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Biomechanical Rationale for Osteosynthesis Stability in Lower Limb Long Bone Fractures Under Multi-directional Loading: A Finite Element Analysis

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Abstract

Background. In the general structure of trauma, fractures of the bones of the lower extremities account for 47.3%, among which diaphyseal injuries occupy the leading place. The choice of the optimal type of fixator (osseous, intramedullary, or external) remains a subject of discussion, since the stability of the "bone – implant" system directly determines the nature of the regenerative process and the timing of consolidation. **Objective:** based on three-dimensional mathematical modeling using the finite element method, to conduct a comparative analysis of the stress-strain state of femur and tibia models with fractures at different diaphyseal levels under four variants of multidirectional loads when using four common osteosynthesis systems. **Materials and methods.** Three-dimensional finite element models of the femur and tibia with transverse diaphyseal fractures in the upper, middle, and lower thirds were constructed. The stability of fixation with a bone plate, Ilizarov external fixator, unilateral rod fixator, and interlocking intramedullary nailing (IMN) was investigated. Four loading options were modeled: Option 1 (pure axial compression), Option 2 (axial compression and frontal bending), Option 3 (axial compression and sagittal bending), Option 4 (axial compression and torsion). **Results.** Bone osteosynthesis provides absolute rigidity (Δ = 0.00 mm) in the 3rd and 4th segments of both bones, but in low fractures of the tibia (n/3), stresses increase to 6.70 MPa, and micromobility – to 0.30 mm. The spike apparatus demonstrates universal isotropic elasticity (Δ = 0.01–0.05 mm) in both segments under all types of loads. The rod apparatus in n/3 fractures reveals critical instability: stresses on the femoral rods reach 38.2 MPa, and the displacement of the tibia fragments is 1.90 mm. The IMN method is the standard for the hip at all levels ($\Delta \le$ 0.15 mm); however, in low fractures of the tibia (n/3) under combined forces, fragment displacement reaches extraphysiological limits of 1.42–1.60 mm. **Conclusions.** The stability of the bone-implant system is determined by the design features of the fixation device and the extent of diaphyseal damage. None of the osteosynthesis methods studied offers absolute mechanical versatility across all bone segments. Intramedullary osteosynthesis is the optimal method, except for fractures in the distal third of the tibial shaft, where a wire-based external fixation device proved more biomechanically stable.

Keywords: osteosynthesis, finite element method, stress-strain state, tibia, femur, stability.

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INTRODUCTION

The structure of musculoskeletal injuries indicates an increase in the number of high-energy injuries to the lower extremities[1].

According to the World Health Organization (WHO), injuries from road traffic accidents, falls from heights, and work-related injuries result in over 1.2 million deaths worldwide each year, and 20 to 50 million people suffer from non-life-threatening injuries with long-term or permanent disability worldwide. Of these, musculoskeletal injuries, particularly lower extremity skeletal injuries, account for over 40% of all emergency hospital admissions in developed countries[2].

According to the WHO Global Report on the Status of Road Safety, fractures of the long bones of the lower extremities are the most typical anatomical consequences of polytrauma and high-energy closed and open injuries[3-5].

In Ukraine, according to official medical statistics of the Center for Medical Statistics of the Ministry of Health of Ukraine and research institutions of the National Academy of Medical Sciences of Ukraine, lower limb injuries consistently occupy a leading place among the causes of temporary and permanent disability of the adult population, second only to circulatory system pathologies. The incidence rate of musculoskeletal injuries in the country is 360–400 cases per 10 thousand population annually[6,7].

In the structure of hospitalized patients with a trauma profile, fractures of long bones (femur and tibia) are one of the leading nosologies, reaching 47.3% of all fractures of the lower extremities. According to domestic studies, the share of isolated diaphyseal injuries in this segment is from 45% to 56%. Considering that more than 72% of victims with diaphyseal fractures of the femur and tibia belong to the most able-bodied age group (from 20 to 65 years), the presented problem acquires not only medical but also significant socio-economic significance[8,9].

Since 2022, the epidemiological and clinical picture of trauma in Ukraine has undergone radical and unprecedented changes in connection with full-scale military aggression. According to consolidated data from military medical institutions and civilian support hospitals, the proportion of combat surgical trauma and the consequences of massive rocket and artillery shelling of peaceful cities have radically changed the structure of hospitalizations. Injuries of the musculoskeletal system have become the dominant type of pathology, accounting for 70% to 82% of all wounds. At the same time, gunshot and explosive injuries of the lower extremities are recorded in more than 65% of cases among the total mass of skeletal trauma[10,11].

The combination of long bone fractures with craniocerebral trauma, acubarotrauma (mine-explosive trauma), and internal organ injuries is observed in 35–42% of hospitalized patients in emergency centers. Such injuries are characterized by a "mutual burden syndrome" and require staged surgical treatment according to the principle of orthopedic damage control – initial rapid fixation with external fixation devices (EX-FIX) with subsequent delayed transition to stable internal osteosynthesis (IMN or plates). The explosive deformations and consequences of road accidents involving heavy machinery were mainly multi-fragmented type C fractures according to the AO/ASIF classification, with the presence of extensive segmental bone tissue defects (from 3 to 12 cm) and periosteal detachment over a significant length of the diaphysis. In addition, the shock wave forms large-scale hidden zones of microstructural bone destruction and soft tissue ischemia around the fracture gap. This causes a critical deterioration of local angiogenesis and consolidation processes, and the frequency of delayed consolidation, formation of false joints, and bone defects under such injuries has increased to 18.4–26.8%; in the case of infected open fractures, it exceeds 35%[12].

Therefore, the stability of bone fragment fixation is a key biomechanical factor that determines the nature of bone tissue regeneration (primary or secondary fusion) and minimizes the risk of complications such as delayed consolidation, formation of false joints, or destruction of metal structures. Today, orthopedics uses a variety of implants, grafts, and devices to treat fractures and bone defects[13-15]. During the patient's locomotor activity, the segments of the lower extremities are subjected to complex mechanical forces. Real functional loads are rarely limited to isolated compression or bending deformations. The most critical in terms of loss of fixation stability are combined force vectors that combine axial pressure with frontal or sagittal displacement, as well as torsion (rotation).

To address these issues and study in detail the stress-strain state of the implant-bone system, mathematical modeling using the finite element method (FEM) was applied. This numerical method is the most adapted for the study of complex biological structures, since it allows for accurately reproducing the geometric features of bones, taking into account the heterogeneity of their physical and mechanical properties, and also for assessing the dynamics of the distribution of internal stresses and deformations under the influence of multidirectional loads[16,17].

Studying the stress-strain state (SSS) of the "bone – implant" system under the action of multidirectional biomechanical factors, depending on variations in patient mass and load, allows us to objectively assess the strength limits of structures. The use of mathematical modeling by the FEM makes it possible to substantiate a differentiated approach to the choice of surgical treatment tactics and optimize the parameters of metal structures (plate thickness, rod diameter, number of locking screws) depending on the fracture level, which is strategically important for accelerating patient rehabilitation and reducing the level of disability in Ukraine.

RESEARCH OBJECTIVE

Using finite element modeling, a comparative assessment of the stress-strain state and stability of osteosynthesis systems of the femur and tibia at different fracture levels under multidirectional functional loads was conducted.

MATERIALS AND METHODS

In the laboratory of biomechanics of the State Institution "Institute of Spine and Joint Pathology named after Professor M.I. Sytenko of the National Academy of Medical Sciences of Ukraine", finite element models of the thigh and lower leg were developed. The materials of the model elements were considered homogeneous and isotropic. The physical and mechanical properties (Young's modulus E and Poisson's ratio ν) corresponded to standard literature data: cortical bone ($E = 18350$ MPa, $\nu = 0.29$), spongy bone ($E = 330$ MPa, $\nu = 0.30$), titanium alloy of implants ($E = 1.1\times10^5$MPa, $\nu = 0.30$)[18].

Loading of femur models was carried out in four ways[18]:

1) the first option consisted of the main load on the thigh, which occurred from the load on the femoral head (P_1), large swivel load (M_1), and proximal femoral load (P_2, R_3);

2) the second option - in addition to the main load, an additional load was applied to the proximal part of the thigh in the frontal area, which was directed inward (P_4, R_5);

3) the third option - in addition to the main load, an additional load was applied to the proximal part of the thigh in the sagittal plane and directed forward (P_6, R_7);

4) the fourth option - in addition to the main load, an additional load was applied to the proximal thigh, which created a twisting effect on the model (P_8, R_9).

Loading of models with tibial fracture was carried out in four ways[19-21]:

1) the first option consisted of the main load of the tibia, which consisted of the load on the proximal part of the tibia (R_1, R_2) and the load on the distal tibia (R_3);

2) the second option - in addition to the main load, an additional load was applied to the proximal part of the tibia in the frontal area, which was directed inward (R_4, R_5);

3) the third option - in addition to the main load, an additional load was applied to the proximal part of the tibia in the sagittal plane and directed forward (R_6, R_7);

4) the fourth option - in addition to the main load, an additional load was applied to the proximal part of the tibia, which created a twisting effect on the model (R_8, R_9), as shown in Fig. 1.

RESULTS

A detailed comparative analysis of the numerical results obtained in the process of mathematical modeling by the finite element method of biomechanical systems "bone – implant" allowed us to establish clear patterns of the distribution of equivalent stresses according to Mises and the values of relative displacements of bone fragments. It was established that these parameters have a pronounced differentiated character depending on the anatomical segment (femur or tibia), the localization of the bone defect (proximal, middle, or distal third of the diaphysis), and the vector of the applied functional load[22].

The numerical values and graphical interpretations of the parameters of the SSS of osteosynthesis models of the femur and tibia under varying loading modes and design features of the fixators are systematized in Tables 1–4 and Figs. 2–13[18].

Analysis of the SSS of biomechanical systems under conditions of cumulative (bone) osteosynthesis revealed a critical dependence of fixation stability on the anatomical level of damage to the diaphyseal segment. In particular, stable fixation of fractures of the middle third of the diaphysis of both the femur and tibia with the use of cumulative plates is characterized by high spatial rigidity of the structure (Table 1; Figs. 2–4).

Table 1. Stress-strain parameters of models of periosteal metal osteosynthesis of the femur and tibia under various loading conditions

Fracture level		Model elements		Option 1 (compression)	Option 2 (frontal)	Option 3 (sagittal)	Option 4 (twisting)
Thigh	in/3	Stress, MPa	plate	1.2	0.8	1.4	1.0
			screw	4.6	4.3	4.4	4.6
			in the bone	1.6	4.0	4.2	4.4
		Movement of debris, mm		0.18	0.16	0.17	0.17
	with/3	Stress, MPa	plate	1.2	0.7	1.4	1.1
			screw	1.5	0.9	1.4	1.5
			in the bone	0.9	0.7	1.3	0.9
		Movement of debris, mm		0.00	0.00	0.00	0.00
	n/3	Stress, MPa	plate	0.8	0.9	1.1	0.8
			screw	4.1	4.2	4.1	4.1
			in the bone	1.5	3.6	3.6	3.8
		Movement of debris, mm		0,17	0.15	0.16	0.15
Shin	in/3	Stress, MPa	plate	1.7	0.6	1.0	2.0
			screw	5.3	4.9	5.7	6.1
			in the bone	1.8	2.5	1.8	1.8
		Movement of debris, mm		0.20	0.17	0.18	0.20
	with/3	Stress, MPa	plate	0.8	0.6	0.6	1.1
			screw	1.6	1.1	1.0	1.9
			in the bone	1.3	0.8	0.8	1.7
		Movement of debris, mm		0.00	0.00	0.00	0.00
	n/3	Stress, MPa	plate	1.0	1.6	1.3	1.0
			screw	5.4	6.7	5.4	5.4
			in the bone	2.5	3.3	2.3	3.1
		Movement of debris, mm		0.20	0.30	0.17	0.20

Under all loading conditions, the linear displacement of the fragments was 0 mm. The level of maximum stresses on the plate (0.6–1.4 MPa) and bone tissue (0.6–1.7 MPa) did not exceed critical limits, and the stress on the screws ranged from 0.9 to 1.9 MPa,

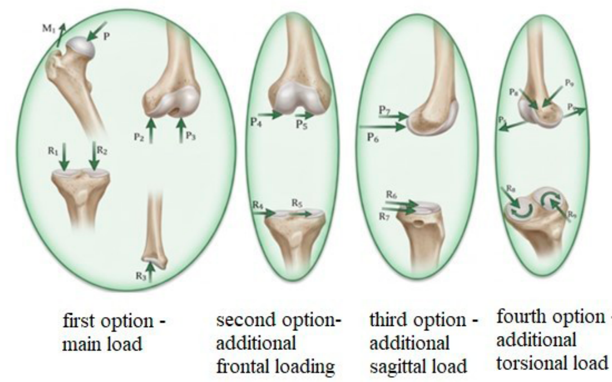

Figure 1. Load diagrams of femoral and tibial segments

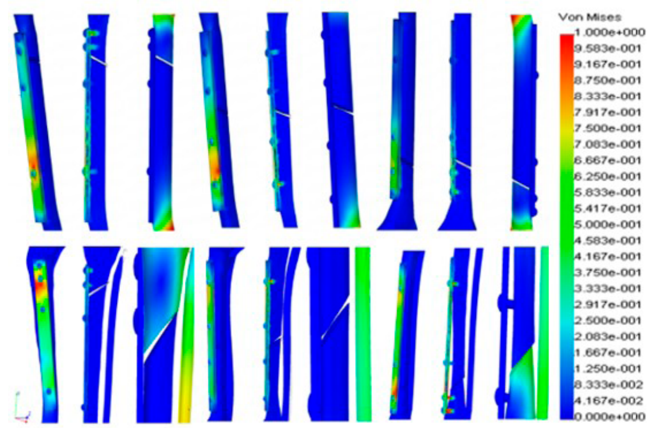

Figure 2. Stress-strain state of models of bone osteosynthesis of the femur and tibia under different loading conditions

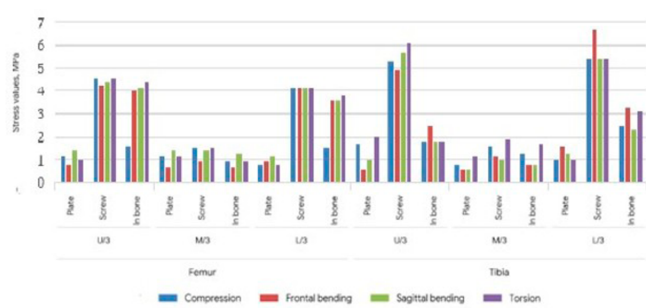

Figure 3. Stress values in the model of the thigh and lower leg when fixing fragments with a plate under the influence of various loads

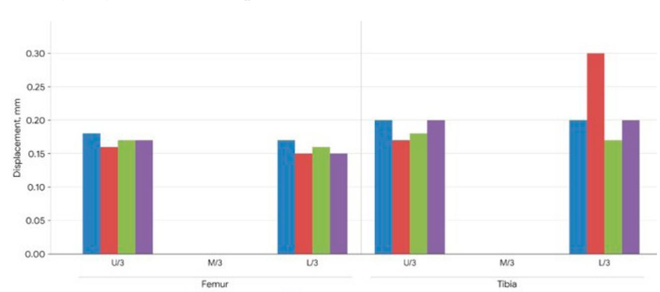

Figure 4. The magnitude of displacements of bone fragments of the model of the thigh and lower leg when fixing the fragments with a plate under the influence of various loads

which ensured biomechanical balance and full compliance with the requirements of stable fixation according to AO standards.

At the same time, moving the fracture zone to the upper or lower third led to a sharp destabilization of the entire system. In fractures of the upper third of the thigh, the peak stress on the screws was 4.3–4.6 MPa, the stress in the bone reached 4.4 MPa, and the displacement of the fragments was 0.16–0.18 mm. A similar picture was recorded in the lower third of the thigh, where the stress on the screws reached 4.2 MPa, and the stress in the bone under the influence of additional moments increased 2.5 times compared to pure compression – up to 3.8 MPa.

The most critical results of bone-on-bone osteosynthesis were obtained for the tibia. A fracture of its upper third was accompanied by stress on the screws of up to 6.1 MPa and movement of fragments of up to 0.20 mm. In the fracture of the lower third of the tibia, the absolute minimum of stability was recorded among all bone-on-bone models: the maximum stress on the screws reached 6.7 MPa, the stress in the bone was 3.3 MPa, and the mutual movement of fragments in the callus formation zone reached a critical 0.30 mm.

This load distribution indicates that the bone plate in marginal diaphyseal fractures is not able to provide stable retention of fragments without contact between them, since all mechanical energy is concentrated on the screws, creating the prerequisites for implant migration, fracture, or the development of bone tissue dysregeneration.

External fixation devices with spikes have demonstrated fundamentally different biomechanical properties, providing high axial stiffness. The displacement of bone fragments under the conditions of use of this device did not exceed 0.05 mm at all studied levels of the femoral and tibial diaphysis. The tension on the supporting titanium rings of the device was consistently low and amounted to 0.1–0.3 MPa for all load variants, which confirms the high structural strength of the external frame, as shown in Fig. 5.

Data on the magnitude of stresses and displacements in the thigh and lower leg models when fixing the EX-FIX fragments on a spike base, depending on the type of load, are given in Table 2 and Figs. 6 and 7.

Table 2. Results of the study of models of the thigh and lower leg when fixing fragments of the EX-FIX on a spike base

Fracture level		Model element		Option 1 (compression)	Option 2 (frontal)	Option 3 (sagittal)	Option 4 (twisting)
Thigh	in/3	Stress, MPa	in rings	0.1	0.1	0.1	0.1
			in the spokes	10.1	30.4	10.9	11.5
			in the bone	1	1.2	1.1	1.2
		Movement of debris, mm		0,04	0.01	0.01	0.01
	with/3	Stress, MPa	in rings	0.1	0.2	0.1	0.1
			in the spokes	5.6	21.8	9.6	6.7
			in the bone	0.9	1.1	1	0.9
		Movement of debris, mm		0.01	0.01	0.01	0.01
	n/3	Stress, MPa	in rings	0.1	0.3	0.1	0.1
			in the spokes	5.1	20.3	8.7	5.3
			in the bone	1	1.6	1	0.9
		Movement of debris, mm		0	0.01	0.01	0.01
Shin	in/3	Stress, MPa	in rings	0.2	0.1	0.2	0.2
			in the spokes	6.2	5.7	6	5.9
			in the bone	1	0.9	1	1
		Movement of debris, mm		0.04	0.04	0.05	0.03
	with/3	Stress, MPa	in rings	0.3	0.1	0.1	0.2
			in the spokes	15.5	16.7	13.3	16.4
			in the bone	1.2	1	1.1	1
		Movement of debris, mm		0.05	0.04	0.05	0.05
	n/3	Stress, MPa	in rings	0.2	0.3	0.1	0.2
			in the spokes	15.3	30.9	15.1	18.8
			in the bone	1.2	1.5	1	1.1
		Movement of debris, mm		0.02	0.02	0.01	0.01

However, the analysis of the stress-strain state of the fixation elements (spokes) revealed areas of significant overload. In the case of a fracture of the upper third of the thigh, the stress on the spokes during axial compression was 10.1 MPa, but when a frontal load was added (option 2), it sharply increased threefold to 30.4 MPa. A similar pattern was observed in the middle third, which was equal to 21.8 MPa, and in the lower third of the thigh, 20.3 MPa.

When fixing the tibia, the highest stability of the implant-bone system was recorded in the upper third of the diaphysis, where the stress on the spokes did not exceed 6.2 MPa. However, in distal fractures (lower third), the stress on the spokes under the action of an additional frontal bending moment increased by twice compared to the pure axial load and reached a peak of 30.9 MPa, as shown in Fig. 5.

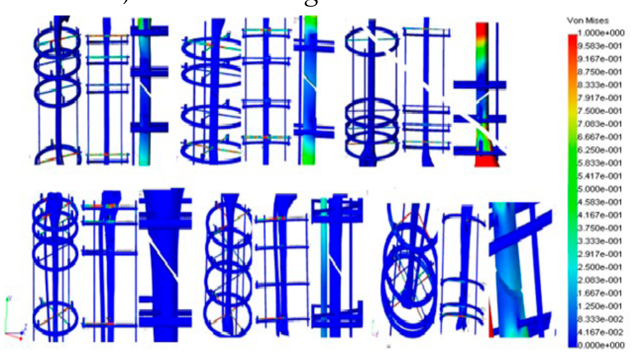

Figure 5. Stress-strain state of osteosynthesis models of femoral and tibial fractures under different loading options with an external fixation device on a spike base

This indicates that although the device provides reliable stabilization and physiological stress distribution, the action of additional bending forces in the frontal plane leads to critical overload of the thin wires. In clinical practice, this can cause elastic deformation and micromobility of the wires in the bone canal, leading to aseptic bone lysis and inflammation of the soft tissues around the metal structure.

The study of models with rod external fixation devices revealed a specific nature of the biomechanical behavior of the system. This stabilization option provides a sufficiently rigid fixation of the fragments, in which the maximum mutual displacements of the fragments do not exceed 0.25 mm under all loads (Fig. 8).

Data on the magnitude of stresses and displacements in the thigh and lower leg models when fixing the EX-FIX fragments on a rod base, depending on the type of load, are given in Table 3 and in Figs. 9 and 10.

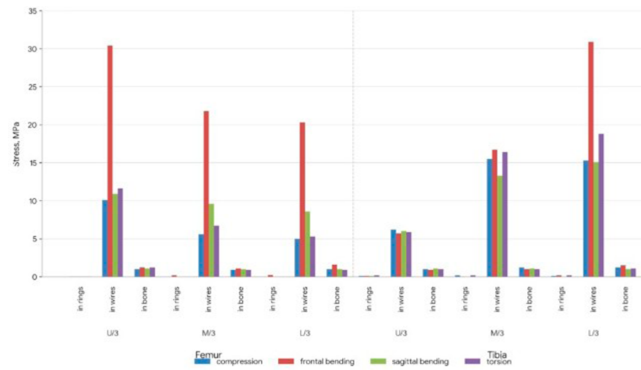

Figure 6. Stress values in models of extra-focal osteosynthesis of EX-FIX on a spike base of femoral and tibial fractures under different loading conditions

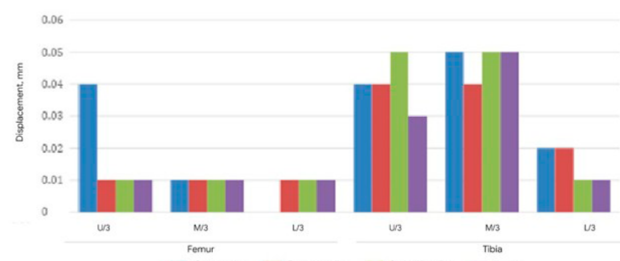

Figure 7. Displacement values in models of extra-focal osteosynthesis of EX-FIX at the spike base of femoral and tibial fractures under different loading conditions

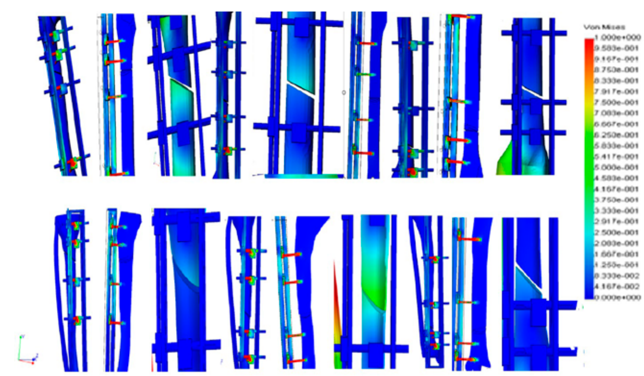

Figure 8. Stress-strain state of models of extra-focal osteosynthesis of EX-FIX on a rod base of femoral and tibial fractures under different loading conditions

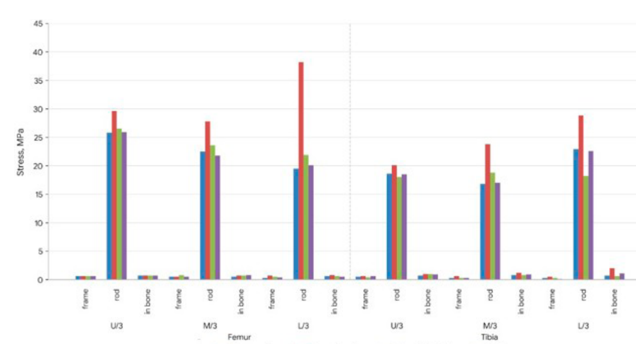

Figure 9. Stress values in models of osteosynthesis of femoral and tibial fractures using a rod-based external fixation device under different loading conditions

Table 3. Results of a study of fracture models of the tibia and femur when fixed with a rod-based EX-FIX

Fracture level		Model elements		Option 1 (compression)	Option 2 (frontal)	Option 3 (sagittal)	Option 4 (twisting)
Thigh	in/3	Stress, MPa	frame	0.6	0.5	0.6	0.6
			rod	25.8	29.6	26.5	25.9
			in the bone	0.7	0.7	0.7	0.7
		Movement of debris, mm		0.10	0.25	0.10	0.15
	with/3	Stress, MPa	frame	0.5	0.5	0.8	0.5
			rod	22.5	27.7	23.6	21.8
			in the bone	0.5	0.7	0.7	0.8
		Movement of debris, mm		0.08	0.25	0.02	0.01
	n/3	Stress, MPa	frame	0.3	0.7	0.5	0.4
			rod	19.5	38.2	21.9	20.1
			in the bone	0.6	0.8	0.7	0.6
		Movement of debris, mm		0.01	0.1	0.00	0.02
Shin	in/3	Stress, MPa	frame	0.5	0.6	0.4	0.6
			rod	18.6	20.1	18.1	18.6
			in the bone	0.7	1.0	1.0	1.0
		Movement of debris, mm		0.07	0.08	0.05	0.12
	with/3	Stress, MPa	frame	0.3	0.6	0.3	0.3
			rod	16.9	23.8	18.8	17.0
			in the bone	0.8	1.3	0.8	0.9
		Movement of debris, mm		0.06	0.08	0.07	0.10
	n/3	Stress, MPa	frame	0.3	0.5	0.3	0.2
			rod	22.9	28.8	18.2	22.7
			in the bone	0.7	1.9	0.6	1.2
		Movement of debris, mm		0.07	0.2	0.21	0.10

The study of models with rod external fixation devices revealed a specific nature of the biomechanical behavior of the system. This stabilization option provides a sufficiently rigid fixation of the fragments, in which the maximum mutual displacements of the fragments do not exceed 0.25 mm for all loading options. The stress on the external frame of the device ranged in the safe range from 0.2 to 0.8 MPa.

However, a high concentration of bending stresses was recorded on the fixation rods. For the femur, the maximum stress on the rods in the upper third of the diaphysis was 25.8–29.6 MPa, in the middle – 21.8–27.7 MPa, and in the lower third, under the influence of frontal bending load, it increased sharply to 38.2 MPa. In the tibia, the stress on the rods also remained significant, reaching a maximum of 28.8 MPa in the lower third of the diaphysis.

A feature of rod fixation is the extremely low level of stress directly in the bone tissue, which ranges from 0.5 to 1.3 MPa. This is significantly lower than in the case of EX-FIX fixation on a spike base and several times lower than in the case of plate osteosynthesis/ The rigid external beam takes on almost all the mechanical stress

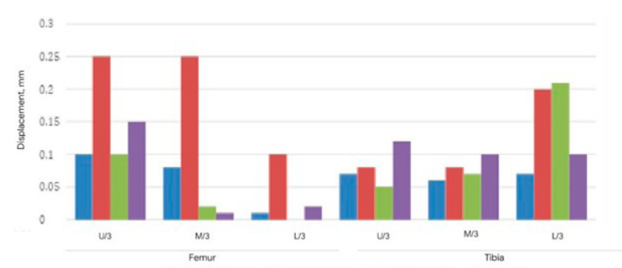

Figure 10. Displacement values in models of osteosynthesis of femoral and tibial fractures using a rod-based external fixation device under different loading conditions

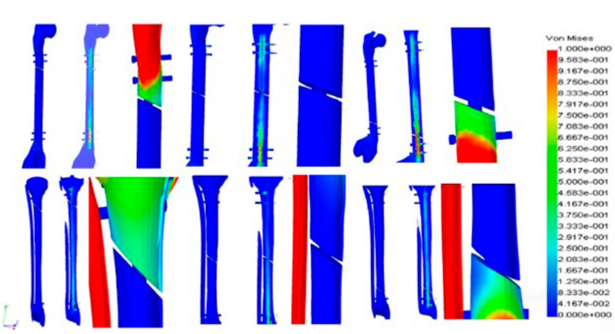

Figure 11. Stress-strain state of interlocking intramedullary nailing models of femoral and tibial fractures under different loading conditions

of the system, completely excluding the bone from the process of pressure transmission.

In the absence of adequate dynamic axial loading on the bone tissue, the processes of reparative osteogenesis are significantly slowed down. This creates a serious clinical risk of delayed union or formation of a false joint and can also lead to aseptic necrosis of the bone around the rods in the area of their contact with the cortex (Fig. 11).

IMN osteosynthesis has demonstrated optimal stress distribution in the implant-bone system in most fracture locations. Due to the location of the rod inside the medullary canal, the load axis coincides with the anatomical axis of the segment, which minimizes destructive bending moments.

Data on the magnitude of stresses and displacements in the thigh and lower leg models when fixing the fragments IMN, depending on the type of load, are given in Table 4 and in Figs. 12 and 13.

Table 4. Results of the study of models of the thigh and lower leg during fixation of fragments with an IMN

Fracture level		Model elements		Option 1 (compression)	Option 2 (frontal)	Option 3 (sagittal)	Option 4 (twisting)
Thigh	in/3	Stress, MPa	in the nail	0.7	2.4	1.2	0.7
			in screws	0.6	1.9	0.9	0.6
			in the bone	0.4	0.8	0.4	0.4
		Movement of debris, mm		0.17	0.2	0.21	0.2
	with/3	Stress, MPa	in the nail	0.7	2.5	1.1	0.7
			in screws	0.6	2	0.8	0.7
			in the bone	0.5	1	0.6	0.4
		Movement of debris, mm		0	0	0	0
	n/3	Stress, MPa	in the nail	0.6	2.6	1.3	0.7
			in screws	0.5	2.3	1	0.6
			in the bone	0.3	1.1	0.4	0.4
		Movement of debris, mm		0.18	0.33	0.25	0.2
Shin	in/3	Stress, MPa	in the nail	3.2	4.2	2.2	2.9
			in screws	3.2	4.2	2.2	2.9
			in the bone	1.9	2	0.6	1.6
		Movement of debris, mm		0.15	0.08	0.12	0.12
	with/3	Stress, MPa	in the nail	1.3	1.5	1.3	1.2
			in screws	1.3	1.5	1.3	1.2
			in the bone	0.4	0.9	0.6	0.6
		Movement of debris, mm		0	0	0	0
	n/3	Stress, MPa	in the nail	1.3	1.5	1	1.3
			in screws	1.3	1.4	1.32	1.3
			in the bone	0.8	0.8	0.6	0.6
		Movement of debris, mm		0.7	1.6	1.31	1.42

In case of femoral fractures, IMN osteosynthesis provides reliable fixation at all levels of the diaphysis. In the middle third, there is no mutual displacement of the fragments, and the stress is distributed harmoniously between the nail (0.7–2.5 MPa), screws (0.6–2.0 MPa), and the cortical layer (0.4–1.0 MPa). At the peripheral levels (upper and lower thirds) of the thigh, there is only a slight increase in the mobility of the fragments to 0.17–0.33 mm, which is within the physiologically permissible range of micromotions.

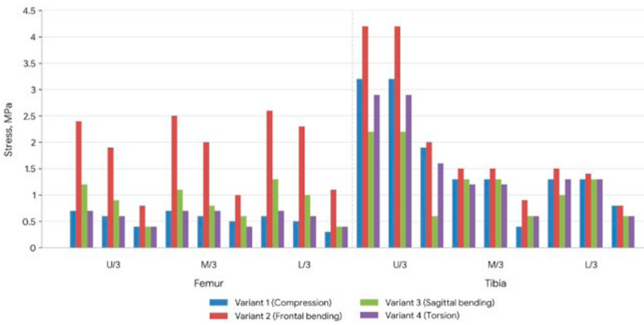

Figure 12. Stress values in models of femoral and tibial fractures under different loading conditions fixed with an interlocking intramedullary nail

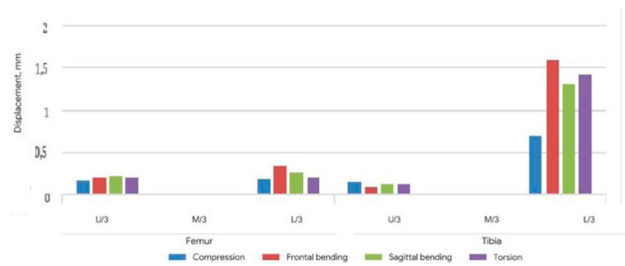

Figure 13. Displacement values in models of femoral and tibial fractures under different loading conditions fixed with an intramedullary locking nail

When fixing the tibia, the IMN also demonstrates high stabilizing properties in the upper and middle thirds of the diaphysis. However, when the lower third is fractured, a sharp destabilization of the system is recorded. The mutual displacement of the fragments during axial compression was 0.70 mm, and under the influence of an additional bending moment in the frontal plane, it increased to a critical 1.60 mm.

This phenomenon is explained by an anatomical discrepancy in the distal segment of the tibia, where the medullary canal significantly expands, passing into the metaphysis. The 10 mm diameter intramedullary nail in this area is devoid of dense cortical support and is fixed cantilevered only on locking screws. This provokes the occurrence of rotational and angular instability of the "bell" type, significantly increasing the risk of fracture nonunion and screw failure.

DISCUSSION

Treatment of patients with lower limb injuries is a very difficult task today, as doctors are increasingly faced not with isolated fractures, but with the consequences of severe tissue destruction. This is clearly confirmed by the clinical data of Tanasiienko et al., who draw attention to the high severity and polystructural nature of skeletal injuries in the civilian population during modern hostilities. Under such circumstances, surgical care should be based on the principles of orthopedic Damage Control[9]. This approach is fully consistent with the strategy proposed by Berwin et al. regarding the mandatory staged surgical management of patients with polytrauma to preserve their lives[12].

However, primary stabilization is not just a temporary solution but a biomechanical foundation. The importance of this choice is emphasized by Guryev et al.: according to their observations, errors in the tactics of limb fixation in multisystem injuries are one of the key factors leading to forced amputations. Our mathematical modeling results complement these conclusions and give them a concrete justification: they show which structures can withstand the combined pressure during the patient's motor activity, and which ones pose critical risks for bone union[6].

When we evaluate regeneration processes, it becomes apparent that mechanical loading directly governs the biological behavior of tissues. The study by Litvishko et al. directly confirms our findings that the stress-strain state of the periosteum and fibrin-blood clot in the fracture zone fundamentally determines the future structural organization of the bone callus. However, the nature of this influence depends on the chosen implant[17].

The excessive rigidity of monolateral rod external fixation devices that we have identified leads to the fact that the stress in the bone itself drops to minimal levels, 0.5–1.3 MPa. This phenomenon of "stress shielding" directly contradicts the biological conditions of normal osteogenesis described by Litvishko et al., since the critical underloading of the bone inhibits mechanotransduction processes[17]. Instead, controlled axial micromobility was recorded in our study with pin fixation (Δ=0.01–0.05 mm), fully complying with the biological principles of these authors, creating a delicate balance of elastic deformations necessary to stimulate regeneration.

Understanding the mechanical limits of implants allows us to see the root cause of unsatisfactory treatment results in practical medicine. A statistical analysis by Popsuyshapka et al., based on a large dataset of medical and social expertise, revealed a high incidence of delayed consolidation and false joints in the treatment of diaphyseal fractures of long bones. This alarming clinical trend finds its clear comparative confirmation in our finite element models[8].

We found that in low tibia fractures (n/3), the popular method IMN osteosynthesis demonstrates critical mutual displacement of fragments to 1.42–1.60 mm due to anatomical widening of the bone canal ("bell effect"). At the same time, osteosynthesis with a plate at the same level causes an escalation of stresses on the screws to a critical 6,7 MPa. Both of these biomechanical facts explain in detail the structure of real complications recorded by Popsuyshapka et al.[8]. Our results prove the failure of the "universal implant" template and force us to reconsider surgical protocols in favor of a differentiated choice of construction depending on the exact topography of the diaphyseal fracture.

CONCLUSIONS

Summarizing the data obtained, the following conclusions can be drawn:

1. Under the conditions of cumulative (bone) osteosynthesis with plates, there is a pronounced eccentricity of the application of axial load, due to the lateral location of the fixing beam relative to the neutral axis of the bone segment. In the absence of primary interfragmentary contact, bending stresses accumulate in the fixing screws located in proximity to the fracture zone. This leads to a complete leveling of the damping function of the bone and blocks the physiological transmission of pressure through the bone tissue.
2. External fixation rod devices exhibit excessive mechanical stiffness. The large distance from the support frame to the bone axis under cantilevered insertion of rods creates a significant force lever, which leads to the concentration of high stresses in the metal rod, while the bone tissue experiences critical underloading. This phenomenon of classical stress shielding prevents osteoregeneration by inhibiting mechanotransduction in the fracture zone.
3. External fixation wire devices provide a rational biomechanical balance: a high level of pretension of thin wires guarantees sufficient axial rigidity in combination with the possibility of controlled microdisplacements of bone fragments. This stimulates the processes of reparative osteogenesis due to elastic deformation of structural elements under axial loading. At the same time, under the conditions of pronounced shear and bending forces in the frontal plane, critical peak overloads of the wires occur, which limit the application of this technology in complex rotational instability.
4. IMN osteosynthesis is the most biomechanically balanced method for diaphyseal fractures of the femur (at all levels) and the middle/upper third of the tibia. The location of the nail along the axis of the bone segment ensures a symmetrical distribution of compressive forces between the implant and the cortical layer, which allows to preserve the damping capacity of the bone and protects the fixation elements from premature failure.
5. Comparative integration of the results of mathematical modeling allows us to reveal the fundamental mechanisms of interaction between bone tissue and metal fixators in the absence of primary stability of fragments.
6. Numerical analysis allowed us to formulate clear clinical recommendations for the selection of metal structures and planning of surgical tactics for combined injuries of the lower extremities.

Prospects for further research.
Further studies are planned to investigate the options for osteosynthesis of the tibia and femur at similar levels and with similar fixators in overweight or obese victims. The results of the studies contribute to the optimal choice of osteosynthesis agents for diaphyseal fractures of the lower extremities against the background of overweight.

Ethics Statement and Conflict of Interest Disclosures

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Ethics Consideration: The authors declare that all the procedures and experiments of this study respect the ethical standards in the Helsinki Declaration of 1975, as revised in 2008(5), as well as the national laws. This study was approved by the Institutional Research Board and Ethics Committee.

Conflict of interest: No known conflict of interest is associated with this publication.

Availability of data and materials: The data used and/or analyzed throughout this study are available from the corresponding authors upon reasonable request.

Competing interests: The authors declared that they have no competing interests.

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