РАЗРАБОТКА СИСТЕМЫ АВТОМАТИЗИРОВАННОГО ПРОЕКТИРОВАНИЯ ДЛЯ ИССЛЕДОВАНИЯ НАПРЯЖЕНИЙ И ДЕФОРМАЦИЙ БИОЛОГИЧЕСКИХ ТКАНЕЙ ДЛЯ ЦЕМЕНТНОГО ЭНДОПРОТЕЗИРОВАНИЯ БЕДРЕННОЙ КОСТИ

COMPUTER-AIDED- DESIGN SYSTEM DEVELOPMENT FOR STRESS AND STRAIN RESEARCH OF BIOLOGICAL TISSUES FOR CEMENT FEMORAL ARTHROPLASTY

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ABSTRACT

The proposed computer-aided-design system gives the chance to traumatologist to receive geometrical models for cement femoral arthroplasty of bone and soft tissues based on computer tomogram together with mechanical properties anisotropy, to carry out multiple geometrical transactions with these models, to calculate at the same time mechanical strains and stresses, to compare calculation results for receiving the best geometrical and settlement models, to keep them in the database.

Key words: mechanical stresses and strains of biological tissues, cement and endoprosthesis; cement femoral arthroplasty; computer-aided-design system of medical appointment.

АННОТАЦИЯ

Предложенная система автоматизированного проектирования дает возможность травматологу получать геометрические модели костных и мягких тканей для цементного эндопротезирования бедренной кости на основе снимков компьютерной томограммы с учетом анизотропии механических их свойств, выполнять различные геометрические операции с этими моделями, вычислять в любой период времени механические напряжения и деформации моделей, сравнивать полученные вычисления для получения наилучших геометрических и расчетных моделей, сохранять эти модели в базе данных.

Ключевые слова: механические напряжения и деформации биологических тканей, цемента и эндопротеза; цементное эндопротезирование бедренной кости, система автоматизированного проектирования медицинского назначения.

INTRODUCTION

Now, one of the most important medical and biological problems is the problem of treatment and rehabilitation of patients with femoral neck fracture or ununited fracture of femoral neck, after infectious diseases of bone tissues. Cement arthroplasty is the most effective in these cases. This type of joints arthroplasty is the most reliable and construction is wear-resistant. Bone cement is used as a shock absorber, which absorbs shocks and evenly distributes the external forces over the entire bone tissue surface.

With the use of computer and information technologies, it was possible to virtually build various treatment options for injured bone tissues, to calculate stress and strain of solid models for purpose of cement femoral arthroplasty using numerical methods for solving problems. Therefore, computer and mathematical modeling of physiological processes taking place in injured bone tissues is an actual task and has important practical significance.

OBJECTIVE RESEARCH

Therefore, it is useful to develop an automated design system that creates solid models of bone and soft tissues, starting with the import of a computer tomogram into the system and ending with a comparative mechanical analysis of models of bone tissues for purpose of cement femoral. The operator in automatic and semi-automatic mode can calculate the maximum loads of bone tissue, allowable deformations inside the bone tissue. In our time there are no existing similar systems. The offered system program is of great theoretical and practical interest for medicine.

The program of medical appointment gives the chance to the traumatologist to carry out multiple geometrical transactions with these models on planning of conducting transaction, to calculate at the same time mechanical tension and deformations, to carry out the comparative analysis of calculation results for identification of the best geometrical and settlement models.

Structure of the developed automated design system.

The proposed software package includes several automated systems or program modules, connected by a single user interface. These packages are the Materialise Mimics - the system which builds solid models based on computer tomogram (Materialise Inc., 2015); Catia - CAD system (Catia Inc., 2015); Ansys Multiphysics - CAE system (Ansys Inc., 2015). The user interface and automatic calling of the program modules are implemented in C++ Builder programming language.

Let's consider in details how each program module works when performing a stationary static strength analysis for the purposes of cement femoral arthroplasty.

Creation of geometric models.

The first user's task is to obtain solid models based on a computer tomogram (Figure 1). Per the threshold values of biological objects, it refers to the models of hard and soft tissues in Mimics package by Materialize (Figure 2). The package includes the SurgiCase Orthopaedics system – the database of models of constructed fixation systems and information database of mechanical properties of hard and soft tissues.

Further export of the model should be in the stereolithographic format *.stl (set of surface triangles) or in the neutral data format *.iges (*.igs). These formats allow to transfer information about geometric and mechanical properties of biological tissues, they are supported by commonly available CAD and CAE systems. After importing files into one of the selected formats, there is a task to generate a solid 3D model. The required functions of the second program module Catia are automatically called up.

There are two ways to generate the 3D model. The first one is when you download the model in *.iges (*.igs) format, then the geometric model is presented as a cloud point. So, the next steps are the following: the construction of a quality finite-element network, "pulling" the surface onto the network, correction of its geometric defects and closing the surface to obtain a solid model (Figure 3). The *.stl alternative is when the model is already represented in the form of finite elements, it is only necessary to correct the geometry defects and perform the surface closure.

Figure. 1. Screenshot of the program interface that creates solid models based on computer tomogram

Figure 2. Selecting the threshold value of biological tissues

Figure 3. Obtain a femur solid model

The first option is more accurate, the 3D model we get is as close as possible to the original object, but there is a probability to get a model with complex geometry, which makes this impossible to build the finite network element in the engineering design package. That's why during the exporting of the model into one of the formats mentioned above, it is necessary to define the priority between the quality of geometric models and problem complexity.

Geometric operations with geometric models for the purposes of cement arthroplasty.

In order to add the biomechanical and anatomical properties of femur bone tissue, it is necessary to perform some geometric operations with the generated 3D models (Figure 4). The femur model is divided into layers of bone tissue and into segments by a secant plane. For cement arthroplasty purposes, the required part of the bone tissue is cut from the model to make a cement model.

Figure 4. Screenshot of the program interface that performs geometric operations with the geometric models

After completing the previous tasks, some additional geometric operations with 3D models are required (Figure 5). To combine the solid models into a single assembly, it is necessary to define auxiliary points, per which the bone tissue model relates to the endoprosthesis model, cement assembly, i.e. in biotechnical system (Figure 7). And then join the models point by point into a models' assembly.

Figure 5. Screenshot of the program interface that performs some additional geometric operations with models

Also, the points are necessary to find the centers of gravity of the sections. If you connect the centers of gravity of bearing joints of lower limbs, then you get a biomechanical line, which is an important clinical reference even at the stage of diseases diagnosing. Accordingly, already at this stage the doctor can understand, what endoprosthesis placement reduces the arising stresses in some parts of the bone. And at the output, there is the transfer to the third software module Ansys, which produces the calculated models (Figure 6).

Figure 6. Assembly of solid models

Creation of calculated models.

In this case a traumatologist works with several modules: Design Modeler (geometric builder), Static Structural (static strength analysis), Engineering Data (material properties

library) (Figure 7). The advantage of such construction is that one calculated model is used to solve a wide range of interdisciplinary problems.

Figure 7. Screenshot of the module interaction for obtaining calculated models

To obtain mechanical properties of materials.

Solid biological tissue is an anisotropic medium, to determine its characteristics we introduce a local coordinate system ($i, j = 1, 2, 3$). The example of its positioning in relation to the tibia is shown in Figure 8. The axes orientation in the coordinate system should be made per the symmetry in the bone tissue structure (Bonfield & Li, 1967; Evans & Thomas, 1973; Mather, 1967; Sedlin, 1965; Yamada, 1970; Knetas et al., 1980; Sevostianov & Kachanov, 2000). For example, for a compact layer, the axis x_1 coincides with the prevailing direction of osteons, i.e. with the direction of longitudinal axis of the bone. The axis x_2 should be positioned along the tangent to the circumference of the section, since the compact bone tissue has a cylindrical structure of lamellar non-osteon bone tissue. The axis x_3 is orthogonal to the axes x_2 and x_3 . For a spongy layer the axis x_1 has the same direction as for a compact layer; the axis x_2 has a transverse direction, the axis x_3 is anteroposterior.

The connection of deformation tensor (1) we represent as a finite incomplete tensor polynomial, consisting only of a linear component:

$$\varepsilon_{ij} = a_{ijkl} \sigma_{kl} \quad (i, j, k, l = 1, 2, 3), \quad (1)$$

where a_{ijkl} – compliance tensors of the 4th rank.

Per the values of the compliance tensor components, we calculate material constants (2)

$$E_i = (a_{iiii})^{-1} \quad (i = 1, 2, 3) \quad (2)$$

and the coefficients of transverse deformation (3)

$$\mu_{ij} = (-a_{jjii}) \div a_{iiii} \quad (i, j = 1, 2, 3; i \neq j), \quad (3)$$

characterizing the elastic properties of solid biological tissue.

Figure 8. Orientation of samples in the tibia diaphysis

Figure 8 shows a traumatologist's sequence of actions in the system to set the anisotropy of the elastic properties of compact and spongy layers of bone tissue, muscle tissue, cement and endoprosthesis. The coordinates to position anisotropy on the axes should be created for each bone segment. The stress tensor components are the basis for calculation of elasticity modulus, transverse deformation modulus and shear modulus. For this purpose, the package includes a knowledge data base with the data from experimental and therapeutic practice, literature references (Sevostianov & Kachanov, 2000; Knetas et al., 1980; Martin & Boardman, 1993; Martin et al., 1998; Duda, 1998; Lotz et al, 1995). Tables 1, 2 show the anisotropic properties of compact and spongy layers of femur bone tissue, they serve as an example of how to work with the package.

Table 1

The anisotropy of the elastic properties of a compact layer of femur bone tissue

The 1st segment from 0 ° to 70 °:							The 2nd segment from 70 ° to 140°:						
	A	B	C	D	E	F		A	B	C	D	E	F
1	(Pa) ▾	(Pa) ▾	(Pa) ▾	(Pa) ▾	(Pa) ▾	(Pa) ▾	1	(Pa) ▾	(Pa) ▾	(Pa) ▾	(Pa) ▾	(Pa) ▾	(Pa) ▾
2	5,015E+13						2	5,303E+13					
3	-1,481E+13	1,1911E+14					3	-1,598E+13	1,2065E+14				
4	-1,556E+13	-7,149E+13	1,3E+14				4	-1,621E+13	-7,161E+13	1,5913E+14			
5	0	0	0	1,11E+13			5	0	0	0	9,7E+12		
6	0	0	0	0	7,28E+12		6	0	0	0	0	7,15E+12	
7	0	0	0	0	0	4,93E+12	7	0	0	0	0	0	4,75E+12

The 3rd segment from 140 ° to 200 °:							The 4th segment from 200 ° to 250 °:		
	A	B	C	D	E	F			
1	(Pa) ▾	(Pa) ▾	(Pa) ▾	(Pa) ▾	(Pa) ▾	(Pa) ▾	Young's Modulus X direction		1,629E+10 Pa
2	5,595E+13						Young's Modulus Y direction		8,37E+09 Pa
3	-1,746E+13	1,0686E+14					Young's Modulus Z direction		6,27E+09 Pa
4	-1,802E+13	1,746E+13	1,2534E+14				Poisson's Ratio XY		0,308
5	0	0	0	1,02E+13			Poisson's Ratio YZ		0,562
6	0	0	0	0	6,28E+12		Poisson's Ratio XZ		0,313
7	0	0	0	0	0	4,7E+12	Shear Modulus XY		4,49E+09 Pa
							Shear Modulus YZ		2,24E+09 Pa
							Shear Modulus XZ		3,06E+09 Pa

The 5th segment from 250 ° to 300°:			The 6th segment from 300 ° to 360 °:		
Young's Modulus X direction	1,876E+10	Pa	Young's Modulus X direction	2,058E+10	Pa
Young's Modulus Y direction	9,66E+09	Pa	Young's Modulus Y direction	7,95E+09	Pa
Young's Modulus Z direction	7,75E+09	Pa	Young's Modulus Z direction	6,25E+09	Pa
Poisson's Ratio XY	0,323		Poisson's Ratio XY	0,3	
Poisson's Ratio YZ	0,767		Poisson's Ratio YZ	0,631	
Poisson's Ratio XZ	0,331		Poisson's Ratio XZ	0,308	
Shear Modulus XY	5,22E+09	Pa	Shear Modulus XY	4,69E+09	Pa
Shear Modulus YZ	2,74E+09	Pa	Shear Modulus YZ	2,5E+09	Pa
Shear Modulus XZ	4,62E+09	Pa	Shear Modulus XZ	3,12E+09	Pa

The femoral head:		
Young's Modulus X direction	1,16E+10	Pa
Young's Modulus Y direction	1,22E+10	Pa
Young's Modulus Z direction	1,99E+10	Pa
Poisson's Ratio XY	0,42	
Poisson's Ratio YZ	0,23	
Poisson's Ratio XZ	0,23	
Shear Modulus XY	4E+09	Pa
Shear Modulus YZ	5,4E+09	Pa
Shear Modulus XZ	5E+09	Pa

Table 2

The anisotropy of the elastic properties of a spongy layer of femur bone tissue

The part of a bone	The area of a bone	The load direction	The elastic modulus $E * 10^7, \text{Pa}$
The head			29,3
The head, the lower condyle		Upper lower	29,23
		Foreground	24,98
		Outer Internal	72,22
The condyle	The outer part		30,35
	The inner part		29,78
The neck		Along the neck	45,89
The trochinter			3,71

For Titanium BT1-00 endoprosthesis it suggests the elastic modulus of $1.1025E+11$ Pa and the Poisson's ratio of 0.32. For bone cement the elastic modulus is $2E+9$ Pa, the Poisson's ratio is 0.3 (Knetas et al., 1980).

Creation of the finite element network for the models.

For users of the software under consideration there is an automated calling of options, which split the model into finite elements (Faustini, 2006; Lee, 1994). High quality of network is required, that's why the quality of the split should be analyzed each time by the built-in tool Ansys Structural (Bruyaka et al., 2010). There must be common nodes on the surfaces of contiguous models. To solve this task the user should combine these models into one assembly. The quality of models' split into finite elements is determined by the placement of groups of elements with approximately the same size by models volume (Figure 9).

Figure 9. The partition of models into finite elements for the purposes of cement femoral arthroplasty and the quality control of the finite element network

Setting contact conditions.

This is a rather important stage, where everything depends on the materials properties. If we expect the bone tissue is a hard material, then the models should be fastened in a way, preventing the penetration of the model of one layer into a model of another layer. For plastic materials, accordingly we should consider the hardness of bone tissue layers and set the tolerance of penetration of one model into another model.

Setting external influences.

It is necessary to consider the influence of pressure, environmental factors and gravity on the bone tissue of muscles attached to the bone.

The values of the distributed force over a certain area, i.e. the pressure values, are represented by the expansion of the force vector $\mathbf{F}$ or pressure $\mathbf{P}$ on the projections onto the coordinate axes of local or global coordinate systems.

The values of projections of the muscle force $\mathbf{F}$ on the axis (4-6) are defined by the following formulas:

$$F_x = \mathbf{F} \cos \alpha = \sqrt{F_x^2 + F_y^2 + F_z^2} \cos \alpha; \quad (4)$$

$$F_y = \mathbf{F} \cos \beta = \sqrt{F_x^2 + F_y^2 + F_z^2} \cos \beta; \quad (5)$$

$$F_z = \mathbf{F} \cos \gamma = \sqrt{F_x^2 + F_y^2 + F_z^2} \cos \gamma, \quad (6)$$

where the angle α forms a vector $\mathbf{F}$ with the coordinate axis x , the angle β - with the axis y , the angle γ - with the axis z . The cosines of these angles are called the directional cosines of the vector $\mathbf{F}$. For them the following formula (7) is valid:

$$\cos^2 \alpha + \cos^2 \beta + \cos^2 \gamma = 1 \quad (7)$$

The pressure coefficient is the ratio of the force $\mathbf{F}$ to the area over which that force is distributed. The formulas for calculating the pressure components P_x , P_y , P_z should be written in a similar way to the forces projections (Forster, 2003; Bean, 1998).

Trauma doctors can take the values of forces acting on bone tissues surface from the database with literary references. For our example, which shows how to work with the

system, we take 650N as the weight of the body without considering the lower limbs by the segmentation method (Dubrovsky, 2003).

Setting the strength properties for the models.

Bone tissue is a brittle material, in comparison to stretching we need significantly more compression efforts for its destruction. Therefore, per the first strength theory, the calculated main normal stresses along the longitudinal direction of bone in mechanical sense should not exceed the ultimate stress limit when stretching: $\sigma_1 \leq \sigma_t$.

Also for the stress-strain analysis of bone tissues, the Mohr-Coulomb strength theory is applicable, per which we compare the main normal stresses of the longitudinal and transverse directions of the bone with the ultimate compressive strength σ_c , considering the safety coefficient K_{safety} :

$$K_{safety} = \left(\frac{\sigma_1}{\sigma_t} + \frac{\sigma_3}{\sigma_c} \right)^{-1}. \quad (8)$$

For the strength analysis of the models, the plastic yield criterion is also applicable, the third strength theory. The maximum shear stress (9) is calculated based on main normal stresses of longitudinal and transverse directions of bone tissue. This maximum value is compared with the yield stress limit of materials:

$$\tau_{max} = \frac{(\sigma_1 - \sigma_3)}{2}. \quad (9)$$

For considering both normal and tangential stresses, there are so-called equivalent stresses or Mises stresses σ_m (10). They are calculated by the formula:

$$\sigma_m = \sqrt{0,5 \left[(\sigma_{11} - \sigma_{22})^2 + (\sigma_{22} - \sigma_{33})^2 + (\sigma_{33} - \sigma_{11})^2 + 6(\sigma_{23}^2 + \sigma_{13}^2 + \sigma_{12}^2) \right]}. \quad (10)$$

Equivalent stresses are compared with the yield stress limit of bone tissue σ_y or the fatigue strength limit for the metals (CADFEM; Martin, 1995).

In the developed automated system, the user selects the strength theory for the materials of the model and their limit values. As bone tissue is a brittle material, we can apply the Mohr-Coulomb strength criterion, per which we compare the calculated values of tensile ultimate strength of materials and compressive ultimate strength of bone tissue of models.

The strength properties of spongy and compact layers in the Engineering Data module are shown in Table 3, together with the Tensile Ultimate Strength and Compressive Ultimate Strength (Knetas et al., 1980; Currey, J. D. 1970; Currey, J. D. 2002; Keaveny, 1994).

RESULTS AND THEIR DISCUSSIONS

The system offers a tool of cutting by a surface, which gives the possibility to look at the distribution of the calculated values by the volume of models in the required places.

In solving thermal and mechanical problems we calculated Equivalent Stress, Equivalent Strain, Vector Principal Stress, Vector Principal Strain, Total Deformation, Normal Stress and Normal Elastic Strain in three directions of the axes, Shear Stress in XY, YZ, XZ planes.

Based on the calculations we see that the largest values of displacements are observed in the middle and upper parts of femoral head and upper half of femoral neck (Figure 10). High stresses in the 3rd and 4th segments of the bone during compression are explained by lesser thickness of the layer in comparison to other segments (Figure 11) (Keaveny, 2001). But when the bone is bent, the maximum stresses are observed in the 1st segment (Figure 12), where the compact layer of bone tissue is comparatively thicker (Dobelis, 1982). Value distribution of stresses and strains is explained by the anatomical structure of layers of bone tissue (Turner et al, 1997; Duda, 1998).

The calculated values do not exceed the limits for stresses and displacements.

Figure 10. The distribution of the values of Shear strain along the direction of the axis Z by the femur volume

Figure 11. The distribution of the values of Normal stress along the direction of the axis Z of the global coordinate system

Figure 12. The distribution of the values of Shear stress (XY Plane) by the femur volume

The calculated results deviate by 1-5% from the experimental, analytical data given in the foreign literature which describe the similar conditions with real bones (Kaorapapong et al., 2011; Bean, 1998; Zienkewicz, 2002), it proves the reliability of the results.

CONCLUSION

The calculated values of stress-strain state of bone tissues during implant arthroplasty are the closest to real ones. With the help of the first software module, a doctor can get virtual solid-state models on the basis computer tomogram, i.e. consider the geometric and mechanical properties of a patient's femur. There is a manual mode of editing the model, which allows to achieve more exact result in construction of 3D models.

The models have a complex geometry, since the objects created by nature have an irregular shape. They tend to have defects and surface roughness. This makes the solution of mechanical problems difficult. That's why solid models require pre-processing of geometric shapes. For 3D modeling, the Catia package was selected. It allows users to make a geometric model as close as possible to the original, and to make some geometric operations for smoothing the surface of the geometric model and simplifying the solution of mechanical and thermal problems. The package includes an essential set of tools for dividing the models into segments and setting up the local coordinate systems to consider the mechanical properties of anisotropy. The program makes it possible to cut out the required volume in a bone tissue model for an implant installation and for cement.

Ansys package was selected as a calculation software module. It provides two-way data exchange with the Catia package, so there is no loss of geometric shape of the models. Therefore, it increases the reliability of the calculated results.

The authors developed a mathematical model for studying the stresses and deformations of bone tissues, considering the anisotropy and heterogeneity of their mechanical properties, as well as various external influences. Thus, the system allows us to obtain the results, which are close to the values characteristic for real bone tissues.

The suggested methodology, the algorithm of work in the system, allows a traumatologist to calculate the stresses and strains of the constructed models for cement femur arthroplasty before it is performed. If the results are unsatisfactory, the doctors can perform a virtual operation a lot of times to get the best results. Undoubtedly, this will significantly improve the quality of medical services.

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