ORIGINAL ARTICLE

CONTENTS

Anatomic-biomechanical substantiation of stabilization of the sacroiliac joint in cases of unstable pelvic injuries with a countersink-compression screw

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ABSTRACT

Aim: Study the mechanism of interaction between the 'sacroiliac joint – screw' system and determine the optimal parameters of the stabilizing structure, the strength of the system connection through computer modeling, and anatomical-biomechanical experiment

Materials and Methods: The optimal parameters of the stabilizing structure for the sacroiliac joint were calculated using software package MathCAD. To validate the results of the numerical modeling, corresponding investigations of mechanical characteristics and determination of stiffness of the studied systems were conducted by an upgraded testing stand, TIRAtest-2151

Results: Optimal dimensions of the stabilizing structure were calculated as follows: a thread length with a diameter of 9 mm ranges from 20 mm to 25 mm, and a thread length with a diameter of 7 mm ranges from 30 mm to 80 mm. The screw body, with a length from 15 mm to 70 mm and a diameter of 4.5 mm, is positioned between two thread portions. Under standard screw connection loading, a region of plastic deformation is observed under low force (≈40 N). Subsequently, elastic deformations are observed up to 900 N, after which the connection fails, and deformation of the stabilizing structure occurs

Conclusions: Resulting from the study the authors revealed that the stiffness of the fixed system with countersink-compression screws increases with the applied load, reaching 67-68% of the stiffness of an undamaged joint.

At all load levels, residual deformations in systems with C1Cc screws are significantly lower than the residual deformations in systems with C2Ct screws, indicating an enhanced deformation reliability of fixation with counter-compressive screws.

KEY WORDS: sacroiliac joint, stabilization, pelvic injuries, compression screw, combined pelvic fractures, operative treatment methods

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INTRODUCTION

Injuries to the pelvic bones constitute 5% to 12% of the total number of traumas. Isolated injuries occur in 7-8.5% of cases, multiple pelvic injuries account for up to 18%, and combined injuries reach up to 36% [1].

Injuries to the pelvic ring result from the high-energy mechanical factor which causes combined damage in various anatomical areas. In 62-87% of cases, such injuries are characterized as "polytrauma" [2].

The injuries of the pelvis combined with internal organs' trauma are observed in 48-80% of cases; with cranial-cerebral traumas – in 25-55%; with closed chest traumas – in 25-44%; with closed abdominal traumas – in 16-55%; with urinary tract injuries – up to 20%; with spinal injuries – up to 14%; with fractures of limb bones – 20-69%; with damage to major vessels and nerves – up to 10% [3].

The mortality rate, depending on the severity of the trauma, reaches 30% and does not tend to decrease. Disability manifests in 22-66% of afflicted individuals, while suboptimal outcomes are documented within the range of 20-74% [4].

The universally recognized classification of pelvic injuries is the Tile-AO/ASIF classification, which is grounded in the concept of pelvic ring stability/instability. It encompasses three types of fractures: Type A, characterized by minimal displacement without disruption of the integrity of the dorsal aspect of the pelvic ring, with an intact pelvic diaphragm capable of withstanding routine physiological loads; Type B, distinguished by rotational instability and vertical stability; and Type C, characterized by both rotational and vertical instability, accompanied by complete disruption of the pelvic ring, encompassing the posterior sacroiliac complex, including the ligaments sacrospinosum and sacrotuberosum [5].

The treatment strategy for these patients is founded upon the principles of "damage control surgery – DCS" [6] and "damage control orthopedics – DCO" [7].

According to the AO/ASIF guidelines, Type A fractures are considered stable injuries and generally do not require surgical intervention. For Type B injuries, characterized by anterior, rotational, and partial posterior instability, stabilization of the anterior segment is typically sufficient, utilizing external fixation devices as customary, taking into account the patient's overall condition. Type C injuries, characterized by both anterior and posterior instability, necessitate stabilization of both the anterior and posterior semirings [8].

The stabilization of the posterior semiring is objectively necessitated; however, the overall condition of patients, particularly in the early stages of hospitalization, and the anatomical peculiarities of the affected area, require minimally invasive technologies and appropriate fixation methods. This forms the basis for further research in this field.

AIM

To investigate the interaction mechanism of the "sacroiliac joint – screw" system and determine the optimal parameters of the fixation construct, as well as assess the strength of the connection system through computer modeling and anatomic-biomechanical experiment.

MATERIALS AND METHODS

A comprehensive analysis of literature on the treatment of unstable pelvic fractures from 2010 to 2023 was conducted using three databases (PubMed, Scopus, and Web of Science). The search focused on keywords such as pelvic fractures, combined pelvic fractures, operative treatment, methods, and means of fixation.

Titles and abstracts were reviewed during the selection process, and potentially relevant articles were assessed for inclusion.

Inclusion criteria: Full-text articles encompassing clinical/anatomic-biomechanical studies.

Exclusion criteria: Case reports, pilot studies, and preliminary investigations were excluded.

Through computer modeling employing formulas to assess the load-bearing capacity of threaded connections, the interaction mechanism of the sacroiliac joint-screw system was investigated. Furthermore, the correlation between the axial force of the threads of small and large diameters in the countersink-compression screw was studied. Optimal areas of load-bearing thread surfaces, the optimal number of turns, and the magnitude of their pitch were determined. Additionally, the necessary lengths of the larger and smaller threads, as well as the entire countersink-compression screw, were identified to create compression and prevent the thread of small diameter from breaking with a larger pitch during the screw insertion into the bone.

In addition, the reserve strength of the sacroiliac joint-screw system connection was determined under static dosed loading. The time spent to create axial force from the total tightening moment of the countersink-compressing screw was also calculated, which allowed the authors to determine the direction and magnitude of compression. All calculations were performed using the MathCAD software package.

The screw interaction with the bone is based on the spatial curve mechanism, where the screw line, is formed by the hypotenuse of a right-angled triangle when projected onto the lateral surface of a cylinder (Fig. 1).

For testing the fixator-bone systems, the authors utilized a universal testing machine TIRAtest-2151. This machine was used to determine the strength and deformation characteristics of materials and objects under tension, compression, and bending.

RESULTS

Based on the anatomical and functional characteristics of the sacroiliac joint, the design of the screw corresponded to the initial parameters: compression force – 1.5-2 kN; joint gap 3 mm; the material strength limit of the bone in cross section – 5 MPa.

CALCULATION OF THREAD STRENGTH

Since the bone has a significantly lower strength limit than the material of the screw, calculations were performed for the strength of the bone thread, as well as plastic deformations in the bone body, known as slipping, specifically in the section of the screw with a smaller diameter. The actual distribution of the load on the threads depends on many factors, most of which are random. Therefore, in practice, the strength calculation of the thread is made considering not actual stresses, but conditional stresses, which are compared with permissible ones. The scheme for calculating the strength of the thread is presented in Fig. 2.

The conditions for thread strength based on crushing stress were determined by the formula:

$$\sigma_{cr} = \frac{\mathbf{F}}{\mathbf{z} \cdot \boldsymbol{\pi} \cdot \mathbf{d_2} \cdot \mathbf{h}} \leq [\boldsymbol{\sigma}_{cr}],$$

where z=H/S – the number of threads with a height H. Substituting the given parameters for the smaller diameter thread, we get:

$$\sigma_{cr} = \frac{6064}{3 \cdot \pi \cdot 0{,}0058 \cdot 0{,}0012} = 92444020Pa.$$

The screwing length H was determined by the inequality:

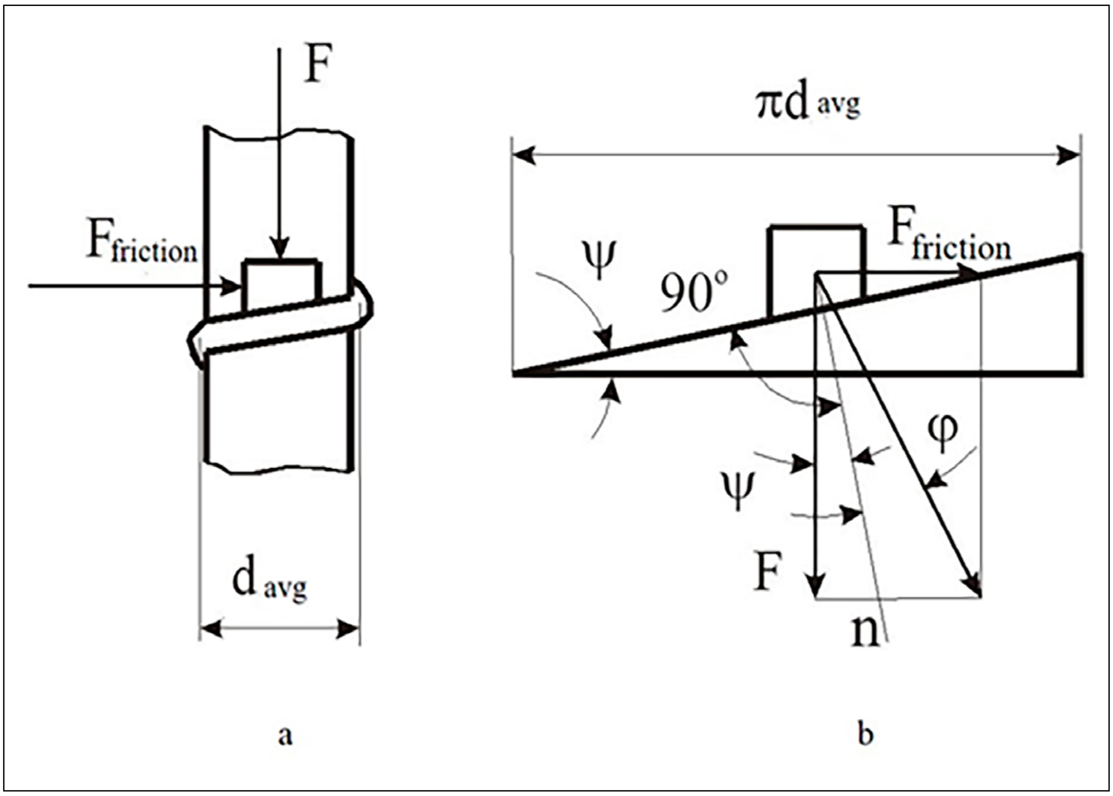

Fig. 1. Mechanism of screw interaction with bone. a) – spatial curve of screw-bone interaction; b) – unwinding of one turn along the average thread diameter; F – compression force; d_avg – average thread diameter, ψ – thread pitch angle, n – radius perpendicular to the direction of the screw, F_friction – friction force component, φ – angle of friction.

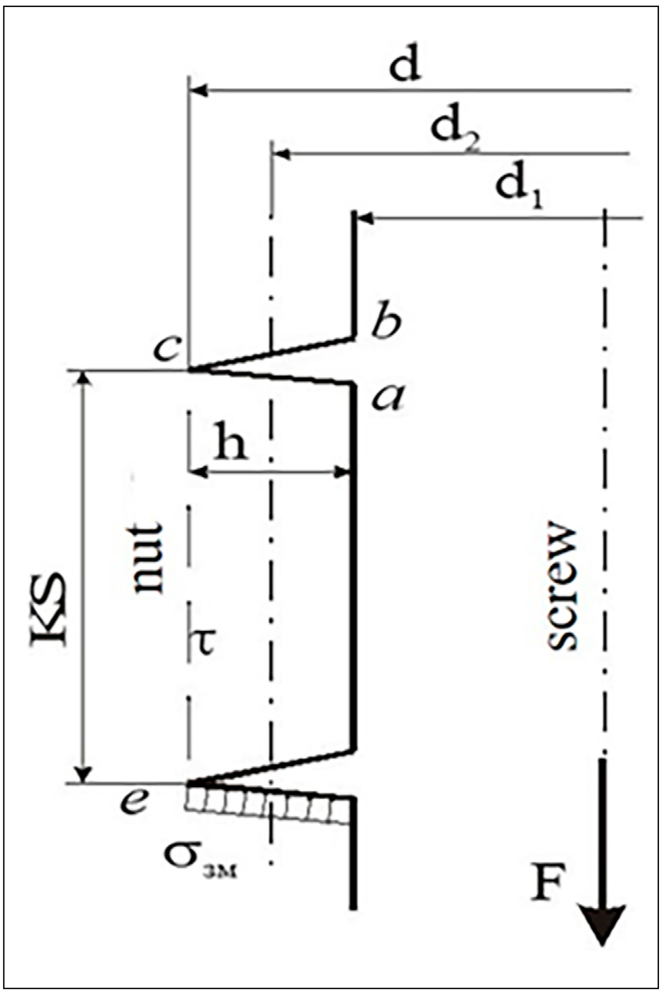

Fig. 2. Scheme for calculating thread strength.

$$H \geq \frac{F}{\pi \cdot d \cdot [\tau_{crosssection}]}$$
.

Substituting the values of the parameters for the smaller diameter thread, we obtain

$$H \geq \frac{1819}{\pi \cdot 0{,}007 \cdot 5\ (10^6} = 0{,}02m.$$

The strength limit for the cancellous bone layer under compression is taken to be one-tenth of the modulus of elasticity of the first kind. Since E=7,8×10^9 Pa, s_{cr}=7,8×10^8 Pa, and the strength limit of the cancellous bone layer varies within the range [$t_{cross\ section}$]=(4,32-12,26)×10^6 Pa.

The strength calculations based on the above formulas show that under the compression condition, even a thread with three turns withstands the maximum calculated force of 6064 N. The calculated crushing stress is 92,444,020 Pa.

Thread cross section calculations showed that with the minimum accepted cross section limit [$t_{cross\ section}$] = 5 x 10^6 Pa and the minimum axial force F = 1819 N, the thread length should be at least 20 mm, which corresponds to 8 turns.

PLASTIC DEFORMATIONS IN THE THREADED PART OF THE BONE

The plastic deformations in the nut significantly affect the strength limit of the threaded connection (Fig. 3).

Due to deformations, the nut increases in transverse dimensions and may "slide" off the bolt with partial cutting of the thread crests. This is particularly character-

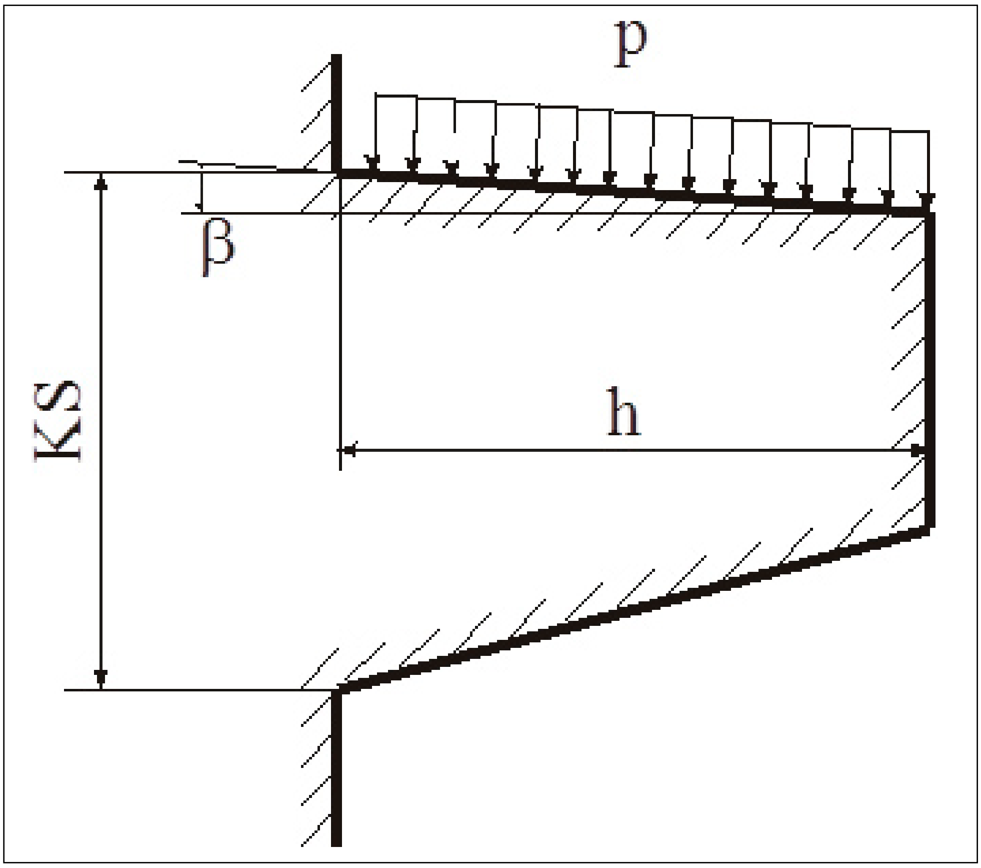

Fig. 3. Load on a thread of an asymmetric profile.

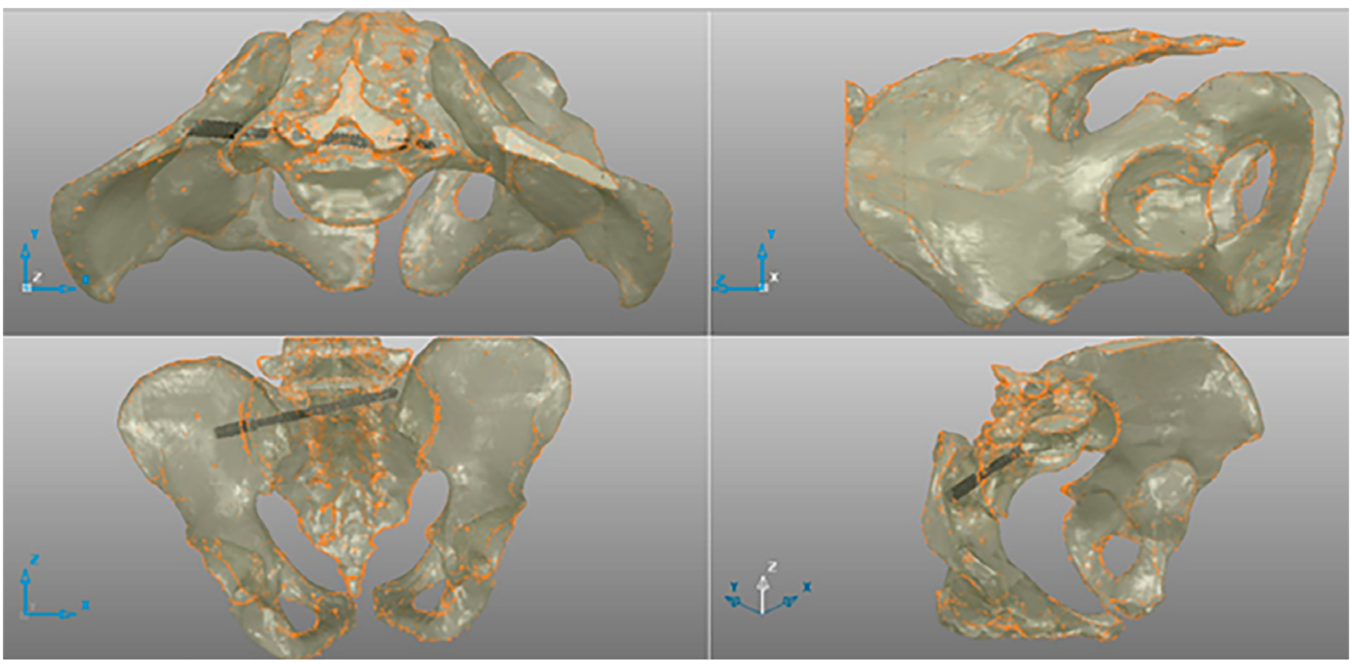

Fig. 4. General view of the stabilization of the sacroiliac joint with a newly designed screw.

istic for thin nuts and for structural components made of lightweight materials. It is evident that such a type of thread disruption will be present in the screw-bone connection as well. Therefore, let's consider under what loads the condition for bone strength against "sliding" will be satisfied.

For a thread of asymmetric profile (Fig. 3), the average radial stress in the nut wall is determined by the formula:

$$\sigma_{av} = p \cdot \frac{h}{S} \cdot tg\beta ,$$

where: p – pressure on the working side of the thread profile, h – height of the thread profile, S – pitch of the

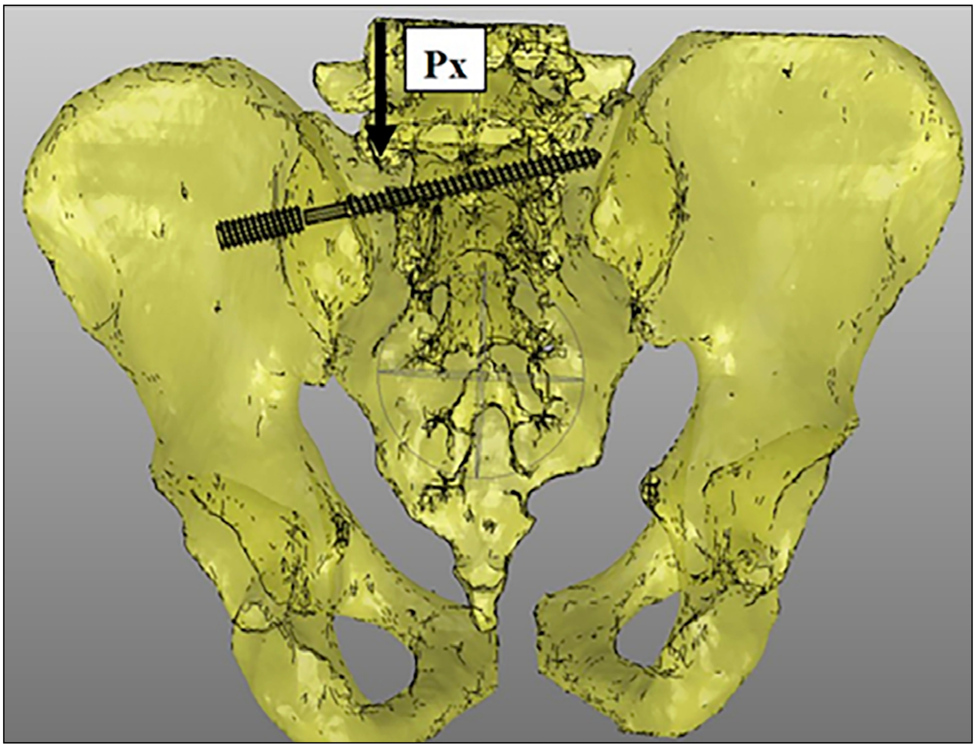

Fig. 5. Load scheme for the connection of the sacral and iliac bones.

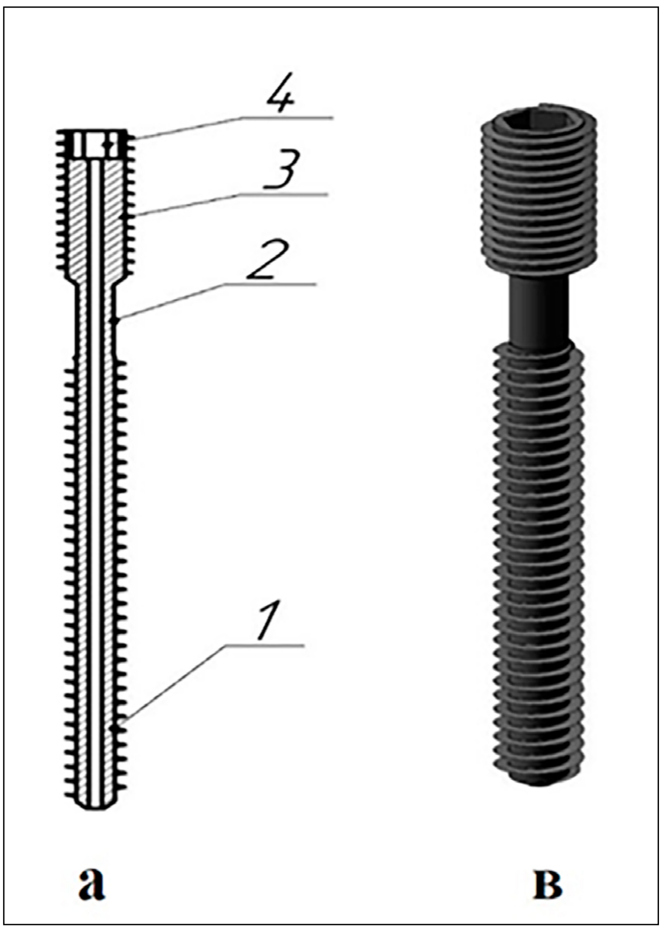

Fig. 6. Schematic representation of the counter-compressing screw, where: a) screw drawing (1 – thread with a diameter of 7 mm; 2 – screw body; 3 – thread with a diameter of 9 mm; 4 – hexagonal hole for the key.), b) overall view of the screw (3D model).

thread, β – angle of inclination of the working side of the thread profile.

Taking the load distribution between the threads as uniform, we find the pressure p on the lateral surface of the thread:

$$p = \frac{\sigma_0 \cdot A_0}{a \cdot z}$$, where

A_0 – is the cross-sectional area of the screw, $a=\pi d_2 h$— s the area of the ring of the thread, z – is the number of turns. Substituting $s_0 \times A_0$ with the force F, we get:

$$\sigma_{av} = \frac{F}{\pi \cdot d_2 \cdot H} \cdot tg\beta,$$

H= where z×S is the height of the nut.

Considering the nut as a ring with a thickness d and an average diameter D_{aver}, we obtain the formula for determining the stress in the ring:

$$\sigma_\theta = \sigma_{aver} \cdot \frac{D_{aver}}{2 \cdot \delta} = \frac{F}{2 \cdot \delta \cdot H} \cdot \frac{D_{aver} \cdot tg\beta}{\pi \cdot d_2}.$$

The strength condition against "sliding" will be expressed by the inequality:

$$\frac{F}{2 \cdot \delta \cdot H} \cdot \frac{D_{aver} \cdot tg\beta}{\pi \cdot d_2} \leq \frac{\sigma_{cross\ section}}{n_{cross\ section}}$$, where:

$s_{cross\ section}$ – is the cross section strength of the nut material, and $n_{cross\ section}$ – is the safety factor for cross section strength.

Hence, the strength of the nut against 'sliding' depends on the angle of inclination of the working surface of the

Table 1. The main design parameters of the screw

Parameters	Measurement unit	Thread of small diameter	Thread of large diameter
External thread diameter	mm	d=7	d¢=9
Internal thread diameter	mm	d_1=4,6	$d¢_1$=7
Mean thread diameter	mm	D_{aver}=5,8	$d¢_{aver}$=8
Thread pitch	mm	S=2,5	S¢=2
Thread height	mm	h=1,2	h¢=1
The angle of the screw line ascent	°	y= 8,6804°	y = 5,0554°

Table 2. Deformation characteristics of the undamaged (H1, H2) and damaged sacroiliac joint with stabilization by the countersink compression (C1Cc) and standard (C2Ct) screws.

Joint type	Δ_{AG}/ mm	P_{AG}/ H	δ-10^3, mm/H	C, H/mm
H1	1	1845	0,542	1845
H2	1	1840	0,543	1840
C1Cc	1	1238	0,8	1238
C2Ct	0,85	885	0,96	1041

Note: Δag – specified compression deformation, mm; V – deformation rate, mm/min; δ – specific deformations, mm/N; Pag – compression forces in the joint at specified deformations Δag, H; C – stiffness of the sacroiliac joint, defined as the ratio of the change in load to the change in deformation within the linear region of the deformation diagram, N/mm.

Table 3. Mean test data for objects under compression

Joint type	Δ_{AG}/ mm	P_{AG}/ H	C, H/mm	ψκ
H1	1	1845	1845	1
H2	1	1840	1840	1
C1Cc	1	1238	1238	0,67
C2Ct	0,85	885	1041	0,56

thread. Therefore, the use of a self-tapping thread with a working angle β≈0 is entirely justified for thin nuts and nuts made of material with a significantly lower strength limit than the screw material under large static loads.

Substituting the values of the parameters for the smaller diameter thread, we obtain

$$\frac{6064}{2 \cdot 0{,}02 \cdot 0{,}02} \cdot \frac{0{,}007 \cdot tg3°}{\pi \cdot 0{,}0058} = 152611 \leq \frac{5 \cdot 10^6}{2} = 2\ 500\ 000 Pa$$

The calculations for bone thread sliding showed that the bone with a thickness of the hypothetical ring of 20 mm and a thread height of 20 mm has a significant safety margin against sliding at the maximum calculated axial force.

Thus, through strength calculations, the conclusion can be drawn that the critical factor when using countersink-compression screws is the strength limit of the cancellous bone layer in cross section.

THE CALCULATION OF THE CONNECTION IN CROSS SECTION

The general view of the stabilization system for the sacroiliac joint using a newly designed screw is presented in Fig. 4.

Since the weight force is practically directed transverse to the screw, only one component of force will act in the connection (Fig. 5). The component PX acts in the direction of cross sectioning the screws and crushing the bone, so the critical factors will be the cross section strength of the screw and the compressive strength of the bone due to the action of component PX

Let's assume that the weight of a person is 80 kg. Therefore, we can assume that the PX component is 800 N. Since the screw is screwed into the sacral and iliac bones, we have a connection of the sacrum-screw and screw-iliac bone without clearances. Therefore, in further calculations, friction between the sacral and iliac bones is not taken into account.

The strength condition of the screw based on cross section stress is determined by the formula:

$$\tau = \frac{4 \cdot P_X}{\pi \cdot d_C{}^2} \leq [\tau],$$

where *dc* – is the diameter of the shank (4.6 mm), [t] – is the allowable cross section stress in the material.

Substituting the given values into the formula, we get:

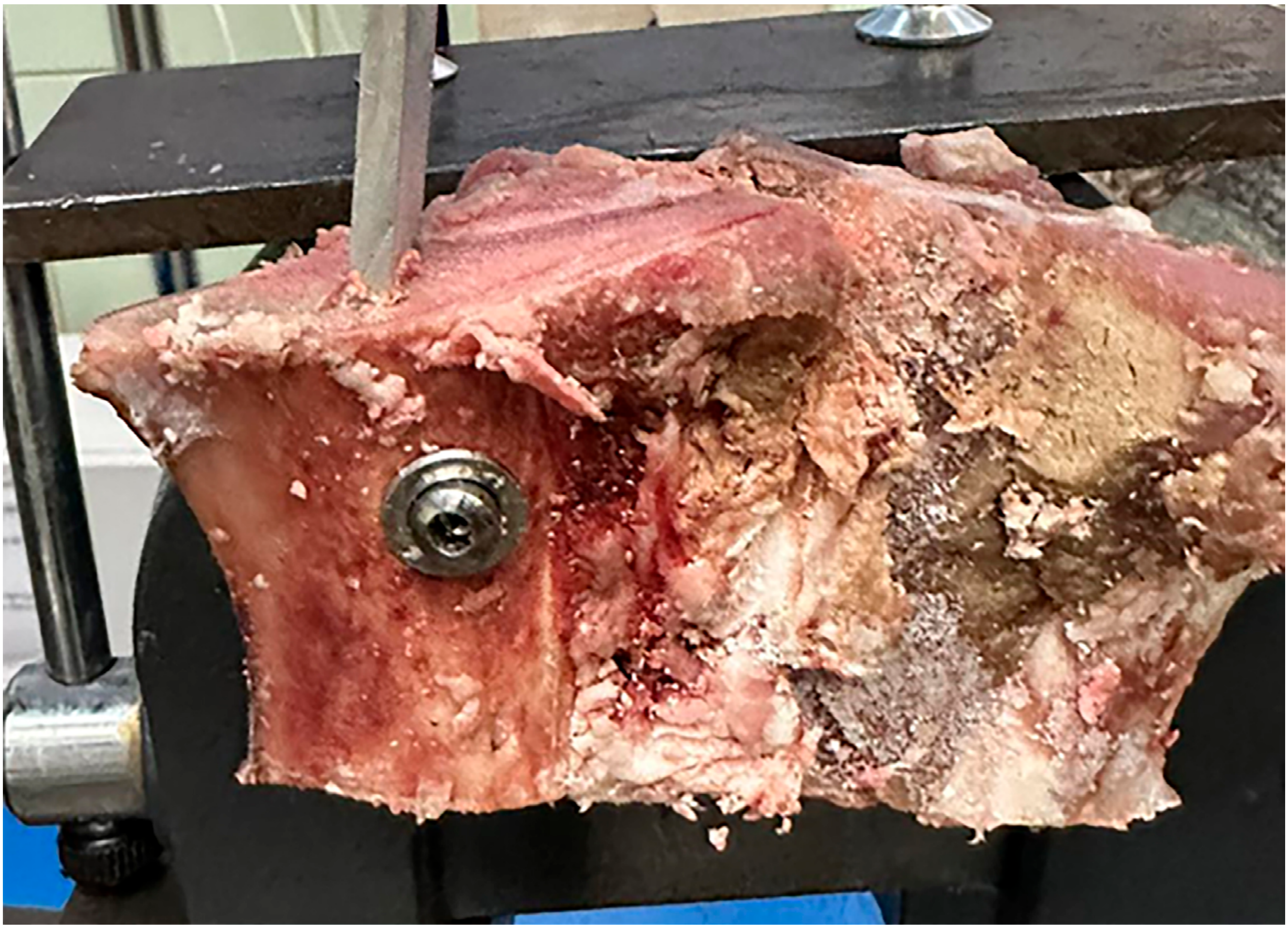

Fig. 7. Secured native specimen of the sacroiliac joint in the research machine.

$$\tau = \frac{4 \cdot 800}{\pi \cdot 0.0046^2} = 48{,}1376\mathrm{MPa}.$$

In this case, the cross section strength limit for hardened steel 95X18 is 770 MPa.

The compression stress, taking into account the assumed uniform distribution through the thickness of the connected parts, is determined by the relation:

$$\sigma_{crush} = \frac{P_X}{d_C \cdot \delta} [\sigma_{crush}], w$$
,

where d – is the thickness of the connected elements.

Calculations are performed for the fracture, as the diameters of the screw in the fracture zone and the bone differ insignificantly, and the thickness of the bone is significantly larger than that of the fracture. In this case, the thickness of the iliac bone is approximately 20 mm, and the diameter of the screw shank is 4.6 mm.

Substituting these values into the formula, we get:

$$\sigma_{crush} = \frac{800}{0{,}007 \cdot 0{,}02} = 5{,}7mPa$$
,

which is within the allowable limits, ranging from 4,32-2,26×10^6 Pa.

The proposed design of the counter-compressing screw and technical characteristics are presented in Fig, 6 and Table 1.

The screw differs from known designs in that it has a central hole with a diameter of 2mm or 1.6mm, depending on the diameter of the leading spike. The pitch of the thread of the smaller diameter is 2.5mm, and the pitch of the thread of the larger diameter is 2mm. The threads of the larger and smaller diameters have a profile known in the technology of a supporting thread. In this case, the rectangular segments of the thread profiles of the larger and smaller diameters are directed towards each other. This type of thread is used in cases where significant unidirectional axial loads are transmitted. Due to the small angle of inclination of the working surfaces of the threads, this thread provides increased efficiency even compared to a trapezoidal thread, while retaining all its advantages.

The length of the thread with a diameter of 9 mm varies from 20 mm to 25 mm, and the length of the thread with a diameter of 7 mm ranges from 30 mm to 80 mm. The body of the screw, with a length from 15 mm to 70 mm and a diameter of 4.5 mm, located between the two threads, does not have a thread.

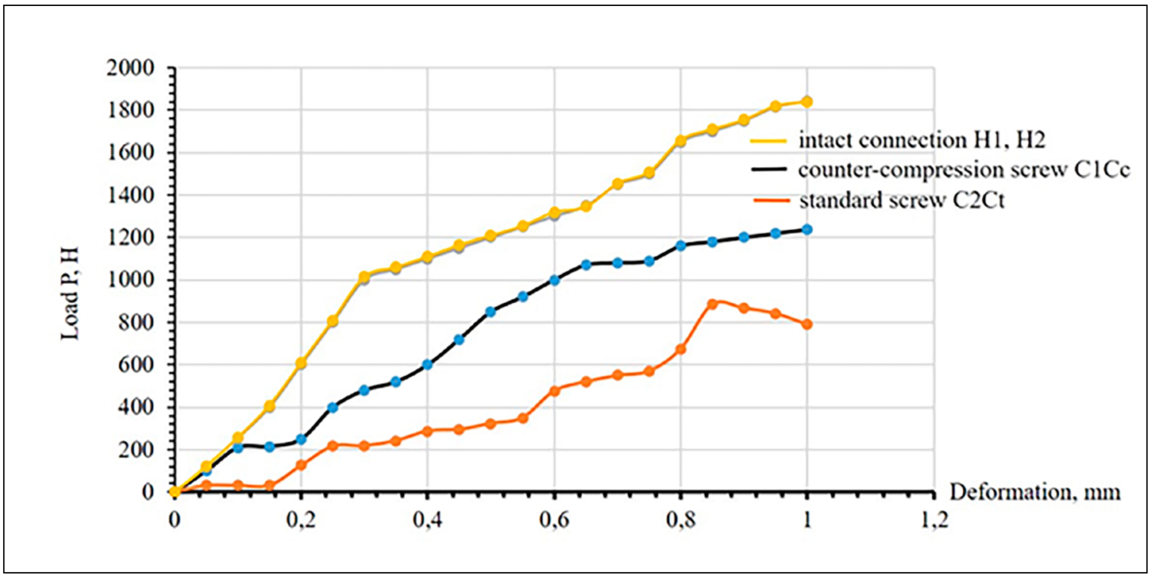

Fig. 8. Deformation diagram for cross section deformation of an undamaged sacroiliac joint (H1, H2), damaged joint stabilized with a countersink screw (C1Cc), and damaged joint stabilized with a standard screw (C2Ct).

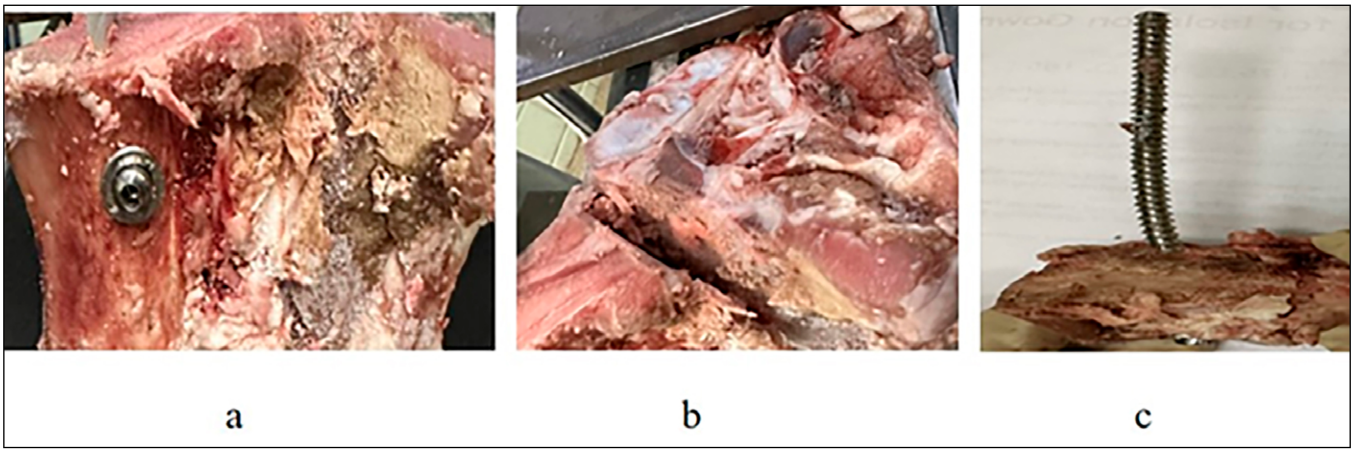

Fig. 9. Destruction of the connection, where: a – damaged sacroiliac joint, connected with a standard screw (C2Ct); b – visualized loss of stability under a load exceeding 900N; c – visible deformation of the screw.

To verify mathematical calculations and for a comparative characterization of the cancellous and countersink screws of our own design for stabilizing the sacroiliac joint, an anatomical-biomechanical experiment was conducted using native preparations of the sacroiliac joint. The anatomical-biomechanical study was carried out at the laboratory of the Department of Normal Anatomy of the Bogomolets National Medical University and the Research Center "Reliability" of the National Technical University of Ukraine "Igor Sikorsky Kyiv Polytechnic Institute" (Fig. 7).

CHARACTERISTICS OF JOINTS UNDER SHORT-TERM SINGLE LOADS

Based on the obtained data, deformation diagrams were constructed for cross section deformation of samples in the vertical direction N1 and N2 (undamaged joints), sample C1Cc (damaged sacroiliac joint, stabilized with countersink screw), and sample C2Ct (damaged sacroiliac joint, stabilized with a standard screw) (Fig. 8).

The loading automatically stopped when the maximum load decreased by the value of Δ2 (100N) or when the displacement/cross section h reached 1mm.

The analysis of the diagrams shows that the deformation of undamaged samples has an elastic character throughout the entire experiment interval. When loading the connection with the countersink screw, there is an initial short elastic deformation segment up to 200N over a length till a cross section of 1mm. A

When loading the connection with the standard screw, a plastic deformation segment is observed almost from the beginning at a small force (≈40N). Subsequently, elastic deformations are observed up to 900N, after which the connection fails, and the stabilizing structure deforms (Fig. 9).

Deformation diagrams in load ranges are not linear. Therefore, the elastic properties of both undamaged joints and joints with compromised integrity but stabilized by both types of screws in the specified force ranges can be characterized by stiffness coefficients C, N/mm (the ratio of the applied load Pmax to the deformation increase Δr) C = Pmax / Δr, where: Pmax – maximum load (N), measured from the deformation diagram; Δr – deformation (mm) corresponding to Pmax.

Quantities inversely proportional to stiffness characterize the flexibility of a system (its ability to deform under applied loads). The flexibility of the specimens was determined based on the calculated stiffness values (specific deformations, mm/N) as quantities inverse to stiffness, denoted as δ = 1/C. This parameter reflects the displacement magnitude resulting from loading the specimen with a force of 1 N. Elastic characteristics of the specimens were determined based on the constructed deformation diagrams (Table 2).

Table 3 presents summarized results of tests on undamaged objects and samples with modeled damage, stabilized by two types of screws. The table also includes data on the change in stiffness due to damage and stabilization by screws compared to undamaged preparations, calculated using the formula: $\psi_к = C^C/C^H$, where $C^c=1/\Delta^c$, $C^H=1/\Delta^H$, with indices "N" representing characteristics of undamaged preparations and indices "C" representing characteristics of preparations with damaged objects fixed by screws.

DISCUSSION

Despite a significant increase in the number of operative interventions for pelvic injuries, especially in combined trauma, the conservative method is used much more often (conservative in 70.4-89.2%; operative in 10.8-29.6%) [9]. However, the conservative method is only possible in patients with stable pelvic injuries. When applying the conservative treatment method for unstable pelvic injuries, unsatisfactory results are observed in 35-66.7%, and mortality is 2.5 times higher than with operative treatment – 21.8% and 8.3%, respectively [10].

The main factor contributing to lethal cases is massive bleeding, which occurs in 80-90% of cases with unstable fractures due to damage to presacral, retrosacral, and paravesical venous plexuses. Unstable pelvic injuries are also accompanied by damage to retroperitoneal muscles (iliopsoas, sciatic, and their fascia), leading to the so-called "chimney effect" – an increase in intrapelvic bleeding, which extends cranially and leads to the development of pelvic and abdominal compartment syndrome [11].

Experimental studies have shown that for every centimeter of diastasis in the symphysis pubis area, the pelvic volume increases by almost 5%, and in the sacroiliac joint area, it increases by 3%. A 5 cm diastasis increases the pelvic volume by up to 20%. All these data suggest the need for the fastest possible reduction of this volume mechanically to reduce blood loss and provide a tamponade effect [12].

The modern concept of treating individuals with unstable pelvic injuries in combined trauma within the first 48 hours requires urgent stabilization, mainly extrafocally, using external fixation devices, C-clamps, or a Hanzc frame, and if possible, performing percutaneous osteosynthesis. The latter type of surgical intervention is recommended when there are urgent indications related to pelvic or intraperitoneal organ injuries (bladder rupture, urethral injury), where bone fragments protrude into the wound, and fixation itself will not be traumatic and prolonged [13]

Stabilization of the pelvic ring in the initial emergency care stage using external fixation devices is the most commonly used method due to its relatively simple application technique [14]. However, in type "C" pelvic injuries involving a complete rupture of the sacroiliac ligaments, fractures of the posterior parts of the iliac bone, transforaminal sacral fractures with vertical displacement of the pelvic bones, fixation of only the ventral part of the pelvic ring does not provide stability to its dorsal part [15]. For the stabilization of the posterior pelvic complex, pelvic C-clamps or a Hanzc frame have become widely used in the early intensive care period [16].

Internal osteosynthesis for fractures of the pelvic bones in patients with combined trauma is implemented as a final stage in cases where life-saving surgeries are performed within or near the pelvis (using a single laparotomy approach). This approach is applied in open and closed multifragmentary fractures with significant displacement, dislocations, and disruption of pelvic organs [17].

Open surgical methods provide good results due to direct visualization and the ability to anatomically align the fractures. However, considering the specificity of this area, the complication rate reaches up to 27%. In addition to iatrogenic trauma to neurovascular plexuses, open osteosynthesis often leads to infectious complications, pelvic hematomas, and secondary hemorrhages. It is essential to consider the traumatic nature of these surgical interventions, which, against the backdrop of a patient's unstable condition, can lead to the so-called "second kick" [18].

Therefore, a large number of studies and implementations are currently being conducted regarding the stabilization of the sacroiliac joint, based on the principles of minimal invasiveness [19,20].

At the moment, there are several methods of internal fixation of the sacroiliac joint: transcutaneous sacroiliac screw, anterior plate, posterior sheath, minimally invasive adjustable plate, and other fixation systems. These methods have a number of advantages and disadvantages, but none of them practically differs in terms of fixation degree [21].

Transcutaneous screws have advantages and are widely used due to minimal additional trauma, making them a promising direction in the treatment of patients with integrity issues in the sacroiliac joint area [22].

Based on their own research, Osterhoff et al. argue that pelvic stabilization using sacroilial screws is sufficient. However, the implementation of this method requires significant expertise and is associated with a high incidence of iatrogenic vascular and nerve injuries. Additionally, both patients and surgeons are exposed to substantial radiation during the performance of such surgical interventions [23].

To minimize complications, a variety of instrumental methods and navigation techniques have been proposed, ensuring visualization during the placement of the fixation screw [24].

Based on the analysis of the effectiveness of percutaneous fixation with sacroiliac screws for stabilizing the sacroiliac joint, it is considered that two screws represent the optimal method of fixation for this location, provided they traverse three cortical layers, and this is regarded as the technique of "central fixation" [25].

However, the frequency of sacral dysmorphism in adults is approximately 30-40%, and in this patient category, the "safe zone" for conducting fixation constructs is 36% smaller than usual [26]. Based on this, Griffin D.R. et al. believe that in such cases, it is challenging to place two sacral screws into the body of the first sacral vertebra. However, according to these same authors, the routine placement of a single screw is considered safe and sufficient for stabilizing the posterior pelvic half-ring [27].

Despite certain controversies in the options for minimally invasive synthesis of the sacroiliac joint, research results indicate their higher effectiveness (around 80%) compared to open synthesis [28]

CONCLUSIONS

1. Percutaneous stabilization of the sacroiliac joint is a minimally invasive fixation method widely used in patients with an unstable overall condition, characteristic of polysystemic and polyorgan injuries. Achieving a sufficient level of stabilization is crucial, making the development and implementation of constructions with optimal characteristics a relevant issue.
2. The results of anatomical-biomechanical research, comparing deformations of intact joints and joints with modeled damage, showed that the rigidity of the system fixed with a countersink-compression screw increases with the load and constitutes 67-68% of the rigidity of the undamaged joint. At all load levels, residual deformations in systems with C1Cc screws are significantly less than residual deformations in the system with C2Ct screws, indicating enhanced deformational reliability of fixation with countersink-compression screws.

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CONFLICT OF INTEREST

The Authors declare no conflict of interest

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