Research Article
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Transpedicular stabilization strategies for thoracolumbar burst fractures: evaluation and improvement in the context of flexion loads

Abstract

Introduction: Burst fractures, characterized by a wide variability in pathomorphological changes, represent one of the most tactically debated issues in modern spinal surgery. The question of treatment strategy is particularly relevant for the thoracolumbar junction area, which, due to its biomechanical features, is especially prone to traumatic injuries. The aim of the study is to investigate the stress-strain state of a lumbar spine model with a burst fracture of the T12 vertebra under different transpedicular fixation options and forward trunk inclination.

Material and methods: We developed a mathematical finite element model of the human thoracolumbar spine, considering a burst fracture of the T12 vertebra. The model also includes a transpedicular stabilization system consisting of 8 screws implanted in the T10, T11, L1, and L2 vertebrae. We simulated four variants of transpedicular fixation using short (monocortical) and long (bicortical) screws, which penetrate the anterior wall of the vertebral body, both with and without two cross-links.

Results: The analysis revealed that the different configurations demonstrated varying stress levels in the analyzed regions of the model. For example, the calculated stress values for the body of the fractured vertebra were 22.6, 25.1, 22.4, and 24.7 MPa, respectively, for models with monocortical screws without cross-links, bicortical screws without cross-links, monocortical screws with cross-links, and bicortical screws with cross-links.

Conclusion: The study provides data on the stress distribution within the lumbar spine model with a burst fracture of the T12 vertebra under various fixation strategies and simulated flexion loading. These findings can aid in clinical decisions regarding the most effective transpedicular stabilization methods to optimize patient outcomes.

Keywords: thoracolumbar junction, spinal trauma, burst fractures, biomechanics, finite element analysis, spinal stabilization

Volume 16 Issue 3 - 2024

Oleksii S Nekhlopochyn,[1] Vadim V Verbov,[2] Ievgen V Cheshuk,[2] Michael Yu Karpinsky,[3] Oleksandr V Yaresko[3]

[1]Department of spinal neurosurgery, Romodanov Neurosurgery Institute, Ukraine
[2]Restorative Neurosurgery Department, Romodanov Neurosurgery Institute, Ukraine
[3]Biomechanical Laboratory, Sytenko Institute of Spine and Joint Pathology, Kharkiv, Ukraine

Correspondence: Oleksii S. Nekhlopochyn, MD, PhD, Senior researcher of Department of spinal neurosurgery, Romodanov Neurosurgery Institute of National Academy of Medical Sciences of Ukraine. Adress: P. Mayborody str. 32., 04050, Kiev, Ukraine, Tel +380 44 483-94-13; Email AlexeNS@gmail.com

Received: May 25, 2024 | **Published:** June 11, 2024

Introduction

In recent decades, the concept of evidence-based medicine (EBM) has effectively become the main vector shaping the development of modern healthcare, both at the theoretical foundation level and in applied clinical aspects. Considering three main factors—results of scientific research, clinical experience of the physician, and patient value preferences—EBM, along with several other advantages, ensures an individualized approach to the patient, improvement of clinical outcomes, reduction of variability in treatment, and justified use of resources.[1] The latter point is particularly significant against the backdrop of progressively increasing economic costs of providing medical care to the population using the most modern, and thus expensive, diagnostic and treatment methods.[2] According to a number of experts, this has key importance in the active promotion of the EBM concept. In fact, EBM to a certain extent ensures a balance between the medical and economic components of healthcare. However, in some cases, it determines the application of the minimally effective treatment method, which is most appropriate for society as a whole, but not necessarily for the individual patient.[3,4]

The application of EBM concept to traumatic spinal injuries presents clear methodological challenges. The endless diversity of radiological characteristics, determined by complex pathomorphological disruptions of various components of the spinal column, creates significant difficulties in the categorization of injuries.[5] It has been noted that in situations where classification systems exhibit low reliability and validity, there is a likelihood of significant heterogeneity within each classification category and, consequently, variability in the treatment of injuries between clinicians, medical centers, and geographic regions.[6]

This issue has been partially addressed by actively integrating simpler and presumably more clinically oriented systems, such as the AO Spine Subaxial Injury Classification System and the Thoracolumbar Injury Classification System, into clinical practice.[7,8] These systems replaced the highly detailed classifications of Ben L. Allen and F. Magerl for subaxial cervical and thoracolumbar regions, respectively.[9,10] However, the significant reduction in classification categories inevitably decreases detail, excluding from analysis a large number of features. Some of these features may fundamentally impact the strategy, tactics, and outcomes of treatment for patients.[11] Moreover, numerous factors significantly influence therapy outcomes in addition to the nature of the injury itself. These factors include age, weight, lifestyle, the presence and severity of osteoporosis, and others.[12] Consequently, forming adequate clinical comparison groups necessary to obtain the most reliable results when evaluating different treatment methods is challenging.[13]

The issue is most pertinent to traumatic injuries of the thoracolumbar junction (TLJ), which is the focus of our interest. Despite the obvious and well-studied biomechanical differences between this zone and other sections of the thoracolumbar spine, most researchers, aiming to increase the volume of clinical groups and thereby obtain statistically more reliable results, prefer to consider the entire region from Th1 to L5 as a whole.[14] This is evidenced by the fact that as of May 2024, only 89 publications dedicated exclusively to traumatic injuries of TLJ are indexed in the PubMed database, while the number of studies describing the characteristics of thoracolumbar spine injuries in general exceeds 2350. This approach results in significant heterogeneity in the findings obtained by researchers.

Moreover, even within the thoracolumbar spine as a whole, certain traumatic injuries pose substantial challenges in determining the optimal treatment strategy from EBM perspective.[15,16] The most relevant and widely known issue to almost every practicing spinal surgeon is the debate over the treatment methods for burst fractures. The discussion involves conservative versus surgical treatment methods, various stabilization options such as long or short constructs, with or without involvement of the injured vertebra, combinations with cementoplasty, partial or complete vertebral body resection performed via anterior or posterior approaches.[17] However, according to a systematic review prepared by the Congress of Neurological Surgeons, there is still no convincing evidence of the superiority of any particular treatment method.[18]

This publication presents a fragment of a study dedicated to examining therapeutic methods for traumatic injuries of TLJ, taking into account its biomechanical and clinical features. We focus on the biomechanical indicators of stabilizing a burst fracture using an 8-screw long segment transpedicular fixation (TPF) without involving the injured vertebra. This method has several clinical advantages—it provides optimal preservation of the spinal axis, promotes quicker consolidation due to reliable fixation, and, if effective, allows for the removal of the system and re-mobilization of the fixed segments.[19]

However, even within the discussed method, there are different implementation options: open installation of the TPF system or minimally invasive installation. The clear advantages of the latter, such as minimizing soft tissue trauma, reducing blood loss, and lowering infection risks, are somewhat offset by the inability to use crosslinks, which, according to several researchers, significantly impact the rigidity and reliability of fixation.[20] This fact can be critically important in the TLJ, as it is the most loaded section of the spine. Additionally, the depth of screw insertion into the vertebral bodies (monocortical or bicortical installation) also affects the load distribution across the fixed segments, according to the literature.[21,22] At the same time, a review of the literature did not reveal studies that specifically examine these parameters of TPF fixation in the context of the TLJ, highlighting the relevance of this research.

Objective

To study the stress-strain state of a thoracolumbar spine model with a burst fracture of the Th12 vertebra under different variants of transpedicular fixation and forward trunk tilt.

Materials and methods

In the Biomechanics Laboratory of Sytenko Institute of Spine and Joint Pathology, NAMS of Ukraine, a mathematical finite element model of the human thoracolumbar spine was created, considering a burst fracture of the Th12 vertebra. The model also includes a transpedicular stabilization system consisting of 8 screws implanted in the Th10, Th11, L1, and L2 vertebrae. To model the burst fracture, the Th12 vertebral body was divided into separate fragments by several planes (Figure 1). The gaps between the fragments were filled with a material that simulates interfragmentary regenerate. The model consists of 44,583 tetrahedral 10-node isoparametric finite elements with quadratic approximation and has 136,472 nodes.

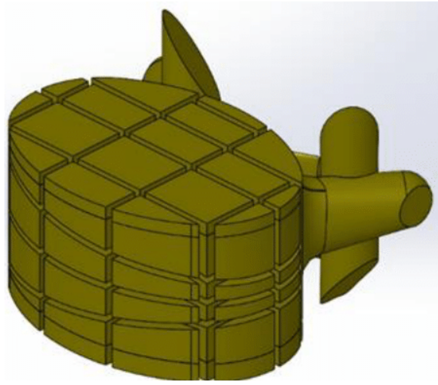

Figure 1 Model of the Th12 vertebra simulating burst fracture.

We simulated four variants of transpedicular fixation using short (monocortical) and long (bicortical) screws, which penetrate the anterior wall of the vertebral body, as well as with and without two crosslinks. It was assumed in the modeling that the material is homogeneous and isotropic. A detailed justification for the appropriateness of this assumption is provided in previous publications. A 10-node tetrahedral element with quadratic approximation was used as the finite element. The mechanical properties of biological tissues, such as cortical and cancellous bone and intervertebral discs, were selected based on literature data.[23,24] For the metal elements, titanium was used, and its mechanical characteristics were chosen according to technical literature.[25] The analysis employed parameters such as E – modulus of elasticity (Young's modulus) and ν – Poisson's ratio. Information on the mechanical characteristics of the materials is presented in Table 1.

Table 1 Mechanical properties of materials used in modeling

Material	Young's Modulus (MPa)	Poisson's Ratio
Cortical Bone	10,000	0.3
Cancellous Bone	450	0.2
Articular Cartilage	10.5	0.49
Intervertebral Discs	4.2	0.45
Interfragmentary Regenerate	1.0	0.45
Titanium VT-16	110,000	0.3

The stress-strain state of the models was studied under the influence of a bending load acting from back to front, simulating forward trunk tilt. The load was applied to the body of the Th9 vertebra and the facet joints. The load application point was chosen to simulate the physiological bending of the spine during forward flexion. The magnitude of the load was 350 N, corresponding to the weight of the upper body. The model had a rigid fixation at the caudal surface of the L5 disc, providing a consistent basis for analyzing the deformation and stress distribution in the upper segments.

For convenience in studying the changes in the stress-strain state of the models depending on the method of transpedicular fixation, the stress magnitude was determined at specific control points (Figure 2). These control points were strategically selected at various critical locations on the vertebrae and the fixation hardware to provide a comprehensive understanding of how different fixation techniques affect the distribution of stress throughout TLJ area. By analyzing the stress at these control points, we aimed to identify potential areas of high stress concentration that could indicate weaknesses in the fixation method or areas prone to failure. The choice of control points was based on both clinical relevance and biomechanical considerations. Clinically, these points correspond to regions commonly associated with complications such as screw loosening, hardware failure, or bone fracture. Biomechanically, they represent regions where the load transfer between the spine and the fixation hardware is most critical.

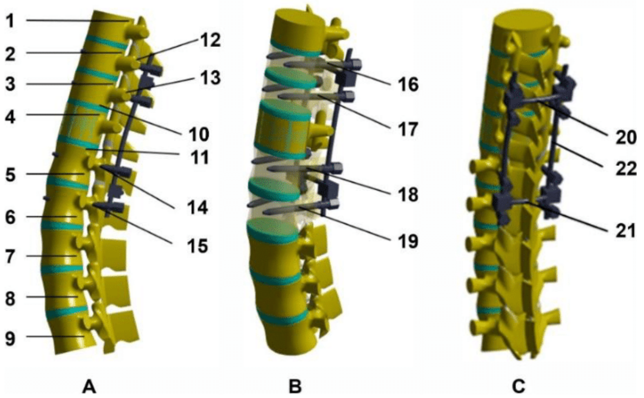

Figure 2 Finite element model in lateral (a), anterolateral (b), and posterolateral (c) view indicating the location of control points: 1 - Th9 vertebral body; 2 - Th10 vertebral body; 3 - Th11 vertebral body; 4 - Th12 vertebral body; 5 - L1 vertebral body; 6 - L2 vertebral body; 7 - L3 vertebral body; 8 - L4 vertebral body; 9 - L5 vertebral body; 10 - Lower endplate of Th11 vertebral body; 11 - Upper endplate of L1 vertebral body; 12 - Screw entry in Th10 vertebral arch; 13 - Screw entry in Th11 vertebral arch; 14 - Screw entry in L1 vertebral arch; 15 - Screw entry in L2 vertebral arch; 16 - Screws in Th10 vertebral body; 17 - Screws in Th11 vertebral body; 18 - Screws in L1 vertebral body; 19 - Screws in L2 vertebral body; 20 - Crosslinks between screws in Th10 and Th11 vertebral bodies; 21 - Crosslinks between screws in L1 and L2 vertebral bodies; 22 – Rods.

The stress-strain state of the models was studied using the finite element method. The criterion for evaluating the stress state of the models was von Mises stress.[26] The modeling was performed using the computer-aided design (CAD) system SolidWorks (Dassault Systemes, France). Calculations of the stress-strain state of the models were carried out using the CosmosM software package.[27]

Results

The application of transpedicular fixation with short screws without crosslinks - Model Modification No. 1 (Figure 3) for the fixation of the Th12 vertebra in cases of burst fractures and forward bending resulted in maximum stress values of 23.1 MPa and 23.6 MPa in the bodies of the L3 and L4 vertebrae, respectively. The stress level in the Th12 vertebral body also remained high at 22.6 MPa. Around the fixing screws, the highest stress value of 11.1 MPa was recorded in the L3 vertebral arches, while the lowest stress value of 3.9 MPa was observed in the Th10 vertebral arches. The most stressed element of the metal construction was identified as the rods, where the stress reached 326.1 MPa. On the fixing screws, the maximum stress level of 42.9 MPa was observed in the L2 vertebra, while the minimum stress level of 18.3 MPa was recorded on the screws in the L1 vertebra. The screws in the thoracic vertebrae were loaded uniformly, as indicated by the stress levels of 21.4 MPa and 23.1 MPa in the Th10 and Th11 vertebrae, respectively.

Replacing the short screws with long screws in the stabilization system without crosslinks - Model Modification No. 2 (Figure 4) allows for a slight reduction in stress levels in the bodies of the intact vertebrae, but the stress in the Th12 vertebral body increases to 25.1 MPa. An increase in stress levels around the transpedicular screws was also observed, with the most significant increase occurring in the L3 vertebral arches, where the stress more than doubled to 23.3 MPa. Conversely, the stress in the Th11 vertebral arches nearly halved, decreasing to 4.9 MPa. The stress levels on all transpedicular screws tend to slightly increase overall. However, a decrease in stress levels on the rods was noted, dropping to 280.2 MPa.

The combination of short fixation screws with crosslinks - Model Modification No. 3 (Figure 5) led to a slight reduction in the maximum stress values at all control points in the model. The most significant stress reduction was recorded in the L3 vertebral body, where the stress decreased from 23.1 MPa to 17.1 MPa. A reduction in stress levels was also observed in all elements of the metal construction, including the rods, where stress levels dropped to 319.2 MPa. The stress levels on the crosslinks were determined to be 2.1 MPa and 4.6 MPa on the upper and lower crosslinks, respectively.

The use of crosslinks in combination with long fixation screws - Model Modification No. 4 (Figure 6), compared to the model without crosslinks, also leads to a reduction in stress values at all control points in the model. This trend applies to both the bony and metal elements of the model, with the exception of the crosslinks, where the stress levels decrease to 1.9 MPa and 5.8 MPa on the upper and lower crosslinks, respectively.

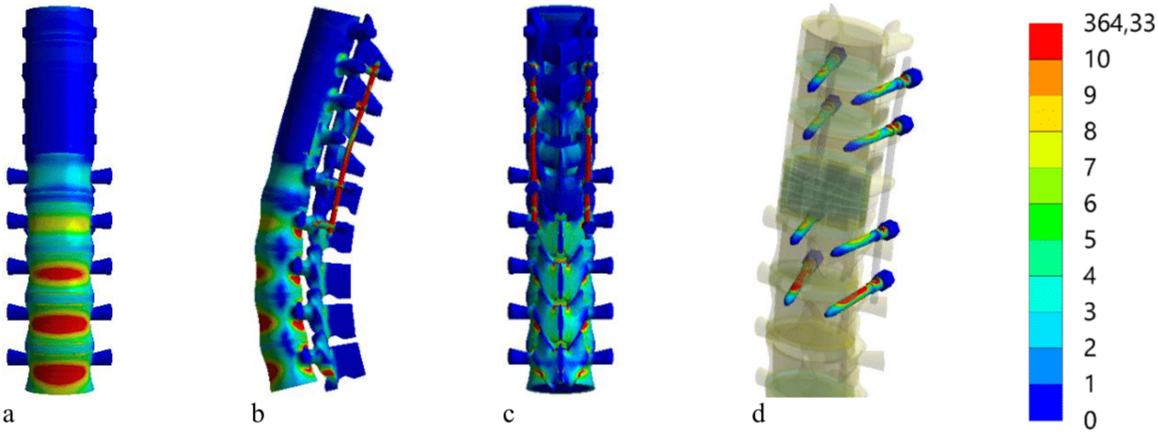

Figure 3 Stress distribution in the thoracolumbar spine model with a burst fracture of the Th12 vertebra under load simulating forward trunk tilt. transpedicular fixation with short screws without crosslinks (Model Modification No. 1): a – front view; b – side view; c – rear view; d – screws.

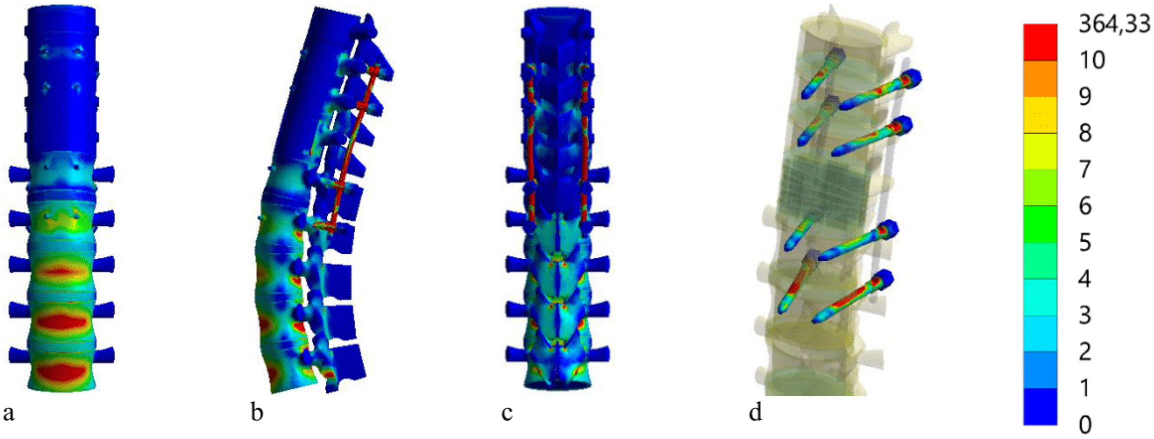

Figure 4 Stress distribution in the thoracolumbar spine model with a burst fracture of the Th12 vertebra under load simulating forward trunk tilt. transpedicular fixation with bicortical screws without crosslinks (Model Modification No. 2): a – front view; b – side view; c – rear view; d – screws.

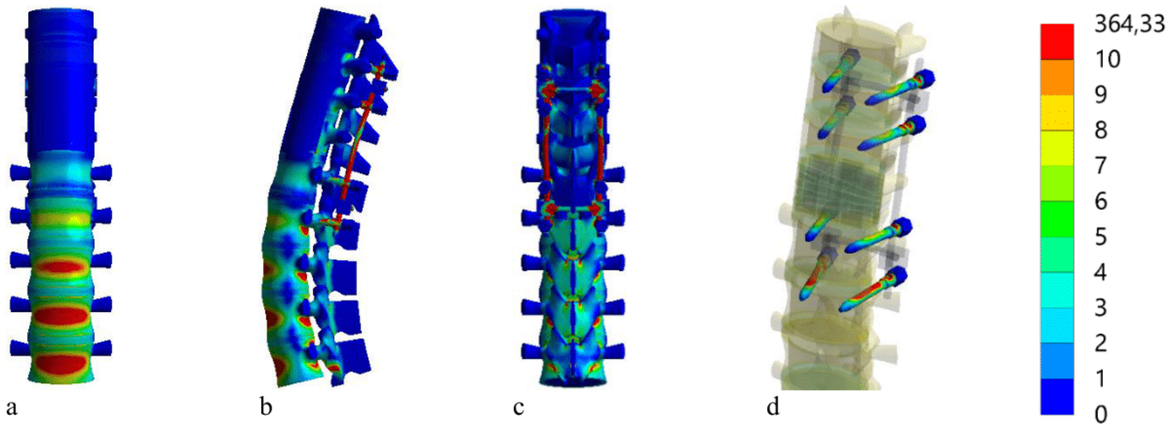

Figure 5 Stress distribution in the thoracolumbar spine model with a burst fracture of the Th12 vertebra under load simulating forward trunk tilt. transpedicular fixation with monocortical screws and crosslinks (Model Modification No. 3): a – front view; b – side view; c – rear view; d – screws.

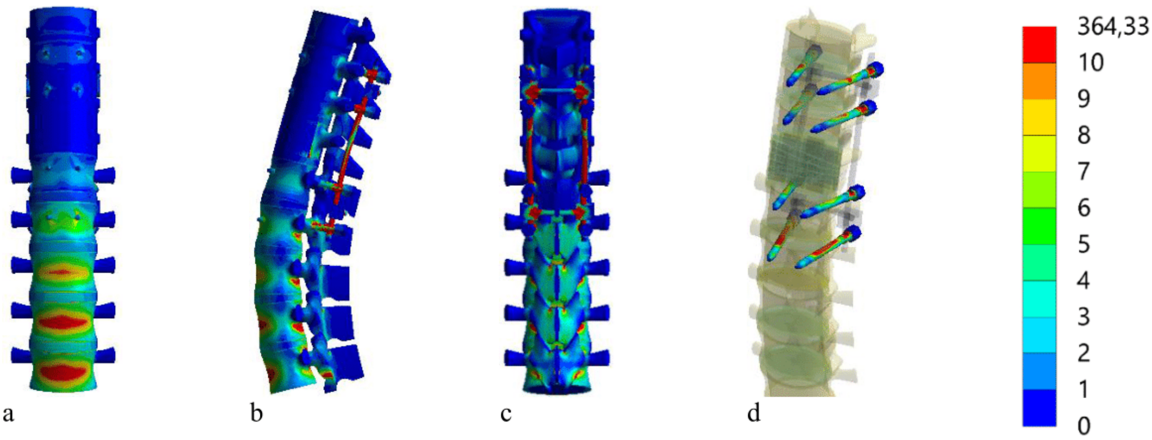

Figure 6 Stress distribution in the thoracolumbar spine model with a burst fracture of the Th12 Vertebra Under Load Simulating Forward Trunk Tilt. Transpedicular Fixation with Bicortical Screws and Crosslinks (Model Modification No. 4): a – front view; b – side view; c – rear view; d – screws.

The data on the magnitude of stress at all control points of the transpedicular fixation models are presented in Table 2.

Upon conducting a comparative analysis of the stress data provided for the different fixation models, the following key observations can be made.

Table 2 Stress under the influence of load simulating forward trunk bending in models of the thoracolumbar spine with a burst fracture of the Th12 vertebra for various options of transpedicular fixation

No		Control points	Stress, MPa			
			Model without crosslinks		Model with crosslinks	
			Short screws	Long screws	Short screws	Long screws
	Model number		1	2	3	4
1	Bone Tissue	Th9 Vertebra Body	1.7	1.4	1.7	1.4
2		Th10 Vertebra Body	9.0	6.2	8.7	6.0
3		Th11 Vertebra Body	6.2	6.6	5.7	6.3
4		Th12 Vertebra Body	22.6	25.1	22.4	24.7
5		L1 Vertebra Body	5.1	4.4	4.7	4.2
6		L2 Vertebra Body	18.7	14.0	15.6	13.2
7		L3 Vertebra Body	23.1	13.9	17.1	13.2
8		L4 Vertebra Body	23.6	18.0	21.6	16.8
9		L5 Vertebra Body	15.1	14.4	14.8	13.8
10		Lower Endplate of Th11	4.2	3.5	4.1	3.4
11		Upper Endplate of L1	9.1	6.4	8.9	6.2
12		Entry of Screws into Arch of Th10	3.9	5.8	3.4	5.2
13		Entry of Screws into Arch of Th11	8.6	4.9	8.2	4.7
14		Entry of Screws into Arch of L1	7.4	9.7	6.9	9.3
15		Entry of Screws into Arch of L2	11.1	23.3	11.1	11.3
16	Metal Constructs	Screws in Th10 Body	21.4	25.4	20.6	26.9
17		Screws in Th11 Body	23.1	27.2	21.1	23.5
18		Screws in L1 Body	18.3	19.4	17.4	18.4
19		Screws in L2 Body	42.9	45.5	38.6	42.8
20		Crosslinks between Th10 and Th11 Screws	-	-	2.1	1.9
21		Crosslinks between L1 and L2 Screws	-	-	4.6	5.8
22		Connecting rods	326.1	280.2	319.2	235.7

Vertebral bodies of unfixed segments (Th9, L3, L4, L5)

The highest stress is recorded in the L4 vertebral body for Model 1 (23.6 MPa). Using long screws in Model 2 reduces the stress by 23.7% to 18.0 MPa. Model 3, which includes crosslinks, shows a stress of 21.6 MPa, an 8.5% reduction compared to Model 1. The combination of long screws and crosslinks in Model 4 results in the lowest stress at 16.8 MPa, a further reduction of 6.7% compared to Model 2 and 22.2% compared to Model 3.

Stress in the L3 vertebral body is highest in Model 1 (23.1 MPa). Model 2 significantly reduces this to 13.9 MPa, a 39.8% decrease. Model 3 shows a stress of 17.1 MPa, a 25.9% reduction compared to Model 1. Model 4 further reduces stress in L3 to 13.2 MPa, 5.0% lower than Model 2 and 22.8% lower than Model 3.

The L5 vertebral body shows stress levels of 15.1 MPa in Model 1. Model 2 reduces this to 14.4 MPa, a 4.6% reduction. Model 3 slightly decreases stress to 14.8 MPa, a 2.0% reduction. Model 4 achieves the lowest stress at 13.8 MPa, 4.2% lower than Model 2 and 6.8% lower than Model 3. The Th9 vertebral body experiences the least stress across all models. Model 1 and Model 3 show a stress of 1.7 MPa. Model 2 and Model 4 both record a lower stress of 1.4 MPa, which is a 17.6% reduction compared to Model 1 and Model 3, maintaining the lowest stress levels.

Vertebral bodies of fixed segments (Th10, Th11, L1, L2)

The highest stress in the fixed segments is observed in the L2 vertebral body across all models. Model 1 shows the highest stress (18.7 MPa). Model 2 reduces this stress to 14.0 MPa, a 25.1% decrease. Model 3 shows a stress of 15.6 MPa, a 16.6% reduction compared to Model 1. Model 4 further reduces the stress to 13.2 MPa, 5.7% lower than Model 2 and 15.4% lower than Model 3. The Th10 vertebral body stress is highest in Model 1 (9.0 MPa). Model 2 shows a reduced stress of 6.2 MPa, a 31.1% decrease. Model 3 records a stress of 8.7 MPa, a 3.3% reduction compared to Model 1. Model 4 achieves the lowest stress at 6.0 MPa, which is 3.2% lower than Model 2 and 31.0% lower than Model 3.

Model 1 shows a stress of 6.2 MPa in the Th11 vertebral body. Model 2 has a slightly higher stress of 6.6 MPa, a 6.5% increase compared to Model 1. Model 3 shows a stress of 5.7 MPa, an 8.1% decrease compared to Model 1. Model 4 records a stress of 6.3 MPa, 4.5% lower than Model 2 and 10.5% higher than Model 3. Stress in the L1 vertebral body is highest in Model 1 (5.1 MPa). Model 2 reduces this to 4.4 MPa, a 13.7% decrease. Model 3 shows a stress of 4.7 MPa, a 7.8% reduction compared to Model 1. Model 4 achieves the lowest stress at 4.2 MPa, 4.5% lower than Model 2 and 10.6% lower than Model 3.

Damaged vertebral body (Th12)

The Th12 vertebral body shows the highest stress across all models. In Model 1, the stress is 22.6 MPa. Model 2 increases the stress to 25.1 MPa, an 11.1% increase compared to Model 1. Model 3 shows a stress of 22.4 MPa, slightly lower than Model 1 by 0.9%. Model 4 records the highest stress in Th12 at 24.7 MPa, which is 9.3% higher than Model 3.

Endplates

The lower endplate of Th11 shows stress levels of 4.2 MPa in Model 1. Model 2 reduces this to 3.5 MPa, a 16.7% decrease. Model 3 shows a stress of 4.1 MPa, a slight 2.4% decrease compared to Model 1. Model 4 records the lowest stress at 3.4 MPa, a 2.9% decrease compared to Model 2 and a 17.1% decrease compared to Model 3.

Stress in the upper endplate of L1 is highest in Model 1 (9.1 MPa). Model 2 reduces this to 6.4 MPa, a 29.7% decrease. Model 3 shows a stress of 8.9 MPa, a 2.2% decrease compared to Model 1. Model 4 achieves the lowest stress at 6.2 MPa, 3.1% lower than Model 2 and 30.3% lower than Model 3.

Screw entry zones

The stress at the screw entry in the Th10 arch is 3.9 MPa in Model 1. Model 2 increases this to 5.8 MPa, a 48.7% increase. Model 3 shows a reduced stress of 3.4 MPa, a 12.8% decrease compared to Model 1. Model 4 records a stress of 5.2 MPa, which is 10.3% lower than Model 2 but 52.4% higher than Model 3. The Th11 arch shows a stress of 8.6 MPa in Model 1. Model 2 reduces this to 4.9 MPa, a 43.0% decrease. Model 3 shows a stress of 8.2 MPa, a 4.7% decrease compared to Model 1. Model 4 records the lowest stress at 4.7 MPa, a 4.1% decrease compared to Model 2 and 42.7% lower than Model 3.

The L2 arch shows the highest stress among the entry zones in Model 1 (11.1 MPa). Model 2 significantly increases this to 23.3 MPa, a 109.9% increase. Model 3 maintains the stress at 11.1 MPa, the same as Model 1. Model 4 reduces the stress to 11.3 MPa, a 117.3% increase compared to Model 1 but a 51.5% decrease compared to Model 2. The stress at the screw entry in the L1 arch is 7.4 MPa in Model 1. Model 2 increases this to 9.7 MPa, a 31.1% increase. Model 3 shows a reduced stress of 6.9 MPa, a 6.8% decrease compared to Model 1. Model 4 records a stress of 9.3 MPa, which is 4.1% lower than Model 2 but 34.8% higher than Model 3.

Screws

Stress on screws in the Th10 vertebral body is 21.4 MPa in Model 1. Model 2 increases this to 25.4 MPa, an 18.7% increase. Model 3 shows a stress of 20.6 MPa, a 3.7% decrease compared to Model 1. Model 4 records the highest stress at 26.9 MPa, a 5.9% increase compared to Model 2 and a 30.6% increase compared to Model 3.

Value on screws in the Th11 vertebral body is 23.1 MPa in Model 1. Model 2 increases this to 27.2 MPa, a 17.7% increase. Model 3 shows a stress of 21.1 MPa, an 8.7% decrease compared to Model 1. Model 4 records a stress of 23.5 MPa, a 13.6% decrease compared to Model 2 and an 11.4% increase compared to Model 3.

Stress on screws in the L2 vertebral body is 42.9 MPa in Model 1. Model 2 increases this to 45.5 MPa, a 6.1% increase. Model 3 reduces the stress to 38.6 MPa, a 10.0% decrease compared to Model 1. Model 4 records a stress of 42.8 MPa, which is 5.9% lower than Model 2 but 11.1% higher than Model 3.

Crosslinks: Model 3 records a stress of 2.1 MPa on the crosslinks between Th10 and Th11 screws. Model 4 reduces this stress to 1.9 MPa, a 9.5% decrease compared to Model 3. On the crosslinks between L1 and L2 screws Model 3 shows a stress of 4.6 MPa. Model 4 increases this stress to 5.8 MPa, a 26.1% increase compared to Model 3.

Rods: The rods experience the highest stress in Model 1 (326.1 MPa). Model 2 reduces this to 280.2 MPa, a 14.1% decrease. Model 3 shows a stress of 319.2 MPa, a 2.1% decrease compared to Model 1. Model 4 achieves the lowest stress at 235.7 MPa, a 15.9% decrease compared to Model 2 and a 26.2% decrease compared to Model 3.

Overall, summarizing the conducted analysis, it should be noted that during forward bending, the stress values in the metal elements of the models differ insignificantly between different transpedicular fixation options. The combination of monocortical screws and crosslinks resulted in the lowest stress values in most of the control points of the models. Comparison of the stress indicators arising at the control points of the model during forward bending suggests that the use of long fixation screws leads to an increase in maximum stress values, both in the bony elements of the model and in the metal construction, compared to models using short screws. The application of crosslinks leads to a reduction in stress levels at all control points of the models, regardless of the length of the fixation screws.

Discussion

As noted earlier, burst fractures are one of the most controversial topics when analyzing traumatic injuries of the thoracolumbar spine. The contentious points arise not only in purely clinical tactical approaches but also in experimental biomechanical studies. A review of the literature shows highly contradictory results regarding the finite element analysis of this pathology.

One reason for such discrepancies is the broad interpretation of the concept of a burst fracture, leading to the use of different modeling principles. For instance, Recep Basaran et al. used a vertebral model with a wedge-shaped deformation of the body, considering only the anterior-superior two-thirds to be damaged.[28] A similar principle, but with resection of the lower part of the vertebral body, was used by Jiangping Xu et al.[29] In both cases, the posterior wall of the damaged vertebral body remains intact.

In contrast, Changqing Li et al. presented finite element analysis data where a burst fracture was modeled by simply removing the lower half of the damaged vertebral body, while the upper part remained completely intact.[30] A similar approach is used in the studies of Hongwei Wang et al.[31] Worawat Limthongkul et al.[32] modeled vertebral damage by reproducing its deformed shape based on CT data, but did not report differences in the elastic properties of the damaged body.[32] Chia-En Wong et al.[33] used a principle of altering the properties of the middle third of the vertebral body to simulate a reduction in supporting function, while the endplates remained undamaged.[33]

This diversity in modeling approaches leads to significant differences in the loading indicators obtained for both the bony elements of the model and the metal constructs, making comparative analysis of the results challenging. In our study, we aimed to replicate two main features that fundamentally distinguish burst fractures from other types of traumatic injuries: the presence of body fragmentation and damage to the posterior wall of the vertebral body-the middle supporting column according to F. Denis' concept.[34] Therefore, despite the certain approximation inevitably present in any finite element studies, our model can be considered as most accurately reproducing the biomechanical characteristics of the damaged thoracolumbar junction with a burst fracture of the Th12 vertebral body.

Moreover, the analysis of the aforementioned publications demonstrates a rather limited range of control points for loading registration. In the vast majority of cases, the main focus is on the elements of the metal construct, which is quite natural since fragmentation or dislocation of these elements leads to the failure of fixation.[35,36] At the same time, the analysis of body loading, particularly the damaged vertebral body, is usually not conducted. This aspect has fundamentally important clinical significance, as it has been noted that even after a successfully performed posterior transpedicular stabilization, there are cases of increasing kyphotic deformity, likely due to the high load on the traumatically fragmented body.

The results obtained in this study demonstrate that flexion loading is a rather unfavorable biomechanical condition for the stabilized thoracolumbar spine region. Significant loads were recorded on the connecting rods; however, the calculated values, even for the most unfavorable configurations in this aspect, are far from the tensile strength limit of VT16 titanium (1030 MPa to 1225 MPa) used in most modern transpedicular systems.[25] Additionally, the comparative results of loading on the transpedicular screws are quite indicative. In our study, for all considered modifications, the highest values were recorded on the screw in the L2 body, almost twice the values noted on other screws. This fact contradicts the results reported in several of the above-mentioned publications but finds its clinical confirmation. Many researchers note that the most distally located screws are characterized by the highest fragmentation frequency, in our case, the caudal-distal screw.

In summarizing the above fragment of the work, it should be emphasized once again the necessity of an individualized approach in treating patients with traumatic injuries of the thoracolumbar spine region, particularly burst fractures of the vertebral bodies in this segment. The appropriateness of using a particular method of transpedicular fixation is evidently dictated by a number of clinical indicators of the patient and prognostic perspectives. For example, in cases where the complex of clinical and morphological signs, such as age, body mass index, gender, comorbidities, degree of osteoporosis, and nature of the vertebral body injury, suggests a relatively rapid consolidation with the prospect of subsequent system dismantling and segment remobilization, a minimally invasive percutaneous stabilization with standard screw lengths, not involving the installation of transverse connectors (model No. 1), is justified. Conversely, in situations where long-term or permanent fixation is anticipated based on the complex of signs, model No. 4-with bicortical screws in combination with two transverse connectors—appears more promising.

Undoubtedly, such statements are rather theoretical and require further clinical confirmation. However, as already discussed, such studies, due to the complexity of their implementation, are a somewhat distant prospect. The data we obtained can already be used in clinical practice to reduce tissue trauma when using minimally invasive techniques and to decrease the incidence of fixation failure caused by both fragmentation of the stabilization system and screw dislocation due to prolonged excessive loading of the bone tissue. Furthermore, the conclusion regarding the appropriateness and characteristics of various types of fixation can only be made by analyzing all loading patterns (compression, flexion, extension, rotation, etc.), which determines the direction of our future research.

Limitations

The model construction assumed the material properties to be homogeneous and isotropic. The poroviscoelastic characteristics of the spinal tissues were not included, as all loads were applied under quasi-static conditions. This approach removes individual patient variations, allowing the focus to remain on the fundamental differences between different fixation methods. As such, this type of simplification is considered appropriate and justified.

Conclusion

The use of long fixation screws during forward inclination causes increased stress levels in the bony elements of the model compared to short screws. Cross-links help reduce the stress magnitude at all control points of the models, regardless of the screw length. When selecting a fixation method, a comprehensive analysis of factors with clinical and prognostic significance is critical.

Acknowledgments

None.

Conflicts of interest

The authors declare that there are no conflicts of interest.

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