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# Biomechanical analysis of short-segment pedicle screw fixation for AO type A fractures at the thoracolumbar junction under extension loading

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*Thoracolumbar junction fractures account for up to 60 % of all spinal injuries, with more than 20 % classified as burst fractures. These injuries are characterized by instability and an increased risk of kyphotic deformity. Short-segment transpedicular fixation, which has been gradually gaining favor as the treatment of choice for burst fractures in this region, undoubtedly reduces invasiveness but can lead to fixation failure if there is no intermediate support in the body of the injured vertebra. Objective. To evaluate the stress-strain state of the thoracolumbar spine with a burst fracture at $Th_{XII}$ under extension loading, considering screw length and the presence of intermediate fixation. Methods. A finite element model of the spine ($Th_{IX}$–$L_V$) with a burst fracture at $Th_{XII}$ was developed. Four variants of short-segment transpedicular fixation were analyzed: short or long (bicortical) screws, with or without additional screws in $Th_{XII}$. A 350 N load was applied to $Th_{IX}$, and stresses were determined at control points using the von Mises criterion. Results. The highest stresses in the connecting rods were observed with long screws without intermediate fixation (337.2 MPa). Introducing intermediate screws decreased the stress in $Th_{XII}$ from 16.2 to 9.8 MPa. Short screws with $Th_{XII}$ fixation distributed loads more effectively, reducing peak stresses. Long screws led to overload at the entry points (up to 12.8 MPa in $L_I$), while their maximum stress reached 95.1 MPa. Conclusions. Intermediate fixation of the injured vertebra reduces stress in both bone structures and the implant, thereby decreasing the risk of correction loss. The most favorable configuration is a combination of short screws with intermediate fixation. Long screws are advisable in cases of osteoporosis, although they increase local stresses — a factor that must be taken into consideration during surgical planning.*

*Переломи грудопоперекового переходу становлять до 60 % усіх ушкоджень хребта, з них понад 20 % є вибуховими. Такі травми характеризуються нестабільністю та ризиком кіфотичної деформації. Коротка транспедикулярна фіксація, що поступово набуває популярності як метод вибору в разі вибухових переломів цієї ділянки, безумовно зменшує інвазивність, проте за відсутності проміжної опори в тілі ушкодженого хребця може призводити до неспроможності фіксації. Мета. Оцінити напружено-деформований стан грудопоперекового відділу хребта з вибуховим переломом $Th_{XII}$ під час екстензійного навантаження залежно від довжини гвинтів і наявності проміжної фіксації. Методи. Створено скінченно-елементну модель хребта ($T_{IX}$–$L_V$) із вибуховим переломом $Th_{XII}$. Досліджено чотири варіанти короткої транспедикулярної фіксації: короткі та довгі (бікіркові) гвинти, із додатковими гвинтами в $Th_{XII}$ чи без них. Навантаження 350 Н прикладали до $T_{IX}$, а напруження визначали в контрольних точках за Мізесом. Результати. Найвищі напруження в з'єднувальних балках спостерігалися за довгих гвинтів без проміжної фіксації (337,2 МПа). Використання проміжних гвинтів знижувало напруження в $Th_{XII}$ з 16,2 до 9,8 МПа. Короткі гвинти з фіксацією $Th_{XII}$ більш ефективно розподіляли навантаження, зменшуючи пікові напруження. Довгі гвинти зумовлювали перевантаження в точках входу (до 12,8 МПа в $L_I$), а їхні максимальні напруження сягали 95,1 МПа. Висновки. Проміжна фіксація ушкодженого хребця знижує напруження в кісткових структурах і металоконструкції, зменшуючи ризик втрати корекції. Оптимальною є комбінація коротких гвинтів із проміжною фіксацією, а довгі доречні за остеопоротичних змін, бо збільшують локальні напруження, що слід ураховувати під час хірургічного планування. Ключові слова. Вибуховий перелом, грудопоперековий перехід, коротка транспедикулярна фіксація, проміжні гвинти, екстензійне навантаження, метод скінченних елементів, біомеханіка хребта.*

**Keywords.** Burst fracture, thoracolumbar junction, short-segment transpedicular fixation, intermediate screws, extension loading, finite element method, spinal biomechanics

## Introduction

Fractures in the thoracolumbar junction, according to various studies, account for approximately 40–60 % of all spinal injuries [1, 2]. More than 20 % of such injuries are classified as burst fractures, which occur mainly as a result of axial loading and are accompanied by damage to the anterior and middle supporting columns of the spine [3]. Such injuries are characterized by pronounced instability of the affected segment, which can lead to the development of kyphotic deformity and prolapse of bone fragments into the spinal canal, increasing the risk of neurological complications [4]. In significant loss of height of the anterior vertebrae, surgical intervention is aimed not only at stabilizing the affected area, but also at preventing progressive kyphosis, which, in turn, can cause secondary compression of the spinal cord [5]. In addition, with the right choice of surgical treatment tactics and timely stabilization, it is possible to achieve indirect decompression of the spinal canal without the need for open access, which helps to reduce intraoperative trauma, reduce blood loss and the risk of infectious complications, as well as accelerate rehabilitation [6, 7]. Despite significant clinical experience, the technique and extent of stabilization for burst fractures remain a subject of debate [8]. In modern practice, short transpedicular fixation has become widely used, which involves the installation of screws in adjacent vertebrae to the damaged one (one above, one below). The main advantages of this approach are less invasiveness and the preservation of a larger number of functionally active spinal motor segments [9]. The technique has demonstrated efficacy in stabilising less burdened thoracic regions; however, its application at the thoracolumbar junction is influenced by both clinical considerations and economic factors. Reducing spinal fusion length in this area can lead to a high risk of early correction loss — up to 50 % according to some studies — and increases the chances of screw loosening or breakage [10]. Long stabilization, covering two levels above and below the fracture, provides better primary stability, but is accompanied by an increase in the volume of surgical intervention and a decrease in mobility due to the immobilization of additional segments. One way to increase the effectiveness of short fixation without increasing its length is to insert additional transpedicular screws into the body of the damaged vertebra, which ensures the formation of a six-screw structure instead of the traditional four-screw [11]. Biomechanical studies show that this approach contributes to an increase in system stiffness, improved retention of kyphotic correction and more uniform load distribution, which reduces peak stresses on each individual screw [12].

One of the key factors in evaluating the biomechanical properties of stabilization methods and predicting their effectiveness is the accurate modeling of loading modes. Most studies focus on flexion, extension, lateral bending, rotation, and axial compression [13, 14]. While flexion is typically considered the most challenging mode for implants, extension loads have their own unique characteristics [15]. Specifically, during extension, similar to flexion, the primary force is applied to the upper screws, while the lower screws experience lower stresses. In the absence of intermediate fixation within the fracture zone, extension may lead to overloading of the connecting beam. Adding two screws to the body of the damaged vertebra creates an additional fulcrum, which provides a more even load distribution, reduces peak stresses in the screws and beams, and improves long-term stability of the correction [16]. In addition, extension affects adjacent motion segments. According to modeling, intervertebral discs above the level of fusion are often subjected to increased loads, which can cause accelerated degeneration and the development of pathological changes [13]. Therefore, the selection of the optimal design for stabilizing burst fractures of the thoracolumbar junction requires a balance between sufficient stiffness to maintain correction and minimizing overload on both implants and adjacent spinal segments, especially during extension movements.

To date, the number of publications on this topic remains limited, highlighting the need for further biomechanical and clinical studies to develop individualized approaches for selecting surgical treatment strategies for patients with burst fractures in the thoracolumbar region.

*Objective:* To investigate the stress-strain state of a mathematical model of the thoracolumbar spine with a $Th_{XII}$ burst fracture under conditions of various short transpedicular fixation configurations during trunk extension.

## Material and methods

The materials of the article were reviewed by the Ethics Committee of the State Institution "Academician A. P. Romodanov Institute of Neurosurgery of the NAMS of Ukraine" (Protocol No. 3 dated 04.05.2018). In the laboratory of biomechanics of the State Institution "Professor M. I. Sytenko Institute of Spine and Joint Pathology of the NAMS of Ukraine", a finite element model of the thoracolumbar spine with simulation of a burst fracture of the $Th_{XII}$ vertebral body was created. A detailed description and characteristics of the constructed model are given in previous publications [17, 18].

To simulate a burst fracture (type A4 according to the AOSpine thoracolumbar spine injury classification system), the $Th_{XII}$ vertebral body was divided into separate fragments by several planes, and the gaps between them were filled with a material that, according to its biomechanical characteristics, corresponds to interfragmentary regenerate [19]. Four options for transpedicular fixation of two adjacent vertebrae (superior and inferior) were considered using short and long screws that passed through the anterior surface of the vertebral body; the option with and without two additional short screws in the $Th_{XII}$ body was also evaluated.

When cconstructing the model, the materials were assumed to be homogeneous and isotropic. The poroviscoelastic properties of the spinal tissues were not taken into account, since all loads were considered under quasi-static loading conditions. Given the significant individual variability of the mechanical and anatomical parameters of the spine, adapting the model characteristics to a specific patient was considered impractical. The main task of this study was to identify the stress distribution under different loading schemes with the possibility of further extrapolation of the results to the general population. Therefore, the selected method for simplifying the model is appropriate.

As the basic type of finite element (FE), a ten-node tetrahedron with a quadratic approximation was chosen, which ensures high accuracy of the modeling results. The choice of the FE type is based on the results of a number of studies that have shown the superiority of ten-node quadratic tetrahedra over four-node linear elements both in terms of calculation accuracy and calculation time [20]. In addition, it has been proven that the number of FEs over 1,700 per vertebral body guarantees an error of less than 0.5 % [21]. Our model consisted of 35,161 ten-node tetrahedral isoparametric FEs and 92,958 nodes, which exceeds 3,000 elements per vertebral body.

The FE mesh was generated using modern software packages that automatically determine the size, type, and number of elements thanks to adaptive algorithms that take into account the curvature of surfaces, the optimal geometry of the model, and prevent the formation of elements of unsatisfactory shape.

The mechanical characteristics of biological tissues (cortical and cancellous bones, intervertebral discs, and articular cartilage) were determined based on literature sources [22–24]. Metal structures were modeled from titanium alloy VT-16, and the characteristics of artificial materials were selected from technical reference books [25]. The mechanical properties used for modeling (Young's modulus (E) and Poisson's ratio (ν)) are given in Table 1.

The above models were investigated under the action of a load simulating the extension of the trunk backwards (from front to back), which was applied to the body of the $T_{IX}$ vertebra and the articular surfaces of its processes. The load value was chosen at the level of 350 N, which corresponded to the mass of the upper half of the trunk [26]. The caudal plane of the $L_V$ disc was rigidly fixed, as illustrated in Fig. 1.

For the convenience of analyzing the influence of different options for transpedicular fixation on the stress-strain state, the values of stresses at fixed control points were determined in each model (Fig. 1).

In particular, among them were the vertebral bodies ($Th_{IX}$–$L_V$), endplates (lower $Th_{XI}$ and upper $L_I$), screw entry and passage points ($Th_{XI}$, $Th_{XII}$, $L_I$), and support beams.

The analysis was performed by the finite element method, using the Mises stress as a criterion for assessing the stress state [27]. The modeling was performed in the SolidWorks computer-aided design system, and the stress-strain calculations were performed in the CosmosM environment [28].

*Table 1*

**Mechanical characteristics of materials used in modeling**

| Material | Young's modulus, MPa | Poisson's ratio |
|---|---|---|
| Cortical bone | 10000.0 | 0.30 |
| Cancellous bone | 450.0 | 0.20 |
| Articular cartilage | 10.5 | 0.49 |
| Intervertebral disc | 4.2 | 0.45 |
| Interfragmentary regenerate | 1.0 | 0.45 |
| Titanium BT-16 | 110000.0 | 0.30 |

## Results

The modeling of four variants of transpedicular fixation for a $Th_{XII}$ burst fracture showed significant differences in the stress distribution and behavior of the structure.

*Short screws without intermediate fixation — model 1 (Fig. 2)*

This variant has a standard short-segment stabilization, when the screws are installed in the vertebra adjacent to the fracture, but without fixation of the fractured $Th_{XII}$ vertebra itself. In this configuration, the load is transmitted through the beams that connect the adjacent vertebrae ($Th_{XI}$ and $L_{I}$), forming a rather long "span" above the fracture. As a result, the maximum stresses in the metal structure are observed precisely in the supporting beams, reaching 361.3 MPa.

Regarding bone structures, the highest stresses were recorded in the body $L_{IV}$ (20.6 MPa), $L_{III}$ (17.5 MPa) and $L_{I}$ (17.4 MPa). In the fractured vertebra $Th_{XII}$, they were somewhat lower (16.2 MPa), which may be a consequence of insufficient retention of fragments. The lowest stresses among the analyzed vertebral bodies were observed in the thoracic region, specifically in $Th_{XI}$, where the stress measured 10.4 MPa.

Regarding the screws, the maximum stresses were observed in the screws of $Th_{XI}$, reaching 79.5 MPa, with somewhat lower stresses recorded in $L_{I}$ at 75.9 MPa. The highest stresses, particularly at the screw entry zones, were found in the arches of $Th_{XI}$ at 15.4 MPa, while in $L_{I}$, this value was slightly lower at 12.4 MPa.

*Long screws without intermediate fixation — model 2 (Fig. 3)*

The use of long screws in adjacent vertebrae (without fracture fixation) creates a stiffer anchorage in the bone, which leads to stress redistribution. The maximum values in the metal structure are reduced at the support beams to 337.2 MPa, indicating a decrease in the overall load on the system.

In the bone structures, the stress level also decreases: 15.7 MPa in the body of $Th_{XII}$, 16.2 MPa in $L_{III}$, and 17.2 MPa in $L_{I}$. However, in the arches of $L_{I}$, where the long screws penetrate deeper into the vertebra, the stress increases to 12.8 MPa. The most significant increase in stress is observed in the screws of $L_{I}$, reaching 95.1 MPa, which is the highest value among all the models. This suggests a potential risk of implant overload in this region.

*Short screws with intermediate fixation — model 3 (Fig. 4)*

Adding intermediate screws to the body of the affected $Th_{XII}$ vertebra significantly changes the nature of the load distribution. The compressed vertebra begins to be not only a passive structural element, but also actively maintain the stability of the segment. It is characteristic that in this configuration the stress in the body of $Th_{XII}$ decreases to 9.8 MPa, almost half as much as in models without intermediate fixation.

In the metal structures, an increase in stress on the supporting beams is observed, reaching 442.3 MPa, indicating a more active involvement of the fractured vertebra in load transfer. Meanwhile, the stresses in the screws themselves are significantly reduced: in the $Th_{XI}$ screws, the stress drops to 66.0 MPa (compared to 79.5 MPa in the model without intermediate screws), and in the $L_{I}$ screws, it decreases to 64.4 MPa (down from 75.9 MPa). A reduction in stress is also noted in the $Th_{XI}$ arches, with the value falling to 8.8 MPa (from 15.4 MPa in the previous model).

Therefore, the inclusion of the affected level in the fixation allows to reduce peak loads at key points of the system, which has a positive effect on the durability of the structure.

*Long screws with intermediate fixation — model 4 (Fig. 5)*

This configuration is the most rigid, as it combines long screws with intermediate fixation. It exhibits the lowest stress levels in the bone structures among all the options. For instance, the stress in the body of $Th_{XII}$ is 9.8 MPa, in $L_{I}$ it is 14.1 MPa, and in $L_{III}$ it is 13.9 MPa.

At the same time, unlike the model with monocortical screws, in metal structures there is a decrease in the maximum stress on the support beams (436.0 compared to 442.3 MPa), which indicates some reduction in stress in the structure.

However, the stress on the $L_{I}$ screws remains relatively high at 75.7 MPa, while the stress on the screws in $Th_{XII}$ is 13.5 MPa, which is only slightly higher than in the previously considered model. This suggests a more uniform distribution of the load, although the overall stiffness of the structure remains high.

The values of stresses at all control points of the models for different options for transpedicular fixation are summarized in Table 2.

When comparing the results of the analysis across the considered models, it's important to focus on several key points that hold the greatest clinical significance.

A detailed analysis of the stress-strain state of the elements of the biomechanical system, especially in the areas of connection of the screws with the bone tissue of the vertebrae, is key to determining

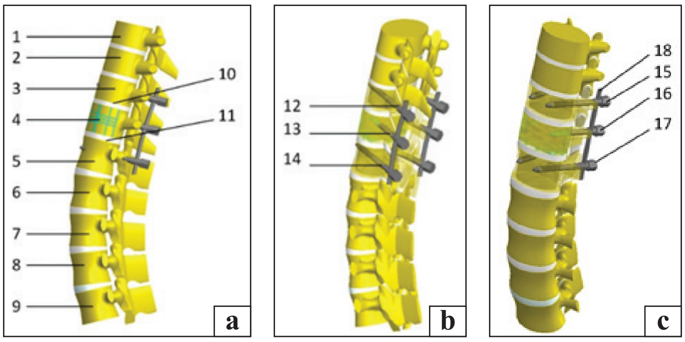


**Fig. 1.** Schematic location of control points: 1 — vertebral body $Th_{IX}$; 2 — vertebral body $Th_{X}$; 3 — vertebral body $Th_{XI}$; 4 — vertebral body $Th_{XII}$; 5 — vertebral body $L_{I}$; 6 — vertebral body $L_{II}$; 7 — vertebral body $L_{III}$; 8 — vertebral body $L_{IV}$; 9 — vertebral body $L_{V}$; 10 — lower endplate of the vertebral body $Th_{XI}$; 11 — upper endplate of the vertebral body $L_{I}$; 12 — entry of screws into the vertebral arch $Th_{XI}$; 13 — entry of screws into the vertebral arch $Th_{XII}$; 14 — entry of screws into the vertebral arch $L_{I}$; 15 — screws in the vertebral body $Th_{XI}$; 16 — screws in the vertebral body $Th_{XII}$; 17 — screws in the vertebral body $L_{I}$; 18 — support beams

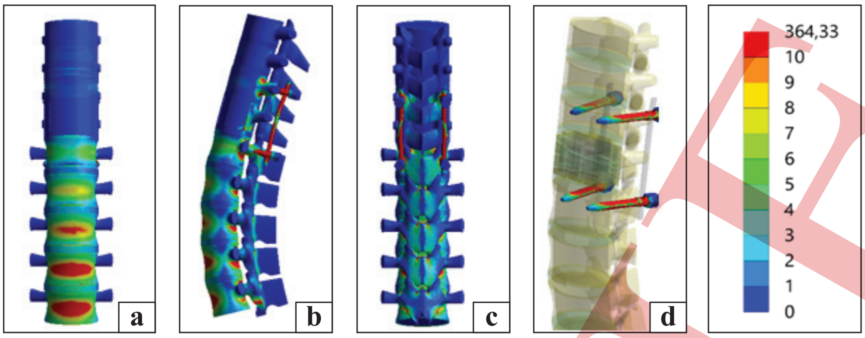


**Fig. 2.** Stress distribution in the model of the thoracolumbar spine with a burst fracture of the $Th_{XII}$ vertebral body during backward bending of the trunk. Transpedicular fixation with short screws without fixation of the $Th_{XII}$ vertebra view: a — front; b — side; c — back; d — screws

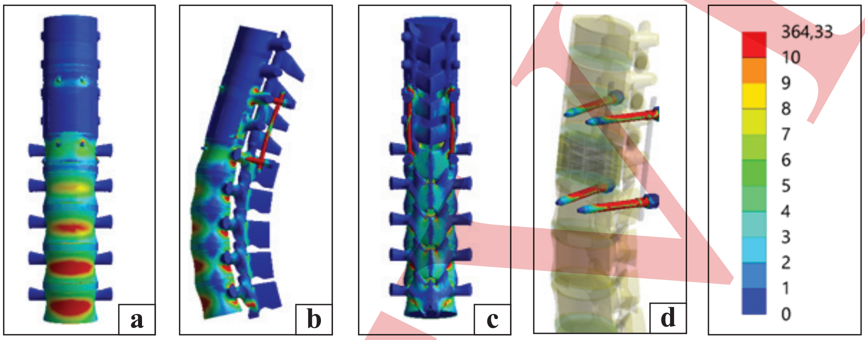


**Fig. 3.** Stress distribution in the model of the thoracolumbar spine with a burst fracture of the $Th_{XII}$ vertebral body during backward bending of the trunk. Transpedicular fixation with long screws without fixation of the $Th_{XII}$ vertebra view: a — front; b — side; c — back; d — screws

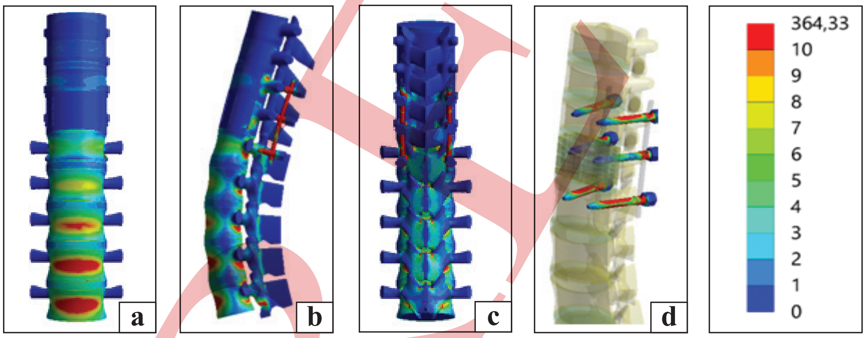


**Fig. 4.** Stress distribution in the model of the thoracolumbar spine with a burst fracture of the $Th_{XII}$ vertebral body during backward bending of the trunk. Transpedicular fixation with short screws with additional fixation of the $Th_{XII}$ vertebra view: a — front; b — side; c — back; d — screws

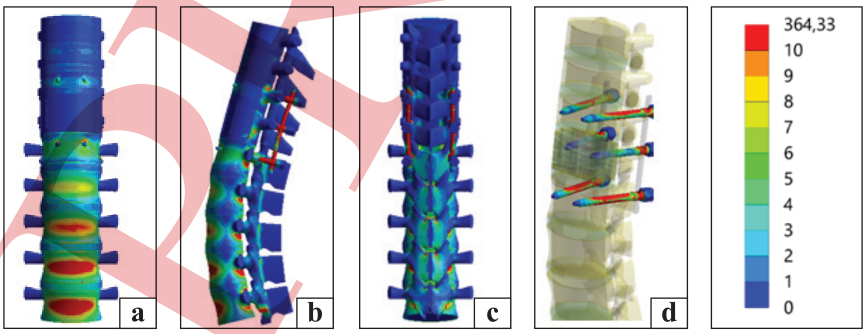


**Fig. 5.** Stress distribution in a model of the thoracolumbar spine with an explosive fracture of the $Th_{XII}$ vertebral body during backward bending of the trunk. Transpedicular fixation with long screws with additional fixation of the $Th_{XII}$ vertebra view: a — front; b — side; c — back; d — screw

the effectiveness of various options for transpedicular fixation in the case of an explosive fracture of the body of the $Th_{XII}$ vertebra. Our study revealed that the highest stress values are concentrated in the places where the screws enter the vertebral arches. In particular, in the model without intermediate screws, the maximum stresses at the places where the screws enter the arch of the $Th_{XI}$ vertebra are 15.4 MPa (short) and 14.5 (long). When intermediate screws are added, these values are significantly reduced to 8.8 and 7.7 MPa, respectively. In the $Th_{XII}$ vertebra, where additional intermediate screws are installed, the stresses at the connection points are 2.5 MPa for both types of screws. In the area of screw entry into the $L_{I}$ vertebral arch, a decrease in stresses was also noted from 12.4–12.8 MPa (model without intermediate screws) to 9.3–10.0 MPa (model with intermediate screws). A detailed analysis of the $L_{I}$–$L_{V}$ vertebral bodies revealed an uneven distribution of stresses with significant differences between individual zones of the vertebral bodies. The highest stresses were recorded in the $L_{IV}$ vertebral bodies (20.6–20.7 without intermediate screws, 18.5–19.1 MPa with intermediate screws) and $L_{III}$ (17.5 without intermediate screws and 13.9–15.1 MPa with intermediate screws). In the $L_{II}$ vertebral body, stresses decrease from 14.7–14.2 MPa in the model without intermediate screws to 12.0–12.5 MPa with intermediate screws. In the $L_{V}$ vertebra, the stresses are the lowest and almost do not change (15.3 without intermediate, 14.8–14.9 MPa with intermediate screws). At the same time, it should be noted that assessing the nature of stress distribution in the most heavily loaded vertebral bodies, particularly the lumbar region ($L_{I}$–$L_{V}$), requires a more detailed and comprehensive analysis. Given the high clinical significance of this information, a more detailed analysis of the stress-strain state of these structures could serve as the basis for a separate, more specialized study. Such an investigation would provide a deeper understanding of the mechanisms behind potential complications and help optimize treatment strategies.

The overall analysis of the obtained data demonstrates that the highest peak stresses in the beams were observed in the configuration of long screws without intermediate fixation (337.2 MPa), which potentially makes this model the most loaded for metal elements. Adding intermediate fixation significantly reduced the stresses in the beams, it was especially effective for the structure with short screws,

*Table 2*

**Tension during backward trunk bending in thoracolumbar spine models with explosive fracture of the $Th_{XII}$ vertebral body in different transpedicular fixation options**

| Control point | | | Stress, MPa | | | |
|---|---|---|---|---|---|---|
| | | | model without intermediate screws | | model with intermediate screws | |
| № | зона | | short | long | short | long |
| 1 | Bone tissue | $Th_{IX}$ vertebral body | 1.6 | 1.5 | 1.5 | 1.4 |
| 2 | | $Th_{X}$ vertebral body | 1.5 | 1.4 | 1.4 | 1.3 |
| 3 | | $Th_{XI}$ vertebral body | 10.4 | 9.8 | 8.3 | 8.2 |
| 4 | | $Th_{XII}$ vertebral body | 16.2 | 15.7 | 9.8 | 9.8 |
| 5 | | $L_{I}$ vertebral body | 17.4 | 17.2 | 15.9 | 14.1 |
| 6 | | $L_{II}$ vertebral body | 14.7 | 14.2 | 12.5 | 12.0 |
| 7 | | $L_{III}$ vertebral body | 17.5 | 16.2 | 15.1 | 13.9 |
| 8 | | $L_{IV}$ vertebral body | 20.6 | 20.7 | 19.1 | 18.5 |
| 9 | | $L_{V}$ vertebral body | 15.3 | 15.3 | 14.9 | 14.8 |
| 10 | | Lower endplate of $Th_{XI}$ vertebra | 3.0 | 3.1 | 2.4 | 2.4 |
| 11 | | Upper endplate of $L_{I}$ vertebra | 8.0 | 7.7 | 8.3 | 8.1 |
| 12 | | Screw entry into vertebral arch $Th_{XI}$ | 15.4 | 14.5 | 8.8 | 7.7 |
| 13 | | Screw entry into vertebral arch $Th_{XII}$ | — | — | 2.5 | 2.5 |
| 14 | | Screw entry into vertebral arch $L_{I}$ | 12.4 | 12.8 | 10.0 | 9.3 |
| 15 | Metal construction | Screw in the vertebral body $Th_{XI}$ | 79.5 | 79.7 | 66.0 | 54.0 |
| 16 | | Screw in the vertebral body $Th_{XII}$ | — | — | 13.2 | 13.5 |
| 17 | | Screw in the vertebral body $L_{I}$ | 75.9 | 95.1 | 64.4 | 75.7 |
| 18 | | Support beam | 361.3 | 337.2 | 442.3 | 436.0 |

where the stresses were 442.3 MPa, but the stability of the structure was significantly improved due to the load distribution by additional attachment points (model 3). Thus, it can be argued that from the standpoint of minimizing stresses in the system, the most favorable configuration is the one with short screws and intermediate fixation, while the most stressed is the model with long screws without intermediate fixation.

## Discussion

*The effect of screw length on the stress level*

The length of the pedicle screws significantly affects the nature of the stress distribution in the structure and bone. Models with long (bicortical) screws demonstrated higher fixation stiffness, which was manifested in an increase in stresses in critical areas of the structure. In particular, the transition from short to long screws (models 1 → 2 and 3 → 4) was accompanied by an increase in the maximum stress in the beams. This is explained by the fact that longer screws are more firmly fixed in the vertebral body and deform less at the point of contact with the bone, due to which more external load is transmitted directly to the metal connecting elements of the system. For example, in modeling without intermediate fixation, additional fixation of the screws in the opposite cortical wall led to an increase in stress in the beams by approximately 30–40 % compared to short screws under the same applied load. Similarly, in the presence of intermediate fixation, long screws also produced higher stresses in the beams, although the relative increase was somewhat smaller (since the presence of an intermediate support partially unloads the connecting elements). Critical zones in the bone—mainly around the screw insertion points (arch roots and adjacent vertebral body) — are also affected by screw length. Models with long screws showed increased local stresses in the pedicle area compared to short ones. This is especially noticeable at the lower supporting vertebra of the construct (in our case, $L_I$), where the load is maximal. In percentage terms, bicortical fixation increased the stresses in the bone at the base of the screw by approximately 15–20 %. Physically, this means that a more tightly fixed long screw fixes the bone tissue more strongly during loading, concentrating the stresses at the screw entry. In contrast, short (monocortical) screws are somewhat more elastic at the fixation point: they do not reach the opposite wall of the vertebra, so the load is partially damped by a small elastic deformation of the bone around the tip of the screw. As a result, the maximum stresses in the design with short screws are slightly lower, and the force distribution is softer, the stresses are more dispersed along the length of the screw and less concentrated near the pedicle. It is worth noting that the difference in the length of the screws affects the distribution of loads in more than one direction. Long screws, providing better grip, improve the resistance of the screw to extraction and can reduce the risk of its micromovement in the bone under load. In the simulation, this is reflected in slightly lower stresses in the distal (anterior) part of the vertebral body, where the tip of the long screw is fixed, the load is perceived there by an additional support point. Thus, the long screw distributes the forces over a larger volume of bone. However, the main "impact" of the load is taken by the screw entry zone into the pedicle, where the bone material experiences more stress. Therefore, in terms of the likelihood of local overload of the bone (which can lead to its resorption or microfractures around the screw), short screws are more gentle.

Thus, increasing the length of the screws increases the stiffness of the entire structure and reduces micromotions in the screw-bone connection, but is accompanied by an increase in maximum stresses in both the metal elements and the bone near the screws. In the context of modeling, this means that long screws fix more effectively, but can create critical overstress zones that should be taken into account when studying the durability of implants and the safety of bone support.

*Effectiveness of intermediate screws*

The introduction of screws into the body of the fractured $Th_{XII}$ vertebra (intermediate fixation) has shown high efficiency in improving load distribution and increasing the stability of the structure. A comparison of pairs of models without and with intermediate fixation (1 → 3 and 2 → 4) demonstrates a common trend: the presence of additional supports in the affected vertebra significantly reduces extreme stresses in the structure and changes the path of load transmission.

First, the intermediate screws take on part of the load that would otherwise fall on the beams and adjacent intact vertebrae. In a model experiment, this was manifested in a decrease in the maximum stresses in the beams under the condition of adding intermediate fixation. For the configuration with short screws, adding screws at $Th_{XII}$ reduced the peak stresses in the connecting elements of the structure by a fairly significant amount (about 5–10 % in our calculations). Although this figure may seem modest, the qualitative effect is very important: the shape of the rod bending changed — instead of one large span between $Th_{XI}$ and $L_{I}$, two smaller spans ($Th_{XI}$–$Th_{XII}$ and $Th_{XII}$–$L_{I}$) were formed, each with its own support point. This means that the rod works with less deflection and carries the load more evenly, reducing the risk of stress concentration in the middle of the span.

Secondly, intermediate screws change the distribution of forces in the vertebrae themselves. A fractured vertebra equipped with screws begins not only to be an object of support, but also actively perceives the load. Our results showed that the stresses in the $Th_{XII}$ body changed in the presence of screws: instead of the load being concentrated on the anterior and middle columns of the fractured vertebra (as is observed in the case of insufficient support), part of the forces were transmitted through the pedicle screws to the beams. This led to a decrease in the peak stresses in the $Th_{XII}$ body in the model with intermediate fixation. In other words, the additional screws strengthened the fragments from the inside, reducing their mutual displacement and the load on the interfragmentary structures. Such a redistribution is positive in terms of preventing further traumatization of the cancellous bone and promoting conditions for fracture consolidation, since the fragments are held more stably. It is also worth noting that intermediate fixation unloads the adjacent intact vertebrae. In the absence of screws in $Th_{XII}$, the vertebrae above and below ($Th_{XI}$ and $L_{I}$) actually bear the entire moment of force associated with fracture stabilization through the beams. This is manifested by high stresses in their bodies and especially in the places of screw attachment. Adding support in $Th_{XII}$ redistributes part of the moment directly to it. As a result, in models with intermediate fixation, the stresses in the bodies of $Th_{XI}$ and $L_{I}$ are somewhat lower, and the maximum stresses near their pedicles are also reduced (as can be seen from the comparison of stresses in the screw insertion zones: in model 3 they are lower than in 1; in model 4 they are lower than in 2). This means that additional fixation protects the neighboring vertebrae from overload, taking part of the efforts on itself.

Thus, the effect of additional fixation of the affected vertebra during extension load simulation is generally positive and multifaceted: intermediate screws reduce stresses in the metal structure, reduce their concentration in bone structures, increase stiffness and stability of the damaged segment, and also contribute to a more uniform distribution of loads throughout the “implant-spine” system.

*Clinical recommendations*

The modeling results have a clear practical reflection, helping to choose the optimal tactics of surgical fixation of a $Th_{XII}$ burst fracture. First of all, the importance of additional fixation of a traumatically affected vertebra has been confirmed. The installation of transpedicular screws in a fractured $Th_{XII}$ significantly increases the initial stability and unloads the implants. Clinically, this means better support of the anterior spinal column and a greater likelihood of preserving vertebral height and correcting kyphosis after surgery. Patients with this fixation are likely to have a lower risk of early deformity or pseudarthrosis of the fracture, since the fragments are securely immobilized. Thus, short-segment fixation with the inclusion of the fracture level (models 3 and 4) seems to be a reasonable choice in the case of burst fractures, this technique has already gained support in clinical practice and our modeling confirms its biomechanical feasibility.

Regarding the length of the screws, the data obtained indicate a certain compromise. Bicortical screw insertion (long screws) provides somewhat better fixation in the bone, which may be critical in cases of osteoporosis or poor bone quality, when there is a risk of screw pullout. In such situations, long screws will help to fix more firmly in the bone tissue and hold the construct; that is, clinically reduce the likelihood of screw pullout or migration under load. However, at the same time, the increased stiffness of the fixation leads to an increase in the loads on the construct itself. Practical conclusion: When using long screws, the surgeon should consider the quality and strength of the supporting beams, opting for those with higher endurance if necessary. Additionally, it is important to account for the need to limit excessive loads on the patient's spine during the postoperative period. Short screws, on the other hand, show lower stress concentrations and may be sufficient in cases where the bone is strong. Clinically, under conditions of sufficient bone mineral density, monocortical screws hold the segment quite reliably and at the same time reduce the risk of damage to the opposing cortical plate and adjacent structures (e. g., vessels, ligaments) during their installation [29]. The results obtained did not reveal a critical advantage over long screws in terms of stress reduction or stability, therefore, from the point of view of the stability/load balance on the structure, short screws with intermediate fixation are the optimal solution in most cases. Long screws can be considered as a means of increasing the reliability of fixation in difficult cases, but their necessity should be weighed against potential risks.

It should be noted that in the context of comparing our data with the results of other studies, there are certain difficulties due to a number of factors. In particular, most of the existing works use simplified damage models that do not take into account all the details considered in ours [16, 30–32]. In addition, other researchers often analyze a much smaller number of control points, which may not allow for a full assessment of the load distribution [14, 33, 34]. At the same time, a certain dissonance is introduced by the differences in the loading modes of the models, which also complicates the direct comparison of the results. Despite these differences, our data are generally consistent with the general trends observed in other studies and reliably reflect the nature of the load in the most critical structures [16, 30, 33, 35].

## Conclusions

The modeling of various options for short transpedicular fixation of the thoracolumbar junction confirms that under extension loading, the best balance between immediate stability and long-term safety of the structure is achieved when the affected vertebra is involved, using the optimal length of screws for a specific patient. This tactic minimizes the risks of failure of the structure and provides conditions for successful fusion, while reducing undesirable consequences for adjacent spinal segments in the future.

**Conflict of interest.** The authors declare the absence of a conflict of interest.

**Prospects for further research.** The nearest prospect is the completion of the analysis of all main loading patterns, which will allow for a comprehensive assessment of the biomechanical behavior of various fixation options. Special attention is planned to be paid to identifying the most critical overload zones in bone structures and elements of metal structures. Based on the obtained data, optimal stabilization configurations will be determined to increase the reliability of fixation and reduce the risk of complications in patients with explosive fractures of the thoracolumbar junction. It is planned to conduct clinical validation of the obtained biomechanical results, as well as develop individualized approaches to choosing a fixation configuration taking into account bone density, fracture morphology, and the degree of destruction of the anterior column. Dynamic modeling, with simulation of the fusion process and prolonged loading, is also considered.

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# BIOMECHANICAL ANALYSIS OF SHORT-SEGMENT PEDICLE SCREW FIXATION FOR AO TYPE A FRACTURES AT THE THORACOLUMBAR JUNCTION UNDER EXTENSION LOADING

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