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Rheological models of trabecular structures joint implants obtained by additive processes

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Abstract. *Features of the structure of trabecular structures for the manufacture of implants of hip and knee joints and rheological models that can be used as a basis for analyzing the dynamics of biomechanical systems "bone-articulated implant" are considered. It is taken into account that the implant itself should be made in the form of a combined set of functional elements (or initial surfaces), the dynamic properties of which are variable and maximally adapted to the properties of the connected bone, which allows to preserve the initial properties of the bones of the joints as much as possible while ensuring the proper strength and reliability of the structure in as a whole The interdependence of the results of surgical intervention with the patient's initial condition, indications for treatment, his activity and possible postoperative complications was analyzed. An optimization function of the process of designing, manufacturing and operational support of implantation, which has probabilistic components, is proposed.*

It is shown that it is most appropriate to use the Burgers model when studying the dynamics of the "bone-articulated implant" components, and the trabeculae density coefficient can be a generalized characteristic of the formed trabecular structures, provided that the geometric parameters match the bone tissue.

Keywords: *trabecular structures, implants of hip and knee joints, dynamic parameters, rheological models, additive processes, provision of functional requirements.*

Relevance and statement of the problem

Joint diseases (arthritis) are among the ten most common diseases in the world [1]. Medical or surgical treatment of patients is traditionally used to fight diseases.

Drug therapy, according to [2], is limited to the symptomatic administration of anti-inflammatory and analgesic agents [3], as well as partially intra-articular and partially systemic treatment with hyaluronic acid, chondroitin sulfate, interleukin-1 receptor antagonists, and glucosamine sulfate. Despite the good results shown in reducing pain, it has not yet been possible to prevent the progression of arthritis with this agent [4].

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More common treatments are surgical treatments such as Drilling with guide elements [5], anterograde/retrograde drilling, and microteaching [6]. These surgical treatments do not involve local cartilage replacement; instead, a series of holes are made through the subchondral border plate. In the case of the guides drilling and its subsequent development in the form of a "micro-rupture", blood can penetrate the cartilage defect in the area of the defect, and thus fibrocytes, mesenchyme stem cells and chondroblasts are carried from the spongy compartment into the cartilage defect. Together with growth factors, these species form a blood clot and differentiate into articular cartilage [5].

Clinical studies have shown a reduction in pain and high joint mobility [7]. However, a problem remains with prolonged rest or immobilization of the joint, during which the regenerative quality of the developing fibrocartilage is generally poor. Due to its structure, this fibrous cartilage is often insufficient for high mechanical loads, especially in the knee joint, and undergoes rapid degeneration, which may require further surgical interventions.

Arthroplasty is the replacement of a joint, such as a knee joint, with an artificial one [8]. When a total natural knee replacement is performed for the first time, it is called a primary total knee replacement, or pTKR. However, a pTKR can fail for a number of reasons, including loosening, poor performance, wear, and infection, and in some cases the pTKR will need to be replaced. Revision total knee replacement, or rTKR, is usually performed to replace such a failed pTKR. Typically involves removing the entire pTKR and implanting a set of new rTKR components into the diseased or damaged bone. Both rTKR and pTKR aim to place the implants in a predetermined position and orientation relative to the native anatomy to restore knee function as best as possible [7].

Despite the efforts of surgeons and researchers around the world, the problem of high-quality selection of implants remains particularly relevant. At the same time, the active development of additive processes and technologies opens fundamentally new opportunities in the manufacture of implants themselves and tools for surgical intervention: now it is possible to take into account the individuality of the anatomy and mechanical load of the joint, which is manifested in the patient's lifestyle, on the basis of previously collected information, to pay attention to the presence of some chronic diseases, as well as bone damage before implantation. However, the use of additive processes and, accordingly, the means of implantation reproduced with their help, requires a detailed and comprehensive study of the dynamics of the products, the conditions of their connection with the patient's biological tissues, because the product can have a structure as close as possible to the natural state.

From these positions, the search for means of effective and adequate description of the behavior of the biomechanical system "bone-implant", the justification of the use of appropriate rheological models is an important scientific and practical task, taking into account the significant need for implants in patients after military operations.

The purpose of the work

Development of a dynamic model of the system "implant-bone tissue of the patient", which would take into account the presence of trabecular structures and functional films and coatings of the contact surfaces.

Theoretical foundations

The effectiveness of surgical treatment of arthritis by partial or complete replacement of damaged tissues depends on many factors, which can be conditionally classified into the following groups [9], [10]: anatomical features of the body (Φ_1); the presence of chronic diseases, joint injuries that are subject to surgical treatment (Φ_2); methods and techniques of surgical intervention (Φ_3); rehabilitation (Φ_4), including the possibility of immobilization of movable joints); methods of initial diagnosis (Φ_5); means of reproduction and construction of components of endoprostheses (Φ_6). In this case, the cause-and-effect diagram in its general form will be similar to Fig. 1. The factor of the maximum duration of the implant's functioning (P_1) is taken as the target function, but the same function can be another, for example, minimal trauma, minimization of the rehabilitation period, etc.

The diagram also shows the second-level factors taken into account, which in turn can also be divided into separate sub-levels. In this case, the objective function will have the form:

$$P_1 \subset \Phi_1 \cdot \Phi_2 \cdot \Phi_3 \cdot \Phi_4 \cdot \Phi_5 \cdot \Phi_6 \to \max . \quad (1)$$

The expression will also be valid for the case when the corresponding components of Φ_i are represented in the form of arrays of parameters (or properties) of the modeled system, to the components of which approaches of set algebra

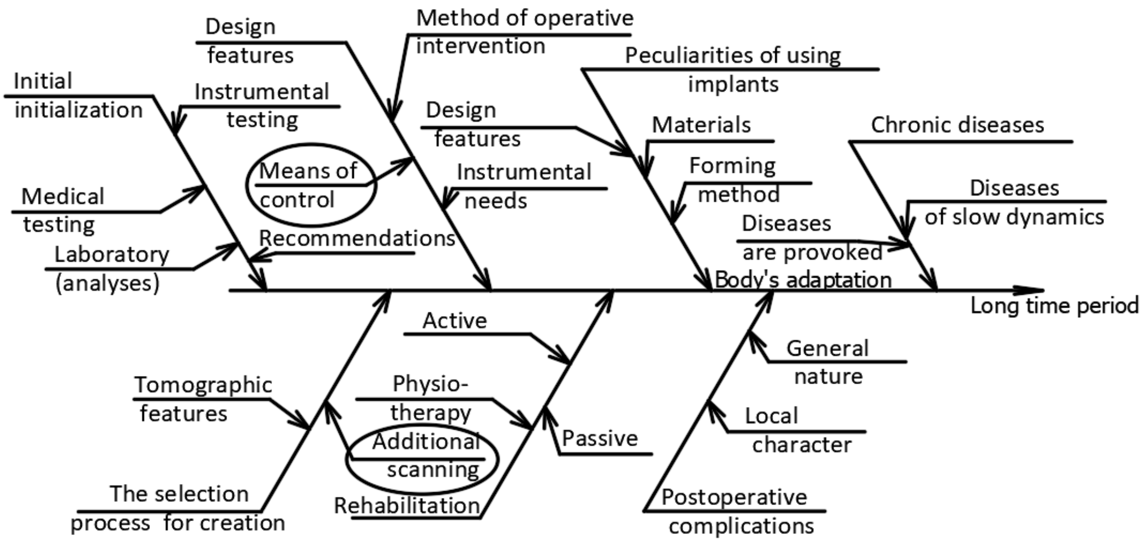


Fig. 1. Cause and effect diagram of the effectiveness of surgical treatment of arthritis

can also be applied. Such components can be both constants (with unchanged properties), and functions, as well as probabilistic ones. That is, for the first two cases we have:

$$K_{F1}^{i} = C; \quad K_{F1}^{j} = f\left(x_1, \ldots x_n\right). \tag{2}$$

For the random components characterizing, for example, the Φ_2 component (the presence of chronic diseases or the mechanical properties of a damaged joint), the tensor of the elastic moduli $\lambda_{ij\alpha\beta}$ of the bone can be taken into account, which will determine the tensor random field.

We will assume that the tensor random field $\lambda_{ij\alpha\beta}(x_i^{(m)})$ is defined within a certain domain v, if each finite system of points $x_i^{(1)}$, $x_i^{(2)}$,…, $x_i^{(n)}$ from the domain v is matched by $N \cdot n$ – dimensional law of probability distribution of values $\lambda_{ij\alpha\beta}$, where N is the number of independent components of the elasticity tensor. The area v itself geometrically corresponds to the shape of the bone, and can be specified by polynomials of higher orders. In the case of isotropy of the properties at the points of the limited space, the tensor λ will be characterized by two (λ and μ) constants, so we will have a $2 \cdot n$ – dimensional probability distribution law for the values λ and μ. In this case, the distribution density will be:

$$K_{F2}^{k} = f_2\left(x_1, \ldots x_n\right) f_2^{n}\left[\lambda\left(x_i^{(1)}\right), \mu\left(x_i^{(1)}\right), \ldots \lambda\left(x_i^{(n)}\right), \mu\left(x_i^{(n)}\right)\right]. \tag{3}$$

Then the optimization function for the Φ_1 component from (1) taking into account (2) and (3) will have the form (provided that the strength of the “Implant-bone” structure must be maximal and uniform:

$$P_{F2} = K_{F2}^{i} \cdot f_1\left(x_1, \ldots x_n\right) \cdot f_2\left(x_1, \ldots x_n\right) f_2^{n}$$
$$\left[\lambda\left(x_i^{(1)}\right), \mu\left(x_i^{(1)}\right), \ldots \lambda\left(x_i^{(n)}\right), \mu\left(x_i^{(n)}\right)\right] \rightarrow \max .$$

The mechanics of the implant elements and the human-implant biomedical system are generally expressed as follows. The “human-implant” system can be defined through the components of the set Φ_1, Φ_2, Φ_6. We consider the components of the sets Φ_3, Φ_4, Φ_5 to be functionally conditioned or independent. So:

$$\Phi_3 \subset \Phi_1 \cdot \Phi_2; \quad \Phi_4 \subset \Phi_1 \cdot \Phi_2 \cdot \Phi_6; \quad \Phi_5 \subset \Phi_1 . \tag{4}$$

Thus, the conditions for modeling the biomechanical system “man-implant” are the components $\Phi_1 \cdot \Phi_6 \cap \Phi_2$. The latter has a probabilistic character, as it changes over time τ:

$$K_{F2}^{k} = \cdots(\tau) .$$

Existing models of joints (similar to hinges and articulations) are based on the properties of the material and the characteristics of the product as a whole, which are unchanged and correspond to compact materials. At the same time, the biomechanical system as a whole has a number of features that need to be taken into account when creating appropriate models.

Thus, the analysis of well-known rheological models given in [11] shows that the Saint-Venant-Coulomb model of an ideal plastic body (a solid body located on a plane, during the movement of which friction is constant and does not depend on the normal force), Fig. 2 *a*, quite often used to describe pathologically changed joints of the human musculoskeletal system.

This model is based on the law of external (dry) friction, according to which there is no deformation if the shear stress is less than a certain value σ_t, which is called the yield point. If the load reaches the yield point, then the deformation of the perfectly plastic body develops, which has no limit, and the flow occurs at any speed.

The value σ_t reflects the strength of the body structure: $\sigma < \sigma_t$, $\gamma > 0$.

When the yield point is reached, the deformation of a perfectly plastic body develops, which has no limit, and the flow occurs at any speed: $\sigma = \sigma_t$, $\gamma > 0$. At $\sigma = \sigma_t$, the structure of an ideal plastic body is destroyed, after which the load resistance is completely absent.

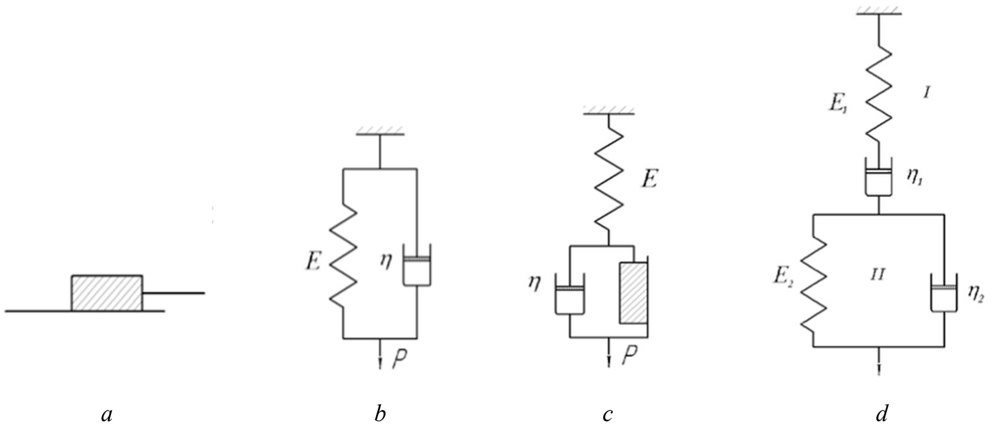


Fig. 2. Model of an ideal plastic Saint-Venant-Coulomb body and the dependence of the deformation of this body on stress (*a*); body of Kelvin – Voigt (*b*); Bingham body model (*c*) and Burgers body (*d*)

However, to describe the behavior of real joints, it is advisable to use more complex rheological models, for example, the Kelvin-Voigt model (a model of a viscoelastic body, which is able to restore its properties after removing the load) or the Bingham model (in which a body that exhibits elastic or elastic properties depending on stress). This is especially relevant for the analysis of contacting surfaces, when the contact itself occurs through the surface film [12].

In the Kelvin–Voigt model, it is taken into account that a viscoelastic body is able to restore its properties after removing the load (elasticity). Such a model is formed as a result of the parallel connection of Hooke's body and Newton's body (Fig. 2 *b*), where the dependence of relative deformation on time is also taken into account. Such models are characteristic of cases when, after removing a constant load, the sample will slowly return to its original shape, which corresponds to an exponential curve. The behavior of the body is described by the following differential equation:

$$\sigma_{\Sigma} = \gamma E + \eta \frac{d\gamma}{dt}, \tag{6}$$

the solution of which for the case of stretching under a constant load allows us to determine $\gamma = \frac{\sigma}{E}\left(1 - e^{-\frac{t}{T}}\right)$, where $T = \frac{\eta}{E}$ is the creep time, $\gamma_{\infty} = \frac{\sigma}{E}$ is the relative deformation.

Bingham's model (Fig. 2 *c*) predicts that at low stresses ($\sigma < \sigma_t$) only elastic deformations develop (similar to Hooke's body). When stress $\sigma > \sigma_t$ is reached, plastic deformation takes place, which grows infinitely (flow of a viscous-plastic body). The mathematical model of a visco-plastic body (the Shvedov-Bingham equation) has the form: $\sigma = \sigma_{\gamma} + \eta \frac{d\gamma}{dt}$, where η' is plastic or structural viscosity, Pa·s; σ_t – yield strength, Pa.

An interesting solution can be the use of Burgers models, the scheme of which is shown in Fig. 2 *g*, and the integral view of the model is as follows:

$$\frac{\eta_2}{E_1}\sigma + \left(1 + \frac{E_2}{E_1} + \frac{\eta_2}{\eta_1}\right)\int_t \sigma dt + \frac{E_2}{\eta_1}\int\left[\int_t \sigma dt\right]dt = \eta_2\gamma + E_2\int \gamma dt. \tag{7}$$

The creep equation of the Burgers model [10] will look like this:

$$\gamma = \frac{\sigma}{E_1} + \frac{\sigma t}{\eta_1} + \left(\frac{\sigma}{E_2}\right)\left[1 + e^{-\frac{E_2 t}{\eta_2}}\right].$$

The above models are simplified versions of connections that do not have important interlayer components or surface films, as a result of which the adequacy of such models is quite limited and requires clarification. At the same time, when considering an implant as a subsystem of long-term operation, the processes of changing individual characteristics of the "implant-bone" system become important, due to: 1) changes in the contact conditions of implant elements (wearing of surface layers, changes in contact conditions, etc.), 2) changes fixation of elements on the patient's bones.

Hip (Fig. 3) and knee (Fig. 4) joints have structural differences, as well as differences in degrees of freedom. For the knee joint, we can consider a system with two rotational degrees of freedom *a* (with a rotation angle α), *b* (rotation angle β) and one linear χ (movement x along the

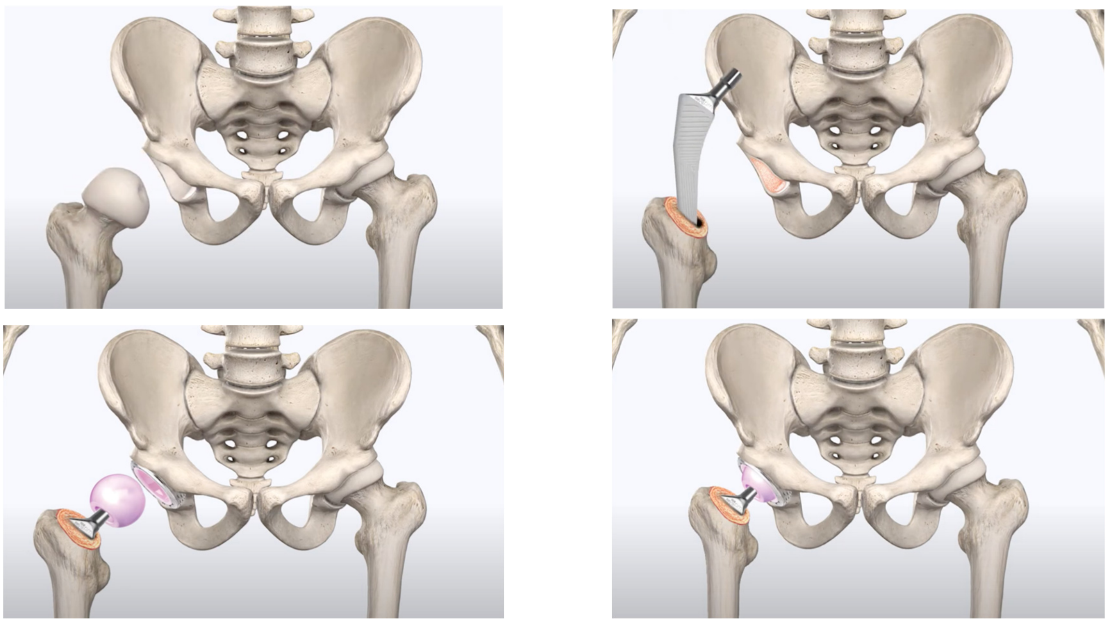


Fig. 3. Implantation of the hip joint (Internet ill.)

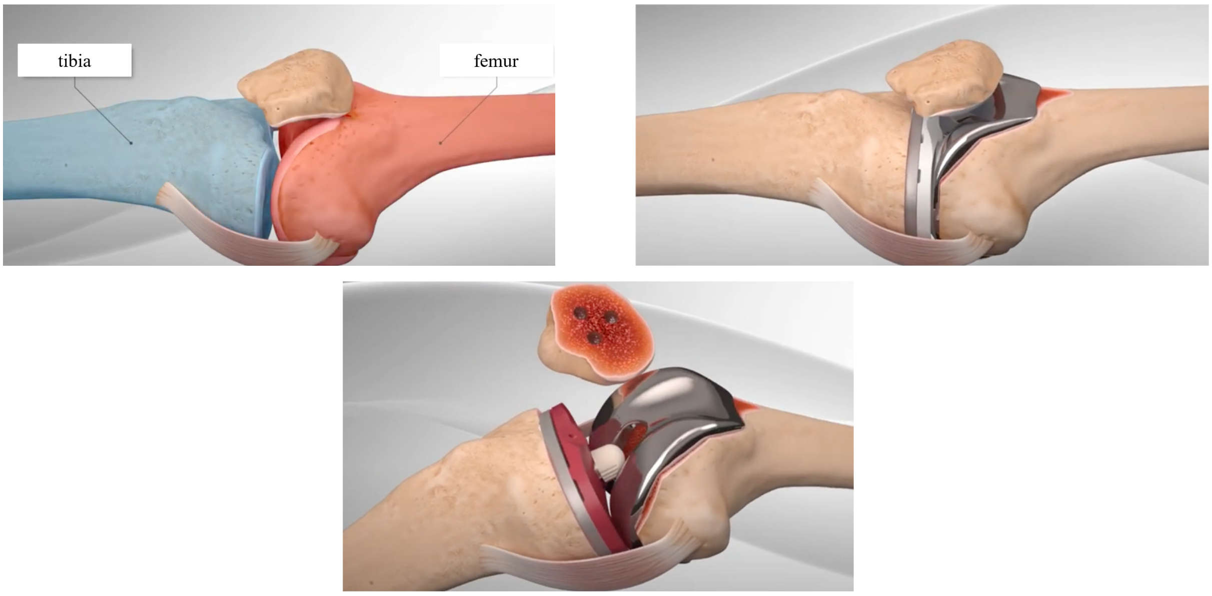
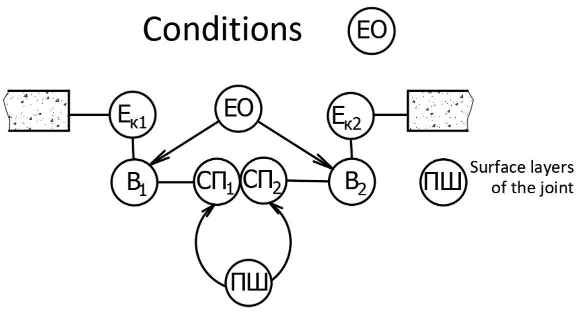


Fig. 4. Implants in the knee joint (Internet ill.)

tibia); the hip joint due to the contact of two conjugate components at an angle δ has two linear orthogonal coordinates – ξ (movement x along the tibia) and ζ (movement z along the patient's spine). We take into account the final viscosity of the coupling components, as well as the presence of an elastic-plastic layer on the contact surface. We will assume that the combination of implant components corresponds to the scheme of Fig. 5 *a*.



a

b

Fig. 5. Combination of implant elements (*a*) and applied levers (*b*)

Here E_{k1}, E_{k2} are elements of attachment to bones; $СП_1$, $СП_2$ – connecting the bone with the implant (cement); B_1, B_2 are power levers for load redistribution. The latter are determined by the type of implant and can be both short and of significant length (in the case of installation in the patient's bone cavity, Fig. 6, *b*).

Then, in the first approximation, you can use the corresponding rheological model of Burgers, adjusting it according to the scheme of Fig. 6.

It is marked here: M_1, M_2 – the points of contact of the mass of the components (implant); c_b, c_o, c_k – given elasticity coefficients of the corresponding components of the implant and interlayers, which are determined based on the rheological models taken into account (6, 7); b_b, b_o, b_k – damping coefficients (including at the "implant component – bone" junction). The influence of the environment (in particular, connective tissue and muscles) is taken into account by the dynamic coefficients c_{fi} and b_{fi}.

Additional conditions in the dynamic model should include considering the kinematic limitation of the movement of the joint pair in the longitudinal direction (for the hip joint – axes ξ and ζ, Fig. 7, which requires taking into account non-linearities of the discontinuous type.

The conjugation contact between endoprosthesis elements 1 and 2 can be determined by using the representation of the contact of bodies with the existing spatial deviations of the surfaces in the form of hemispherical convexities, the normal of which are defined as $n_1, n_2, ...n_i$.

Such bulges are determined by the waviness of the surfaces formed by additive processes. Determining the deformations of such elements is reduced to solving the Hertz problem for the contact of two bodies with curvature r_i,

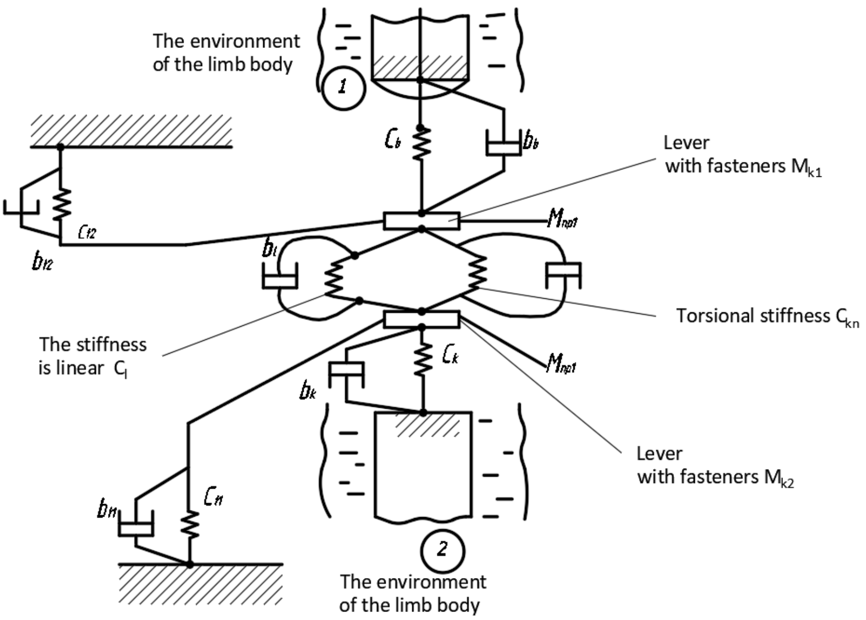
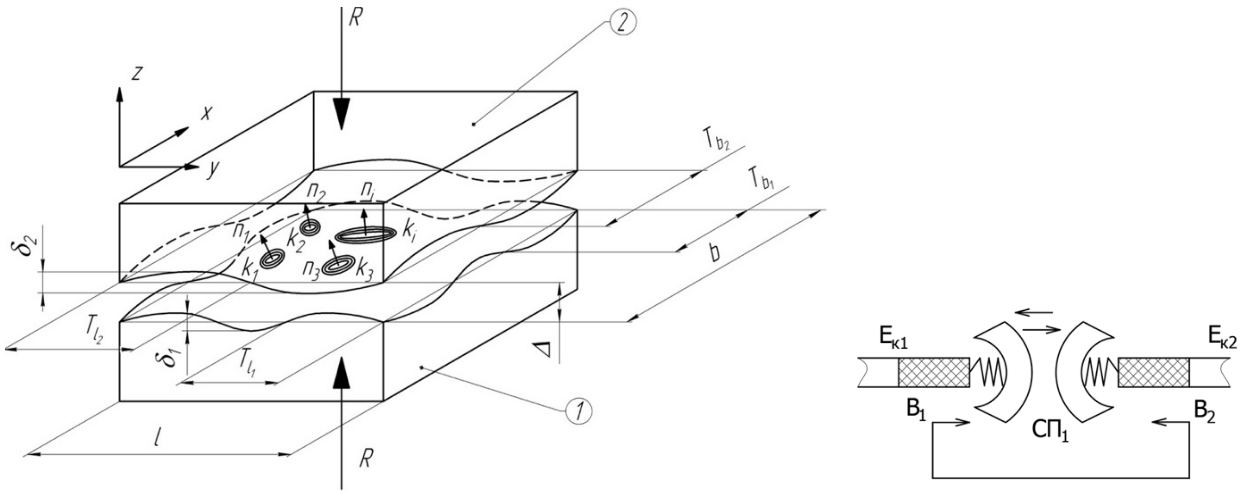


Fig. 6. Joint model with implants according to Burger's assumptions



a *b*

Fig. 7. Contact of the mating surfaces in the joint (*a*) and kinematic restriction of the movement of the joint pair (*b)*. Here *E* is the reduced modulus of elasticity of the surface layer of the contacting elements

which is determined by the location of the contact points on the plane $b \times l$, and protrude above the surface by the amount δ_1 and δ_2, and the contacting surfaces themselves are loaded with a compressive force R.

Then, for the simultaneous number of contact points $n = N$, the force R will be divided between the contact points, and at one-point k_i, $(k_i = R/n)$

$$k_i = \frac{4}{3} E^* \sqrt{r} d^{\frac{3}{2}}. \quad (8)$$

Here d is the penetration depth of the contacting surfaces at the point of contact, E^* will be determined by the equation:

$$\frac{1}{E^*} = \frac{1-\nu_2^2}{E_1} + \frac{1-\nu_2^2}{E_2},$$

and the reduced radius r will be determined by the radii of the surface projections, $\frac{1}{r} = \frac{1}{r_1} + \frac{1}{r_2}$. This makes it possible to determine the distribution of stresses on the contact sections according to the maximum value $p_0 = \frac{2}{\pi} E^* \sqrt{\frac{d}{r}}$:

$$p = p_0 \left(1 - \frac{r^2}{a^2}\right)^{\frac{1}{2}}. \quad (9)$$

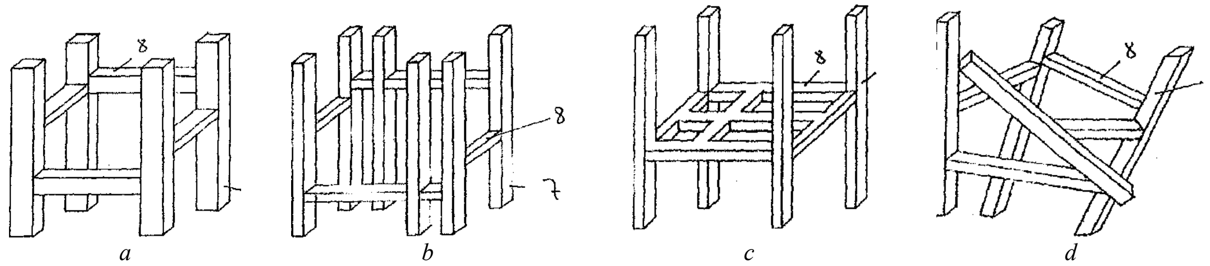


Fig. 9. Trabecular structures of orthogonal (*a–c*) and spatial (*d*) execution (according [7])

To understand the magnitude of the loads on the patient's bone and to determine the rational conditions for the formation of the properties of the surface layers, a comparison of the response of the biomechanical system to the external load is necessary. Such a load can be a test impact R (described according to the Heaviside function), and the reaction - the maximum value of the resulting stresses σ (or deformations γ_∞) and the rate of impact energy absorption in the subcritical region. The equations of movement of the implant elements in the attachment to the operated bones at the moment of loading the joint will correspond to the D'Alembert principle, taking into account the rheological models of contact in the articulation of the implant, as well as in the two points of fixation of the implant on the bone:

$$m_{k1}\frac{d^2x_1}{dt^2}=R-c(x_2-x_1)-b\frac{d}{dt}(x_2-x_1),$$
$$m_{k2}\frac{d^2x_2}{dt^2}=R-c(x_2-x_1)-b\frac{d}{dt}(x_2-x_1). \quad (10)$$

And for the rotation (α) of the conjugate surfaces of the implant, the equation will have the form

$$J_k\frac{d^2\alpha}{dt^2}=T-c(\alpha)-b\frac{d}{dt}(\alpha).$$

Coordinates l_ξ (movement *x* along the tibia) and l_ζ (movement *z* along the spine of the patient) are recalculated according to the scheme of force interaction of the implant in the patient's skeleton,

$$l_\zeta = x_2\cos(\delta_1),\ l_\xi = x_2\sin(\delta_2). \quad (11)$$

Elastic $c(x_2-x_1)$ and dissipative $b\frac{d}{dt}(x_2-x_1)$ components correspond to the case of application of the Kelvin–Voigt rheological model, (6); for other cases (Burgers model), taking into account nonlinear frictional forces and plastic deformation of implant components, the component of equations (10) $F=c(x_1-x_2)+b\frac{d}{dt}(x_2-x_1)$ will take the form:

$$F_b=\frac{c_bc_k}{c_b+c_k}(x_2-x_1)+\frac{b_bb_k}{b_b+b_k}\frac{d}{dt}(x_2-x_1)-$$
$$-[\mu N]sign\frac{d}{dt}(x_2-x_1). \quad (12)$$

Gradient printing was performed by forming the trabecular structures of the implant body in the form of forms recommended [4] and shown in Fig. 9.

Parameters of reproduced structures (Fig. 10): thickness of orthosupports δ, mm; trabecula height h, mm; the dimensions of the trabecula in the projection of the applied load $b\times b$, mm; for trabeculae of a spatial form – the angles in the plane γ and φ, as well as the angle ε relative to the load axis; the cross-section of the elements that make up the trabeculae is $d\times d$. Connection conditions: all reproduced trabeculae have the same area of connected elements both in the layout projection, i.e. dimensions $b\times b$, and in height. The change in the parameters of elasticity and plasticity, which will accordingly affect the deformability of the structure as a whole, is provided by the height of the trabecula h, as well as the cross-section of its elements $d\times d$. At the same time, the reduction or increase of the crossing occurs symmetrically relative to the central axis of the connecting element. The change limits of the corresponding components and examples of printed trabeculae are given in Table 1.

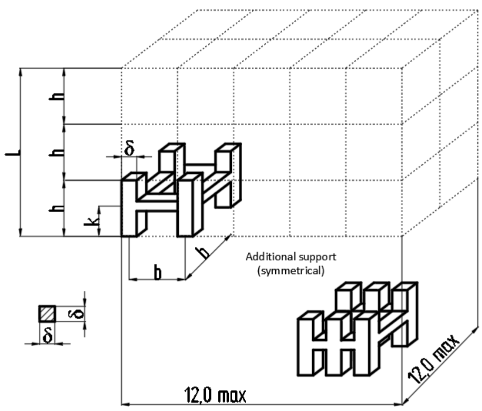
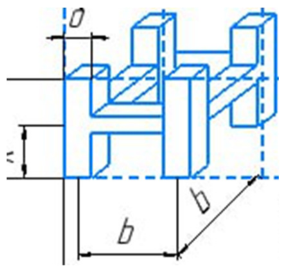
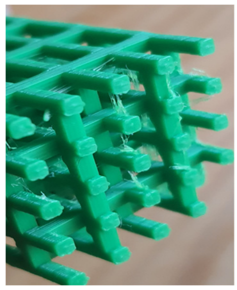
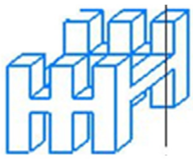
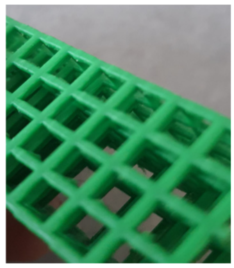


Fig. 10. Parameters of trabecular structures and their combination in a demonstrator product for mechanical tests

Table 1. Typical trabecular structures reproduced by means of 3-D printing

No	Characteristics, mm		Element sketch	Printed demonstrator
	δ	0,4…1,3		
	b×b	1,0…4,8		
a)	h	1,1…2,5		
	k	0,3…0,8		
	l	12,0…18,0		
	δ	0,4…0,8		
	b×b	1,0…3,2		
b)	h	1,1…2,5		
	k	0,2…0,4		
	l	12,0…15,0		
	δ	0,4…0,6		
	b×b	1,0…3,2		
c)	h	1,1…2,0		
	k	-----		
	l	12,0…18,0		
d)	δ other according at previous	0,4…1,3		

A 3D FDM printer of the KLEMA-2 type was used to print trabecular structure demonstrators, see Table 2, as well as an improved PRUSA printer with a high-temperature head (printing temperature up to 450 °C, which is necessary for working with REEK-type plastics); bone structures were studied on the demonstrators using a 3D REVOPOINT scanner with appropriate software; analysis of structures was carried out using an electronic microscope of the REM-106 type, mechanical tests were performed on tearing machines P-20 as well as on a test bench to determine low- and multi-cycle fatigue of control samples. typical characteristics of printers are given in the Table 3.

For the demonstrators, materials were used in accordance with the recommendation [7], namely: polymers that are not biodegradable, in particular polyamide (PA), polyether ketones, in particular PEK [polyether ketone], PEKK [poly-(ether ketone ketone)], PEEK [polyetheretheretherketone], polyethylene (PE), in particular UHMWPE [ultra-high molecular weight polyethylene], or, for example, bioresorbable polymers, in particular PCL [poly-caprolactone].

Discussion of research results

At the first stage, model experiments were conducted on the developed dynamic model (10), (11), based on rheological models (6), (7), which were reflected in the corresponding components (12) of the differential equations of motion.

As the initial load for the dynamic system, the impulse shock load of the demonstrator *F* was chosen, and the response of the system was determined by the movement of the implant components at the connection point and by the resulting stress amplitudes on the contact surfaces *R*(*t*). Variable parameters were selected: the shape of the trabeculae and their geometric parameters, which determine the dynamic coefficients c and b; properties and thickness of surface films. The obtained transient processes in the form of reactions to the external load are shown in the Table 4.

Modeling conditions and reproduced transient processes of various trabeculars selected as demonstrators according to Table 1. The load function is shock, the duration of shock contact is 0.01 s. VT-5 Titanium was chosen as the material for research, the physical and mechanical properties of which (as a compact material) are available in Solid Works libraries.

The results of modeling made it possible to draw several important conclusions.

The use of Bingham's rheological models is appropriate when modeling trabecular structures that have high stiffness, density (due to parameters c_b , c_o , c_k ,b_b , b_o , b_k) and are closer to a solid compact body. The Voigt model showed worse convergence and the presence of residual stresses in the studied body.

Mathematical models based on Burgers' proposition proved to be the most adequate, satisfactorily described the change in the state of the studied body, and were more resistant to the accepted algorithmic steps of integration.

A comparison of the dynamic patterns of shock damping (3) and (5) shows that the greater damping capacity of trabeculae allows for faster damping of oscillatory phenomena on the contact surfaces of the joint model; the absence of surface films and their replacement with a compact material can cause the opening of the contact pair.

Table 2. Used printers for 3-D printing

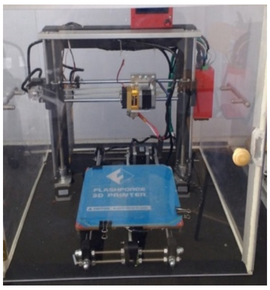
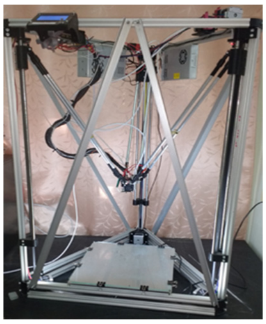
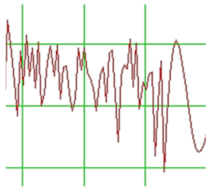
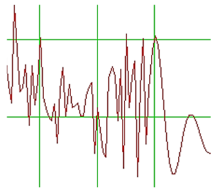
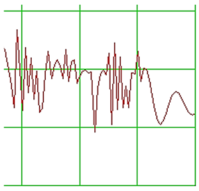
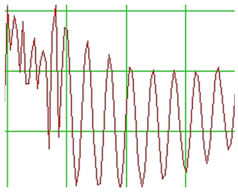
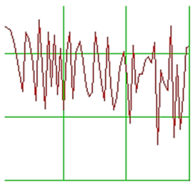
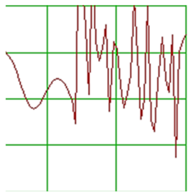
No	General appearance	Features of the extruder	Application
PRUSA I3		Direct extruder, filament heating up to 430 °C, water cooling	Printing with high-temperature reinforced plastics, direct extruder*
DELTA		Bowden extruder, filament heating 250 °C, air cooling	Printing of simple axisymmetric shapes, Bowden extruder*
KLEMA		Bowden extruder with a short supply link, filament heating up to 230 °C, air cooling	Printing of high-precision products, direct extruder*

Table 3. Comparative characteristics of printers

Characteristics	KLEMA	DELTA P600	PRUSA I3
Working area, mm	220×220×250	600×600×600	250×250×250
Extruder temperature, °C	230	290	230
Desktop temperature °C	90	120	60
Number of extruders and type	1	2	1
Extrusion speed, mm/s	120	200	200
Filament diameter, mm	1.75	1.75	1.75
Chamber	Closed	Opened	Opened

Table 4. Transient processes during instantaneous step loading of various demonstrators of trabecular structures

Trabecular structure parameters	Transitional process	Trabecular structure parameters	Transitional process
(1) rectangular initial, with parameters according to the Bingham's model		(2) rectangular, elastic system, Bingham's model	
(3) rectangular with a high coefficient damping, Burgers model		(4) oblique structure, loss of strength according to the Kelvin–Voigt model	
(5) rectangular with low damping capacity, Burgers model		(6) rectangular with high rigidity of the Bingham's model	

Better dynamic characteristics revealed orthogonal trabecular structures (when comparing, for example, structures (3) and (4)), which indicates the expediency of further use of orthogonal structures.

A comparison of shock load damping patterns of existing structures was carried out using the Burgers rheological model for modeling. It was established that despite the practically identical amplitudes of the resulting stresses, their damping rate is much lower for a compact body, and the practically complete disappearance of oscillations (at the level of 5 % of the stresses upon impact) is observed during the time $\tau = 0.3...0.6$ s, which is greater than the simulated results with trabecular structures.

Thus, the expediency of using trabecular structures is determined not only by medical indications and anatomical properties of human bone tissue, but also by better dynamics of absorption of external mechanical influences, mainly of an impact nature. On the other hand, the geometrical parameters of trabeculae significantly affect the controlled dynamic characteristics, which requires additional research on model demonstrators of trabecular structures.

The use of additive processes for the reproduction of implants also allows to perform gradient printing, focusing on the change of elastic properties, for example, along the surface of the connection, as well as based on the requirements of strength and reliability, which will make it possible to maximally adapt the stress state of the hip part of the implant to the stress state of the patient's bone during its movement and rest.

A generalized characteristic of the formed trabecular structures can be the density coefficient of the trabecular structure, defined as:

$$K_{\text{v}} = \frac{4b^2\left(h + l - b\right)}{l^2 h}$$

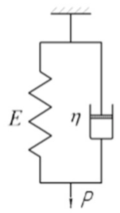
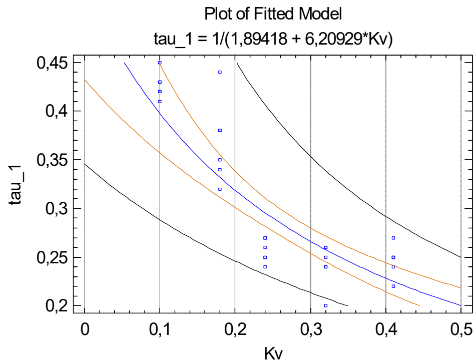
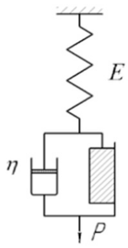
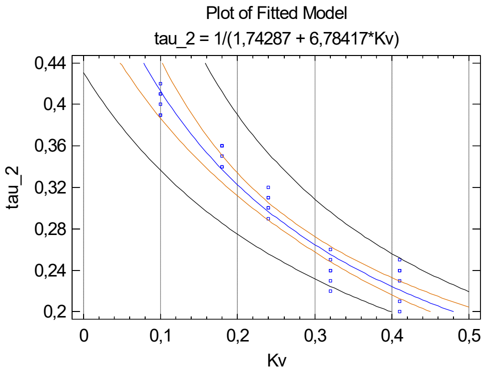
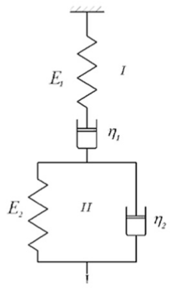
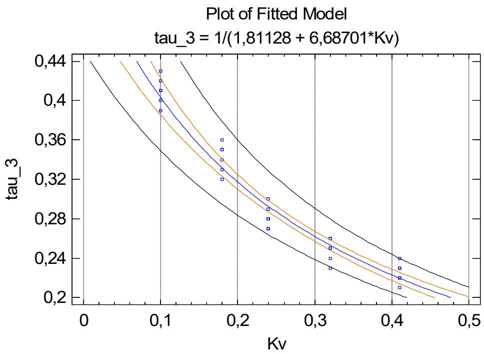
b, l, h – geometric parameters determined according to Fig. 9, Fig. 10.

Then, in accordance with the schemes of trabeculae execution, their geometrical characteristics and elastic-plastic properties of the material used in their manufacture, the most effective operating conditions of the product can be obtained based on the determination of individual dynamic parameters of the biomechanical system as a function of K_v, v, E, G, etc.

According to the rheological models taken into account, trabecular structures were identified (in particular, Table 1, option a, b) for uniaxial loading by compressive force with removal of elasticity and creep curves during mechanical tests of the demonstrators.

The analysis of simulation results is illustrated in Table. 5. The statistical sample was formed based on calculations of transient processes on the models of Voigt, Bingham and Burgers at 10 points of the duration of the transient process (at the same time, it was considered that constancy is achieved under the condition that the instantaneous amplitude of A_p decreases to 10 % of the initial

Table 5. Duration of the transient process during dynamic tests of printed demonstrators of different densities K_v

Rheological model	The duration of the transition process	Comments
Voigt model 		The correlation coefficient $R = 0.843$ Demonstrator transition time $\tau_1 = \dfrac{1}{1,89 + 6,21 K_v}$
Bingham's model 		The correlation coefficient $R = 0.946$ Demonstrator transition time $\tau_2 = \dfrac{1}{1,74 + 6,78 K_v}$
Burgers model 		The correlation coefficient $R = 0.972$ Demonstrator transition time $\tau_3 = \dfrac{1}{1,81 + 6,68 K_v}$

values). Since the transient process is damping and corresponds to a hyperbola, as a model of the functional conditioning of the duration τ, s, the density coefficient of the trabecular structure K_v is the regression equation of the form $\tau_i = \dfrac{1}{b_0 + b_1 K_v}$.

The plasticity change at the point of contact is taken into account in the Bingham model. At the same time, the peculiarities of the interaction of the implant components and the damping and elasticity coefficients, in particular, as well as the scheme of connecting the elements to each other (based on the contact conditions at individual points according to (8), (9)) were specified. The statistical processing of the results and the model are shown in Table 5, third row. The next line shows the results of processing according to the Burgers rheological model, based on the fact that the model corresponds to the structure of Figs. 6, and the dynamic parameters E_1, E_2, η_1, η_2, reflect c_b, c_o, c_k – reduced elasticity coefficients of the corresponding implant components and layers, b_b, b_o, b_k – damping coefficients taking into account the “implant component – bone” junction).

Conducting a variance analysis between the array of results obtained by calculations based on the Bingham and Burgers model proves that with a probability of 0.95 it can be assumed that such models are similar and can be used to perform dynamic studies of spatial trabecular structures.

In general, the regularity of the change of parameters is preserved, but the scattering of parameters for the Burgers model is the smallest, which ensured the maximum correspondence, estimated by the correlation coefficient *R*, between the calculated values of the amplitudes A_p^i and the values A_m^i obtained during modeling. In general, the controlled parameters of the duration of transient processes differ little from each other (no more than 15...20 %), however, the reproduction of dynamic shock load phenomena is more stable precisely when using the Burgers model.

So, we have that, in general, the models of Bingham and Burgers can be used almost equally when describing trabecular structures.

At the same time, the Burgers model is more accurate, as it allows you to also consider the contact with the bone tissue in the form of an additional elastic-plastic articulation.

The formulated conclusions are based on the results obtained during the testing of demonstrators made of PET and PLA plastics, however, they may also apply to other materials with similar physical and mechanical properties.

Conclusions

Features of the structure of trabecular structures for the manufacture of implants of hip and knee joints and rheological models that can be used as a basis for analyzing the dynamics of biomechanical systems “bone-articulated implant” are considered.

It is shown that the expediency of using trabecular structures is determined not only by medical advantages and anatomical properties of articulated human bone tissue, but also by better dynamics of absorption of external mechanical influences, mainly of an impact nature.

The interdependence of the results of surgical intervention with the patient’s initial condition, indications for treatment, his activity and possible postoperative complications was analyzed. An optimization function of the process of designing, manufacturing and operational support of implantation, which has probabilistic components, is proposed.

It is shown that it is most appropriate to use the Burgers model when studying the dynamics of the “bone-articulated implant” components, and the trabeculae density coefficient can be a generalized characteristic of the formed trabecular structures, provided that the geometric parameters match the bone tissue.

The use of additive processes for the reproduction of implants also allows to perform gradient printing, focusing on the change of elastic properties, for example, along the surface of the connection, as well as based on the requirements of strength and reliability, which will make it possible to maximally adapt the stress state of the hip part of the implant to the stress state of the patient’s bone during its movement and rest.

At the same time, the geometric parameters of the trabeculae significantly affect the controlled dynamic characteristics, which requires additional research on model demonstrators of trabecular structures.

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Реологічні моделі трабекулярних структур суглобних імплантатів, отриманих адитивними процесами

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Анотація. *Розглянуто особливості будови трабекулярних структур для виготовлення імплантатів кульшових і колінних суглобів та створення реологічних моделей, які можуть бути покладені в основу аналізу динаміки біомеханічних систем “кістково-суглобовий імплантат”. При цьому враховується, що сам імплантат має бути виготовлений у вигляді комбінованого набору функціональних елементів (або вихідних поверхонь), динамічні властивості яких є мінливими та максимально адаптованими до властивостей сполучної кістки, що дозволяє досяжно зберегти вихідні властивості кісток суглобів із забезпеченням належної міцності та надійності конструкції в цілому, обумовлюючи взаємозалежність результатів оперативного втручання з вихідним станом хворого, показаннями до лікування, його активністю та можливим післяопераційним періодом. Проаналізовано ускладнення. Запропоновано оптимізаційну функцію процесу проектування, виготовлення та експлуатаційного забезпечення імплантації, яка має ймовірнісні складові.*

Показано, що при дослідженні динаміки компонентів «кістково-зчленованого імплантату» найбільш доцільно використовувати модель Бюргерса, а коефіцієнт щільності трабекул може бути узагальненою характеристикою сформованих трабекулярних структур за умови відповідності геометричних параметрів кісткової тканини.

Ключові слова: *трабекулярні структури, імплантати кульшових та колінних суглобів, динамічні параметри, реологічні моделі, адитивні процеси, забезпечення функціональних вимог.*