Original article

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Short-segment stabilization techniques for burst fractures of the thoracolumbar junction: a finite element study under lateral flexion

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Introduction: Burst fractures of the thoracolumbar junction (TLJ, T10–L2) are common spinal injuries associated with a high risk of neurological complications. Transpedicular fixation is one of the most effective treatment methods; however, the optimal choice of fixation configuration remains unresolved. This study aims to analyze the stress-strain state of various short-segment transpedicular fixation configurations for Th12 vertebra burst fractures under lateral flexion loading.

Materials and methods: A finite element model of the Th9–L5 spinal segment with a simulated Th12 burst fracture was created. Four fixation configurations were considered: M1 – short screws in Th11 and L1 (without intermediate screws), M2 – long screws in Th11 and L1 (without intermediate screws), M3 – short screws in Th11 and L1 with intermediate screws in Th12, and M4 – long screws in Th11 and L1 with intermediate screws in Th12.

The models were analyzed using *CosmosM* software, assessing equivalent von Mises stress at 18 control points. Loads simulated physiological lateral trunk bending.

Results: Models with long screws (M2, M4) demonstrated lower maximum stresses in connecting rods (315.5–321.0 MPa) compared to short screws (324.8–324.9 MPa). The inclusion of intermediate screws (M3, M4) significantly reduced stress in the fractured Th12 vertebra (by up to 28%), in adjacent vertebral endplates (by 18–25%), and at screw entry points into vertebral arches (up to 28%). The lowest fixation screw stresses were observed in the model with long and intermediate screws (up to 38% lower compared to the baseline model M1). However, intermediate screws minimally influenced stresses in the connecting rods (up to 1.2%).

Conclusions: The optimal short-segment transpedicular fixation configuration is the use of long screws in adjacent vertebrae combined with intermediate fixation in the fractured vertebra (M4). This approach provides optimal load distribution, reduces the risk of construct failure, and preserves mobility of adjacent segments. Long screws improve overall system stiffness, while intermediate screws effectively stabilize the damaged segment and significantly unload critical areas of the construct and adjacent anatomical structures.

Keywords: *burst fractures; thoracolumbar junction; transpedicular fixation; short-segment stabilization; finite element modeling; lateral flexion; intermediate screws*

Introduction

Injuries of the thoracolumbar junction (TLJ) are among the most common spinal traumas. Approximately 50% of all vertebral fractures occur in the Th10–L2 segment [1].

The biomechanical characteristics of the TLJ—specifically, the transition from the relatively rigid thoracic to the more flexible lumbar segment—predispose this region to a higher risk of injury [2]. Such traumas carry considerable clinical significance, as a substantial proportion of patients (20–50%) experience associated neurological deficits [3]. Given the risk of spinal cord injury and subsequent disability, TLJ injuries warrant special attention and timely management.

Among TLJ injuries, burst fractures of the vertebral bodies are of particular concern. A burst fracture constitutes a severe form of spinal trauma characterized by fragmentation of the vertebral body with disruption of the anterior and middle spinal columns and retropulsion of bony fragments into the spinal canal [4]. These fractures typically result from high-energy mechanisms, such as falls from height or motor vehicle accidents, and frequently lead to compression of neural structures [5]. Burst fractures are prone to progressive kyphotic deformity and neurological complications; therefore, surgical treatment is often indicated. This includes spinal stabilization through internal fixation and, when necessary, decompression of the spinal canal (direct or indirect) [6, 7].

Transpedicular fixation is a widely accepted method for surgical stabilization of TLJ fractures [8]. Two principal strategies are employed: long-segment fixation (involving two or more vertebrae above and below the fracture level) and short-segment fixation (involving only one vertebra above and below the fracture) [9]. Long-segment constructs provide superior initial stability and more effectively prevent post-traumatic kyphosis [10]. However, they require immobilization of a greater portion of the spine, resulting in reduced mobility of additional segments and increased surgical invasiveness [11]. Short-segment constructs are less invasive and help preserve motion in a larger portion of the spine, but have historically been associated with a higher risk of fixation failure (such as rod breakage, screw migration, or pullout) and secondary deformity [12]. According to the literature, conventional short-segment fixation of thoracolumbar burst fractures results in loss of kyphotic correction >10° or other stabilization-related complications in approximately half of the cases [13].

To enhance the reliability of short-segment transpedicular fixation systems, various technical methods have been employed to strengthen the construct. In particular, the additional use of transverse connectors increases the rigidity of the system under rotational loads. However, the placement of intermediate screws directly into the body of the injured vertebra is considered a more effective method for reinforcing short fixation [14]. As early as 1987, it was demonstrated that including the injured vertebra in the fixation construct significantly increases its stiffness: resistance to bending and axial loading rises by 84–160% compared with the standard configuration that does not include such screws [15]. Numerous biomechanical studies have confirmed that the presence of an intermediate screw improves construct stability and protects the anterior spinal column by reducing the load on the implants [14]. Clinical observations also indicate the advantages of fixation involving the damaged vertebra. This approach allows for better maintenance of deformity correction and reduces the incidence of stabilization failure during the postoperative period compared to traditional short fixation without intermediate screws [16].

Most biomechanical analyses of transpedicular fixation systems have focused on the behavior of constructs under flexion/extension and axial loading. However, lateral bending (lateroflexion) is an equally important component of implant loading, since lateral forces generate pronounced asymmetric deformations within the construct and stress concentration in certain fixation elements [17]. The limited data regarding fixation system behavior during lateroflexion do not allow for a precise assessment of their strength reserve under such conditions. Therefore, the study of the stress–strain state of fixation elements under lateral spinal bending remains a relevant and important research direction.

The authors conducted a comparative analysis of short-segment transpedicular fixation variants for burst fractures in the TLJ, using intermediate screws inserted into the body of the injured vertebra and without them. In addition, the length of the main screws inserted into the adjacent intact vertebrae (monocortical versus bicortical fixation) was also taken into account.

The study examined only one model of burst fracture with short-segment transpedicular fixation without reconstruction of the anterior supporting column. This approach was chosen because other methods of surgical stabilization—such as corpectomy with a telescopic cage or hybrid techniques—had been previously analyzed and partially published by the authors. The presented findings constitute a component of a large, multi-stage study devoted to the biomechanical evaluation of various stabilization strategies for traumatic injuries in the TLJ region.

Objective: to analyze the stress–strain state of various options for transpedicular fixation in a T12 burst fracture under conditions of lateroflexion (lateral bending).

Materials and methods

Model of the spine and fixation options

A finite element model of the thoracolumbar spine segment (Th9–L5) with a burst fracture of the T12 vertebra was developed. The T12 vertebral body was modeled with a destructive defect to represent the presence of bone fragments and structural damage. For this purpose, a region of reduced stiffness was incorporated into the model to simulate the interfragmentary regenerate. Transpedicular fixation was used to stabilize the injured segment. Four fixation configurations were considered:

- **model 1 (M1)**: short (monocortical) screws inserted into the Th11 and L1 vertebral bodies without intermediate screws in the fractured T12 body (standard four-screw fixation).
- **model 2 (M2):** long (bicortical) screws inserted into the Th11 and L1 vertebral bodies without intermediate screws in the T12 body.
- **model 3 (M3):** short (monocortical) screws inserted into the Th11 and L1 vertebral bodies with intermediate screws placed in the T12 body (six-screw fixation).
- **model 4 (M4):** long (bicortical) screws inserted into the Th11 and L1 vertebral bodies with intermediate screws in the T12 body.

Short screws were confined within the vertebral body, whereas long screws penetrated the anterior cortical wall to enhance fixation rigidity (bicortical placement).

It is well established that the use of transverse connectors positively influences the load distribution within the fixation system by reducing stress

This article contains some figures that are displayed in color online but in black and white in the print edition.

concentration and increasing overall construct stiffness. Their application is justified in cases where biomechanical studies or clinical observations indicate a risk of reaching the mechanical strength limit of certain components of the stabilization system — both metallic and bone structures.

However, in the framework of this study, a fixation configuration corresponding to a minimally invasive (percutaneous) surgical procedure was simulated. In such interventions, the placement of transverse connectors is technically infeasible. Therefore, the effect of transverse connectors was not evaluated.

Materials and their properties

All biological tissues and implant elements in the model were considered homogeneous and isotropic. The mechanical characteristics of the materials — Young's modulus (E) and Poisson's ratio (ν) — were selected based on literature data and technical documentation [18–20]. The mechanical properties of the materials used in the simulation are summarized in ***Table 1.***

Finite element network and software

The geometric model of the spine was constructed using the computer-aided design (CAD) system SolidWorks (Dassault Systèmes). For the strength analysis, the CosmosM software package was employed, which implements the finite element method (FEM) [21]. The discretization of the model was performed using solid tetrahedral elements with 10 nodes (quadratic displacement field approximation). This level of mesh refinement ensures more accurate calculations of the stress–strain state of both the spinal structure and the implants [22].

Loading and boundary conditions

The model was loaded according to a scheme simulating a lateral bending of the torso. To achieve this, a bending moment was applied to the upper part of the model (the region of the Th9 vertebral body and its corresponding articular processes) through a lateral force of 350 N. This value approximately corresponds to the weight of the upper half of the human torso, creating a physiological load on the thoracolumbar region of the spine during lateral bending. The lower base of the model—specifically, the caudal (inferior) surface of the L5 intervertebral disc—was rigidly fixed (immovable support condition) to reproduce the influence of pelvic support. Thus, the simulation reproduced realistic boundary conditions: the upper vertebrae were subjected to gravitational loading during flexion, while the lower end of the model remained stationary [23].

Evaluation of the stress–strain state

The analysis of stresses and deformations in the models was performed using the von Mises equivalent stress criterion [24,25]. This approach enables the evaluation of maximum stress intensity in both bone structures and fixation elements for each stabilization method. To compare the effectiveness of the constructs, the stress levels were determined in 18 control points within key areas of the model — including the bodies of vertebrae adjacent to the fracture and the elements of the metal fixation system ***(Fig. 1)***:

- ***vertebral bodies*** Th9, Th10, Th11, Th12, L1, L2, L3, L4, L5 (control points 1–9, respectively);
- **endplates of the vertebrae adjacent to the damaged one**: the inferior endplate of Th11 (point 10) and the superior endplate of L1 (point 11);
- **entry zones of the screws** into the vertebral arches Th11 (point 12), Th12 (point 13), and L1 (point 14);
- **screws** within the vertebral bodies Th11 (point 15), Th12 (point 16), and L1 (point 17) (for models without intermediate fixation, points 13 and 16 are absent, as no screws were inserted into Th12);
- **connecting rods** (fixation system bars) (point 18).

This approach made it possible to identify stress concentration zones for each fixation configuration and to compare them, which is important for the biomechanical assessment of stabilization reliability [26].

A separate modeling of the unfixed state was not performed, since in the case of a burst fracture without stabilization, the structure loses its mechanical integrity, making an accurate stress calculation impossible.

Table 1. Mechanical characteristics of materials used in the simulation

Material	Young's modulus, MPa	Poisson's ratio
Compact bone tissue	10 000	0,30
Cancellous (spongy) bone tissue	450	0,20
Articular cartilage	10,5	0,49
Intervertebral disc	4,2	0,45
Interfragmentary regenerate	1,0	0,45
Titanium (VT-16 alloy)	110 000	0,30

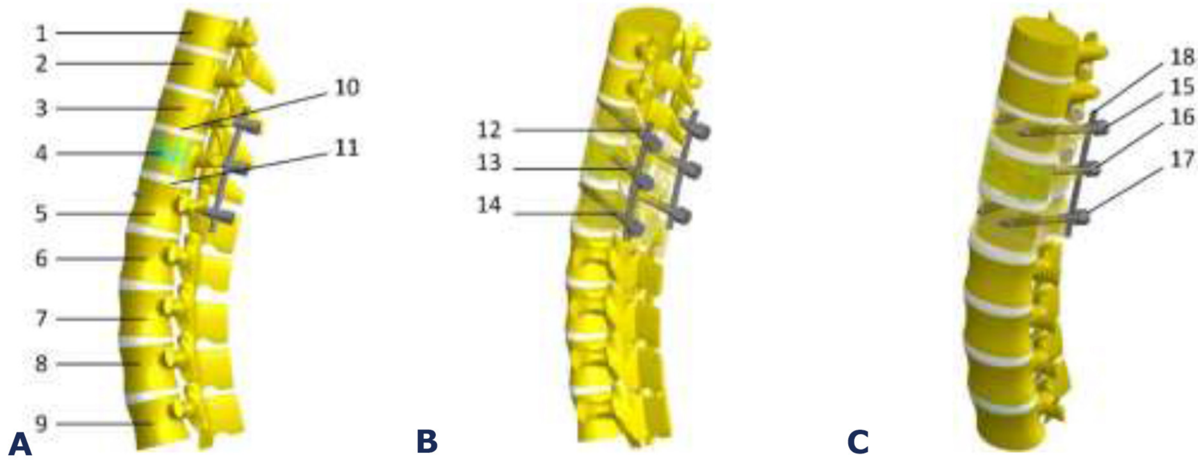

Fig. 1. Schematic arrangement of the control points of the models (description in the text): A – lateral projection; B – posterior-lateral projection; C – anterior-lateral projection

Results

For each control point, we analyzed how the stress level changes when transitioning from one fixation model to another.

Stress in vertebral bodies (control points 1–9)

Upper and lower segments (Th9, Th10, L2–L5). In vertebrae distant from the fixation zone, stresses remain low and nearly identical across all configurations. Specifically, in the vertebral bodies Th9 and Th10, the stress values range from 1.0 to 1.4 MPa and show negligible differences between the models. In the lumbar L2–L5 vertebrae, the stresses range from 12.1 to 17.5 MPa, decreasing when longer and intermediate screws are used (for example, in the body of vertebra L3 — down to 12.1 MPa). This is an expected result, as regions far from the site of injury and implantation are influenced mainly by external loading and depend little on the configuration of the fixation device.

Vertebrae adjacent to the fracture (Th11 and L1). In the vertebrae adjacent to the injured Th12, where screws were placed, a moderate reduction in stress was observed as the fixation was improved. The baseline model (short screws without intermediate ones) shows stress values of 7.9 MPa in the Th11 vertebral body and 18.9 MPa in the L1 body. The use of longer screws (without intermediate ones) reduces these stresses to 7.8 MPa (Th11) and 18.0 MPa (L1). Adding intermediate screws into the Th12 vertebral body (short screws with intermediates) further unloads the Th11 vertebra (down to 7.0 MPa), but slightly increases the stress in L1 (up to 19.0 MPa). The lowest stress levels in adjacent vertebrae were recorded when combining long screws with intermediate ones: 6.8 MPa in Th11 and 17.1 MPa in L1.

Overall, the use of longer screws and intermediate fixation contributes to a reduction of stress in the bodies of the adjacent vertebrae. The effect of intermediate screws is more pronounced for Th11 (~ 14% reduction) than for L1. When both factors—long and intermediate screws—are combined, a cumulative effect occurs, providing the lowest stress levels in Th11 and L1.

Injured Th12 vertebra. The affected Th12 vertebral body exhibited the most pronounced differences among all bony structures across the tested models. Without intermediate fixation, the Th12 vertebral body experienced maximum stresses of 12.3 MPa (short screws) and 12.1 MPa (long screws). The use of intermediate screws significantly reduced the load on the vertebra: stresses decreased to 9.1 MPa (short screws with intermediates) and 8.8 MPa (long screws with intermediates). Thus, intermediate fixation of the Th12 vertebra reduces stress by 27–28%. The screw length alone, without intermediate fixation, had a relatively minor effect (a decrease of only 1–2%), but when combined with intermediate screws, the maximum reduction reached 28%. Summary data on stress distribution in the bodies of the main vertebrae for all fixation models are presented in ***Table 2.***

A visual representation of the stress distribution on the model elements, depending on the type of fixation system, is shown in ***Fig. 2.***

Stress on the endplates (points 10–11)

Changes in stress within the endplates of the Th11 (caudal) and L1 (cranial) vertebrae exhibit smaller amplitudes and less consistent trends, as the cartilaginous tissue of the intervertebral discs partially compensates for differences between fixation schemes due to a similar load transfer mechanism.

The inferior endplate of the Th11 vertebra (point 10) without intermediate fixation demonstrates stresses of 4.3 MPa (short screws, M1) and 4.2 MPa (long screws, M2). The presence of screws in the Th12 vertebral body leads to a distinct reduction in load on the inferior endplate of Th11 — to 3.8 MPa (short screws with intermediate fixation, M3) and to a minimum of 3.2 MPa (long screws with intermediate fixation, M4). This difference indicates effective load reduction in this area due to stabilization of the injured Th12 segment with intermediate screws. Compared with the baseline model (M1), the maximal load decrease amounts to approximately 25.6% (from 4.3 to 3.2 MPa).

Table 2. Equivalent stress (MPa) in the vertebral bodies of T11, Th12, and L1 for four fixation models under lateral flexion

Control point	Short screws without intermediates	Long screws without intermediates	Short screws with intermediates	Long screws with intermediates
Th11 vertebral body	7,9	7,8	7,0	6,8
Th12 vertebral body	12,3	12,1	9,1	8,8
L1 vertebral body	18,9	18,0	19,0	17,1

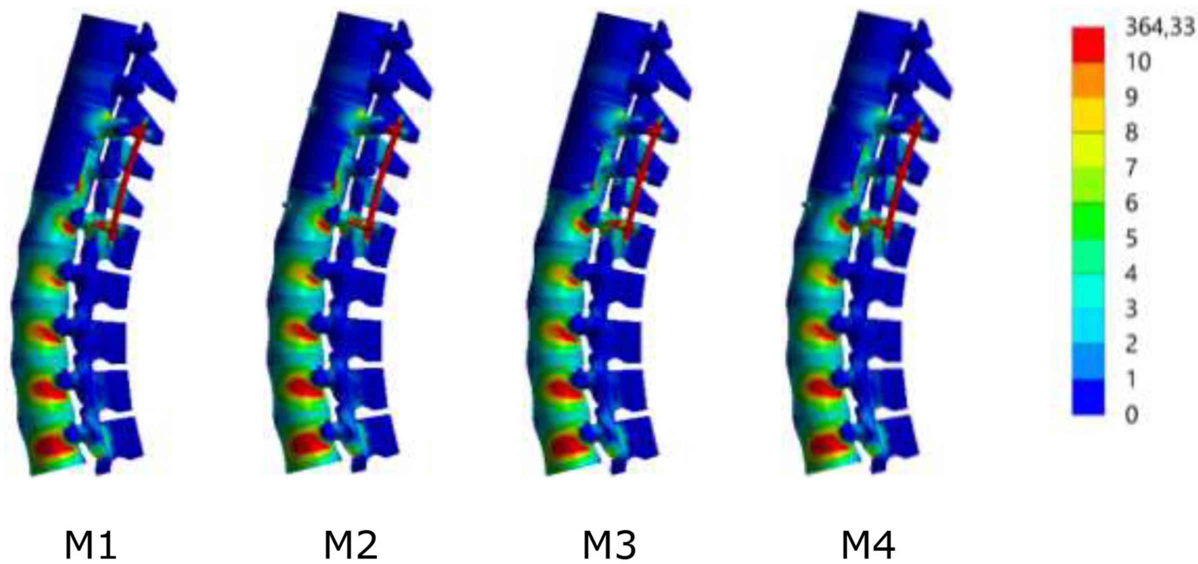

Fig. 2. Stress distribution in models (M1–M4) of the thoracolumbar spine with a burst fracture of the Th12 vertebral body during lateral flexion

The superior endplate of the L1 vertebra (point 11) shows a slightly smaller amplitude of stress variation, though the downward trend is clearly evident. The highest load was recorded in the model without intermediate screws and with short screws (M1) — 8.2 MPa; slightly lower in the model with long screws without intermediate fixation (M2) — 8.1 MPa. The addition of intermediate screws significantly reduced the load on the superior endplate of L1: in the model with short and intermediate screws (M3), the load decreased to 7.0 MPa. In the configuration with long and intermediate screws (M4), the lowest value was recorded — 6.7 MPa, corresponding to an 18.3% reduction compared with the baseline configuration (M1). Thus, the use of intermediate screws substantially decreases the load on the superior endplate of the L1 vertebra, improving structural stability and reducing the risk of cartilage tissue damage.

Overall, endplate stress varied by 1.1–1.5 MPa depending on the model. The most effective load reduction was observed in the model with long screws combined with intermediate fixation.

Stress distribution in screw entry zones (points 12–14)

Arch of the Th11 vertebra (point 12). At the screw entry site in the Th11 vertebral arch (the point of load transfer from the screw to the bone), a gradual decrease in stress was observed when transitioning from short screws without intermediate fixation (7.5 MPa, M1) to long screws without intermediates (7.3 MPa, M2). A more pronounced reduction in stress was recorded with the use of intermediate screws: for short screws with intermediate fixation (M3), the stress decreased to 5.8 MPa (a 22.7% reduction compared to M1). The model with long screws and intermediate fixation (M4) demonstrated a slight increase in stress to 6.1 MPa; however, it remained substantially lower (by 18.7%) than in the baseline configuration. Thus, the presence of intermediate screws provides a significant unloading effect on the Th11 vertebral arch, while the screw length moderately affects the stress distribution.

Arch of the Th12 vertebra (point 13). This area was analyzed only in models with intermediate screws (M3, M4). The stress at the entry point of the intermediate screws into the damaged Th12 vertebral arch remained low: 3.7 MPa for short screws (M3) and slightly lower (3.5 MPa) for long screws (M4). These findings indicate that additional intermediate screws do not generate significant local overstress in the fractured segment, confirming their safety and effectiveness for stabilizing the injured vertebra.

Arch of the L1 vertebra (point 14). The stress in the screw entry zone of the L1 vertebral arch showed a downward trend when moving from short screws without intermediates (14.9 MPa, M1) to long screws without intermediates (13.2 MPa, M2). The inclusion of intermediate screws in the Th12 vertebral body further reduced the stress: in the model with short and intermediate screws (M3), it reached 13.4 MPa, close to the M2 value. The lowest stress level was recorded in the configuration with long and intermediate screws (M4) — 10.7 MPa, which is 28.2% lower than in the baseline model (M1). This demonstrates that the combination of long and intermediate screws provides the most effective unloading of the screw entry zone in the inferior adjacent vertebra, which is crucial for preventing bone tissue damage.

Stress in screws (control points 15–17)

Screws in the Th11 vertebral body (point 15). In the model with short screws and without intermediates (baseline model), the stress value was 47.5 MPa. The use of long screws without intermediate fixation resulted in a slight reduction in stress (to 46.9 MPa), likely due to improved screw anchorage in the bone. The addition of intermediate screws (short screws with intermediates) further reduced the stress in the Th11 vertebral body screws to 39.2 MPa, while the combination of long and intermediate screws achieved the minimum value of 29.3 MPa — 38% lower than in the baseline model. Therefore, the use of intermediate screws ensures the most substantial reduction of stress in the screws of the Th11 vertebral body, particularly when combined with long screws.

Screws in the Th12 vertebral body (point 16). Data for this point were analyzed only for models that included intermediate screws. The stress in the intermediate screws was the lowest among all screws in the construct: 12.6 MPa (short screws with intermediate fixation) and 10.3 MPa (long screws with intermediate fixation). The low stress values can be explained by the fact that these screws primarily serve a stabilizing function for the injured vertebral fragments, bearing a smaller portion of the overall torque. The length of the main screws in the adjacent vertebrae had only a minor effect on the load of the intermediate screws in the Th12 vertebral body, although longer screws demonstrated slightly better (by about 18%) unloading of these intermediate structural elements.

Screws in the L1 vertebral body (point 17). The screws in the lower adjacent vertebra (L1) also exhibited considerable variation in loading depending on the fixation configuration. In the baseline model (short screws without intermediate fixation), the stress reached a maximum of 50.1 MPa. The use of long screws without intermediate fixation reduced the stress to 45.3 MPa (a decrease of approximately 10%). The addition of intermediate screws further reduced the stress on the screws in the L1 vertebral body—to 48.6 MPa (short screws with intermediate fixation) and a minimum of 39.9 MPa (long screws with intermediate fixation), which is 20% lower than in the baseline configuration. Thus, as in the case of the Th11 vertebral screws, the lowest stress levels were recorded for the combination of long screws with intermediate fixation.

Baseline values of equivalent stress in the main structural elements and in the "metal–bone tissue" contact zones are presented in ***Table 3.***

Stress in the connecting rods (point 18)

The connecting rods (longitudinal bars of the fixation system) experience the highest stress among all structural elements, as they effectively bear a significant portion of the load by bypassing the damaged spinal segment. The durability of the entire construct depends on how efficiently the rods are unloaded since exceeding the fatigue limit may lead to metal fatigue fractures.

Without intermediate fixation (only 4 screws): The maximum stress (σ) in the rods during lateral flexion reached 324.9 MPa in the model with short screws (M1). This was the highest recorded stress among all analyzed configurations, typically concentrated between the screws where the bending moment is maximal. Using longer screws (M2) slightly reduced the stress peak to 315.5 MPa, a decrease of about 2.9% compared to M1. Thus, in the absence of an intermediate support point, screw length has only a minor influence on rod stress.

With intermediate screws into the Th12 vertebral body, the changes were modest. Adding an intermediate support in the short-screw model (M3) did not significantly alter rod stress, yielding a maximum of 324.8 MPa, practically identical to the baseline model M1 (difference <0.1%). The configuration with long screws and an intermediate support (M4) reduced the maximum stress to 321.0 MPa, or 1.2% lower than the baseline (M1). The difference between models M3 and M4 was minimal (≈1.2%), indicating that in this configuration, the main contribution to rod unloading is provided by the screw length rather than the presence of an intermediate support.

Table 3. Equivalent stress (MPa) in the main structural elements and "metal–bone tissue" contact zones (screw entry points, screws, rods) for the four models

Structural element	Short screws without intermediates	Long screws without intermediates	Short screws with intermediates	Long screws with intermediates
Screw entry in Th11 (arch)	7,5	7,3	5,8	6,1
Screw entry in Th12 (arch)	–	–	3,7	3,5
Screw entry in L1 (arch)	14,9	13,2	13,4	10,7
Screws in Th11	47,5	46,9	39,2	29,3
Screws in Th12	–	–	12,6	10,3
Screws in L1	50,1	45,3	48,6	39,9
Connecting rods	324,9	315,5	324,8	321,0

From the standpoint of construct longevity, the most reliable configurations were those with long screws—regardless of intermediate screw presence—since the maximum rod stresses were lowest (315.5–321.0 MPa). Given the typical yield strength range for Ti-6Al-4V titanium alloy implants (760–800 MPa), the observed stress levels (315.5–324.9 MPa) account for only about 40% of this threshold [27]. This aligns with international standards (ASTM F136, ISO 5832-3) and literature data on the mechanical properties of this material, confirming a substantial safety margin. Even relatively small differences in absolute stress values can positively affect long-term structural integrity and fatigue resistance, enhancing overall reliability during prolonged clinical use.

An alternative rod material is the cobalt–chromium (CoCr) alloy, characterized by greater stiffness compared to titanium. Its use may be appropriate when maximum construct rigidity is required—particularly in cases of severe anterior column destruction. However, the higher rigidity of CoCr rods is potentially associated with increased contact stresses at fixation sites and a risk of stress concentration within the bone tissue, which may have clinical implications.

In the context of the present model, Ti-6Al-4V was selected as a more "compliant" and physiologically compatible material to simulate a standard surgical stabilization scenario in the TLJ.

In summary, within the analyzed models, intermediate screws did not contribute significantly to rod unloading (stress reduction <1%), unlike their substantial positive effect on other control points (injured Th12 vertebral body, endplates, and screw entry zones in Th11 and L1). This indicates that the primary role of intermediate screws lies in local stabilization of the injured segment and redistribution of local loads, whereas the global load on the rods is predominantly influenced by the main screw length. Therefore, the combined use of long and intermediate screws provides the most optimal balance of stress distribution across all structural control points.

Discussion

Effectiveness of intermediate screws in a fractured vertebra

The addition of intermediate screws into the body of the fractured vertebra significantly increases the rigidity and stability of short-segment fixation. Biomechanical studies have demonstrated that six-screw constructs (one level above and one level below the fracture, plus screws inserted into the fractured vertebra) markedly limit the mobility of the injured segment compared to the standard four-screw configuration [28]. In particular, modeling studies have shown a reduction in flexion and lateral bending angles with the inclusion of "index-level" screws in short constructs, resulting in approximately a 25% increase in stiffness [29]. In contrast, the difference for long-segment (two-level above/below fracture) osteosynthesis was minimal, indicating that the addition of intermediate screws is most effective in short-segment systems. Therefore, incorporating the fracture level into fixation substantially reinforces the construct and reduces the risk of angular correction loss even under flexion loads.

The improvement in fixation achieved through intermediate screws has also been clinically confirmed. A prospective study involving 80 patients found that short-segment fixation including the fractured vertebra provided better correction and maintenance of kyphotic deformity, whereas "skipping" the fractured level resulted in progression of kyphosis (a mean difference of 29% compared to constructs including the injured level) [14]. The incidence of implant breakage was higher in the group without screws placed in the fractured vertebral body. A recent meta-analysis further confirmed that the addition of screws into the fractured vertebra enhances biomechanical stability, facilitates better height restoration, reduces loss of correction, and decreases the rate of implant failure and fragmentation [30]. Consequently, many authors recommend supplementing short-segment transpedicular fixation with screws inserted into the fractured vertebra in cases of unstable wedge or burst fractures. This approach reduces the likelihood of recurrent vertebral collapse and hardware failure [28]. Although intermediate screws significantly strengthen short constructs, in cases of highly unstable injuries (AO type C fracture-dislocations), even enhanced short fixation may still provide inferior rigidity compared to long-segment constructs [13]. Therefore, for multi-column fractures with pronounced instability, the use of longer fixation is advisable, while for most burst fractures in the TLJ region, short-segment fixation reinforced with screws at the fracture level remains an effective and biomechanically justified approach.

Effect of screw length

The length of pedicle screws, particularly the use of bicortical (long) screws with perforation of the anterior cortical plate, influences both the fixation strength and the distribution pattern of mechanical loads. Bicortical anchorage is traditionally considered stronger due to the engagement of the opposite vertebral wall. Experimental studies have demonstrated that the pullout force required for screw extraction may be ~ 40% higher with bicortical fixation compared to shorter screws [31]. This effect is especially relevant in cases of osteoporosis, where longer screws significantly improve implant retention within the bone [32]. A "stiffer" anchor, represented by a long screw, alters the stress distribution within the construct. Finite element modeling of lateral flexion has shown that short (monocortical) screws generate lower peak stresses in the fixation system compared to long bicortical screws — at all control points, the applied load was slightly higher when longer screws were used. Therefore, a longer screw, by more firmly "linking" the rod to the vertebra, bears a greater share of the bending moment, which may lead to higher localized stress around its entry point. In contrast, shorter screws exhibit a degree of compliance, redistributing part of the load through the connecting rods. Importantly, although the difference in stress distribution is not critical, it does exist and may become significant under extreme loading conditions.

The use of bicortical screws, however, is associated with an increased risk of injury to adjacent anatomical structures (vessels, organs) in the event of perforation of the anterior cortical wall. Therefore, when selecting screw length, the surgeon must consider the patient's individual anatomy and the necessary balance between fixation stability and procedural safety [33].

Changes in stress distribution in rods, screws, and adjacent vertebrae

The analysis of the stress–strain state of the construct under lateral flexion revealed a characteristic pattern of load distribution among its elements. Maximum stresses typically occurred in the terminal (especially caudal) screws, as they bear the greatest portion of the bending load. For instance, in a short-segment fixation model of an L1 vertebral fracture, the highest equivalent stress was concentrated in the lower screws of the construct [34]. Similar findings were obtained in our study. The addition of intermediate screws altered the pattern: the load was partially redistributed from the terminal screws to the fracture level and the rods. According to modeling data, the inclusion of screws in the fractured vertebra reduced stress in other screws of the construct. Thus, in a six-screw model, lower equivalent stresses were recorded in the transpedicular screws compared with a four-screw system [28]. At the same time, the stress in the rods slightly increased, as the stiffer "coupling" of the segment required the rod to absorb more of the bending moment. Hongwei Wang et al. reported that the stress in the connecting rods with intermediate screws reached a level similar to that observed in long-segment fixation, whereas without intermediate screws, the load on the rod was slightly lower [35]. The reduction of stress in screws when intermediate screws are used is a positive factor, since overstressing of these elements and their surrounding bone often leads to fatigue failure or implant loosening.

An important aspect is the load distribution in the body of the injured and adjacent vertebrae depending on the fixation configuration. If the affected vertebra is not included in the construct (i.e., screws are inserted only into the neighboring vertebrae), a portion of the bending moment is still transmitted through its body despite its minimal contribution to load transfer. In cases where intermediate screws are used, the load on the damaged vertebra is further reduced, resulting in moderate stress formation [34]. This not only decreases the load on the injured segment and thereby the likelihood of secondary deformation but also stimulates bone remodeling in accordance with Wolff's law, promoting better fusion and healing [36].

Regarding adjacent, unaffected segments, studies have shown that overly rigid constructs can increase the load on adjacent intervertebral discs. Specifically, the maximum intradiscal pressure in the upper adjacent segment is usually higher than in the lower one [37], which may indicate a potential risk of overloading the superior neighboring disc. Therefore, shorter fixation with intermediate screws, which preserves the mobility of more motion segments, potentially exerts a lesser impact on adjacent levels compared with excessively long and rigid constructs. When selecting the optimal fixation configuration, it is crucial to maintain this balance—ensuring sufficient rigidity for effective stabilization of the fracture while avoiding excessive load "shielding," which could negatively affect both the implants and the surrounding anatomical structures.

Risks of structural overload during lateral flexion

Lateral flexion of the spine creates asymmetric loading on the implanted system: one rod is subjected to compression, the opposite one to tension, while the pedicle screws on the side of inclination experience significant bending and torsional moments. This loading mode is critical for the construct, as it concentrates stress on individual components. If the fixation strength or configuration is insufficient, lateral flexion may lead to local overload, resulting in fatigue cracking of the rod or gradual loosening of screws in the bone. Historically, short four-screw constructs have often been associated with implant failures (such as rod fracture, bending, or screw pullout) under loading, particularly in the lateral projection. The addition of intermediate screws significantly reduces the risk of such complications by redistributing the load: as mentioned above, the stresses on the most loaded screws decrease, while the overall stiffness of the construct increases, thereby limiting excessive displacement and cyclic loading [28]. Clinical series have demonstrated that the use of six-screw constructs substantially lowers the incidence of metalwork failure, although it does not completely eliminate it. For example, Liao et al. reported approximately 11% of implant failure cases even when fracture-level screws were used (3 out of 27 patients) [38, 39]. In another long-term follow-up study of short fixation with intermediate screws, the rate of implant breakage or loosening reached 16.7% [30]. These data suggest that overload may still occur even in reinforced systems, particularly if the patient resumes significant physical activity early or has osteoporotic changes.

Significant loading during lateral flexion occurs on the rod opposite the direction of inclination, which is subjected to tensile forces. In this context, the role of transverse connectors between the rods is crucial. The addition of at least one cross-link enhances the spatial rigidity of the construct and synchronizes the performance of the left and right rods, preventing their excessive divergent bending. Modeling studies have shown that the presence of a transverse connector reduces stress levels in screws and rods under lateral loading. Specifically, the combination of short screws with a transverse link demonstrated the lowest critical stress among tested configurations, making it biomechanically optimal under lateral flexion [17]. Thus, to reduce the risk of construct overload, it is advisable to use intermediate screws and, when possible, equip the system with transverse connectors. These measures improve load distribution and increase the mechanical strength reserve of the implants under lateral loading. This approach, however, involves a certain trade-off between biomechanical efficiency and clinical feasibility, as such procedures are often performed using minimally invasive techniques. The installation of transverse connectors requires an open surgical stage, which may potentially offset the benefits of a minimally invasive approach. Therefore, when selecting the optimal surgical strategy, it is essential to balance the required construct stability with the degree of invasiveness, considering both clinical conditions and patient needs.

Comparison of short and long fixation systems: practical insights

Biomechanical and clinical evidence indicates that a properly configured short-segment transpedicular fixation can provide effective stabilization of burst fractures in the TLJ while preserving more motion segments. A short construct (four screws) with the addition of screws into the fractured vertebra achieves stiffness comparable to a long-segment system and significantly exceeds that of a standard short construct without intermediate fixation points [29]. This configuration demonstrates a favorable stress distribution — minimal critical stress in the screws among fixation options, acceptable load sharing in the rods, and inclusion of the fractured vertebral body in bearing the load, which potentially enhances osteogenesis [34]. Clinically, this results in better preservation of vertebral height and less progression of deformity without increasing complication rates [14]. Thus, reinforced short-segment fixation allows limiting the number of instrumented levels, which is especially important in younger patients to maintain spinal mobility and prevent unnecessary disability of adjacent segments.

Long-segment systems (extending at least two levels above and below the fracture) provide maximum stability due to multiple fixation points and wide load distribution. They remain the method of choice in cases of extreme fracture instability, significant osteoporotic bone involvement, or multilevel injuries, where short fixation—even with intermediate screws—may not ensure adequate correction retention. However, this stability comes at the cost of losing motion in additional spinal segments and potentially increasing stress on uninvolved adjacent levels.

Comparative analyses of treatment outcomes reveal that, in the absence of indications for extensive stabilization, short constructs that include the fractured vertebra are not inferior to long systems in clinical results, while offering the advantage of reduced surgical invasiveness and better preservation of the spine's anatomic and functional integrity.

Therefore, according to contemporary approaches to managing burst fractures in the TLJ, preference should be given to the most conservative yet sufficiently stable fixation methods. Short-segment transpedicular systems with intermediate screws represent a balanced strategy that ensures adequate stability under lateral bending and other load conditions, while minimizing the risks of implant overloading and adjacent-segment degeneration. This approach is supported by both biomechanical research (including numerical modeling and cadaveric testing) and clinical observations with adequate postoperative follow-up.

For most type A (burst) injuries in the Th10–L2 region, short-segment transpedicular fixation involving the fractured vertebra provides the necessary stability with a smaller fixation span. Only in cases of extreme instability or insufficient bone quality should long-segment constructs be preferred to protect against mechanical overload under all conditions.

Conclusions

Based on the analysis of stresses at 18 control points across four models, the following conclusions can be drawn:

1. **Intermediate fixation of the fractured vertebra (Th12)** plays a decisive role in load redistribution during lateral flexion. The addition of two intermediate screws into the Th12 vertebral body significantly reduces stresses in the critical elements of both the fixation construct and the spine itself.

2. **Long transpedicular screws** (inserted deeper into the adjacent vertebral bodies) generally improve the biomechanical performance of the fixation system, although their effect is less pronounced than that of the intermediate screws.

3. **The combined use of long screws and intermediate fixation (M4)** provides the most optimal stress distribution. This model demonstrated the lowest stress values in nearly all control points. Therefore, M4 can be considered the most balanced option for short-segment spinal fixation in Th12 vertebral injuries, as it combines the advantages of both approaches: intermediate support at the fracture level substantially unloads the construct, while long screws ensure the safe realization of this support through uniform stress distribution.

4. **Practical implications.** The analysis indicates that, in the treatment of burst fractures in the TLJ, preference should be given to short-segment fixation involving the fractured vertebra (use of intermediate screws). This approach provides mechanical stability comparable to traditional long-segment fixation (two levels above and below the fracture) while requiring fewer implants. The length of the transpedicular screws should be maximized within the patient's anatomical limits, particularly when using intermediate screws, to prevent local overstressing. Both factors contribute to reducing loads on implants and bone structures, potentially improving clinical outcomes—such as a lower risk of implant failure, better conditions for fracture healing, and preservation of spinal segment mobility.

Disclosure

Conflict of interest

The authors declare no conflict of interest.

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