Application of a digital prototype for CT-based bone strength analysis

Oleg Gerasimov
N.I. Lobachevsky Institute of Mathematics and Mechanics
Kazan (Volga region) Federal University
Kazan, Russian Federation
valeriy.karasikov@gmail.com

Karina Sharafutdinova
N.I. Lobachevsky Institute of Mathematics and Mechanics
Kazan (Volga region) Federal University
Kazan, Russian Federation
KaRSharafutdinova@stud.kpfu.ru

Ramil Rakhmatullin
N.I. Lobachevsky Institute of Mathematics and Mechanics
Kazan (Volga region) Federal University
Kazan, Russian Federation
RRRakhmatulin@stud.kpfu.ru

Tatyana Baltina
Institute of Fundamental Medicine and Biology
Kazan (Volga region) Federal University
Kazan, Russian Federation
tvbaltina@kpfu.ru

Maksim Baltin
Institute of Fundamental Medicine and Biology
Kazan (Volga region) Federal University
Kazan, Russian Federation
baban.bog@mail.ru

Artur Fedianin
Institute of Fundamental Medicine and Biology
Kazan (Volga region) Federal University
Kazan, Russian Federation
artishock23@gmail.com

2022 VIII International Conference on Information Technology and Nanotechnology (ITNT) | 978-1-6654-8571-5/22/$31.00 ©2022 IEEE | DOI: 10.1109/ITNT55410.2022.9848693

XXX-X-XXXX-XXXX-X/XX/$XX.00 ©20XX IEEE

The application of image data is one of the main approaches to creating digital prototypes of inhomogeneous medium elements. The article discusses the method of finite element (FE) analysis based on the use of computed tomography (CT) data of bone organs subjected to external loads. Based on the distribution of the material density in the volume, the geometry of the sample was restored by constructing an FE mesh. The calculations were made on the basis of a three-dimensional isoparametric FE of a continuous medium with a linear approximation of the geometry and displacement field. The data on the medium permeability were taken into account when integrating the local stiffness matrix. The purpose of the study is to describe the general technique for constructing a numerical model that allows static calculation of objects with a porous structure based on their CT data. Kinematic loading was used as boundary conditions. The analysis of the obtained results was carried out by constructing computational and post-processing grids. The qualitative assessment of the simulation results was based on the values of the normalized energy error. In the model problem, a three-point bending of a pig anterior bone sample was considered. The results obtained by numerical calculation were compared with data from a full-scale experiment. The relative error of the results ranged from 3 to 15%. The area of crack formation, determined from a full-scale experiment, was explained by the results of numerical simulation, which showed the occurrence of values exceeding the yield strength for bone material in this area. The obtained estimates reflect not only the effectiveness of the proposed approach, but also the consistency with bending test data. Moreover, the method turns out to be less resource-intensive and more efficient in terms of calculation time relative to existing approaches.

Keywords—digital prototype, numerical modelling, bone tissue, porous structure, image segmentation, finite element analysis, inhomogeneous medium.

I. INTRODUCTION

At the moment, numerical simulation is applicable in various fields of scientific research. In this case, one of the most promising methods for describing the behaviour of inhomogeneous medium elements is modelling based on data from images of the samples under study [1]. This approach allows us to consider the problems of multi-connected porous materials, which is relevant in clinical practice, as it can have a significant impact on the treatment quality [2]. The determination of the stress-strain state is the main task in assessing the restructuring of bone tissue, in particular, in the development of an individual implant [3]. Currently, there are multiple approaches to reconstructing the model mechanical properties based on image data. The first of them involves the approximation of inhomogeneous material properties by constructing a distribution using the mean intercept length method. In this case, the physical relations are determined by the elastic constant tensor and the fabric tensor [4]. Another method includes the reduction of anisotropy to orthotropy by calculating constants from the data of numerical experiments [5, 6, 7, 8]. In this study, a third approach is presented, based on taking into account the properties of the material by the method of weighted integration of the stiffness matrix of each finite element (FE) of the mesh.

The use of computed tomography (CT) data is one of the main approaches to creating images of the studied area. This operation implies the construction of a digital prototype (a 3D value array corresponding to the medium permeability in the current voxel). Similar values can be interpreted based on the quantitative Hounsfield X-ray density scale [9]. Modern methods of CT make it possible to obtain comprehensive data not only about the state of bone tissue, but also about its interaction with the implant [10, 11, 12]. Thus, on the basis of CT data, segmentation of the research area can be performed, which turns out to be a complex process, since difficulties arise in restoring not only the organ geometry, but also its internal structure. It is worth noting that the geometry of the sample obtained in this way must be consistent for the possibility of superimposing computational and post-processing grids, on which the accuracy of modelling directly depends. The main approach to segmentation of the original image is stratified separation [13]. In this case, for example, it is possible to separate the cortical layer and the spongy bone tissue. Subsequently, each individual medium is used in modelling with a variety of mechanical parameters [14, 15]. Due to the labour-intensive nature of this segmentation method, modern research is aimed at developing optimal methods for carrying out this procedure [16]. On the one hand, the greatest accuracy of calculations can be achieved by modelling each micro-volume of a continuous medium with one FE [17]. However, in this case, the number of calculations performed sharply increases. Because of this, it becomes necessary to enlarge the FE size in such a way that each local volume corresponds to a certain number of voxels [12, 18]. In this case, each voxel can be considered as a quadrature point neighbourhood of the local stiffness matrix (LSM) integration. Thus, the questions of determining the number of voxels within one element, as well as the applicability of the FE method in everyday clinical practice in general [19], remain open.

The purpose of the work is to implement the method of static calculation of porous structure samples. The simulation was based on the construction of a three-dimensional isoparametric linear FE of a continuous medium based on CT data.

II. MATERIALS AND METHODS

A. Implementation of a CT-Based Finite Element

Let us consider a method for describing the volume of a test sample based on its CT data. In this case, the value of each voxel corresponds to a certain micro-volume of the medium and represents the average value of the Hounsfield unit. Usually, based on such values, three-dimensional images in the gray-scale spectrum can be built. Accordingly, the dimensions of each micro-volume are Δx, Δy and Δz, which are usually equal to each other and are determined by the CT scanner resolution. The FE method can be chosen as a calculation technique to describe the behaviour of the studied objects. In this case, the geometry of the samples is divided into an FE mesh. In this work, the algorithm provides for the construction of a three-dimensional eight-node FE with a linear approximation of the geometry and displacement field. In the case of inhomogeneous medium, the elasticity matrix can be rewritten in the form:

$$[D(\{r\})] = [D^B] \cdot \omega(\{r\}), \quad (1)$$

where $[D^B]$ is the elasticity matrix of a solid material, $\omega\{r\}$ is a scalar function of the inhomogeneity defined by the digital prototype, $\{r\}$ is the vector at the point where the elasticity matrix is set. A detailed description of the general relations for the weighted integration of the LSM was presented previously [20]. In the final form, the element stiffness matrix can be determined using the following relation:

$$[K^E]=\iiint[B(\xi,\eta,\zeta)]^T[D][B(\xi,\eta,\zeta)]\omega(\xi,\eta,\zeta)|J(\xi,\eta,\zeta)|d\xi d\eta d\zeta, \quad (2)$$

where (ξ,η,ζ) are the local coordinates of the element, $[B(\xi,\eta,\zeta)]$ is the matrix connecting the reduced elastic strain vector and the nodal displacement vector, $|J(\xi,\eta,\zeta)|$ is the Jacobian determinant of the coordinate transformation.

To determine the average value of stresses over the element volume, the following relation can be used [20]:

$$\bar{\sigma}^E = V_E^{-1}\iiint\{N\}^T\{\sigma_0^E\}dV^E, \quad (3)$$

where $\{N\}$ are shape functions, $\{\sigma_0^E\}$ is the vector of approximation coefficients, V^E is the element volume.

The energy error is defined as follows:

$$\bar{E}^i = \tfrac{1}{2}\iiint\{\Delta\sigma\}^T[D]^{-1}\omega^{-1}\{\Delta\sigma\}dV^E, \quad (4)$$

where $\{\Delta\sigma\}$ is the stress error vector. The element strain energy can be determined by the equation:

$$\bar{U}^i = \tfrac{1}{2}\iiint\{\sigma\}^T\{\varepsilon\}dV^E. \quad (5)$$

As a result, the energy error is normalized using the strain energy:

$$\tilde{E}^i = [\bar{E}^i / (\bar{U}^i + \bar{E}^i)]^{1/2} \cdot 100\%. \quad (6)$$

Thus, the described approach allows us to take into account the inhomogeneity of medium based on the stiffness matrix. The stress-strain state analysis should be carried out in terms of averaged stress and strain. The quality of the results can be evaluated according to the normalized value of the energy error. A more accurate formulation of the approach and convergence estimates of numerical calculation are presented in previous publications [21, 22].

B. Segmentation and Meshing

Let us introduce a method for determining the properties of a bone organ based on the its CT data. In this case, each value of the input data array corresponds to the optical density in the current micro-element of the volume. Based on such a quantity, both elastic constants and ultimate stresses can be calculated:

$$\rho = a(\rho) + b(\rho) \cdot HU, \quad (7)$$

$$E = a(E) \cdot \rho^{b(E)}, \quad (8)$$

$$[\sigma] = a(\sigma) \cdot \rho^{b(\sigma)}, \quad (9)$$

where the coefficients *a* and *b* can be obtained from experiments or taken from the literature and determined by the type of bone material or a specific organ.

Bone is a inhomogeneous porous material with an irregular structure. Because of this, the segmentation of the initial sample geometry implies the presence of a different distribution of bone substance in each volume element. Thus, the imposition of the computational grid on the selected area determines the method of CT data segmentation and assumes the presence of both a cortical plate and a spongy substance in each mesh element. As a result, the proposed approach, based on the construction of an FE mesh, makes it possible to take into account the anisotropy of the material at the level of modelling a separate FE according to CT data. Fig. 1 shows a 3D model of a digital bone organ prototype. In the image of the longitudinal section (Fig. 2), the area with the highest density (the cortical plate, the outer contour) and the area corresponding to the cancellous structure (the inner part of the bone) can be distinguished.

Fig. 1. 3D representation of CT data for the anterior bone of a pig.

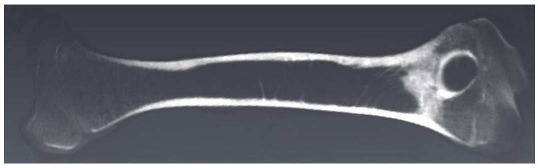

Fig. 2. A slice of the digital prototype image in the longitudinal direction.

Currently, despite the popularity of image segmentation approaches to restore the geometry of samples, there are difficulties in the study of bone organs due to their irregular geometry and complex internal structure. The proposed method makes it possible to take into account the anisotropic properties of the material indirectly, without separating the cortical and spongy regions. In this regard, a regular grid can be constructed for the entire space of CT data. In this case, the process of binarization of the original array of values makes it possible to carry out the procedure for removing FEs with a relatively low content of bone substance, which determines the process of restoring the geometry of the original sample from data on the distribution of material density in the volume of the study area. The grid obtained in this way will be referred to as the computational grid. Kinematic and/or static boundary conditions can be applied to the reconstructed bone geometry. Further, after calculating the LSM for each FE, the standard procedure for assembling the global stiffness matrix is carried out. Based on the nodal displacement values obtained from the FE calculation, the stress and strain vectors can be received, averaged over the volume of each mesh element. From the computed results, principal values and directions, as well as von Mises stress and strain, can be determined. The stress-strain state and CT data were compared by introducing an additional grid, which will be referred to as the post-processing grid. Such a mesh can be built as a contour to the processed CT data and is the result of interpolation of the values obtained from the computational grid.

C. Full-Scale Experiment

The experiments were carried out on six Vietnamese male swine weighing 15–20 kg. The animals were placed in separate cages under standard laboratory conditions with unlimited access to food and water and a 12 h day/night cycle. The protocol of the experiment, including anaesthesia, surgery, postoperative care, testing, and euthanasia, was approved by the Animal Care Committee of Kazan State Medical University (protocol #5 of 20 May 2020) [23]. All experimental procedures were carried out in accordance with the standards to minimize the suffering of animal and the size of experimental groups. The animals were included in the experiments after an adaptation period of at least 7 days. The experiments used animal bones after spinal contusion injury. The animals were sacrificed on the 8th week after spinal cord injury. Euthanasia was performed under deep anaesthesia by overdosing of an inhalation anesthetic (Isoflurane). The bones of the fore and hind limbs were extracted.

The bones were scanned using CT. The system of micro/nanofocal X-ray control for CT and 2D control Phoenix V | tome | X S240 was used for scanning. The system is equipped with two X-ray tubes: a microfocus with a maximum accelerating voltage of 240 kV with a power of 320 W and a nanofocus with a maximum accelerating voltage of 180 kV with a power of 15 W. For primary data processing and creation of a volumetric (voxel) model of the sample based on X-ray images, datos|x reconstruction software was used. The sample fixed in the holder was placed on a rotating table of the CT chamber at the optimal distance from the X-ray source. The survey was performed at an accelerating voltage of 90-100 kV and a current of 140-150 mA.

The CT data size was within 1 GB, the resolution of the scanning device was 0.2 mm. The total number of values from the image – 7523. Based on the size of the CT scan area, a regular FE mesh was built. The calculated area was determined by removing elements corresponding to a low content of bone substance (less than 5%). Consequently, the original CT data were pre-binarized in accordance with the specified threshold value. This procedure implied equating to 0 those array elements whose values turned out to be less than the threshold, and to 1 – greater than or equal. The calculation of the mechanical properties of the material was based on the use of CT data already processed in this way. Young's modulus for pure bone tissue can be calculated from (8) and was 30 GPa and Poisson's ratio – 0.3. In the calculations, the simulation of the sample for a three-point bending was considered. The upper and lower surfaces of the distal and proximal sections of the bone were fixed in displacements (marked with black lines in Fig. 3A, B). Kinematic boundary conditions in the form of displacements were applied to the upper surface of the bone diaphysis (marked with red arrows in Fig. 3A, B) and were equal to 1, 2, 3 and 4 mm, respectively. The evaluation of the reliability of the obtained results was based on the values of the calculated equivalent force.

The calculation was performed on the basis of the developed software in the C++ programming language (g++ compiler). Due to the resource-intensive method of numerical integration based on CT data, the use of OpenMP parallel solution technology was ensured, which made it possible to compute the calculation matrices for each FE on a separate logical core of the processor. Numerical simulation was carried out on the basis of a computer of the following configuration: CPU – AMD Ryzen 5 2600 Six-Core Processor 3.40 GHz, RAM – G. SKILL AEGIS 16 GB DDR4 16GISB K2 Kit (2x8GB) 2933 MHz, motherboard – MEG Z690 ACE. Visualization of computational and post-processing grids was performed in the ParaView system. The total calculation time was about 15 minutes.

III. Results and Discussion

The calculations were carried out using two computational grids with element sizes of 10 and 5 mm, respectively. The analysis of the normalized error was carried out for the results obtained on the basis of both meshes. Particular attention was paid to the normalized error values in the elements with the maximum von Mises stress values. Fig. 3A, B shows the distribution of normalized error values (computation grid) for the anterior bone sample, and Fig. 3C, D – von Mises stress (post-processing grid). Averaging the values over the FE volume, taking into account the CT data, determines a different interpretation of the results obtained. In this case, the spatial distribution of stress values in the element is unknown, but can be averaged from the digital prototype data. Thus, the appearance of falsely overestimated and underestimated values is due to the uneven distribution of bone material in the volume of each element, which explains the need to calculate and appropriately evaluate the normalized error. The results of calculations based on two FE meshes showed that the largest energy error is achieved in boundary elements with a relatively low content of bone tissue. This effect is explained by a decrease in the stiffness of the element due to the weighted integration of the LSM. The error value in the regions of interest turned out to be less than 50%, while the maximum values correspond to the zone of application of kinematic boundary conditions, in the other regions the error is less than 25%.

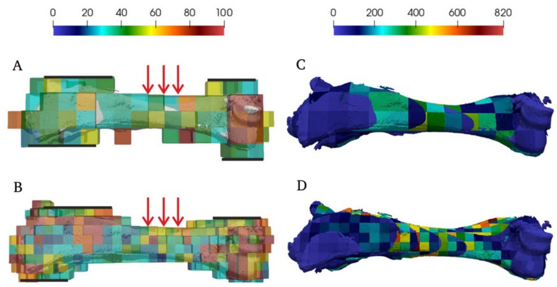

Fig. 3. Modelling results: A, B – computational grid (normalized error (%)); C, D – post-processing grid (von Mises stress (MPa)).

Consider the results corresponding to a mesh with a FE size of 5 mm and a displacement of 4 mm. The minimum energy error (20%) and the maximum von Mises stress (500 MPa) are achieved in the load application area (Fig. 3B, D). The 2nd component of the principal stresses was used to estimate the equivalent force in the area of application of the displacements. The reaction to the equivalent force was calculated by multiplying the integral of the stress vector in the area of load application to the corresponding area. In this case, the 1st and 3rd principal stress components describe the stress state in the bone. The resulting streamlines constitute an orthogonal system. It is important to clarify that in the region of maximum von Mises values, the 1st component of the principal stresses reaches its maximum value (400 MPa), and the 3rd component – minimum value (-400 MPa), which explains the occurrence of a crack in a full-scale experiment, since the values exceed the yield strength [24, 25]. The relative error between the values of the equivalent force and the load obtained as a result of the full-scale experiment ranged from 3 to 5%. The inaccuracy of the results can partly be explained by the approximate determination of the boundary conditions.

The modelling results showed that the proposed area segmentation approach and the subsequent numerical calculation based on weighted integration of the LSM of each FE make it possible to obtain reliable results, despite the simplified method of processing CT data by binarizing the initial array by a given threshold. Thus, the nodal displacements obtained by the proposed approach are determined not only by the properties of the material, but also by its distribution in the volume. Previously, it was shown that the volume fraction of the bone substance and the material distribution have a much greater influence on the mechanical properties than other morphological parameters. In this regard, the method of estimating the distribution of properties based on tensor approximation is currently popular. In practice, this approach implies a preliminary calculation of such a tensor at the initial stage and its further application in constructing a mechanical model and in carrying out a numerical calculation. The method proposed in this work makes it possible to take into account the structural features of the material indirectly by weighted integration of the LSM of each FE. This method is not only less resource-intensive, but also more efficient in terms of calculation time.

Thus, the proposed method allows the use of separate CT data sections. In this case, based on the initial dimensions of the region, the regular grid is constructed. Further, according to the data on the distribution of the material density, the geometry of the sample is restored, which is an area of interest from the point of view of calculations. Boundary conditions can be applied to the computational grid obtained in this way, which, as a result, makes it possible to evaluate the strength of bone organs according to individual CT data. Based on the proposed approach to numerical modelling, typical loading cases can be considered. The quality of the obtained results can be assessed on the basis of the normalized energy error. The analysis of the received data can be automated based on the values of the normalized error and stress invariants (von Mises stress).

IV. CONCLUSION

The article presents one of the possible approaches to the numerical modelling of inhomogeneous medium elements based on their CT data. An algorithm for segmentation of initial images and a method for constructing a computational grid based on the peculiarity of the distribution of the material mechanical properties in the test sample volume are considered. A three-dimensional isoparametric linear FE of a continuous medium is constructed taking into account CT data. The implementation scheme of the numerical model is described, the concepts of computational and post-processing grids are introduced. The reliability assessment of the obtained results was based on the normalized energy error values. Three-point bending was modelled by the proposed method. The calculation results were compared with the data of a full-scale experiment. The resulting solution showed the effectiveness of the developed approach and its applicability to the study of porous bone structures.

ACKNOWLEDGMENT

The results are obtained under support from the Kazan Federal University Strategic Academic Leadership Program ("PRIORITY-2030").

FUNDING

This work was supported by the scholarship of the President of the Russian Federation for young scientists and graduate students (SP-4182.2022.4).

REFERENCES

[1] F. Marwa, E. Y. Wajih, L. Philippe, M. Mohsen, “Improved USCT of Paired Bones Using Wavelet-based Image Processing,” IJIGSP, vol. 10(9), pp. 1–9, 2018. DOI: https://doi.org/10.5815/ijigsp.2018.09.01.

[2] A. A. Kichenko, V. M. Tverier, Y. I. Nyashin, A. A. Zaborskikh, “Experimental determination of the fabric tensor for cancellous bone tissue,” Russ. J. Biomech., vol. 15(4), pp. 66–81, 2011.

[3] P. V. Bolshakov, R. M. Tikhilov, A. V. Mazurenko, “Numerical study of the influence of the degree of undercoverage of the acetabular component reinforced by screw on the load-bearing capacity of hip joint endoprosthesis,” Russ. J. Biomech., vol. 22(1), pp. 15–24, 2018.

[4] O. Sachenkov, R. Hasanov, P. Andreev, Y. Konoplev, “Determination of muscle effort at the proximal femur rotation osteotomy,” IOP Conf. Ser. Mater. Sci. Eng., vol. 158(1), article 012079, 2016.

[5] G. V. Fedotenkov, A. V. Gritskov, D. Y. Levitskiy, Y. A. Vahterova, Y. Sun, “Timoshenko beam and plate non-stationary vibrations,” INCAS Bull., vol. 13, pp. 41–56, 2021. DOI: https://doi.org/10.13111/2066-8201.2021.13.S.5.

[6] G. V. Fedotenkov, D. I. Makarevskii, Y. A. Vahterova, T. Q. Thang, "The inverse non-stationary problem of identification of defects in an elastic rod," INCAS Bull., vol. 13, pp. 57–66, 2021. DOI: https://doi.org/10.13111/2066-8201.2021.13.S.6.

[7] N. V. Kharin, O. V. Gerasimov, P. V. Bolshakov, A. A. Khabibullin, A. O. Fedyanin, M. E. Baltin, T. V. Baltina, O. A. Sachenkov, “Technique for determining the orthotropic properties of the bone organ according to computer tomograph,” Russ. J. Biomech, vol. 23(3), pp. 395–402, 2019.

[8] A. Ridwan-Pramana, P. Marcian, L. Borak, N. Narra, T. Forouzanfar, J. Wolff, “Finite element analysis of 6 large PMMA skull reconstructions: A multi-criteria evaluation approach,” PLoS ONE, vol. 12, article e0179325, 2017. DOI: https://doi.org/10.1371/journal.pone.0179325.

[9] J. Y. Rho, M. C. Hobatho, R. B. Ashman, “Relations of mechanical properties to density and CT numbers in human bone,” Med. Eng. Phys., vol. 17(5), pp. 347–355, 1995.

[10] P. Bolshakov, I. Raginov, V. Egorov, R. Kashapova, R. Kashapov, T. Baltina, O. Sachenkov, “Design and optimization lattice endoprosthesis for long bones: Manufacturing and clinical experiment,” Mater., vol. 13(5), article 1185, 2020. DOI: https://doi.org/10.3390/ma13051185.

[11] P. V. Bolshakov, O. A. Sachenkov, “Destruction simulation for the inhomogeneous body by finite element method using computed tomography data,” Russ. J. Biomech., vol. 24(2), pp. 248–258, 2020.

[12] L. Giovannelli, J. J. Rodenas, J. M. Navarro-Jimenez, M. Tur, “Direct medical image-based Finite Element modelling for patient-specific simulation of future implants,” Finite Elem. Anal. Des., vol. 136, pp. 37–47, 2017. DOI: https://doi.org/10.1016/j.finel.2017.07.010.

[13] P. Marcian, J. Wolff, L. Horačkova, J. Kaiser, T. Zikmund, L. Borak, “Micro finite element analysis of dental implants under different loading conditions,” Comput. Biol. Med., vol. 96, pp. 157–165, 2018. DOI: https://doi.org/10.1016/j.compbiomed.2018.03.012.

[14] O. Gerasimov, N. Kharin, O. Vorobyev, O. Sachenkov, “Determination of the mechanical properties distribution of the sample by tomography data,” J. Phys. Conf. Ser., vol. 1158, article 022046, 2019. DOI: https://doi.org/10.1088/1742-6596/1158/2/022046.

[15] G. Hettich, R. A. Schierjott, H. Ramm, H. Graichen, V. Jansson, M. Rudert, F. Traina, T. M. Grupp, “Method for quantitative assessment of acetabular bone defects,” J. Orthop. Res., vol. 37(1), pp. 181–189, 2018. DOI: https://doi.org/10.1002/jor.24165.

[16] T. Moriya, H. R. Roth, S. Nakamura, H. Oda, K. Nagara, M. Oda, K. Mori, "Unsupervised segmentation of 3D medical images based on clustering and deep representation learning," Prog. Biomed. Opt. Imaging – Proc. SPIE, vol. 10578, article 1057820, 2018. DOI: 10.1371/journal.pone.0188717.

[17] A. Tveito, K. H. Jaeger, M. Kuchta, K.-A. Mardal, M.E. Rognes, "A cell-based framework for numerical modeling of electrical conduction in cardiac tissue," Front. Phys., vol. 5(5), article 48, 2017. DOI: https://doi.org/10.3389/fphy.2017.00048.

[18] E. Nadal, J. J. Rodenas, J. Albelda, M. Tur, J. E. Tarancon, F.J. Fuenmayor, "Efficient finite element methodology based on cartesian grids: application to structural shape optimization," Abstr. Appl. Anal., article 953786, 2013. DOI: https://doi.org/10.1155/2013/953786.

[19] M. Viceconti, M. Qasim, P. Bhattacharya, L. Xinshan, "Are CT-Based Finite Element Model Predictions of Femoral Bone Strengthening Clinically Useful?" Curr. Osteoporos. Rep., vol. 16, pp. 216–223, 2018. DOI: https://doi.org/10.1007/s11914-018-0438-8.

[20] O. V. Gerasimov, N. V. Kharin, A. O. Fedyanin, P. V. Bolshakov, M. E. Baltin, E. O. Statsenko, F. O. Fadeev, R. R. Islamov, T. V. Baltina, O. A. Sachenkov, "Bone Stress-Strain State Evaluation Using CT Based FEM," Front. Mech. Eng., vol. 7, article 688474, 2021. DOI: https://doi.org/10.3389/fmech.2021.688474.

[21] O. V. Gerasimov, D. V. Berezhnoi, P. V. Bolshakov, E. O. Statsenko, O. A. Sachenkov, "Mechanical model of a heterogeneous continuum based on numerical-digital algorithm processing computer tomography data," Russ. J. Biomech., vol. 23, pp. 87–97, 2019.

[22] O. Gerasimov, N. Kharin, E. Statsenko, D. Mukhin, D. Berezhnoi, O. Sachenkov, "Patient-Specific Bone organ modeling using CT based FEM," Lect. Notes Comput. Sci. Eng., 2021, in press.

[23] F. Fadeev, A. Eremeev, F. Bashirov, R. Shevchenko, A. Izmailov, V. Markosyan, M. Sokolov, J. Kalistratova, A. Khalitova, R. Garifulin, R. Islamov, I. Lavrov, "Combined Supra- and Sub-Lesional Epidural Electrical Stimulation for Restoration of the Motor Functions after Spinal Cord Injury in Mini Pigs," Brain Sci., vol. 10(10), p. 744, 2020. DOI: https://doi.org/10.3390/brainsci10100744.

[24] T. D. Crenshaw, E. R. Peo, Jr., A. J. Lewis, B. D. Moser, "Bone Strength as a Trait For Assesing Mineralization in Swine: A Critical Review of Techniques Involved," Sci. J. Anim. Sci., vol. 53(3), pp. 827–835, 1981. DOI: https://doi.org/10.2527/jas1981.533827x.

[25] K. Imai, "Computed tomography-based finite element analysis to assess fracture risk and osteoporosis treatment," World J. Exp. Med., vol. 5(3), pp. 182–187, 2015. DOI: https://doi.org/10.5493/wjem.v5.i3.182.