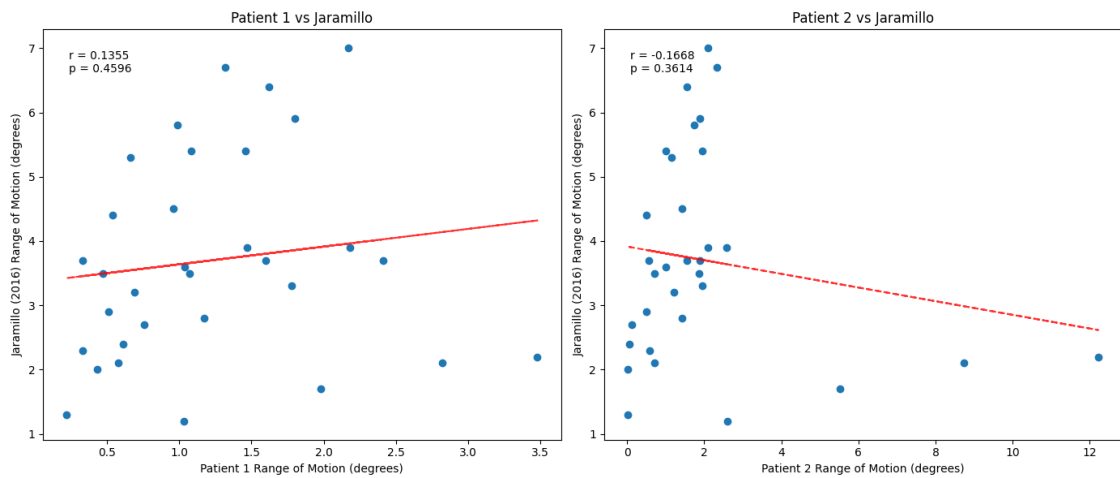


Figure 19: Pearson's correlation coefficient data for patient 1 and 2 against Jaramillo's experimental data for the total spinal segment L4-S1.

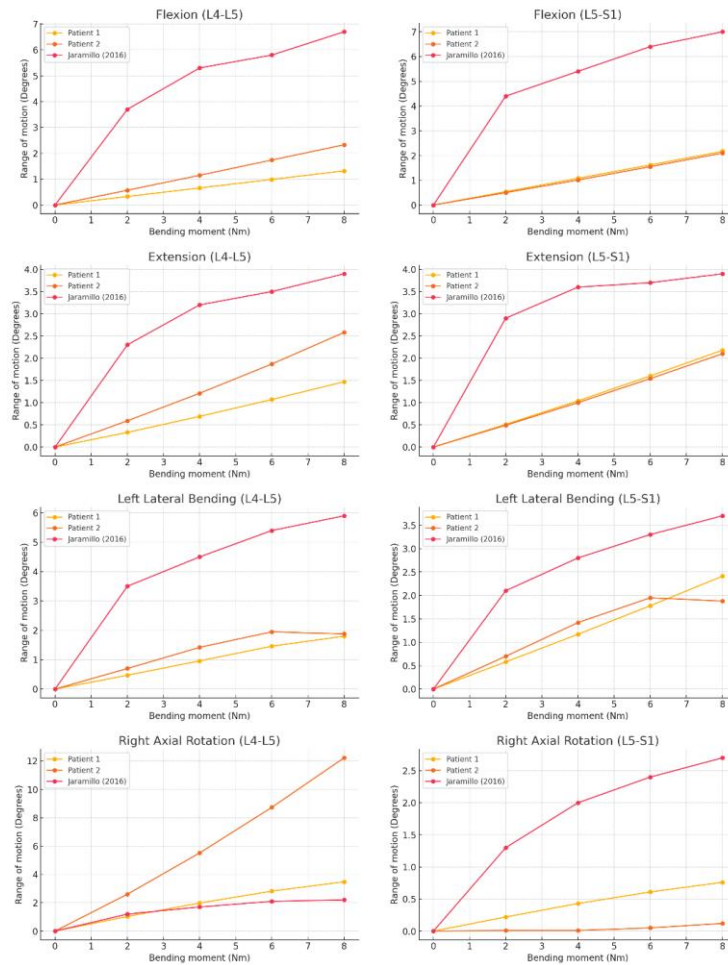


Figure 20: Comparison of the finite element model L4-S1 patient 1 and 2 range of motion data to Jaramillo's experimental spine data: validation of the trends of RoM

4.2.1.2 Fracture Model Validation

To analyse the fracture healing models von Mises stress analysis and validation were completed with a comparison to the finite element spondylolysis L3-S1 model generated by Sairyo (2005)(114). All three models had a lysis defect located in the left pars interarticularis at the L5 vertebra and analysed stress under loading conditions of 400N of axial compression and moments of 10.6Nm for 6 DoF. Analysis of maximum von Mises stress in the cortical and cancellous bone is displayed in Figure 21, for the intact and granulation cases. A comparison of the intact and spondylolysis cases shows a significant increase in von Mises stress in the cortical bone after fracture. For the granulation data the loading modes which resulted in the greatest von Mises stress were flexion, extension, left lateral bending (ipsilateral) and right axial rotation (contralateral) with values of 21.38, 18.89, 12.91 and

18.98 MPa respectively. These trends of maximum stress mirror Sairyo's results which produced maximum von Mises stress values of approximately 14, 13, 29, 8, 25 and 40 MPa for flexion, extension, lateral bending (left, right) and axial rotation (left, right) respectively. Lateral bending right data for patient 2 was not included in the Figure 21 data analysis, due to the negative Jacobian errors produced resulting in a terminated simulation. Analysing the distribution of von Mises stress from Sairyo's model revealed that the highest stress concentrations were consistently located at the dorsal and ventral walls across all bending moments. The locations of stress are replicated in both patient 1 and 2 as represented in the colour plots in Table 23A. For the granulation model, maximum von Mises stress was located at the contralateral ventral wall for extension and flexion. For lateral bending and axial rotation increased von Mises stress values were distributed across the contralateral pars interarticularis with maximum values located at the dorsal wall. For the intact models, the stress distribution was generally uniform across the posterior arch of the vertebra, with only minor increases at the ventral edges of the pars interarticularis bilaterally. Patient 2 presented with increased von Mises stress distribution and magnitude on the contralateral side of the fracture with migration in stress to the posterior pedicles and lamina regions. For right lateral bending patient 2 also had increased stress distribution into the inferior articular process due to collision and intersection between the L5 and S1 facet joints which resulted in the negative Jacobian error.

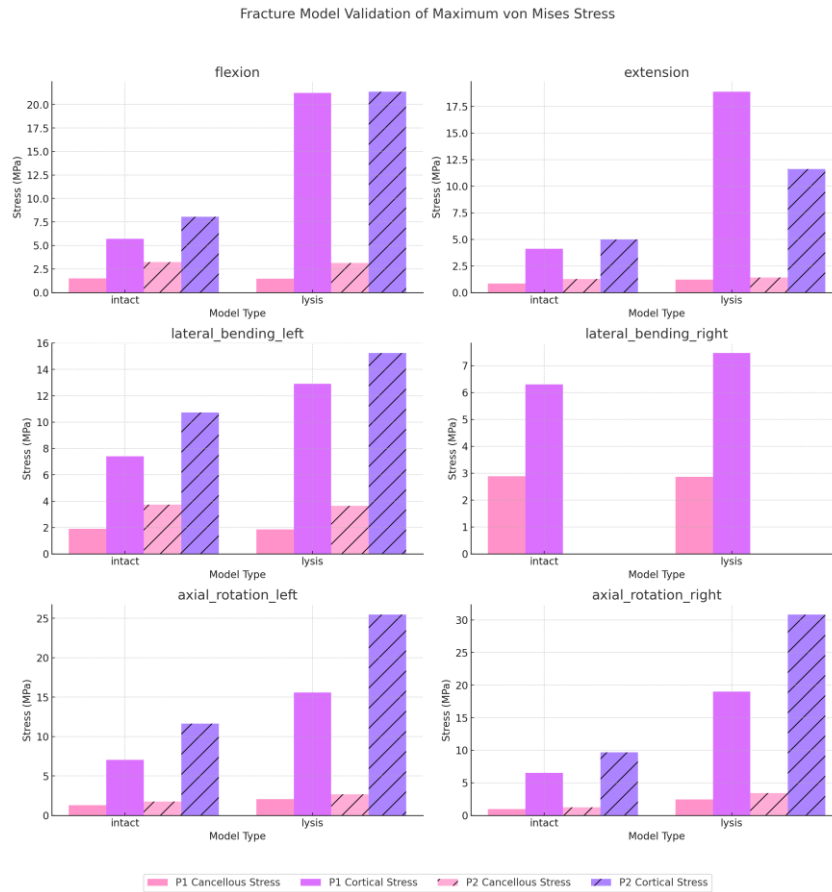


Figure 21: Von Mises stress data for the cortical and cancellous regions of L5 for intact and granulation models for patient 1 and 2.

4.2.2 Discussion:

The finite element models in this study were validated against experimental and computational data for intact and fractured spinal models that were available in literature. Validation of a lumbosacral spinal model is challenging due to the multi-component interactions of the vertebrae, discs, ligaments and facet joints and structures not included in this model such as the muscles of the lower back, which play a key role in controlling motion and mechanical response under loading. The aim of this study is to produce a FE framework to inform conservative treatment, that is flexible and scalable for patient data input. This produces further challenges for validation due to interindividual variability such as differences in anatomy, loading and tissue properties which should be considered in clinical application. Such variability makes it difficult to validate the framework using a limited number of datasets. Challenges in direct validation arose during data collection, particularly for recent spondylolysis models, including those that incorporate the S1 vertebra. As a result,

analysis of intact and fractured models was completed separately in an effort to validate the framework by comparison with similar models/specimens and loading conditions.

The analysis of the intact model's range of motion outputs revealed that both patient models exhibited lower ranges of motion and stiffer responses to loading compared to the experimental results. The stiffness of the model was reflected in the linear increase of motion of the range of motion compared to the expected initial steep increase in motion presented in Jaramillo's experimental results. Range of motion validation was challenging given the simplicity of the ligament and intervertebral disc components in this model. These two structures are essential in the spine's deformation response under load, significantly contributing to the non-linear behaviour observed in load-deflection curves for the total range of motion. The non-linearity arises from the straightening and reorientation of the fibres, leading to an increase in stiffness following this process.

Validation of these soft tissues is complex and multifactored due to their time-dependent behaviour and interactions with the surrounding structures. Recent studies, such as Turbucz (2022)(145) and Newman (2020)(146), modelled the soft tissue collagen fibres using a unique force-strain curve to capture the initial "toe" region, followed by a linear increase in stiffness. These curves are influenced by the initial elongation/ stretch or length of the ligaments or annulus fibrosus fibres, their stiffness, and the percentage strain. Replicating the ligament length in patient-specific models can be challenging. Ligament length can be obtained from, literature, experimental data or through measurements from CT imaging. However, experimental measurements may not reflect in situ lengths due to differences in mechanical testing setups or tissue changes after dissection. Imaging-based measurements can also introduce inaccuracies due to variations in the subject's posture during scanning. With these inaccuracies over or underestimation of the length of the ligaments may be introduce laxity or tension for certain ligaments. Minster (2023)(147) developed a model to account for this variation testing a range of ligament lengths to ensure the ligaments were in realistic ranges of elongation to account for the mechanical variation. The inclusion of fibre material behaviour in these models is crucial for capturing the characteristic range of motion observed in experimental studies. Therefore, to improve the response of the spine to motion this should be captured in future iterations of this framework to enhance the clinical applicability of the model for conservative treatment. If the range of motion can be accurately captured, then

patient-specific movements can be analysed to predict the implications on the fracture healing or progression.

Von Mises stress was selected as the criterion for fracture model validation due to its ability to highlight potential failure locations in the bone, which is crucial for preventing the progression of spondylolysis. The patient 1 and 2 models produced comparable von Mises results of location and magnitude to Sairyo's(114) computational model. The predicted stress locations were met, with the cortical bone of the contralateral pars interarticularis experiencing increased stress concentrations with an acute defect. The concentration of stress at the dorsal and ventral walls of the pars interarticularis region demonstrates how unilateral fractures can progress to bilateral fractures with repeated stress in this area. Motion trends which aggravated the contralateral pars were also consistent with current spondylolysis models with extension, left lateral bending and right axial rotation causing the largest von Mises stress. Flexion also produced higher von Mises magnitude results than expected likely due to the model stiffness. The contralateral pars compensates for the reduced structural integrity of the vertebrae leading to asymmetrical high concentrations of stress during these motions. However, by achieving bony union from mature bone recovery, the symmetrical stress distribution seen in intact models can be restored. Therefore, conservative treatment plans should closely monitor these elevated stress levels, tailoring recovery to promote healing and restore balanced stress distribution, ultimately preventing further fracture progression.

Where the exact range of motion or von Mises stress magnitudes were not captured, the overall data correlation remained positive and replicated the trends of the experimental data, particularly after excluding outliers. Using correlation as an analysis of validation is a superior method to trying to replicate exact numerical values between two different models and their outputs. Even though this study is comparing replicable loading conditions on a spinal model to the validation models, variations in material parameters, spinal geometries and boundary conditions will influence the results. A high correlation, however, indicates that the model reliably captures the trends of range of motion, stress and behaviour of the system, even with biological or methodological variability. For this application, numerical variability is expected between patients due to different pathologies, spinal presentations or physiological factors. When considering clinical applications such as conservative treatment

programs, the focus is not necessarily to restore RoM to a precise numerical value, but instead suggest a range of safe motions. This will aid understanding of how various movements and loads might impact the healing or progression of spinal injuries. In addition to differences seen between validation models, variation was demonstrated between the two patient models analysed in this study. As the underlying framework, including loading and boundary conditions, and material properties, remained consistent, the only variable was the patient-specific anatomical variation, further demonstrating the need for correlation as a comparative validation metric. For example, the second patient model produced increased rotations and von Mises stress across all loading modes showing the sensitivity of the outputs to changes in geometry. Differences in the patient models include disc height and size depending on the vertebral endplate geometries. The disc size was consistently smaller in the second patient model, likely due to the participant's age, though their exact age was not known. Disc height differences meant that there were differences in the facet joint alignment as the disc degeneration reduces the distance between the facet joints increasing the pressure between the surfaces. There was also an observable difference in the seven-process width length and orientation between the models which will influence ligament connection, length and elongation. The ligaments would also have been affected by the postural changes noticeable in the scans where patient 2 was leaning laterally. Accommodations were made to reduce this rotation in the model to reduce the effects of laxity and tension.

These anatomical differences between the models will have contributed towards the difference in magnitude of the range of motion and stress as well as the errors in the lateral bending and axial rotation of the second model. For patient 2 during left lateral bending there was a collision between the facet joints causing a slippage of the vertebra affecting the magnitude of the results and increasing the stress in the facet joint region. The difference in results due to spinal anatomy shows the importance of patient-specific models when considering conservative treatment frameworks. Anatomical differences in soft tissues or bony structures lead to variations in results, making a one-size-fits-all model inadequate for capturing the diversity within the population. This variability also extends to patient characteristic factors such as tissue degeneration, age-related changes, and pre-existing conditions that also influence spinal biomechanics. Therefore, showing the clinical need for a flexible patient-specific framework to tailor an individual's treatment to their unique

condition. This not only improves the likelihood of successful treatment but also prevents future injury or complications.

When analysing the range of motion trends, the results seemed to diverge past the point of the neutral zone, where the bending of the spine reaches the threshold initiating increasing resistance to deformation. However, for conservative treatment, physical therapy exercises within the neutral zone are recommended to stabilise and protect the fracture zone from shear forces that can damage the healing tissue or the surrounding structures, such as the contralateral pars. Therefore, this model is suitable for simulating low loading conditions typical of early conservative treatment, where range of motion and von Mises stress are both validated.

Further validation of the model would be valuable for assessing the various stages of fracture healing examined in this study. However, to the author's knowledge, no previous models have been developed on the mechanics of spondylolysis fracture healing, and there has been a reduction in the number of recent studies focused on this condition.

4.3 Sensitivity Analysis

4.3.1 Results

4.3.1.1 Material Properties Sensitivity

Ligament Material Properties

The sensitivity results for the L4-S1 range of motion and L5 von Mises stress in the cortical and cancellous bone from the linear regression analysis of the ligament stiffness trials are displayed in Figure 22 for the six loading modes. For patient 2's range of motion results, sensitivity values were negative with the inclusion of trial 3 showing the inverse relationship between increased ligament stiffness and range of motion. Patient 1 was mostly in agreement apart from for axial rotation right which initially was at the highest overall sensitivity value of 1.83, but it slightly decreases as more trials are accumulated with increasing stiffness of the ligaments.

Results from the cumulative sensitivity for von Mises show that for cortical and cancellous stress, axial rotation left and right had the highest sensitivity to ligament stiffness with maximum values of 0.76 and 0.59 for patient 1 during trials 1-2. The overall trends for von Mises were low with a slow decrease in sensitivity with the addition of trials and the seven-

ligament stiffness (Young's modulus) values increasing. An outlier was observed in patient 2, where a notable increase in sensitivity occurred for right axial rotation. The values significantly rose from -0.08 to 0.34, with ligament stiffness exerting greater influence as additional trials were incorporated. Cancellous von Mises stress presented with low and flat sensitivity trends displaying the lack of influence of stiffness on this structure.

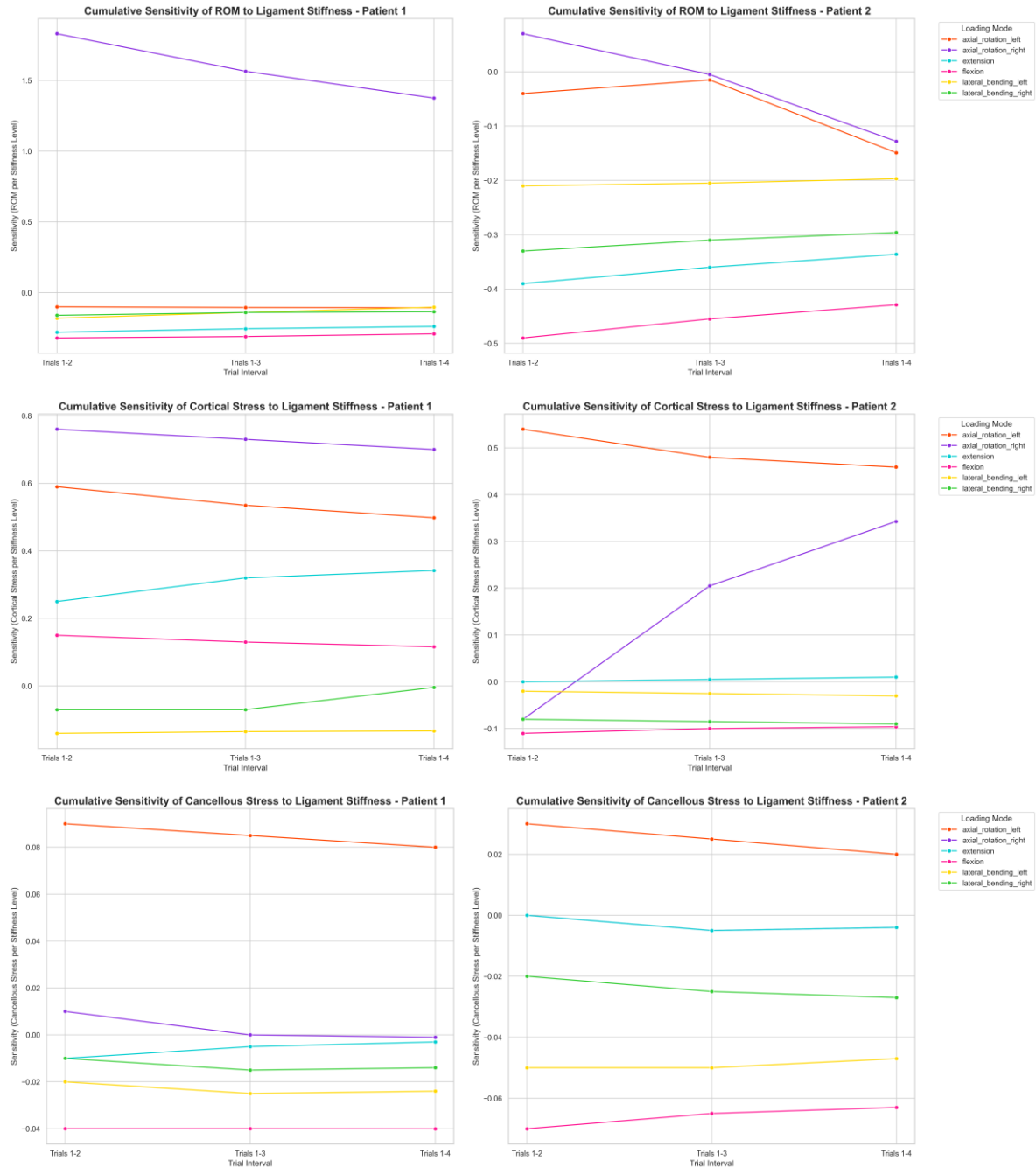
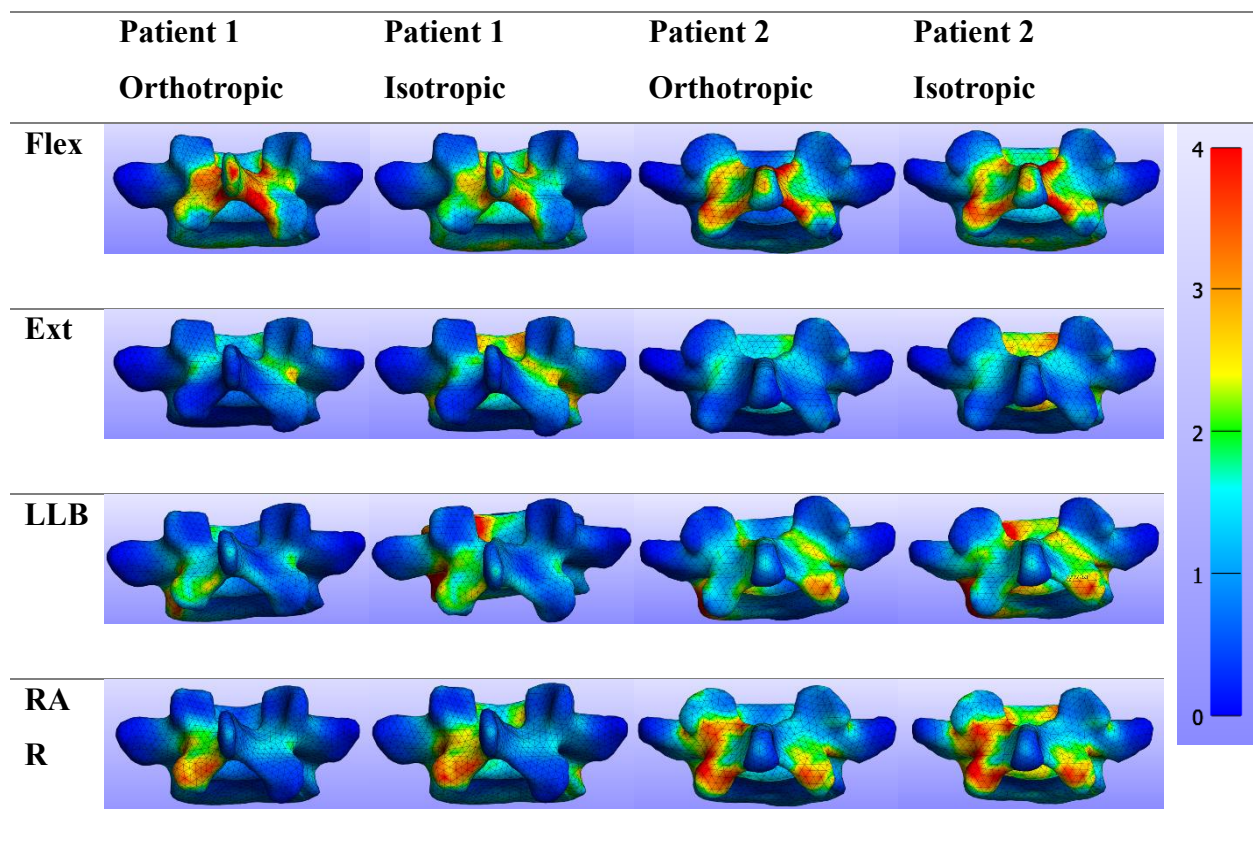


Figure 22: Sensitivity results from the linear-regression analysis of ligament stiffness and von Mises stress or Range of motion for patient 1 and 2.

Vertebrae Material Properties

The sensitivity analysis for the material model comparison of isotropic and orthotropic intact vertebra values is displayed in the form of von Mises stress colour maps in Table 21. Analysis of the colour maps and the magnitude of von Mises stress clearly shows that the Isotropic models produced increased stress throughout the vertebra. For the patient 1 and 2 models under flexion loading the magnitude of cortical stress was 10.95MPa and 12.24 MPa in comparison to 5.69 MPa and 7.5MPa for isotropic and orthotropic models respectively. The flexion loading mode had the most visible difference in stress concentration area, with small areas of increased stress located around the posterior lamina compared to the orthotropic model which had increased distribution in the pars and lamina area. Extension loading produced the smallest values of 6.39, 7.22, 3.96 and 3.75MPa for patient 1 and 2 for isotropic and orthotropic models respectively. Extension produced the most consistent areas of stress between the isotropic and orthotropic models which were localised to the pars interarticularis at the dorsal and ventral walls primarily. Lateral bending left produced the greatest difference in stress magnitude between the material models with values of 12.91 18.71, 6.11 and 9.28MPa for patient 1 and 2 for isotropic and orthotropic models respectively. Areas of increased stress for this loading mode were focused on the pars interarticularis and inferior articular processes with isotropic models presenting with a larger area of stress distribution. Finally, the axial rotation right loading mode produced the most comparable maximum stress magnitudes, with values of 6.8, 7.81, 5.73, and 7.05 MPa for patients 1 and 2 isotropic and orthotropic models respectively. The distribution of load was consistent between the isotropic and orthotropic models with primary concentrations of stress located across the left pars interarticularis walls and both inferior articular processes. Between the patient models there were noticeable differences in the stress location and magnitude, with patient 2 producing larger von Mises stress values for all identical loading modes to patient 1. Both patient models frequently produced high-stress values at the pars interarticularis, however, patient 2 also presented with more frequent presentations of stress in the inferior articular processes and outer lamina walls.

Table 21: Vertebrae material isotropic and orthotropic material model analysis for loading modes using colour maps



Intervertebral Disc Material Sensitivity

The sensitivity analysis of the intervertebral disc material model on range of motion outputs for the L4-L5 and L5-S1 segments, along with comparison validation study data and Jaramillo's (2016)(142) experimental study, is presented in Figure 23. Overall, the Mooney-Rivlin model resulted in a reduction of range of motion values for all loading modes and lumbar segments in comparison to the Neo-Hookean models and the experimental data. Similar to the Neo-Hookean model, the Mooney-Rivlin model also produced a linear range of motion curves and did not capture the desired non-linear increasing spinal stiffness with increasing load that was desired. The Pearson's correlation coefficient r values, displayed in Figure 24, when compared to Jaramillo's experimental data, produced values of 0.4798 and -0.2690 for patients 1 and 2, indicating a moderate positive correlation and a weak negative correlation. This is a minor difference from the original Neo-Hookean r values of 0.1355 and -0.1668.

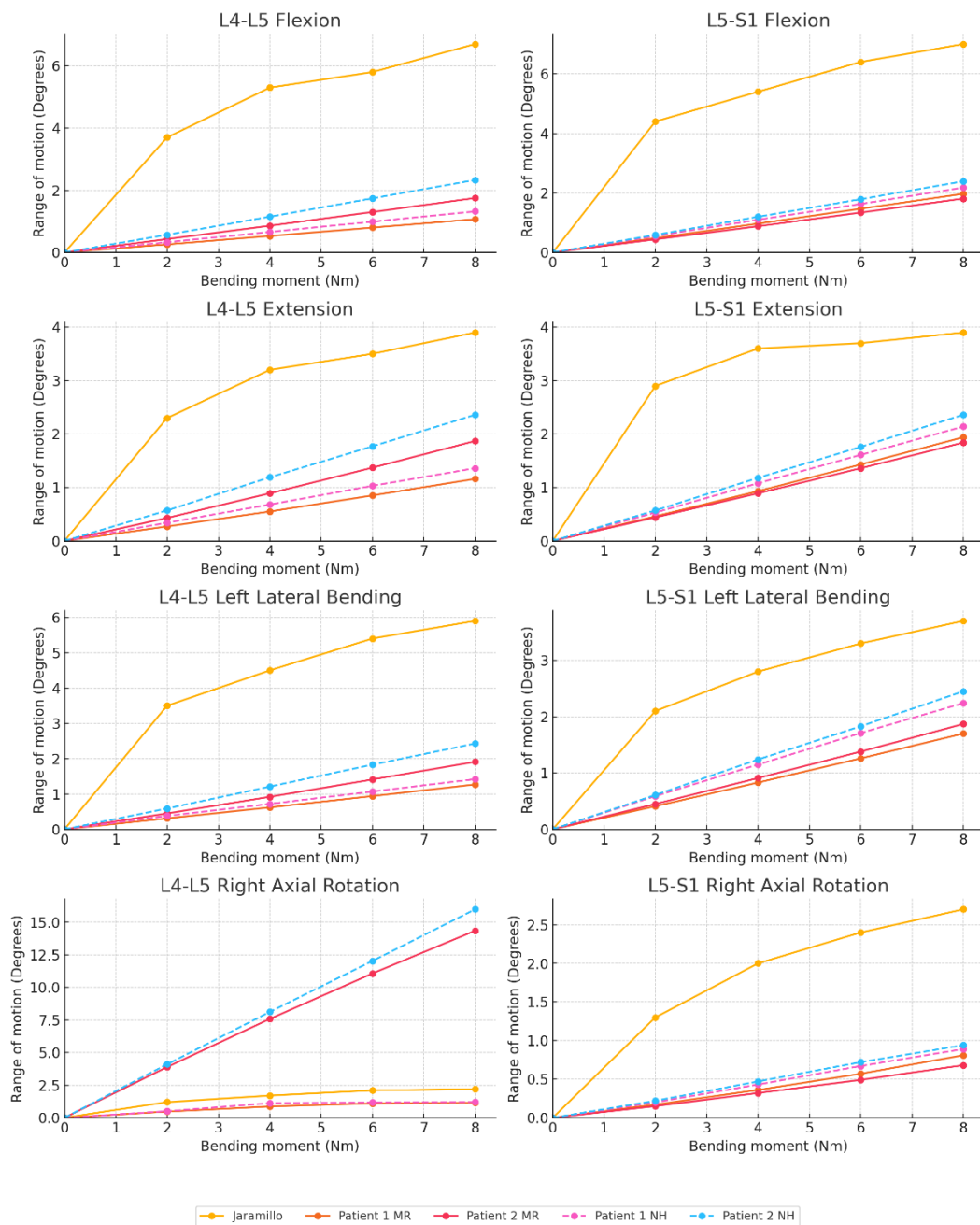


Figure 23: Comparison of the intervertebral disc material properties for L4-S1 patient 1 and 2 range of motion data to Jaramillo's experimental spine data, for material models Mooney-Rivlin (MR) and Neo-Hookean (NH).

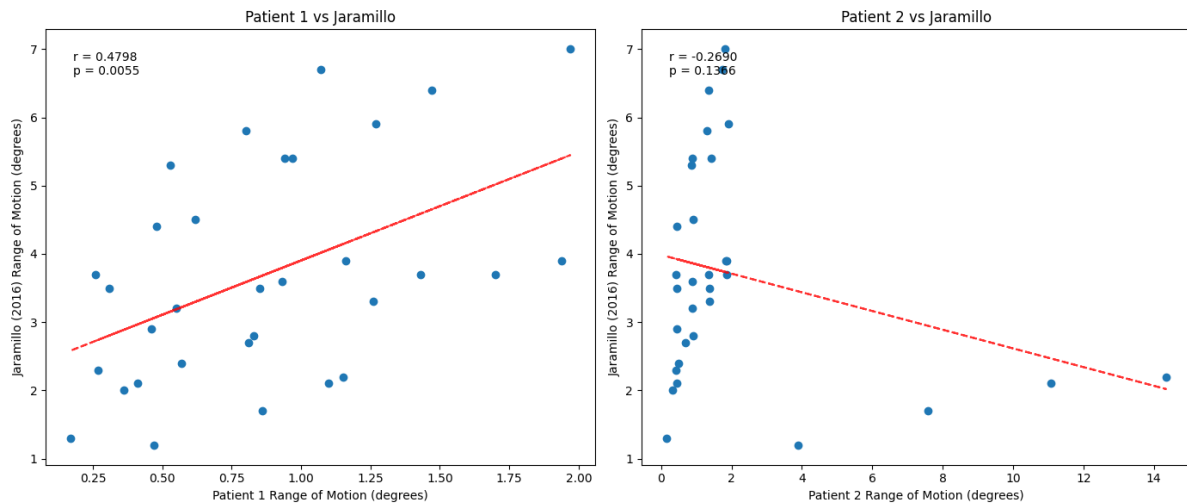


Figure 24: Pearson's correlation coefficient data for intervertebral disc sensitivity analysis against Jaramillo's experimental data for the total spinal segment L4-S1.

4.3.1.2 Fracture Characteristics

Fracture sensitivity was completed for lysis location and width, where location was analysed for two unilateral fractures in the left and right pars interarticularis, with the width of a left unilateral fracture being altered between 1-4mm. Analysis of cortical and cancellous von Mises stress for patients 1 and 2 were completed for the location analysis, displayed in Figure 25. For flexion left there are consistently elevated von Mises stress averaging at 20.07MPa for cortical stress and 2.38 MPa for cancellous stress. In contrast, extension shows variability in the results, with right-sided fractures producing lower cortical stress values of 8.67 MPa compared to 15.25 MPa for left-sided fractures. Trends from the validity study were reproduced, showing that ipsilateral lateral bending and contralateral axial rotation resulted in increased von Mises stress values for both left and right fractures. For left fractures, there was a 68.50% and 75.85% increase in left lateral bending and right axial rotation, respectively, compared to right fractures. Conversely, right fractures exhibited a 108.03% and 75.85% increase in right lateral bending and left axial rotation, respectively, compared to left fractures. Analysis of left unilateral fracture widths was completed for the loading modes that produced the highest cortical stress as seen in Figure 26. Results were consistent for the cancellous bone with an average von Mises stress of 2.31, 1.54, 2.46, 2.73MPa for flexion-extension, lateral bending and axial rotation for patients 1 and 2. However there was variation between the two patients for cortical von Mises stress. Patient 1 exhibited a slight increase in

stress, reaching 12.33 MPa in flexion and 17.14 MPa in extension, and a decrease to 12.40 MPa and 15.70 MPa for lateral bending and axial rotation, respectively, after a fracture width of 2mm, where the stress values plateaued thereafter. Patient 2 produced a spike in results at 2mm before decreasing to a consistent stress value at 3mm, with an increase in results of 19.91, 15.95 and 26.34MPa for flexion, lateral bending and axial rotation and a decrease in extension to 7.93MPa.

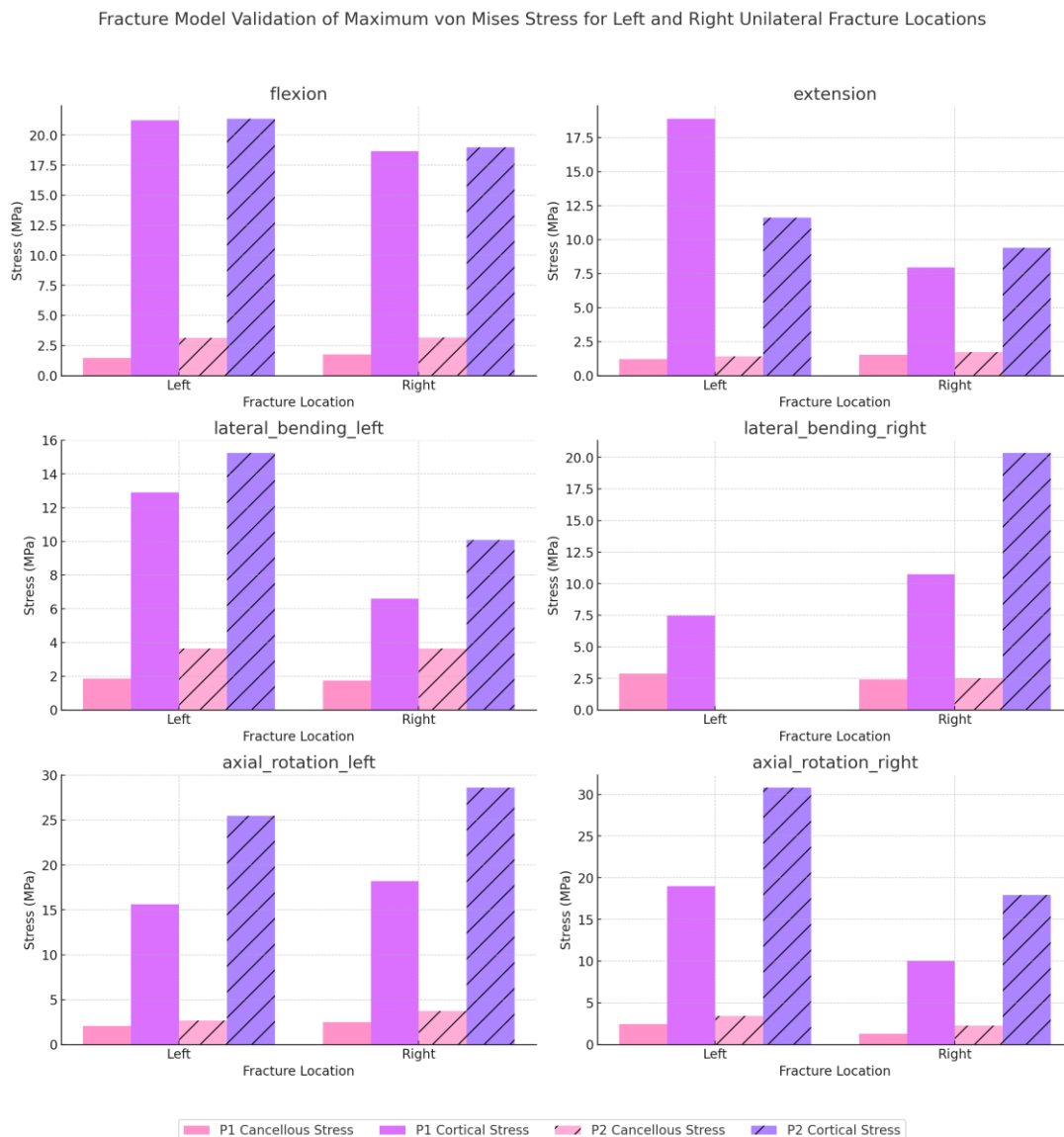


Figure 25: Fracture location sensitivity analysis for patient 1 and 2 across 4 loading modes.

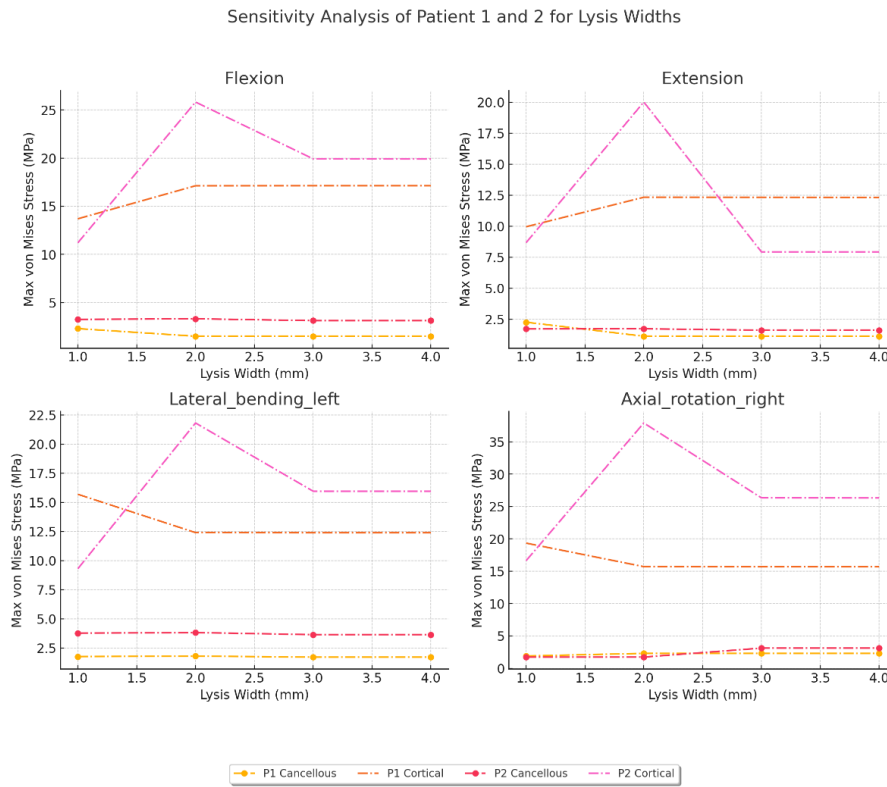


Figure 26: Fracture size sensitivity analysis for 1-4mm fracture widths for patient 1 and 2 across 4 loading modes.

4.3.2 Discussion

Sensitivity analysis was performed to analyse the change in output with variations of input parameters for four components of the spine. The first parameter of interest was the ligament material Young's modulus. For simplicity, this model represented all seven spinal ligaments as a group of linear springs. However, after validation analysis, this approach was theorised to contribute to a reduced range of motion for lateral bending and flexion-extension and reduced the expected response of increased stiffness to increasing movement. The effect of increasing Young's modulus of the springs was analysed across three trials, each representing a higher ligament stiffness level, and compared to the parameters from trial 1. A reduction in stiffness or ligament Young's modulus could not be performed due to limitations in the lateral bending and axial rotation loading modes. These resulted in L5-S1 articular process intersections, leading to negative Jacobian errors and simulation failure, as movement restrictions at higher loads were insufficient. For the range of motion outputs, a predominantly negative sensitivity was found across all loading modes except for axial rotation right for patients 1 and 2. These trends resulted from the reduction of range of

motion with increased ligament stiffness, which if this was a non-linear model where ligament Young's modulus increased with strain, indicates that the correct spinal response would occur.

In both models, axial rotation RoM showed variability, with an initial increase in axial rotation right for patient 1 and axial rotation left for patient 2. However, as the trials progressed, ligament stiffness had a reduced influence on axial rotation. This suggests that other stabilising structures, likely the facet joints, became the primary contributors to restricting motion in this loading mode. This shift in load distribution was supported by the increase in cortical bone von Mises stress during axial rotation and extension with increased ligament stiffness. These loading conditions are less influenced by ligamentous restraint, resulting in an increased load share on the facet joints. This load is transferred through the vertebra's posterior arch, particularly in extreme movements such as hyperextension, where impingement or rotation of the inferior processes causes increased lamina stress (148). Conversely, in flexion and lateral bending, increased ligament stiffness led to a decrease in von Mises stress, as the multiple posterior arch ligaments absorbed a greater share of the load and reduced the forces transferred to the cortical bone.

Flexion, extension and lateral bending range of motion sensitivity remained constant throughout the trials suggesting stability in the model's results. Unlike axial rotation, these motions are primarily restricted by the ligaments, which with the linear model used in this study, behaved consistently under the loading conditions. This response was also captured by the intervertebral disc sensitivity analysis where a linear increase in range of motion was found in both neo-Hookean and Mooney-Rivlin models due to the lack of non-linear fibre incorporation. The change in the material model did not have a visible effect on capturing the resistance to deformation behaviour of the annulus fibrosus fibres or ligaments. Both models exhibited a behaviour where the total range of motion resembled that of the neutral zone (NZ, shown in Figure 6). Therefore, whilst stability in the results is observed, further refinement with non-linear fibre models may be necessary to ensure spinal movement accuracy, particularly at extreme ranges of motion and loading conditions. As the Mooney-Rivlin model did not produce an improvement in the range of motion, the original neo-Hookean model was used for the conservative treatment framework models, to reduce computational time and complexity. This decision was maintained for the other elastic soft tissue sections of

the facet joint cartilage and the fracture/ fracture healing elements. This material model is the simplest of the hyperelastic models (149), however this is advantageous for this study with the multiple component model that required prolonged simulation time.

Capturing the vertebral and fracture biomechanical responses was crucial to the production of an accurate healing model of spondylolysis. Therefore, multiple sensitivity analyses were performed to analyse the effects of bone models and fracture presentation. The effect of isotropic and orthotropic models was presented clearly in the colour map analysis of the L5 vertebra and in the increase in the magnitude of stress in the isotropic models. Isotropic models consider the bone to be homogeneous in all directions which is not structurally accurate with the rods and plates(150) structure of the cancellous bone. As a result, the isotropic model led to an overestimation of von Mises stress and displayed broader, less specific areas of stress concentration in the colour maps compared to the orthotropic model, which better captures the directional properties of bone tissue. As this study aimed to develop a biomechanical framework for optimising patient treatment, overestimations in stress could lead to misguided clinical decisions. Inaccurate stress predictions may result in treatments that overcompensate for the weakness of the vertebra misguiding and potentially unnecessarily prolonging treatment. One of the key challenges in current conservative treatment for spondylolysis is low patient compliance, often due to the lengthy nature of treatment plans. Therefore, the isotropic model was not a suitable model for this study and the orthotropic material model was used for L4, L5 and S1.

Analysis of the unilateral fractures provided promising results for model accuracy for the location study. Left and right unilateral fractures were analysed for P1 and P2 where both models produced the expected result of increased von Mises stress in ipsilateral lateral bending and contralateral axial rotation. This was consistent with the findings from the comparative analysis of the validation study against computational spondylolysis models(114), indicating that the model accurately captures the biomechanical trends under various loading conditions. Fracture width analysis presented differences in von Mises stress between cortical and cancellous bone. Cancellous bone exhibited consistently low stress magnitudes across varying fracture widths, while cortical bone showed higher and more variable stress levels, especially in patient 1. For patient 1 there was an increase in stress at a width of 2mm which then remained consistent over the tests. This 2mm threshold marks the

transition from a fracture that can be managed conservatively to one that is considered severe (151). Therefore, this spike in stress is representative of the progression of the condition's severity. In contrast patient 2 did produce an unexpected result, showing an overall drop in von Mises stress beyond the 2mm width during extension tests. As this study is analysing the healing of spondylolysis fractures, the final chosen width for the conservative treatment framework tests was 4mm. This choice was made for two key reasons. First, the 4mm width produced a plateau in von Mises stress during the sensitivity analysis, reducing the likelihood of stress spikes observed at the 2mm mark. The second consideration was for an increase in fracture geometry due to callus formation in the region of the healing of the pars interarticularis area. The callus formation typically expands around the fracture site, therefore making the 4mm width more representative of the physiological changes during recovery.

Compared to the disc and ligament configuration in this model, the vertebra and fracture geometry are highly patient-specific, as the original geometries were derived from patient imaging. The unique response to loading between patients can be seen in the vertebra material study between patients 1 and 2. Analysis of the intact L5 vertebra under loading of patient 1 produced localisations of stress primarily on the left posterior regions of the lamina and pars area with migration to the left inferior articular process during lateral bending and axial right rotation. In contrast, patient 2 presented with a more balanced and elevated distribution of stress in the ventral regions of the pars interarticularis and outlining the posterior and outer walls of the articular processes. Additional differences in the models were displayed in the fracture location and width studies where patient 2 experienced a reduction in von Mises stress for the moments of extension compared to patient 1. Patient 1 presented a reduced sensitivity to fracture widths in comparison to patient 2. The presentation of the results suggests that the patient-specific differences between patient 1 and patient 2 models in spinal presentation, and hence response to fracture, produce a greater difference in results than other sensitivity inputs, such as ligament stiffness or material laws of the discs or vertebrae. This difference highlights the need for patient-specific biomechanical analysis in the treatment and prevention of spondylolysis.

The locations of stress accumulation, particularly in the pars interarticularis, could predispose these areas to stress fractures under repetitive or high-loading conditions, leading to microdamage or fracture propagation. This sensitivity analysis suggests that these vulnerable

locations depend on the anatomical differences and material property differences of the vertebra. By identifying these patient-specific stress patterns clinicians can better target preventative strategies or conservative treatment strategies and durations, tailored to each patient's biomechanical vulnerabilities.

Chapter 4: Part 2, Conservative treatment framework

Conservative treatment analysis was completed for the patient 1 model after thorough examination of the models during validation, sensitivity analysis and convergence analysis. Based on the sensitivity analysis and verification, patient 1 demonstrated stronger Pearson's correlation coefficient trends relative to experimental spinal data and successfully completed all trials without simulation failure due to negative Jacobian errors across all loading modes. To produce a complete framework for all 6 DoF this was crucial and therefore patient 1 was the chosen model to produce the conservative treatment framework. Patient 1 therefore was analysed for all 6 models of an intact pars interarticularis and 5 models of fracture healing phases: granulation, fibrous tissue, cartilage, immature bone and mature bone. These models were simulated under a progressive series of loading conditions of a reference body load, standing, walking and moderate running to obtain the outputs range of motion and von Mises stress in the cortical cancellous and lysis sections. Each model was analysed with incrementally increasing bending moments for the 6 DoF, for 0-8Nm paired with the compressive body load.

For each activity analysed, a dataset of 180 simulations was produced. Range of motion data is displayed in Figures 27, 29, 31, and 33 for the 6 DoF, maximum von Mises stress is displayed in Figures 28, 30, 32, and 34 with three subplots for the cancellous, cortical and lysis stress, stress location colour maps are shown in Tables 25A, 26A, 27A, and 28A in the appendices.

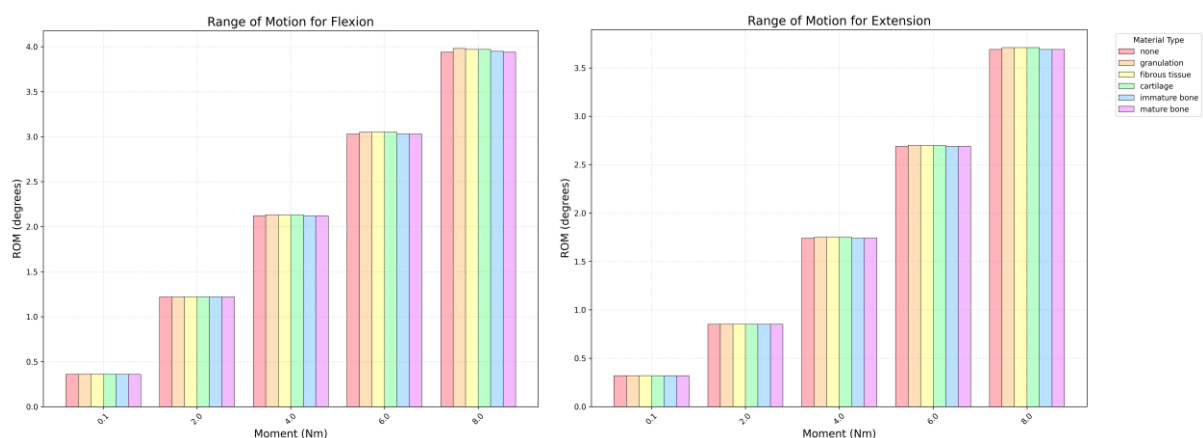
4.4 Results

4.4.1 Low intensity activity

4.4.1.1 Baseline 100N

To produce a low activity and reference activity analysis, a 100N axial compressive body load was analysed for the 6 degrees of freedom. The range of motion increased steadily with the rising bending moment across all degrees of freedom. The most significant increase in

motion and variation between material types occurred during axial rotation, with left and right rotations reaching 6.75° and 6.02° at an 8Nm bending moment for the granulation material model. For flexion, extension and lateral bending left and right the range of motion was consistent across the pars interarticularis material types for each of the bending moments reaching a maximum of 3.71°, 3.98°, 4.6° and 4.59° respectively. Analysis of the von Mises stress outputs show that across the three sections of L5 bone, there is a non-linear escalation in von Mises stress with increasing applied moments, which is especially pronounced in the cortical region. There is distinct variation between the material types with fracture healing progression, with the von Mises stress in the cortical bone decreasing from granulation to mature bone with an average maximum of 11.81MPa to 4.07MPa. Conversely, in the lysis section, the von Mises stress increases from 0 MPa to an average maximum of 2.45 MPa across different loading modes. Similar to the range of motion, the axial rotation in both left and right loading modes generates the largest increase in von Mises stress in the cortical bone, with the granulation material type producing maximum stresses of 16.12 MPa and 17.17 MPa, respectively. Maximum values of stress were consistently increased for the granulation material type for flexion, extension and lateral bending left and right with values of 9.56, 10.08, 10.26 and 7.14MPa, respectively. From the colour maps these peaks of stress increase are visible in the ventral pars interarticularis edge and along the pars interarticularis posterior wall for lateral bending and axial rotation. Axial rotation resulted in the largest area of increased stress which dissipated into the lamina and pedicle regions. Pure axial compressive loads of 100N did not produce visible changes across the vertebral region for all material types.



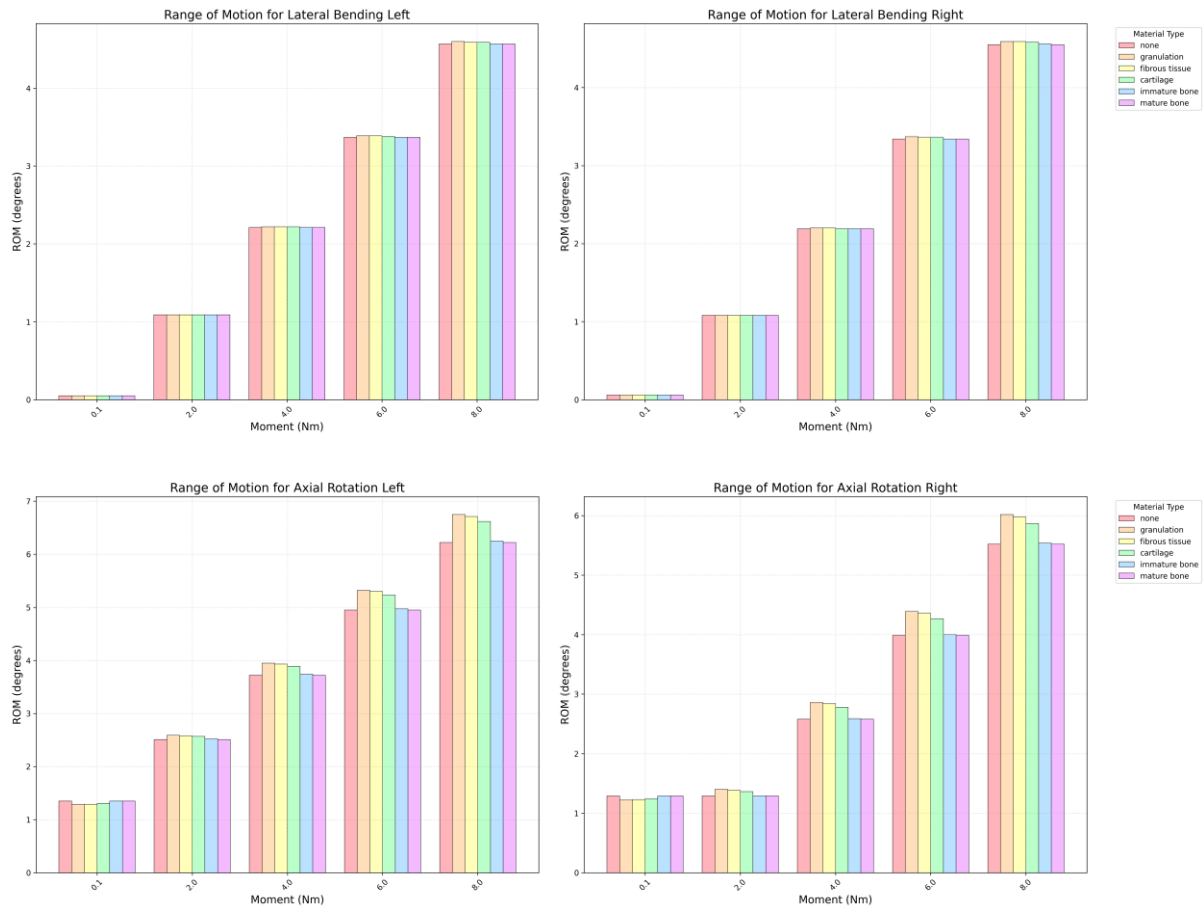
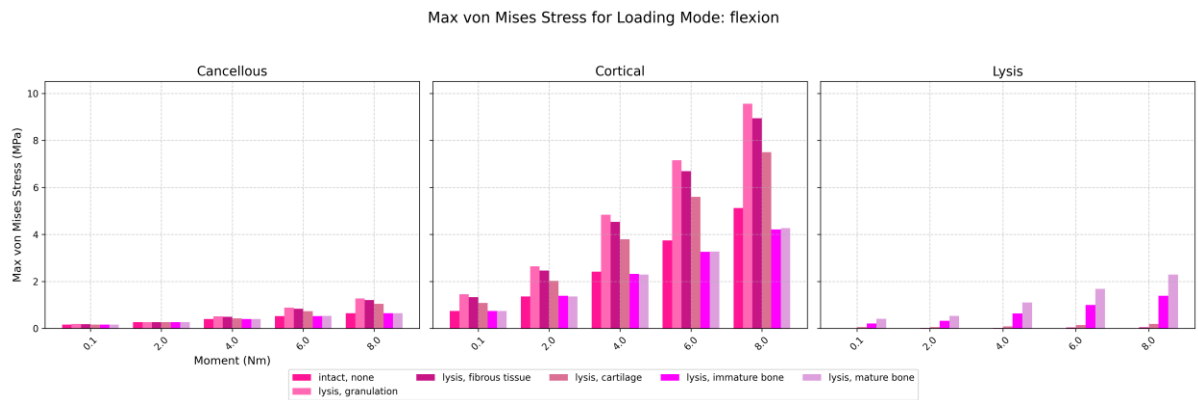
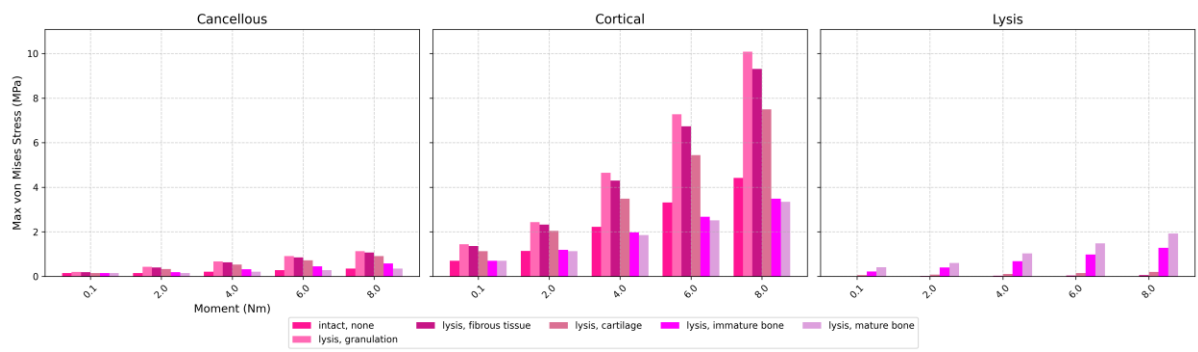


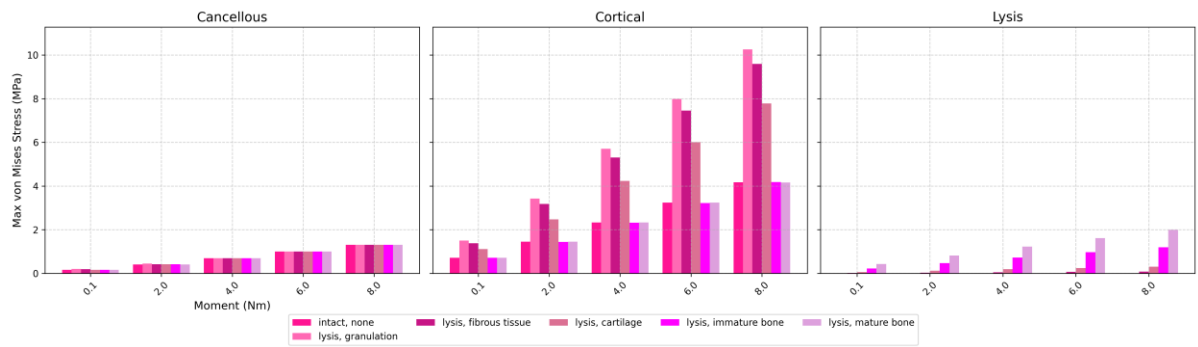
Figure 27: Range of motion analysis of the L4-S1 spinal segment for low intensity activity of a body load of 100N.



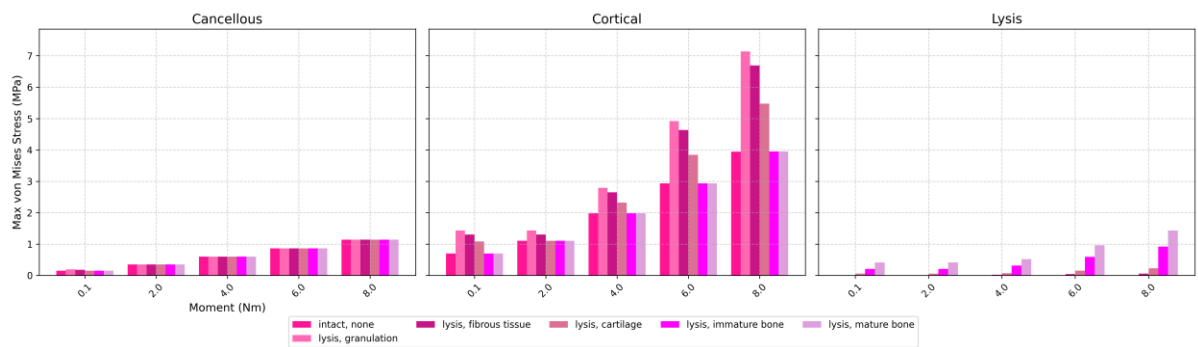
Max von Mises Stress for Loading Mode: extension



Max von Mises Stress for Loading Mode: lateral_bending_left



Max von Mises Stress for Loading Mode: lateral_bending_right



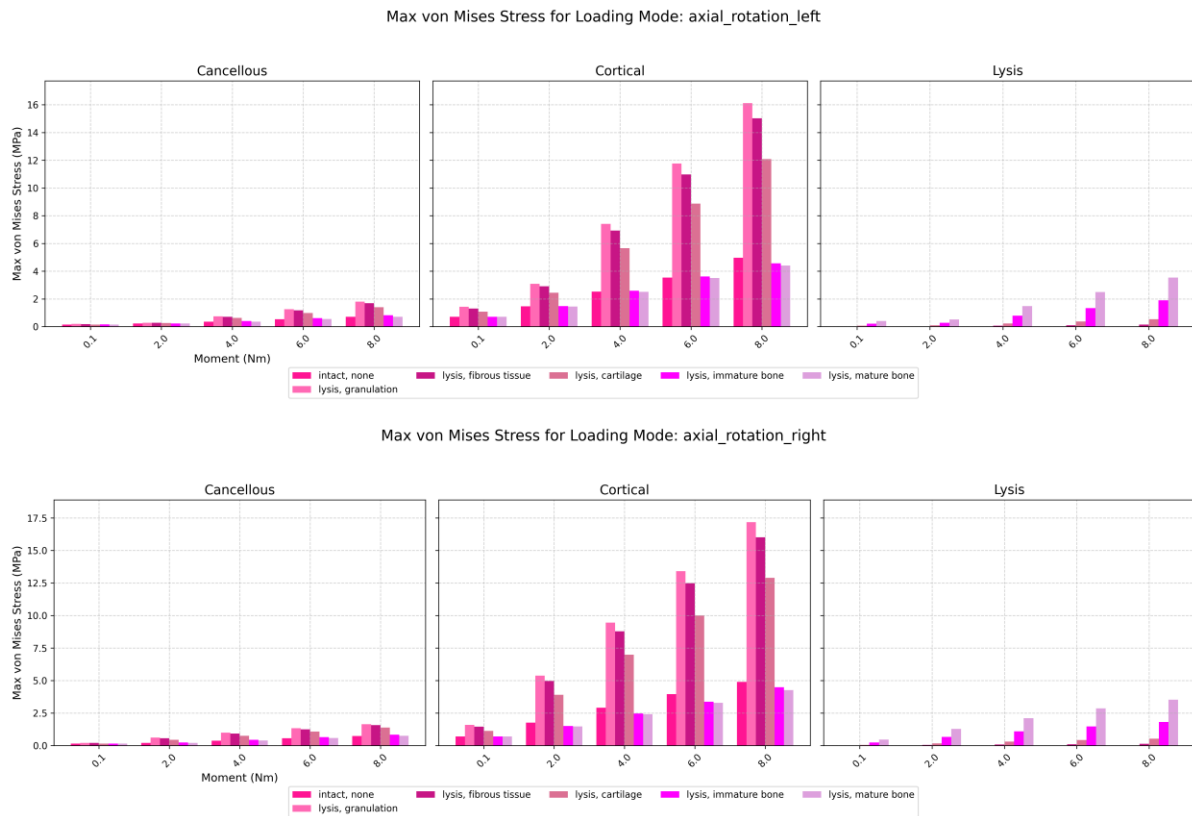


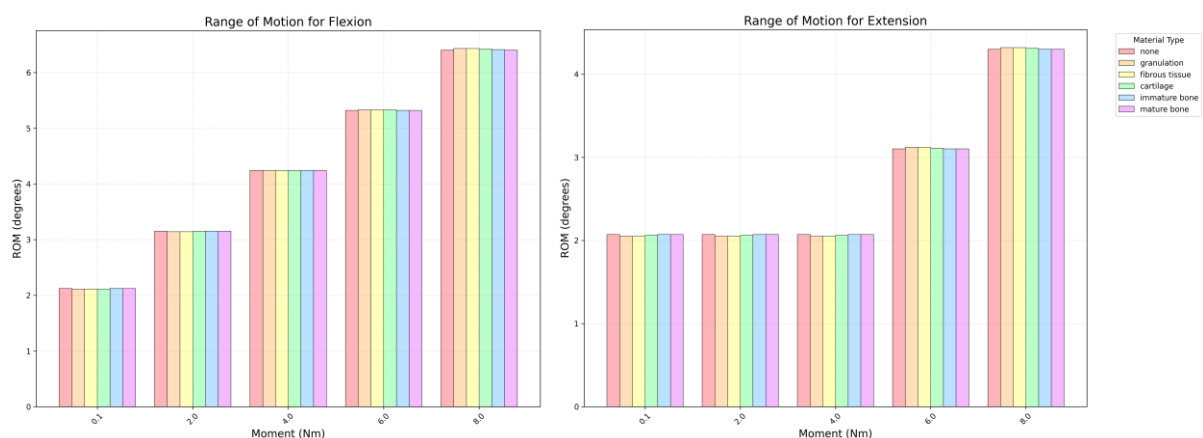
Figure 28: Maximum von Mises stress analysis of the cortical, cancellous and lysis L5 segments for low intensity activity of a body load of 100N.

4.4.2 Medium intensity activity: Activities of daily living

4.4.2.1 Standing 530N

This section represents a simple activity of daily living: of standing upright with an axial compressive load of 530N, combined with the 6 degrees of freedom. This body load produced varying trends in range of motion increase for the 6 DoF loading modes. Lateral bending to the left and right displayed a linear increase in motion to a maximum of 5.32° and 5.35° , respectively at 8Nm. While flexion and axial rotation left produced a more gradual rise to a maximum range of motion of 6.42° and 17.97° , respectively at 8Nm. In contrast, during extension and right axial rotation, certain regions exhibited a plateau in motion. The average range of motion for axial rotation was 11.18° across all applied moments, while extension increased from 2.06° between 0.1-4Nm to 4.31° at 8Nm. For all loading modes, there was little variation in the range of motion between the lysis material types with only small variation seen in axial rotation right with a decrease in motion in the early stages of healing. As observed in the low-intensity activity results, the von Mises stress outputs for standing

reveals a non-linear increase in stress across the three sections of the L5 vertebra as applied moments increase. This pattern is evident in both the cortical region during flexion, extension, left lateral bending, and right axial rotation loading modes. The granulation models at 8Nm loading produced the highest von Mises stress values, measuring 14.47, 10.83, 17.69, and 21.68 MPa, respectively. In comparison, right lateral bending and left axial rotation produced a reduced and constant variation in maximum von Mises stress values across all material types with average maximum values across all moments in granulation of 8.17, 8.66MPa. Stress in the cancellous regions followed a consistent pattern to the 100N load, showing minimal variation across both materials and moments within the specified range. In contrast, von Mises stress in the lysis regions remained consistent across moment loads but produced a linear increase as healing progressed, reaching an average maximum of 2.67 MPa in the mature bone phase. The analysis of the colour maps presented in Table 26A reveals an increased stress distribution during left lateral bending and right axial rotation. In axial rotation, the highest stress magnitudes are dispersed across the right posterior lamina and pars, while in lateral bending, the maximum stress values remain concentrated within the pars region. Flexion and extension at this stage currently present with maximum stress located at the ventral pars wall, with pure compression starting to show signs of increased stress in this area as well.



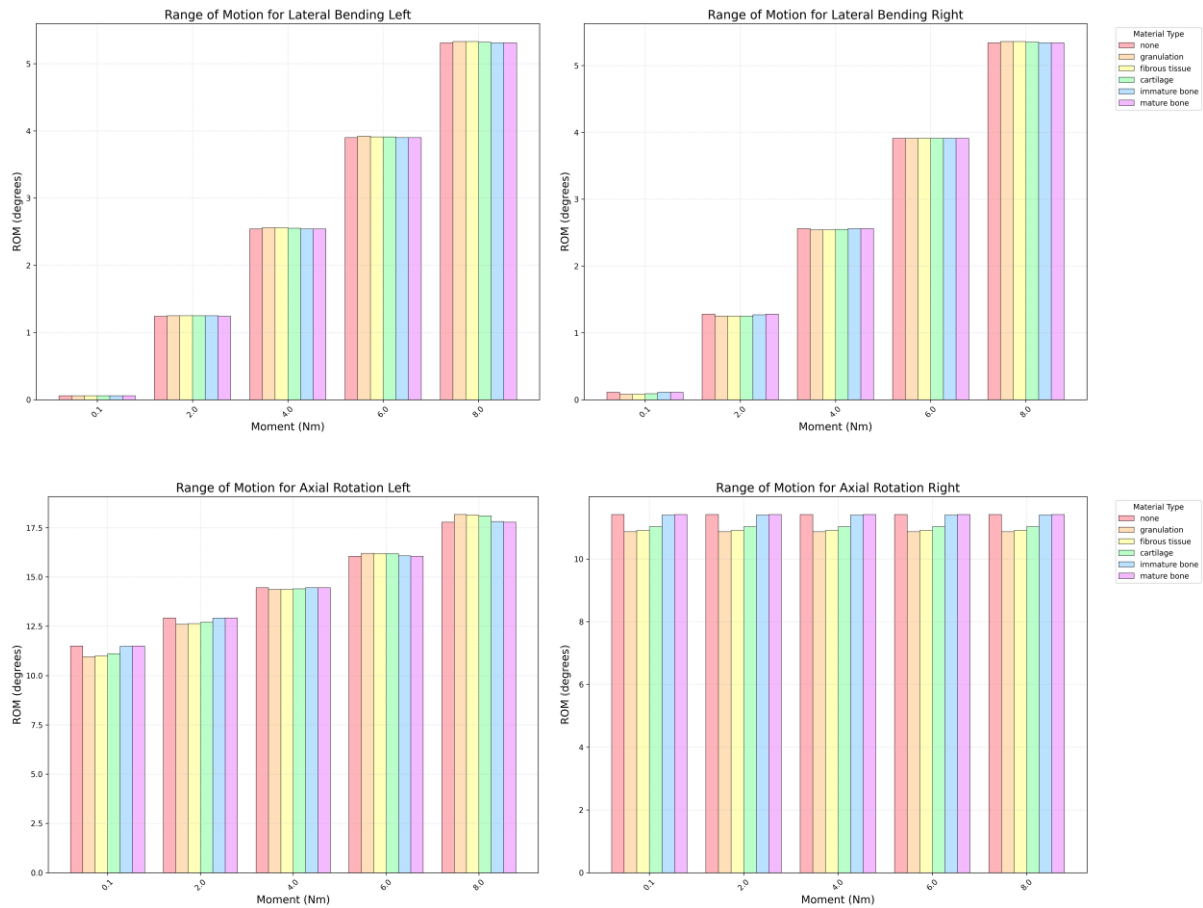
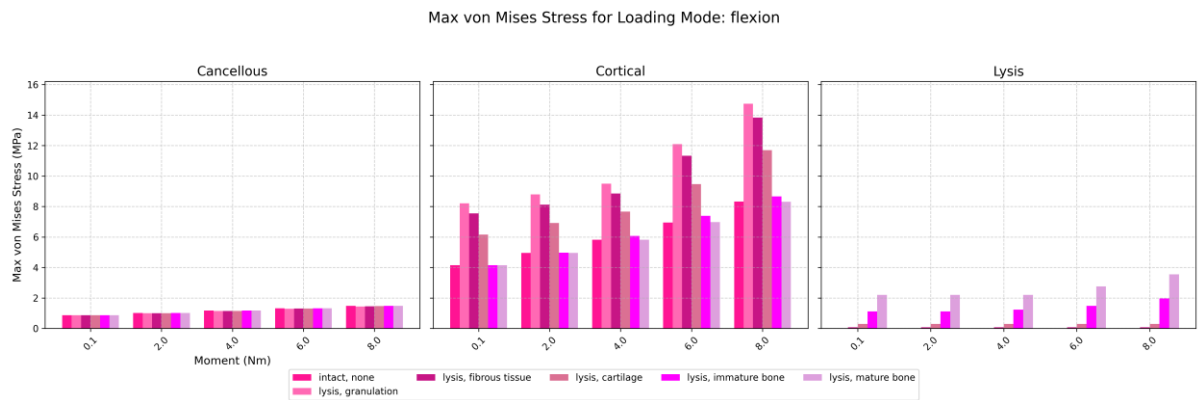
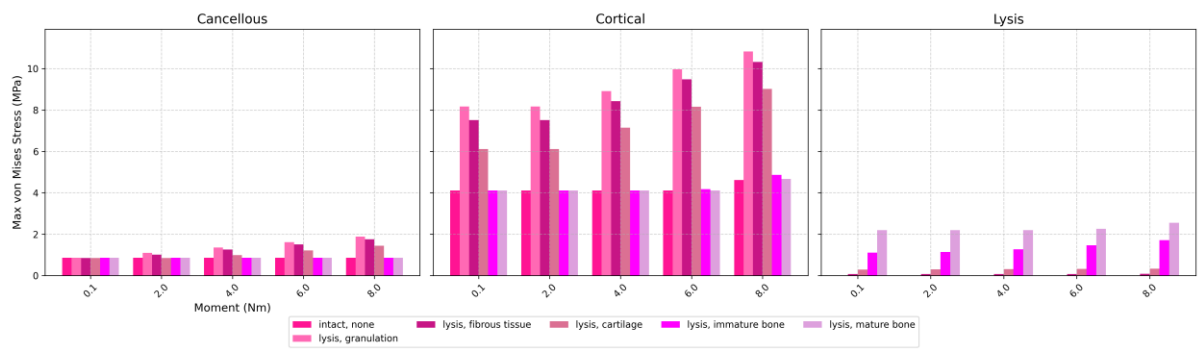


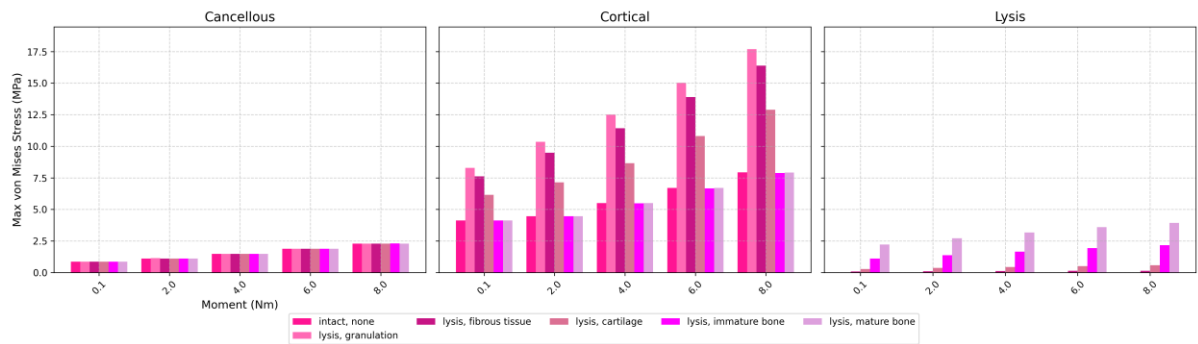
Figure 29: Range of motion analysis of the L4-S1 spinal segment for low intensity activity of a body load of 530N.



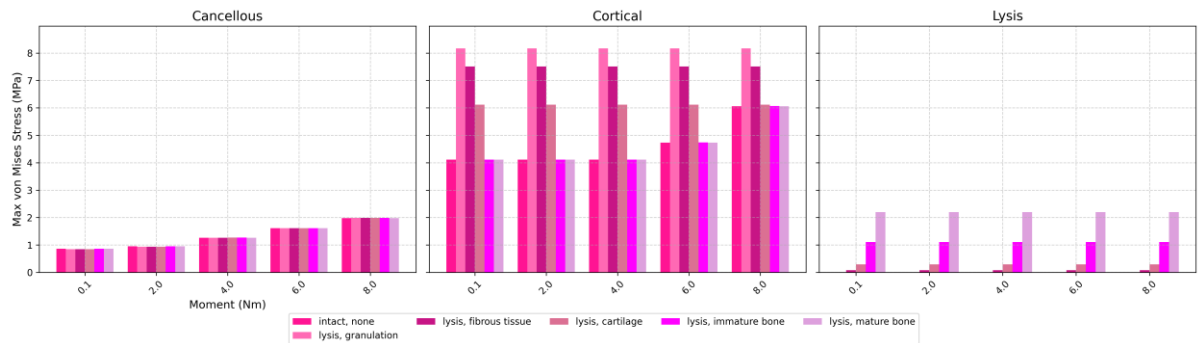
Max von Mises Stress for Loading Mode: extension



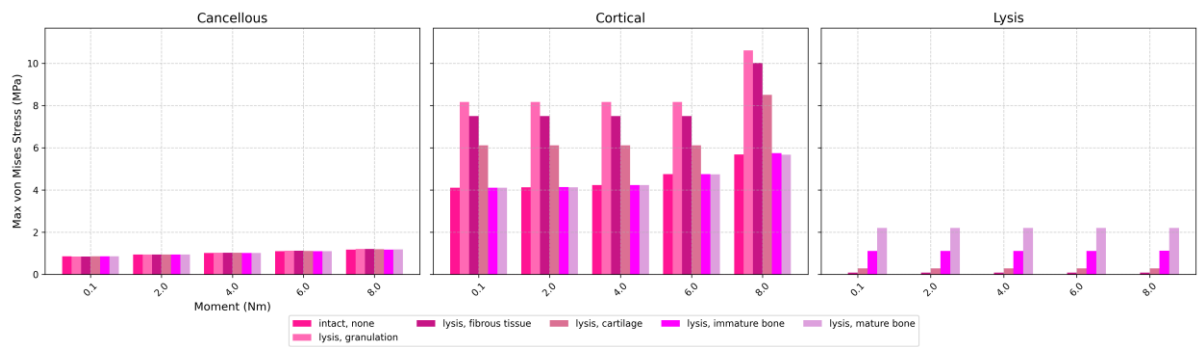
Max von Mises Stress for Loading Mode: lateral_bending_left



Max von Mises Stress for Loading Mode: lateral_bending_right



Max von Mises Stress for Loading Mode: axial_rotation_left



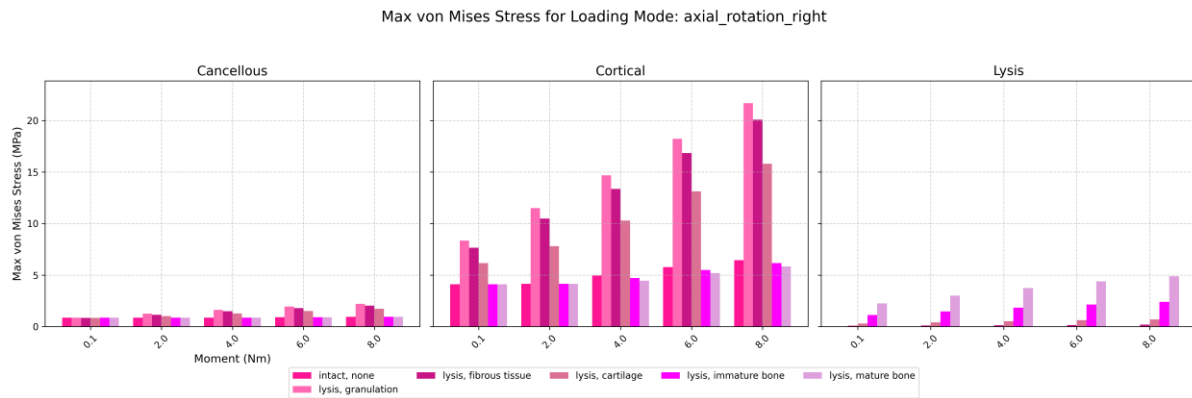
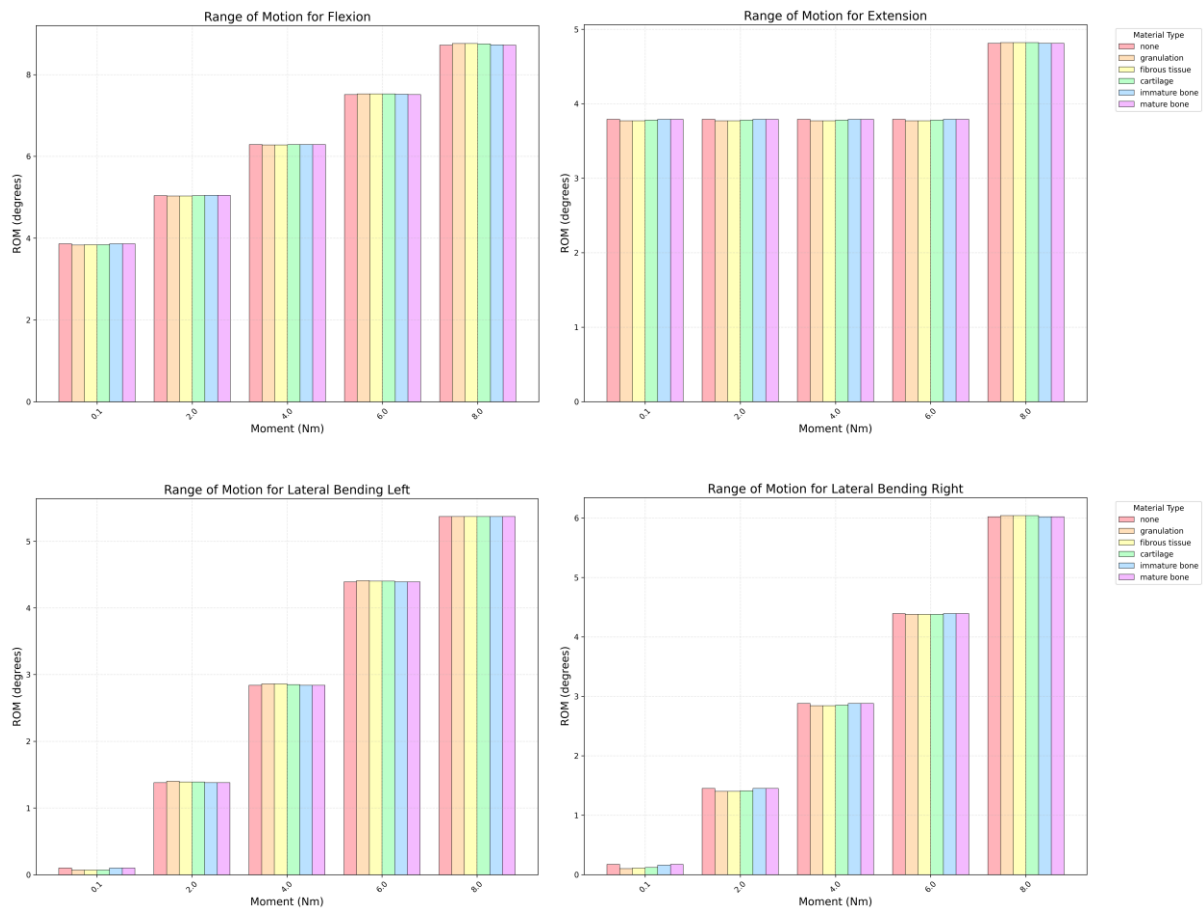


Figure 30: Maximum von Mises stress analysis of the cortical, cancellous and lysis L5 segments for low intensity activity of a body load of 530N.

4.4.2.2 Walking 824N

This section explores a simple activity of daily living: walking with an axial compressive load of 824N. This analysis incorporates the 6 degrees of freedom associated with human movement. This body load produced comparable trends of range of motion increase for the 6 DoF loading modes to the standing activity. Left and right lateral bending displayed a linear increase in motion to a maximum of 5.37° and 6.04°, respectively at 8Nm. While flexion and axial rotation left produced a more gradual rise to a maximum range of motion of 8.76° and 39.49°, respectively at 8Nm. In contrast, during extension and right axial rotation, certain regions exhibited a plateau in motion. The average range of motion for axial rotation was 31.8° across all applied moments, while extension increased from 3.78° between 0.1-6Nm to 4.82° at 8Nm. In agreement with the standing activity, all loading modes showed minimal RoM variation between lysis material types, with a slight decrease in right axial rotation during early healing stages. The von Mises stress outputs for this activity, reveals a nonlinear increase in stress across the three sections of the L5 vertebra as applied moments increase for flexion, lateral bending left and axial rotation right. The granulation models at 8Nm loading produced the highest von Mises stress values, measuring 20.4, 23.42, and 24.27 MPa, respectively. In comparison, right lateral bending and left axial rotation and extension produced a reduced and constant variation in maximum von Mises stress values across all material types with average maximum values across all moments in granulation of 13.32, 13.32, and 13.46 MPa. Stress in the cancellous regions remained consistent with minimal

variation across materials and moments. In contrast, von Mises stress in the lysis regions remained steady across moments but increased linearly with healing, reaching an average maximum of 4.34 MPa in the mature bone phase. Colour map analysis of Table 27A, revealed increased maximum von Mises stress in the granulation phase, with stress for RAR and LLB spreading across the right posterior arch and minimal dissipation into the left lamina. Compression, flexion, and extension stress remained focused on the pars, with flexion beginning to spread across the lamina.



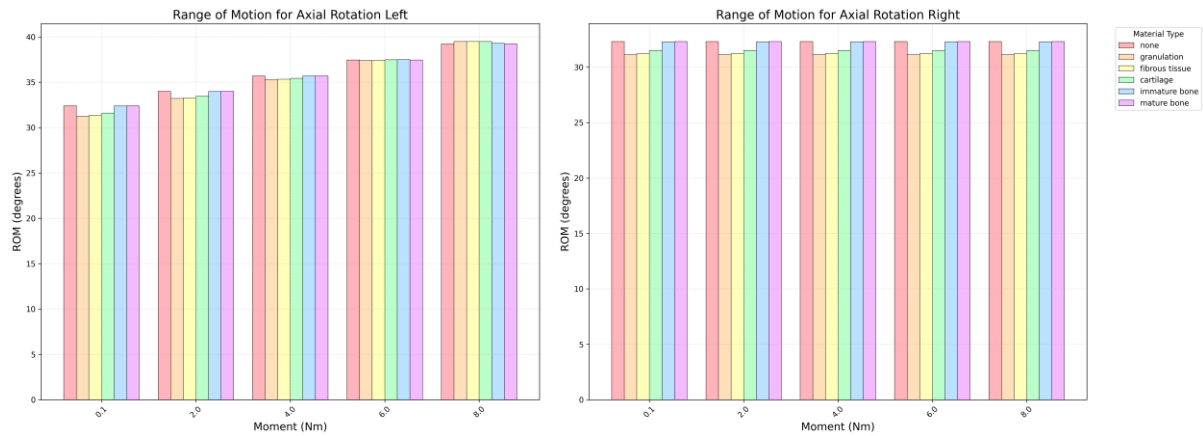
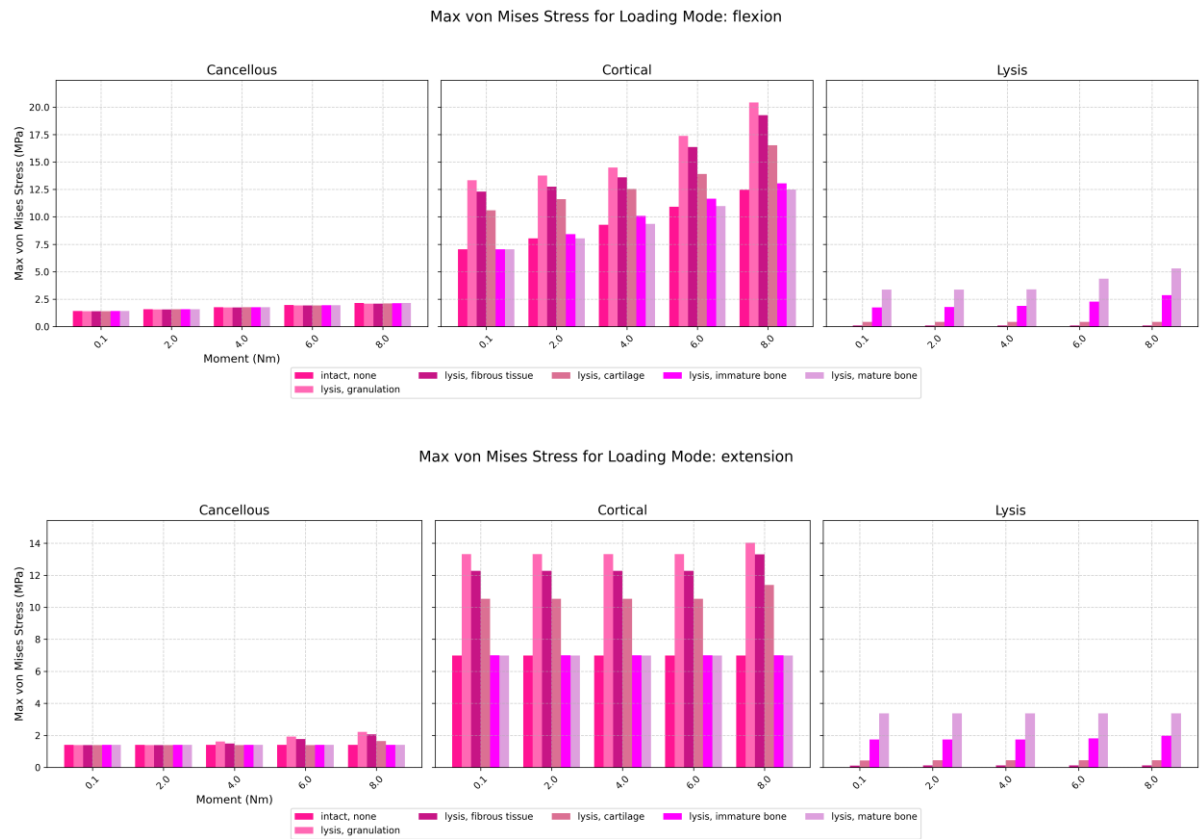
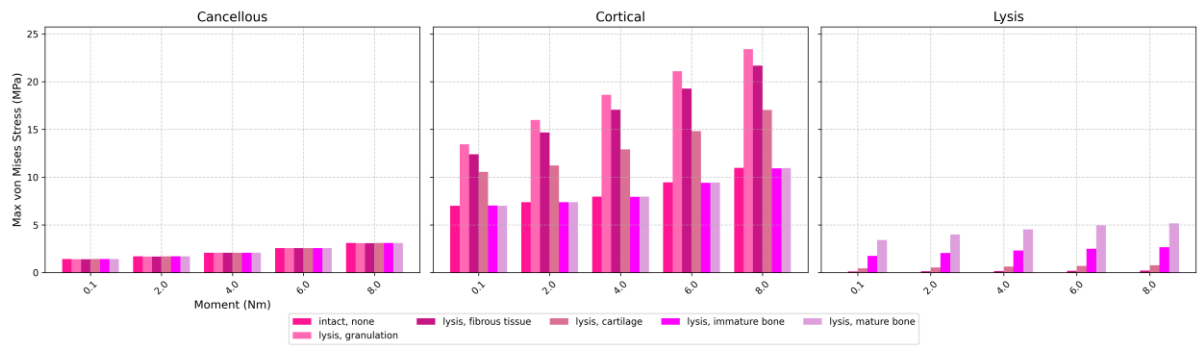


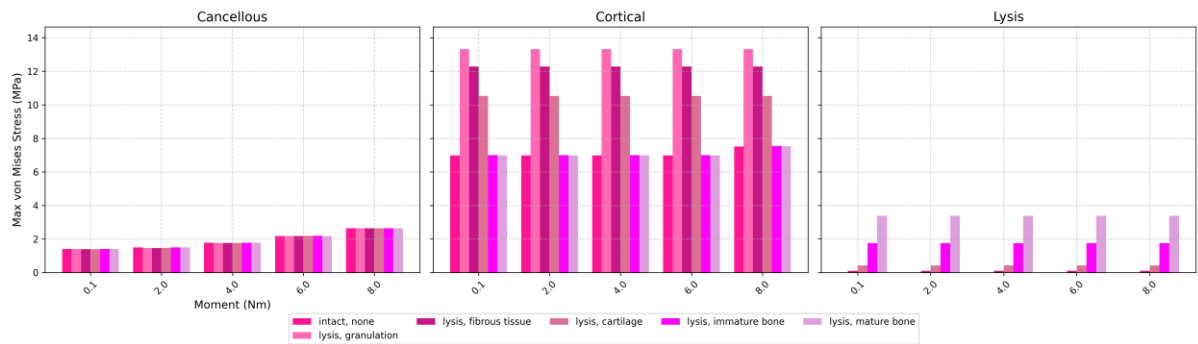
Figure 31: Range of motion analysis of the L4-S1 spinal segment for low intensity activity of a body load of 824N.



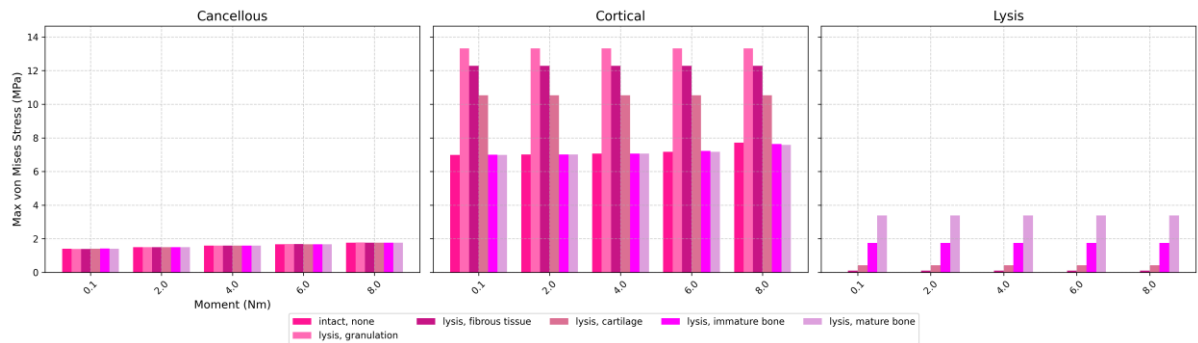
Max von Mises Stress for Loading Mode: lateral_bending_left



Max von Mises Stress for Loading Mode: lateral_bending_right



Max von Mises Stress for Loading Mode: axial_rotation_left



Max von Mises Stress for Loading Mode: axial_rotation_right

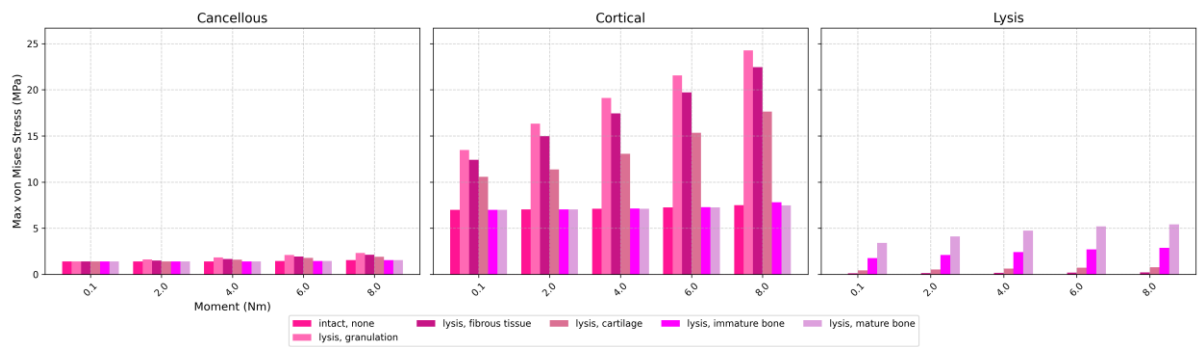


Figure 32: Maximum von Mises stress analysis of the cortical, cancellous and lysis L5 segments for low intensity activity of a body load of 824N.

4.4.3 High intensity activity: Sports activities

4.4.3.1 Moderate running 1060N

This section represents an activity of increased intensity of moderate running with an axial compressive load of 1060N combined with the 6 degrees of freedom. Left and right lateral bending and flexion displayed a non-linear gradual rise in motion to a maximum range of motion of 5.32°, 4.88°, and 11.08° respectively at the successful maximum moment. Right lateral bending produced a negative Jacobian error at 8Nm and therefore outputs of RoM and von Mises stress were not obtained. In contrast, extension and left and right axial rotation produced a consistent RoM of 5.68°, 76.84, and 77.56° across all bending moments with minimal fluctuations between material types for all successful simulations. Von Mises stress increased nonlinearly across L5 with rising moments in flexion, left lateral bending, and right axial rotation. At 8Nm, the highest stress values were 26.94, 28.88 and 23.8MPa, respectively, excluding a 42.62 MPa outlier in the lateral bending cartilage model. In comparison, right lateral bending, left axial rotation, and extension showed lower and consistent von Mises stress, with maximum average values of 17.27 MPa in the granulation phase. Stress in the cancellous regions remained consistent averaging at 2.23MPa with minimal variation across materials and moments. In contrast, von Mises stress in the lysis regions remained steady across moments but increased linearly with healing, reaching an average maximum of 5.99 MPa in the mature bone phase. Analysis of the colour maps in Table 28A, showed that loading modes LLB and RAR displayed high regions of von Mises stress located across the total posterior lamina area, except for the fractured area during the early healing phases. Compression, flexion, and extension also presented with this pattern to a lesser extent, primarily affecting the right pars interarticularis.

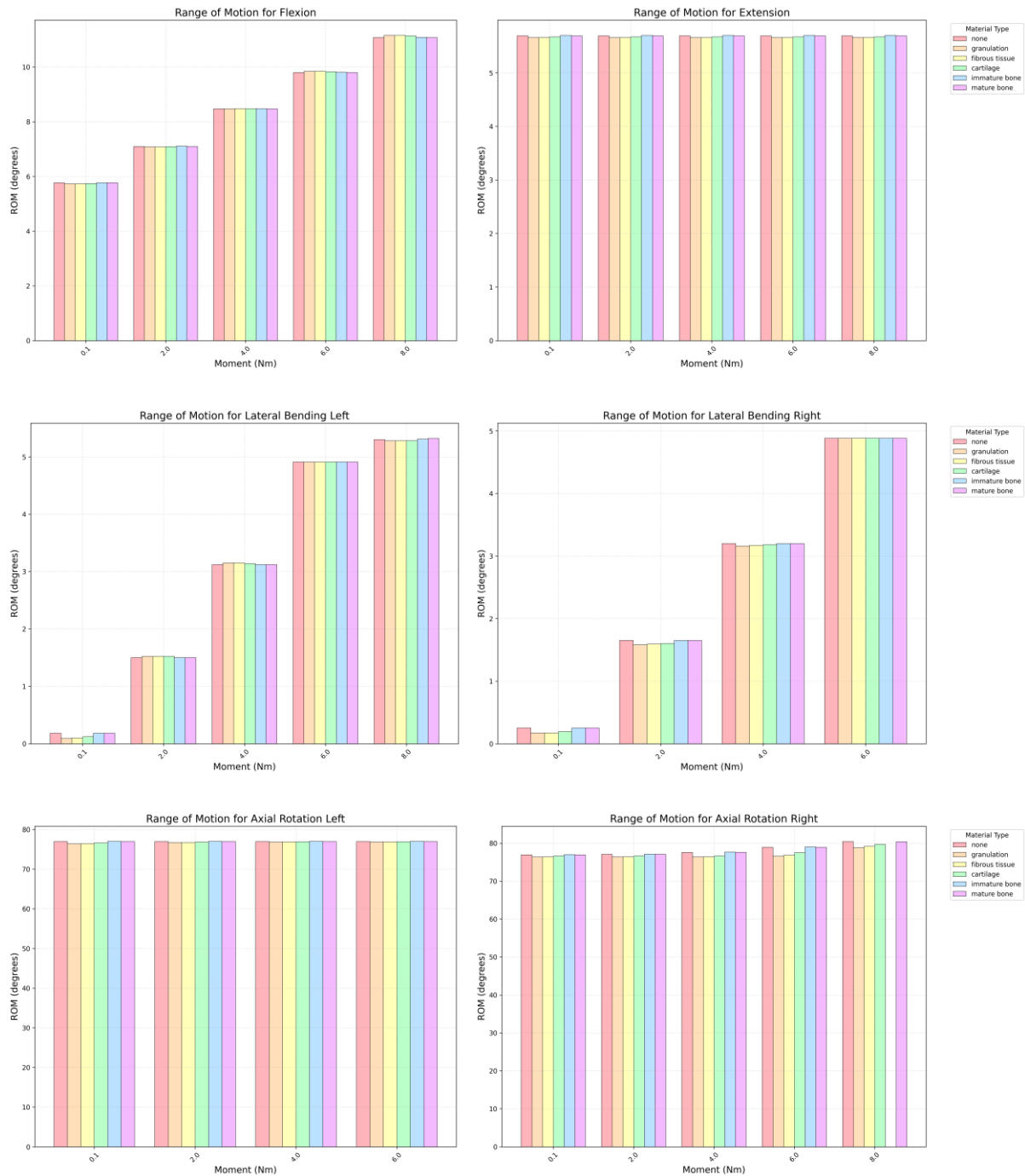
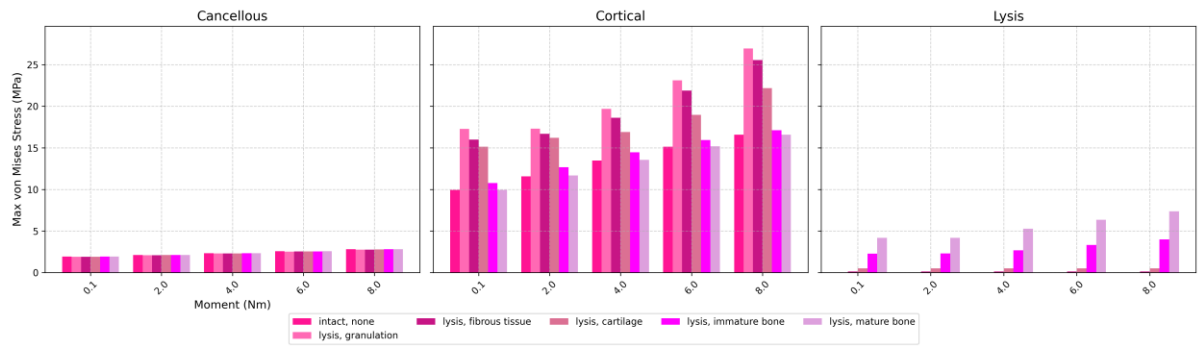
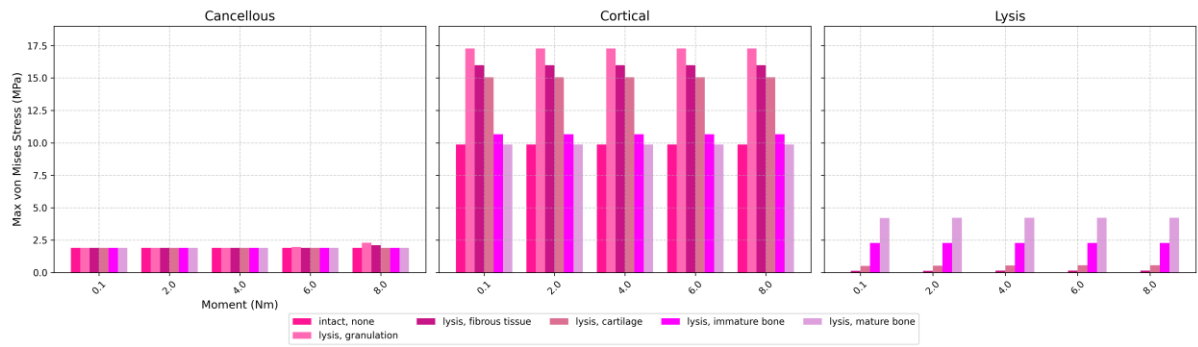


Figure 33: Range of motion analysis of the L4-S1 spinal segment for low intensity activity of a body load of 1060N.

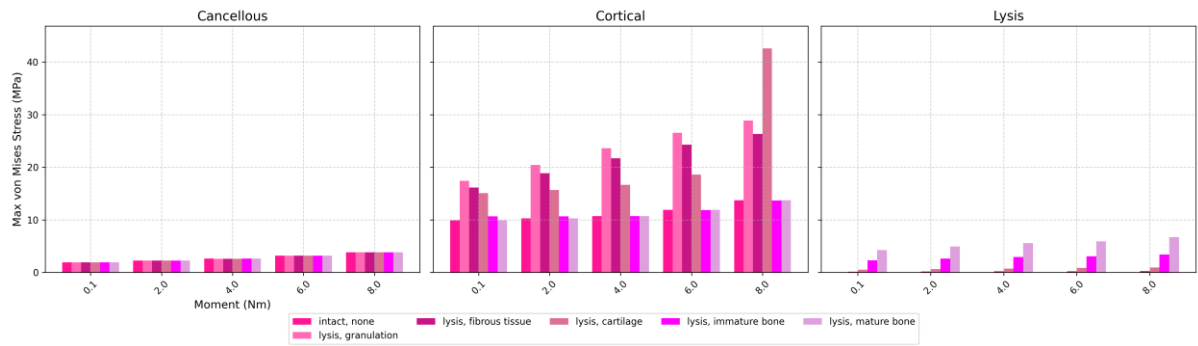
Max von Mises Stress for Loading Mode: flexion



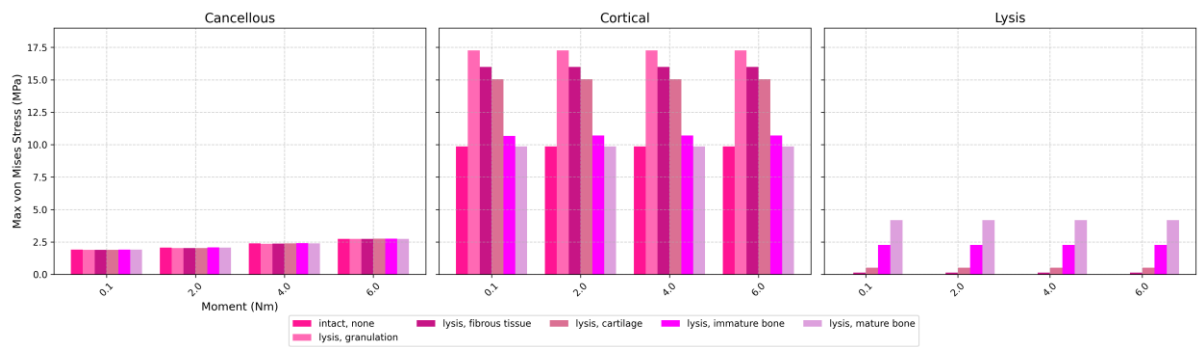
Max von Mises Stress for Loading Mode: extension



Max von Mises Stress for Loading Mode: lateral_bending_left



Max von Mises Stress for Loading Mode: lateral_bending_right



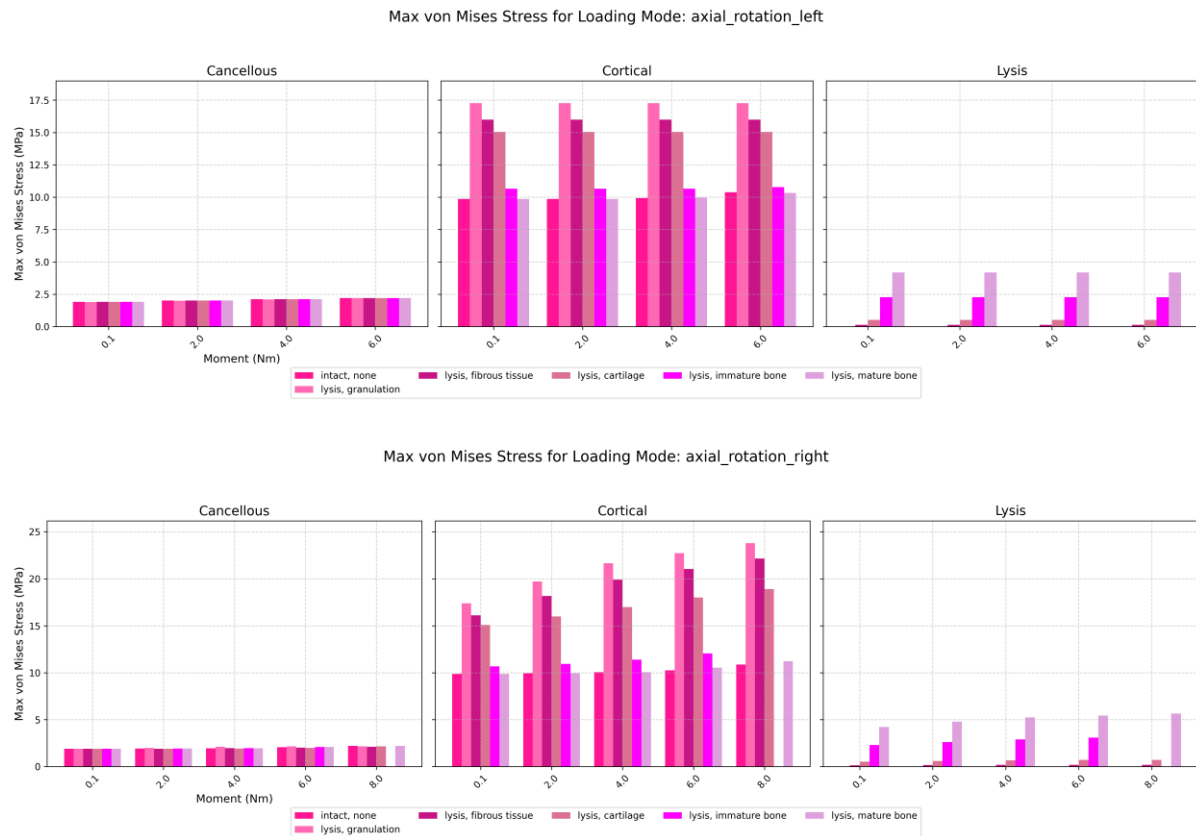


Figure 34: Maximum von Mises stress analysis of the cortical, cancellous and lysis L5 segments for low intensity activity of a body load of 1060N.

4.5 Discussion

The purpose of this study was to analyse the patient-specific spinal framework and provide an informed rehabilitation strategy that promotes optimal recovery while minimising the risk of further injury or complications during the healing process. Activity-specific analysis was conducted using the framework with the patient 1 spinal model to simulate activities of varying intensities. For each activity, the spine was examined across the different stages of fracture healing to provide a detailed understanding of how various loading conditions affect spinal biomechanics. To demonstrate the potential of the framework, simulations were carried out for progressively intense activities under a range of axial compressive loads, starting from a baseline of 100N. This body load is the lowest possible compressive load the lumbar spine can withstand without buckling of the ligaments (152), making it a suitable baseline for assessing spinal mechanics. Activities were then analysed with axial compressive loads

representative of standing (530N), walking (824N), and moderate running (1060N). These were combined with increasing moments across six degrees of freedom (0.1-8Nm).

For the patient 1 model, the loading moments that produced the greatest von Mises stress were flexion, ipsilateral bending and contralateral rotation in the presence of a unilateral fracture. These findings align with Haj-Ali's (2019)(110) analysis of unilateral fractures, which also identified these loading modes as causing the greatest stress on the contralateral pars. Flexion of the spine generates compressive forces on the anterior vertebral body and tensile forces on the posterior arch. For the patient 1 model, the compromised left pars caused a greater imbalance of load in the contralateral pars, which must compensate for the weakened structure. As axial compressive loads increase, this instability worsens, requiring more tensile force to rotate the posterior structure, resulting in higher stress at the ventral wall as the load shifts forward during flexion. With the increased intensity of the activities, the stress in this location increased non-linearly showing the potentially damaging effects of flexion under increased compressive load. Although both this and Haj-Ali's (2019)(110) study identified flexion as a motion that induces fracture risk, findings from the conservative treatment literature review suggest otherwise stating that flexion is a low-risk motion. Conservative treatment analysis (91,92) recommends during rehabilitation the use of flexion permitting hybrid LSOs to grant the patient more movement. These contradicting recommendations could be linked to the fracture stage that is analysed in each study. For this model, fractures of 4mm width were analysed to encompass the total pars interarticularis area to accommodate for any swelling of the fracture due to capsulation. Therefore, compared to the early-stage fracture recommendations that permit flexion, the progressive fractures this model focuses on, with increased risk of motion-induced stress, highlights flexion as a hazardous motion that should be restricted. Bending moments of extension did not produce the elevated stress levels that were found in similar computational models discussed in the literature review (108,111). This may be due to the stiff behaviour observed during extension, preventing hyperextension from occurring, and therefore not replicating the pincer presentation between the articular processes.

Ipsilateral (left) lateral bending produced the highest overall von Mises stress during moderate running activity due to the tensile forces acting on the contralateral pars during this motion, which are exacerbated by the addition of axial compression during high-intensity

activities. Axial rotation generated the highest von Mises stress during low to moderate activity, driven by the torsional forces as the fractured pars rotates toward the intact pars. All three motions suggest potential further degeneration of both the intact and compromised pars, with reduced shock absorption in these areas due to the fracture. This is demonstrated by the high-stress levels observed during the granulation, fibrous tissue, and cartilage phases of fracture healing. Additionally, the resultant increased stress indicates a growing area at risk of potential bilateral spondylolysis progression, with posterior pars stress increasing from standing to moderate running.

For each specific activity, the patterns of von Mises stress and range of motion exhibit interesting trends between the bending moments. In the baseline 100N activities, the von Mises stress and RoM outputs provide a comparable linear increase across all 6 DoF. This is consistent for the flexion and extension loading modes across all activities for all bending moments, where the outputs closely correspond to one another. For instance, for extension, von Mises stress begins to level off in moderate and high-intensity activities, the RoM similarly reflects this reduction in the rate of increase as the moment intensifies.

However, this is not the case for the lateral bending and axial rotation results for the compressive loads of 530-1060N. During lateral bending, the RoM consistently increases from 0.1 to 8Nm across all activity loads. For ipsilateral bending the von Mises stress mirrors this pattern, however, for contralateral bending the von Mises stress stabilised across all bending moments. These trends suggest that while lateral bending motion can continue to increase, contralateral bending does not generate elevated stress in the current range of motion. Conversely, for axial rotation as the intensity of activity increases, the RoM for axial rotation reaches a steady state with increasing bending moment. Ipsilateral rotation mirrors this pattern by becoming progressively more constant across the bending moments, but contralateral stress continues to increase even with RoM stagnation. The decoupling of stress and RoM indicate that for axial rotation, the stagnation in RoM coupled with increasing stress suggests that the spine has reached its maximum rotational capacity. This means that despite additional bending moments, the vertebrae cannot rotate further, but the applied force continues to induce stress on the structures. These results demonstrate the close relationship between RoM and von Mises stress. With the analysis of this specific model's biomechanical patterns, we can better understand the limits of both RoM and stress, which is crucial for

optimising conservative treatment. This knowledge helps to establish safe movement thresholds, reducing the risk of further injury while guiding rehabilitation strategies tailored to the spine's mechanical capacity.

To fully analyse the progression of spinal fractures and optimise rehabilitation strategies, it is crucial to analyse the von Mises stress across different healing phases, particularly after the previously discussed decoupling between the trends of RoM and stress. In all tests (excluding outliers), cortical and lysis stress exhibited an inverse relationship, where when cortical stress decreased, lysis stress increased, as spondylolysis healing progressed. In the initial granulation phase, cortical stress significantly increases in comparison to the intact model, with the magnitude and distribution of the stress depending on the activity intensity. In the walking and moderate running simulations, the maximum von Mises stress was concentrated along the isthmus, with increased stress observed on the ventral edge during flexion and compression, and on the dorsal edge during lateral bending. However, as the healing progressed to cartilage and mature bone stages, the stress distribution balanced between the left and right sides, demonstrating how the posterior arch was at reduced risk of secondary fracture. For the higher-intensity activities, there was also increased von Mises stress neighbouring the lysis region in the posterior lamina indicating the potential for compromised healing. From the cartilage to mature bone regions the pars start to experience higher levels of stress, where this does demonstrate the improved balance of loading in the spine, there is also a chance for regression and injury recurrence especially from the identified problematic loads of flexion, ipsilateral bending and contralateral rotation. Range of motion when considering all loading modes was consistent across the healing phases. This result is expected for unilateral fractures because the stability of the posterior arch is maintained by the intact contralateral side, preventing the excessive segmental motion or slipping commonly seen in bilateral fractures (110).

Stress distribution and magnitude analysis is an ideal measurement that provides valuable insights into the biomechanical risks associated with certain activities. However, von Mises stress is not easily measurable in clinical settings nor widely understood by most patients, particularly as this condition is prevalent in adolescents. Lumbar range of motion, on the other hand, is a common metric assessed regularly to measure treatment progress during rehabilitation using spinal measurement techniques such as goniometry or tape measurements

(153). As the range of motion correlates with stress, whilst being easier to evaluate, it can serve as a practical guideline and proxy metric by setting motion limits for each stage of healing.

There were limitations with this study, particularly visible in the results of the range of motion. Across all loading modes the axial rotation RoM was higher than previously validated spinal finite element models. For example, Wiczenbach's (2023)(143) analysis of a FE L1-L5 spine under a 720N compressive load and 5.5Nm of bending demonstrated a torsional RoM of 7°. In comparison, the L1-S1 model in this study, under a similar load of 824N compression and 6Nm bending, exhibited a torsional RoM of 37.44°. It was also observed that, at higher intensity loads, the range of motion for extension and axial rotation consistently stabilised, indicating spinal stiffening with increasing loads. However, this spinal stiffness is premature at this level of body load, as greater spinal motion would be anticipated under even higher loads, as shown in Chastain's (2023)(44) experimental analysis of lumbar spines. The stabilisation of the range of motion could be due to the intervertebral disc material choice of a neo-Hookean model as this material model is not ideal for capturing accurate elastic material behaviour at large strains(154). However, the overestimation in the magnitude of the range of motion could be due to the previously discussed lack of ligament and disc fibre stiffening behaviour due to the simplicity of the model. The absence of modelled lumbar musculature likely contributes to the two issues of increased axial rotation of the spine and the occurrence of intersection errors resulting in negative Jacobians from facet joint collisions. To address these problems, incorporating the lumbar multifidus muscle would be beneficial to increase the stability of the facet joints from the connections of the muscle between the spinous process (155) and facet capsular ligaments.

While this framework does not yet perfectly capture absolute range of motion, due to simplifications chosen within this model, the model still identified motions which generated the largest increase in von Mises stress, with physiologically relevant magnitudes and locations. Biomechanical stress analysis, across the fractured L5 vertebrae, highlighted motions of flexion, ipsilateral bending and contralateral rotation as high-risk activities that impose the greatest stress at the contralateral pars, particularly for early stages of healing like granulation and fibrous healing. Even though the absolute values of range of motion may be overestimated, the relative trends in stress distribution, across the five healing phases, provide

a valuable guide. The analysis allows the outputs of the model to infer safe motion thresholds by correlating increased stress with the degree and direction of movement, thereby identifying high-risk planes and magnitudes of motion. This analysis supports the clinical relevance of the framework by tailoring rehabilitation strategies that balance necessary mobility, informed by peaks of stress, that minimise the risk of further injury or compromising spondylolysis healing. Based on these results a treatment plan for patient 1 can be proposed as follows.

The findings suggest that in the initial two-week healing period, of granulation and fibrous tissue phases, activities should be limited to low intensity, with minimal essential moderate intensity activities of daily living. Ideally, this should be a period of relative rest to prevent the increased stress observed in most loading conditions during the granulation phase. Since complete rest is unrealistic, activities involving up to 530N of compression with minimal bending can be recommended, allowing for limited movements such as left axial rotation and right lateral bending to facilitate everyday tasks. As cartilage healing progresses, ideally at the 1–4-month interval, interventions of conservative treatment such as physical therapy and orthoses could be used. Orthoses could aid with spinal stabilisation to prevent the increased degrees of lateral bending flexion and axial rotation which, in the cartilage stage, still provide a risk of posterior arch stress, particularly in activities which increase axial compression. Muscle-building exercises, which are essential for conservative treatment aimed at improving spinal stability, should be guided by favouring contralateral bending and ipsilateral rotation. When incorporating moderate-intensity activities, bending moments should be kept below 4Nm, as stress levels begin to increase significantly beyond this threshold. Towards the end of treatment at the 3–6-month period, when bony healing is progressing, a gradual increase in high-intensity activities may be reintroduced to prepare the patient for a return to full activity. This is especially relevant for athletes, as their movements often involve complex combinations of bending and compression. For instance, weightlifting exercises, such as lifting a 10kg barbell, generate significant compressive forces of up to 2632 N (144) along with repetitive motions of flexion and extension, making it essential to build resilience in these motions with tailored rehabilitation.

Chapter 5: Conclusion and Model Outlook

5.1 Lumbar spinal model analysis

Computational modelling is a valuable tool for analysing spondylolysis of the lumbar spine due to the rapid simulation of a wide range of model material properties, loading and boundary conditions tailored to a patient's specific spinal and fracture presentation. This non-invasive biomechanical analysis allows for precise fracture modelling and to provide predictions of fracture progression before deterioration occurs, thereby enabling timely intervention to prevent further complications.

One of the key objectives in developing this computational patient-specific spinal framework was to ensure its usability for future researchers or clinicians who may have limited computational experience. This model is designed not only for biomechanical accuracy, but also for ease of use, enabling efficient generation of lumbar spinal segment models from patient imaging. Spinal segment model construction ensured the integration of patient specific data through the importing process of vertebral geometries from STL formats after image segmentation. Ligament, intervertebral disc and facet joint components are quickly defined and created with the anatomy of the vertebral landmarks and simple variables that control the size, shape and quantity of these structures. Ligament construction through selecting bounding box locations over relevant vertebral landmarks is a quick method of forming ligaments, however, the spacing between each ligament and exact node connection is variable depending on the node spacing of the model. Future iterations of this framework may consider moving from this spring technique to three-dimensional continuous elements. This feature would remove the variability due to mesh variations that was seen in the sensitivity analysis.

Another feature was to ensure customisable material properties, loading conditions, and analysis parameters, allowing users to tailor simulations to simulate patient-specific properties and replicate real-world activities. The variables in this model were configured in a chain simulation format where a sequence of parameters could be defined so that multiple simulations could be run and stored within an accessible data structure. This approach allowed for efficient management of simulations and could quickly iterate through conditions without needing to make manual adjustments. This reduces the time and effort required to

perform multiple simulations, making the workflow significantly more efficient and user-friendly.

An important implementation was the addition of simple visualisation tools, which enhance usability by providing clear and immediate interpretation of results, making it more accessible for users across various disciplines. Each of the 6 degrees of freedom spinal simulation visualisations are displayed in Table 22. After each simulation, the model can be easily imported into software like FEBio Studio for visual analysis of the von Mises stress and displacement outputs measured in this study. Output data was also formatted in several analysis formats such as text files with all variables and outputs or CSV files for statistical analysis.

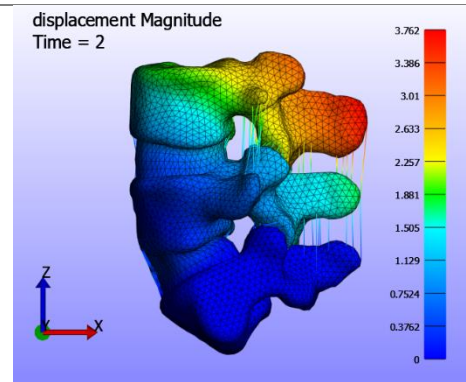
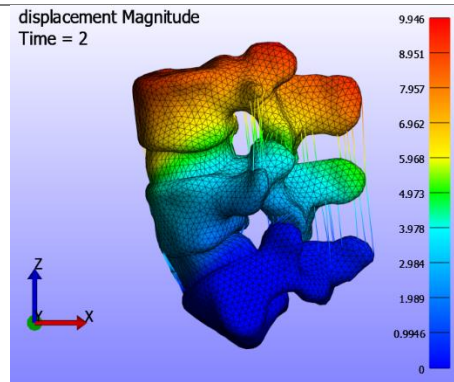
This framework did have limitations due to specific finite element parameters selected. One of these limitations was selecting linear tetrahedral elements for the model mesh, over higher order or hexahedral elements. These alternative mesh configurations would aid with capturing complex geometries and deformation patterns with improved convergence rates and reduced sensitivity to deformation. However, to reduce computational time or intervention required to produce a suitable mesh, linear tetrahedral elements were selected.

Additional limitations arose from assumptions about the source of the model geometries. It was assumed that the initial CT imaging consisted of no spinal pathologies and that the patients were scanned in a natural posture, which would not introduce any pre-bending or tension on the discs or ligaments. Additional assumptions included the geometric and interpatient uniformity of certain structures that were not directly extracted from imaging, which may have influenced the results. For example, it was assumed that the cortical bone maintained a constant thickness throughout the vertebrae and that the intervertebral disc, and its components, were uniformly shaped in all directions. Additionally, the cartilage layer of the facet joints was also considered uniform. While these assumptions aided with the efficiency of the framework, this could have introduced limitations to the degree of patient-specificity of the models.

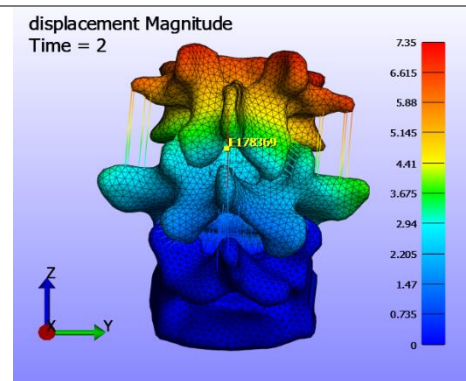
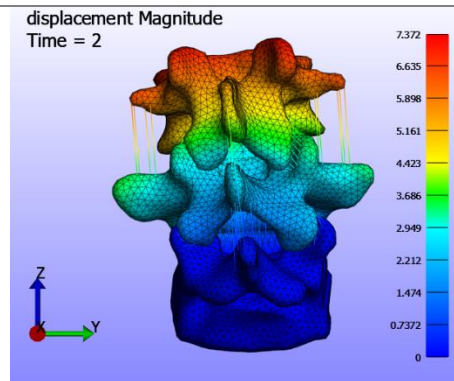
Table 22: Model summary visualisation of 6 degrees of bending of the spine

Model Load

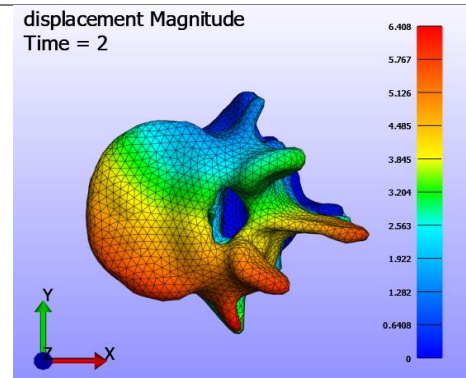
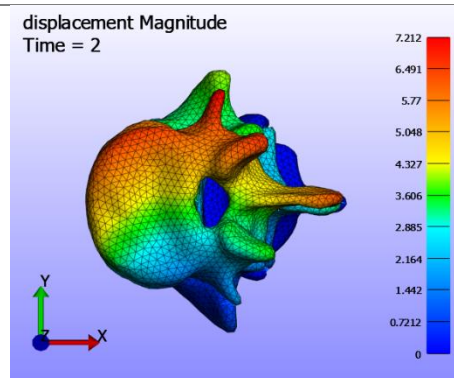
**Axial
compression +
flexion/extension**



**Axial
compression +
Lateral bending
left/right**



**Axial
compression +
Axial rotation
left/right**



5.2 Conclusion

The primary aim of this study was to develop a patient-specific lumbar spinal finite element framework to provide informed conservative treatment from a biomechanical perspective for spondylolysis. To develop this framework two core sections were covered. Firstly, model development, where the lumbar spinal framework was carefully constructed with a focus on creating a user-friendly and flexible system, adaptable to patient data and spondylolysis

cases. To analyse this construction for suitability in clinical applications, validation steps were performed through mesh convergence, model validation, and sensitivity analysis to ensure the framework's accuracy, reliability, and clinical applicability. These steps were performed to analyse two patient models, derived from imaging data.

Global convergence analysis of the total spinal segment showed that this model was robust across a mesh refinement of node distance of 3.05-2.75mm where outputs of total range of motion and L5 segmental displacement met the 5% error tolerance. The local von Mises stress magnitude and location were more sensitive to mesh size, as finer meshes better captured the contours of the vertebral surface. Output sensitivity could also be attributed to soft tissue reconstruction for each specific mesh which is an inherent flaw with the construction process in this model, chosen to enhance model flexibility.

Sensitivity analysis and model validation with analysis against experimental data, revealed that the maximum von Mises stress distribution occurs at the outer dorsal and ventral walls of the contralateral pars interarticularis. The loading modes associated with increased fracture risk were identified as flexion, ipsilateral bending, and contralateral rotation for all patient models, fracture locations and fracture widths analysed. However, statistical analysis of the segmental range of motion, of both patient models' data, revealed only a slight to moderate positive Pearson's correlation in comparison to existing computational models. The resultant range of motion was increased for axial rotation and decreased for lateral bending and flexion-extension, which is assumed to arise from the simplicity of the soft tissue models, specifically the collagen fibre behaviour of the disc and ligaments and the absence of muscle modelling. The results indicated that the model exhibited an overall stiffer and linear response to motion, missing the transition from minimal linear resistance to the rapid increase in non-linear resistance, even with the analysis of multiple soft tissue material models and parameters. However, despite the limitations highlighted from the chosen construction methods, von Mises stress location, magnitude, and range of motion trends were still physiologically relevant in comparison to established finite element models.

The second section of the study took the developed model to simulate progressive stages of fracture healing, from granulation to mature bone, to analyse biomechanical metrics of range of motion and von Mises stress, across loading conditions that simulate progressive intensities of activities. Patient 1 was analysed for activities of standing, walking and running,

which were produced with combinations of increasing axial compressive loads and 6 bending DoF. This analysis found that across the healing phases, cortical von Mises stress produced elevated results for granulation, increasing non-linearly with axial loads up to 1060N. The highest stress was observed during flexion and ipsilateral bending, reaching 26.94MPa and 28.88MPa. As healing progresses, the maximum von Mises stress in the lysis region increases, and the stress distribution shifts from being concentrated in the pars and lamina to a more dispersed pattern once mature healing is achieved.

Range of motion outputs, while not directly mirroring the patterns of von Mises stress, showed a strong correlation with stress distribution as long as the maximum planar RoM was not exceeded (as seen in axial rotation). Therefore, this study suggests using RoM as a proxy metric for vertebral stress and fracture risk. Von Mises stress has been highlighted by this study as an ideal metric for assessing spondylolysis progression and potential treatment complications. However, as stress cannot be measured directly in clinical settings, RoM is utilised as a measurable metric to indirectly assess vertebral stress and associated risks due to their strong correlation. By identifying stress-inducing activities based on RoM limits, treatment strategies can focus on avoiding motions that elevate stress, optimising conservative treatment.

This study provides insights on how to develop conservative treatment strategies from a patient-specific biomechanical perspective. Existing treatment methods often rely on patient-symptom-reporting to guide the initiation and progression of treatment, which can result in ineffective rehabilitation plans that may not ensure adequate fracture healing. Rehabilitation programs often lack biomechanical evidence, leading to a reliance on qualitative assessments rather than data-driven approaches to optimise rehabilitation. Therefore, with the foundational work provided by this study, assessing the potential for FE models in patient-specific biomechanical analysis, future research should aim to further develop this framework, to aid with advancements in the treatment of spondylolysis.

5.3 Future Work Recommendations

5.3.1 Fracture healing modelling

In this study, fractures were modelled by selecting a group of elements in the pars interarticularis area and assigning specific material models and properties from literature to simulate the fracture healing development. However, this model is very simple in construction and therefore could be developed to produce a more physiologically representative fracture. Potential models could include mechanobioregulatory or bioregulatory healing fracture simulations by incorporating the cellular response to loading. This could be analysed for the loading modes highlighted in this study to further progress the understanding of the effects of rehabilitation activities on spondylolysis fractures to detect potential failure. Models could also analyse the formation and anatomical structure of the soft callus which develops in the inflammation/fibrous healing phases, as this process alters the anatomy of the pars affecting the biomechanical stress distribution response to load, which was not captured in this study. Additionally, capturing the specific cortical and cancellous structure for bone remodelling could provide additional information on bony healing times and mechanical effects. Dynamic analysis that simulates the fracture healing over time could provide an informed time scale for rehabilitation progression and optimise the timing for reintroducing varying levels of physical activity.

5.3.2 Activity development

Another potential future direction could be to improve the modelling of the activities. This study took axial compressive values from literature to analyse a progression of simple activities of standing to running, with additional progressive bending to represent a range of motion for each activity. However, the amount of combined bending and compression was not strictly quantified in the literature and therefore a small range was selected. To improve activity representation range of motions of the body and the respective corresponding loading should be measured and quantified for a range of activities from activities of daily living to specific sporting requirements. This should include experimental measurement of kinematic data, such as joint angles, muscle forces, and ground reaction forces during various activities, to capture the loading effects of each spinal component on the resultant force through the spine. This would produce a more detailed approach than the axial compression and bending

moment technique used in this study. The experimental research should be conducted for a large demographic, considering factors such as age, gender, and fitness level, to tailor rehabilitation and training programs to different population characteristics. Each activity analysed in this model was also analysed for a singular motion for static analysis. However, spondylolysis fractures tend to be induced due to repetitive loading leading to fatigue-based fractures. Therefore, a study that analyses cyclic or repetitive loading patterns, to capture the cumulative effects of mechanical stress on the fracture site could provide a long-term representation of conservative treatment activities at the fracture site. This approach could allow for a model that predicts damage accumulation at the fracture site potentially compromising the bony union of the fracture.

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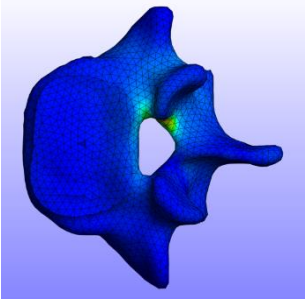
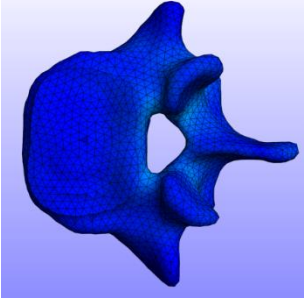
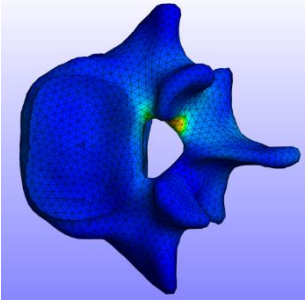
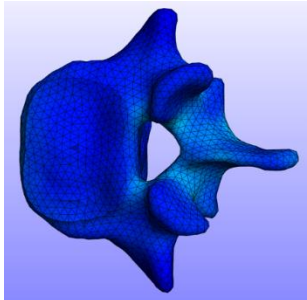
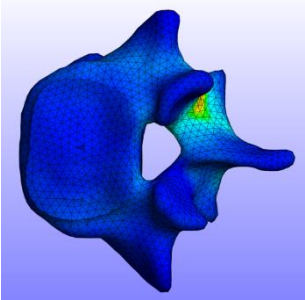
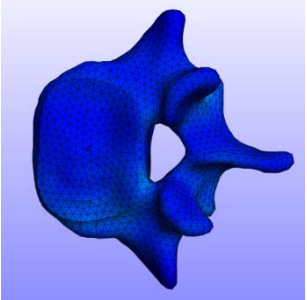
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Figure 35A: Flow chart of the model considerations and potential parameters detailed before construction.



Table 23A: Fracture model validation of von Mises stress locations across the vertebra with maximum values of 15MPa for granulation models and 12MPa for intact models

Patient 1 Lysis model of the L5 vertebra		
	Granulation von Mises stress (MPa)	Intact von Mises stress (MPa)
Extension		
Flexion		
Lateral Bending Left		

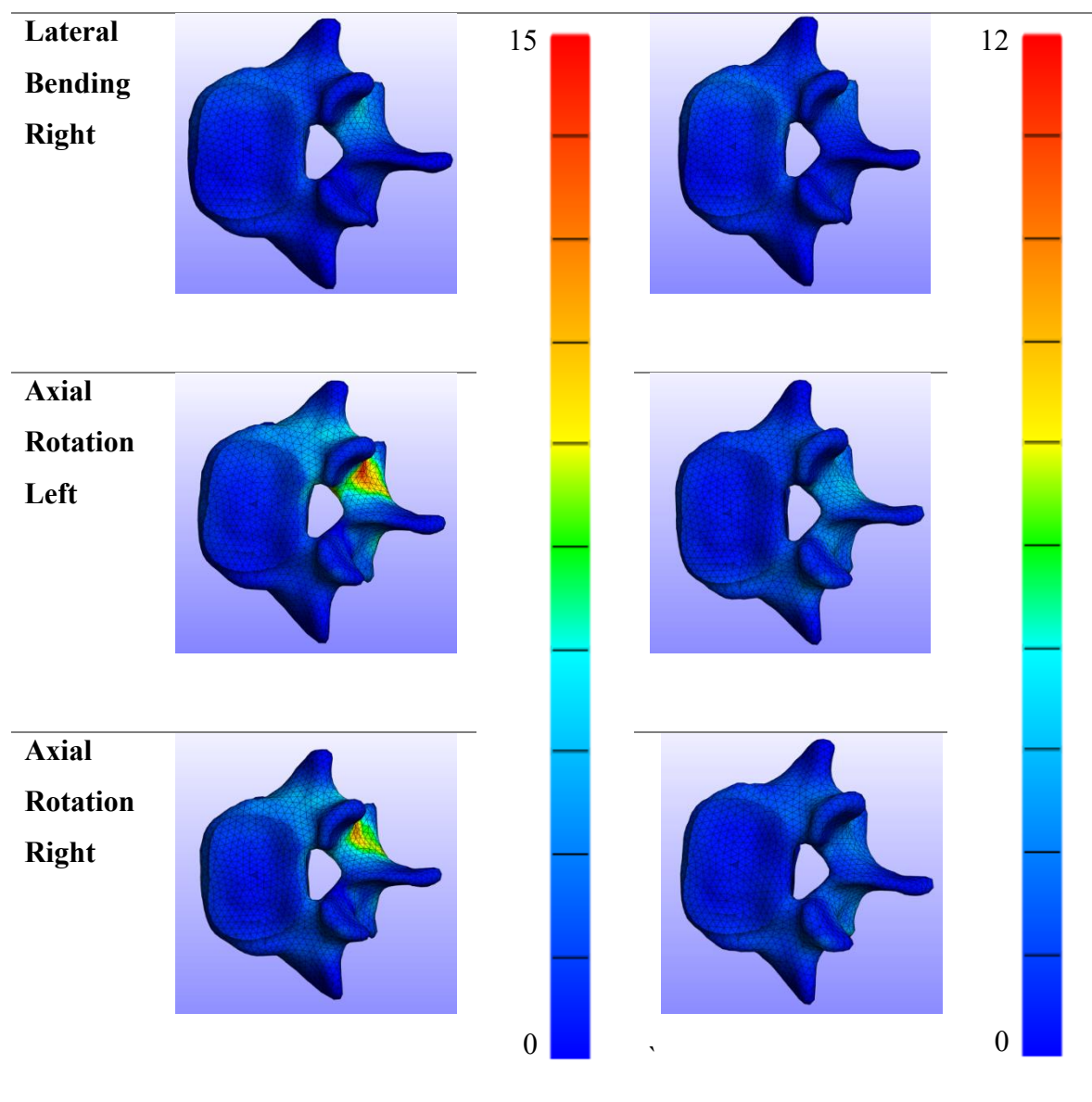
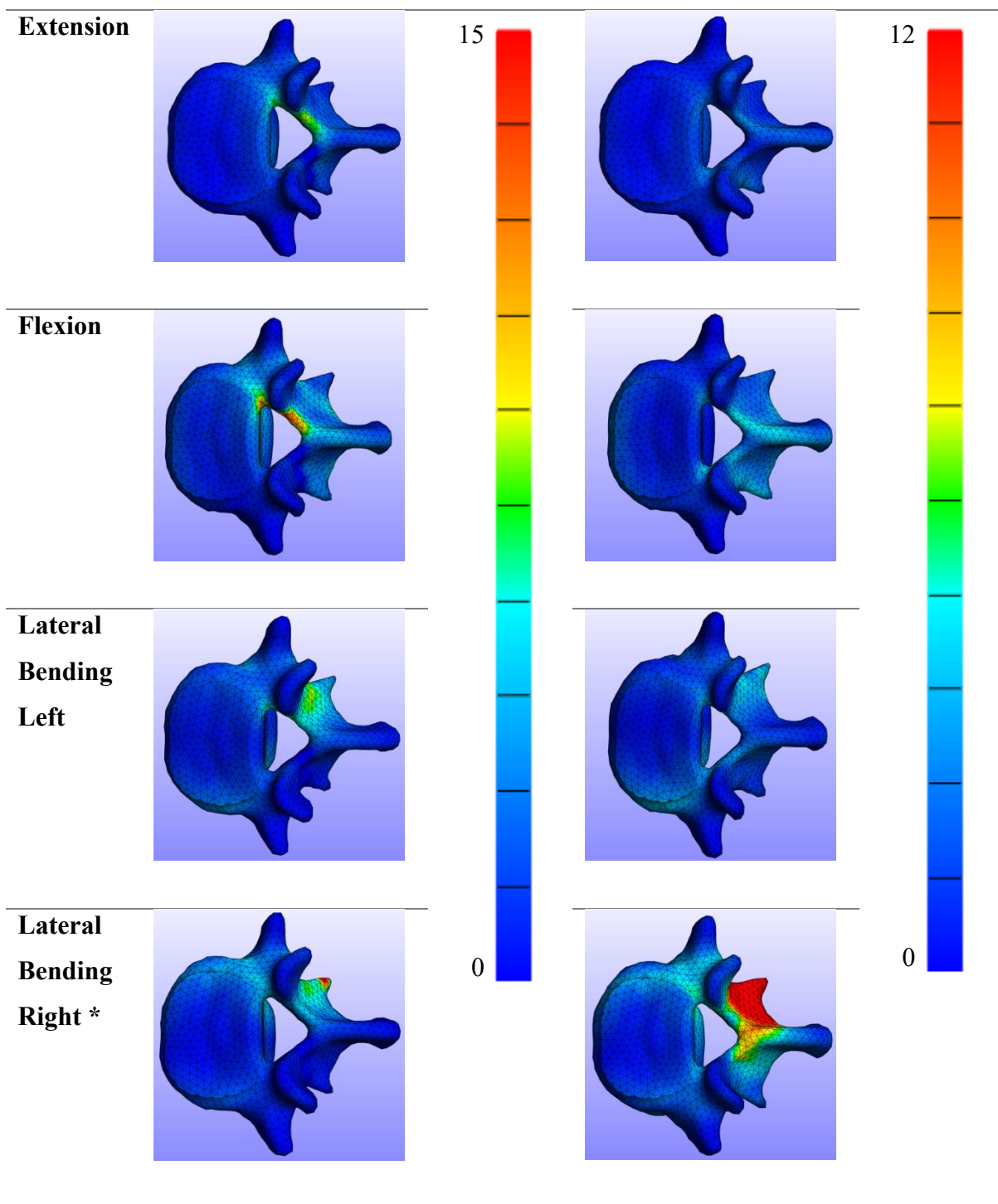


Table 24A: Fracture model validation of von Mises stress locations across the vertebra with maximum values of 15MPa for granulation models and 12MPa for intact models

Patient 2 Lysis model of the L5 vertebra	
Granulation von Mises stress (MPa)	Intact von Mises stress (MPa)



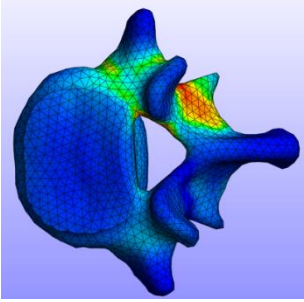
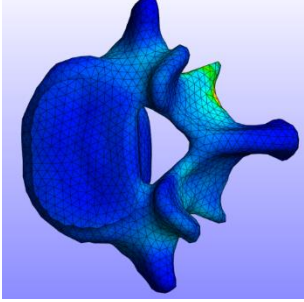
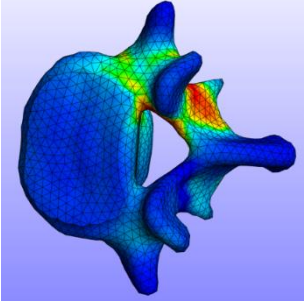
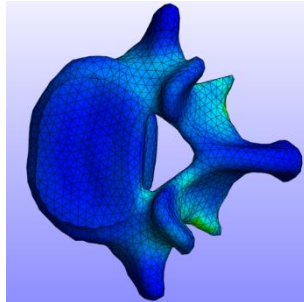
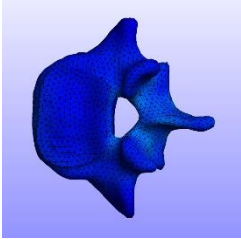
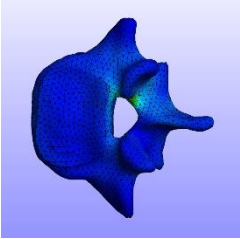
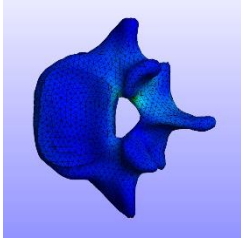
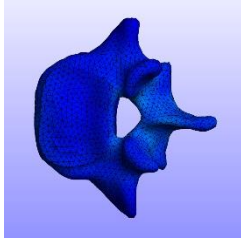
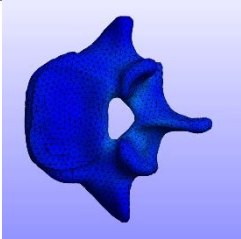
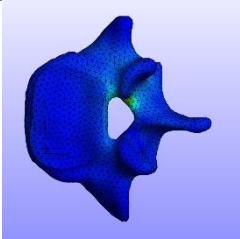
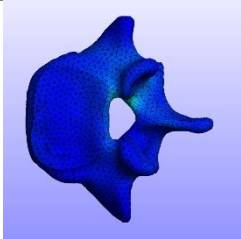
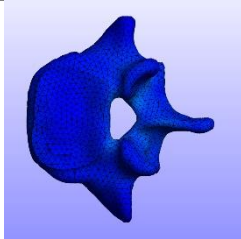
Axial Rotation Left		
		

Table 25A: Impact of five bending loading modes combined with 100N, and pure compression alone, on spondylolysis healing phases

100N	Intact	Granulation	Cartilage	Mature Bone
Flex				
Ext				

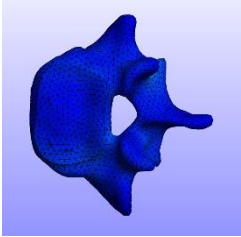
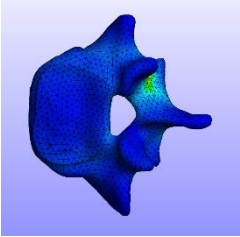
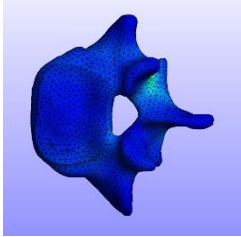
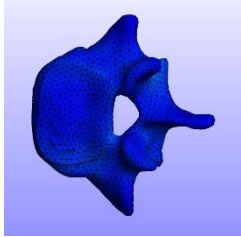
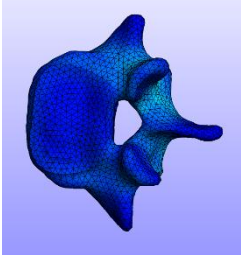
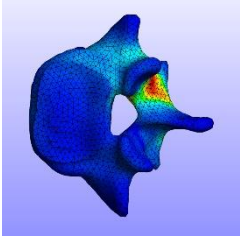
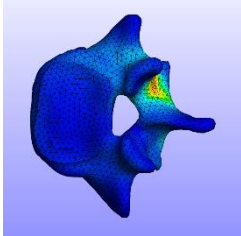
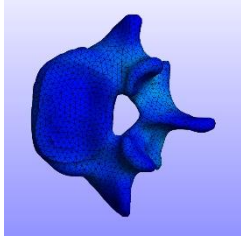
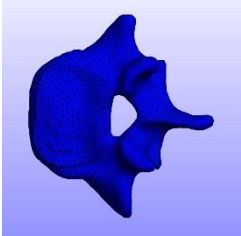
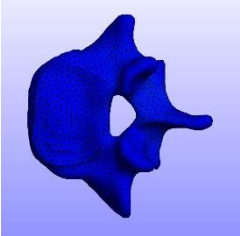
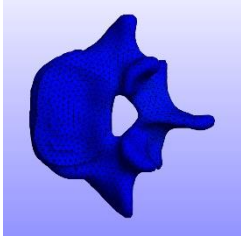
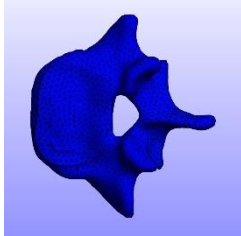
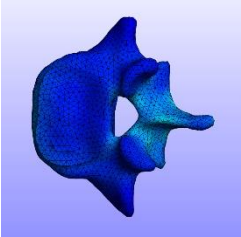
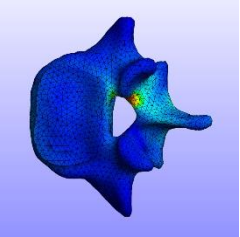
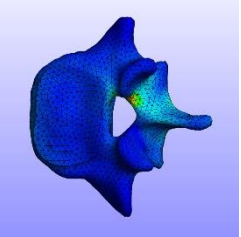
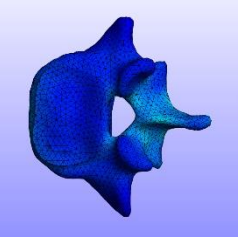
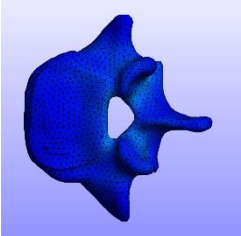
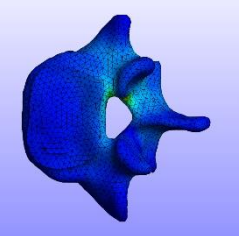
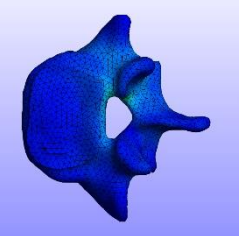
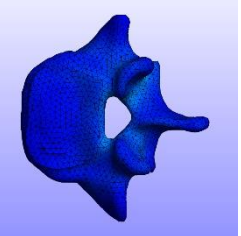
LLB				
RAR				
Comp ressio n				

Table 26A: Impact of five bending loading modes combined with 530N, and pure compression alone, on spondylolysis healing phases

530N	Intact	Granulation	Cartilage	Mature Bone
Flex				
Ext				

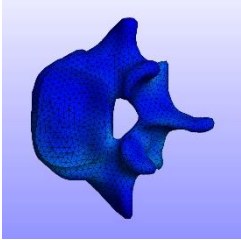
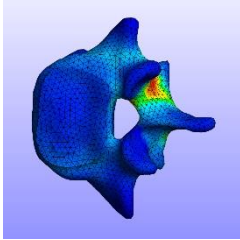
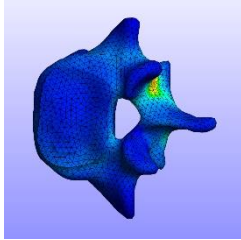
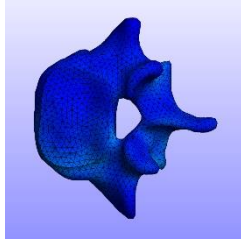
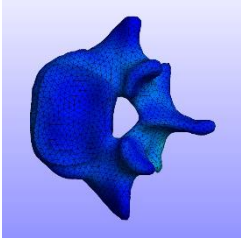
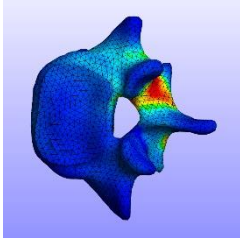
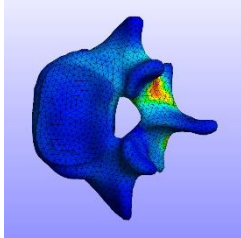
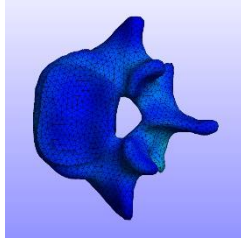
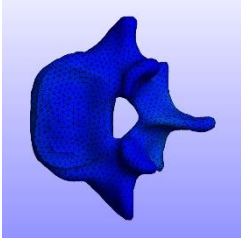
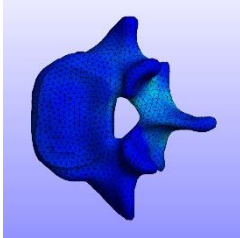
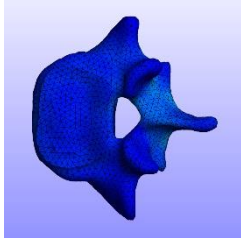
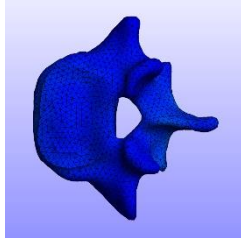
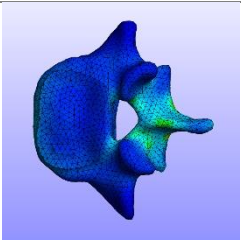
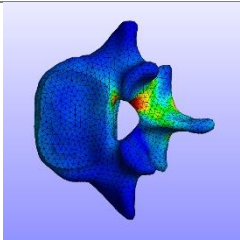
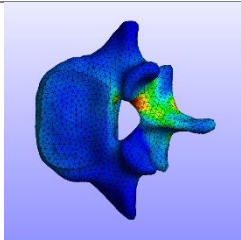
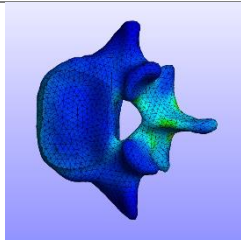
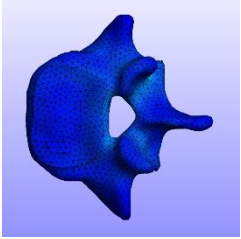
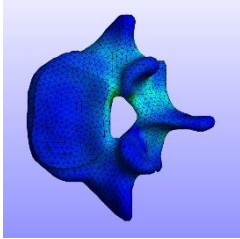
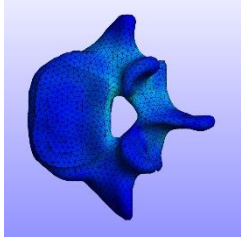
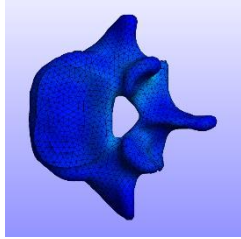
LLB				
RAR				
Compression				

Table 27A: Impact of five bending loading modes combined with 824N, and pure compression alone, on spondylolysis healing phases

824N	Intact	Granulation	Cartilage	Mature Bone
Flex				
Ext				

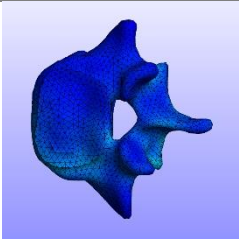
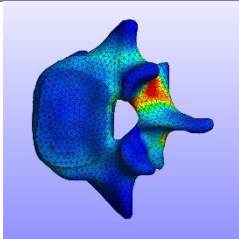
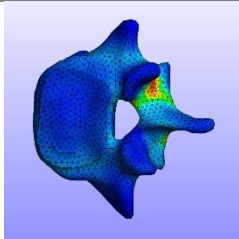
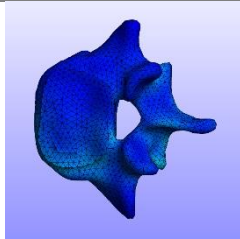
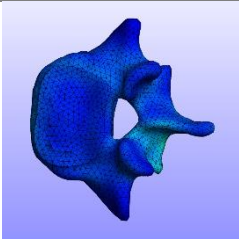
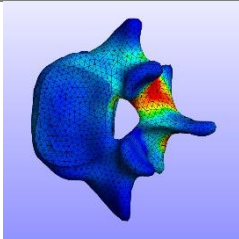
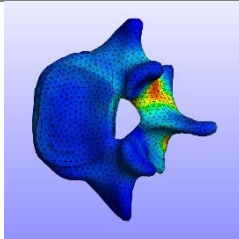
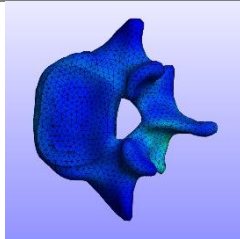
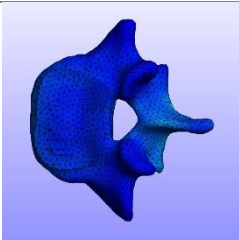
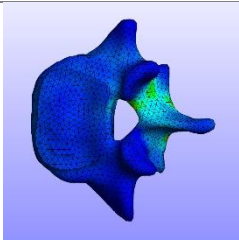
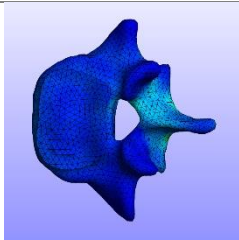
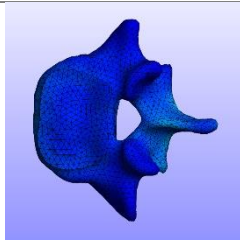
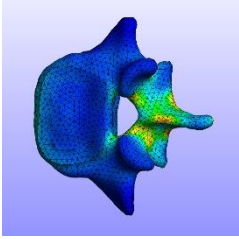
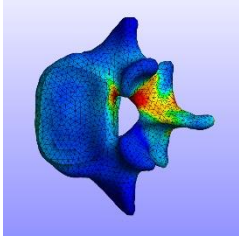
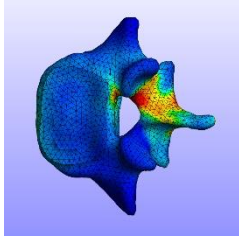
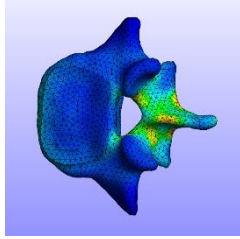
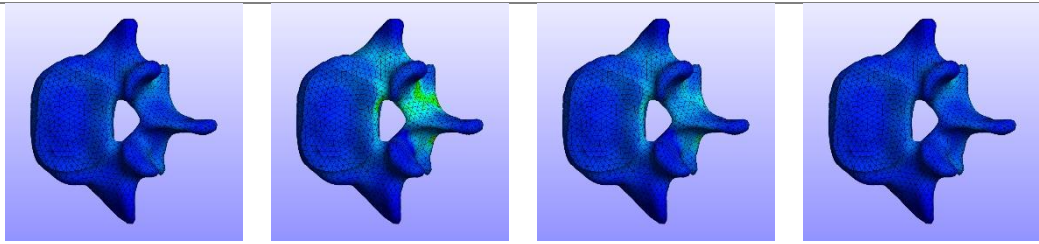
LLB				
RAR				
Compression				

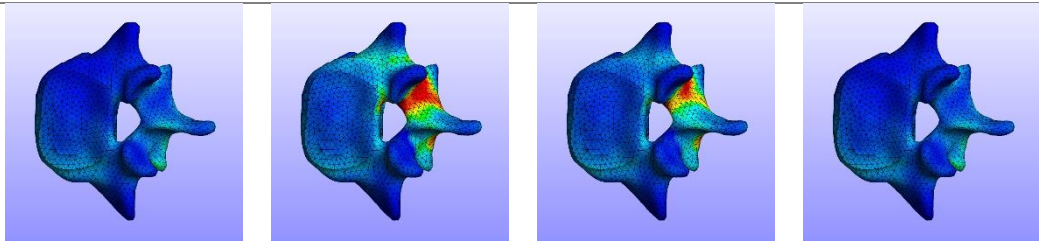
Table 28A: Impact of five bending loading modes combined with 1060N, and pure compression alone, on spondylolysis healing phases

1060N	Intact	Granulation	Cartilage	Mature Bone
Flex				

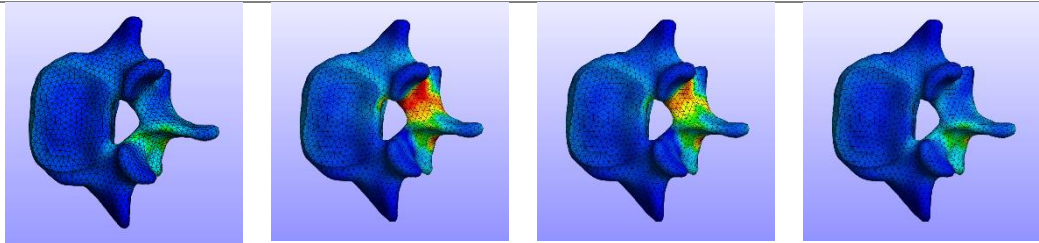
Ext



LLB



RAR



**Comp
ression**

