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Left ventricular assist devices of a new generation

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Left ventricular assist devices of a new generation

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Abstract

The most competitive left ventricular assist devices (LVAD) in the world were identified in this article. 3D models of two samples were constructed for pumping blood based on existing pumps. The models were imported into the software package, where the flow of blood through these models was modeled. For this, a computational grid was constructed. According to the results of calculations, the main characteristics of the samples were obtained, as well as pictures of the distribution of velocities and pressures in the flow part. The calculation used a number of basic hydraulic equations that form a mathematical model.

Introduction

LVADs are becoming increasingly popular in medicine. They resolve the problem of heart failure and other heart diseases. Such devices can be divided into 3 generations. The most modern pumps of the 3rd generation have a continuous flow, are most compact and cause minimal damage to blood bodies. [1]-[4]

As part of this study, two centrifugal pumps to support the left ventricle of the heart were calculated.

The size of the first pump is 69mm in diameter, 35mm in height, weight – 475g. A flow rate of 7L/min was obtained at 135mmHg pressure with 8 W of power consumption. At a rotational speed of 2000-5500rpm, 10L/min can be produced. [4]

The second pump is made of titanium with an anti-thrombogenic coating and weighs 370 g. It has dimensions of 55x64 mm. Impeller with a diameter of 36 mm open type, designed to prevent blood stagnation in the working chambers. A flow rate is 12 L/min against 100 mmHg pressure at 2600 rpm. The hemolysis rate is 0.005 ± 0.002 g/100 L. [5]

The main objective of the study is to improve the characteristics of devices in comparison with prototypes, as well as reduce their effects on the human body.

The calculations of two prototypes were carried out in order to further improve them on parameters such as head and efficiency. In modern blood pumps, the efficiency is very low. Therefore, one of the objectives of the study is to improve the pumps in this parameter as much as possible.

Mathematical model

The prototypes for the calculated pumps are the Evaheart and HeartMate III LVADs (Fig. 1, 2).



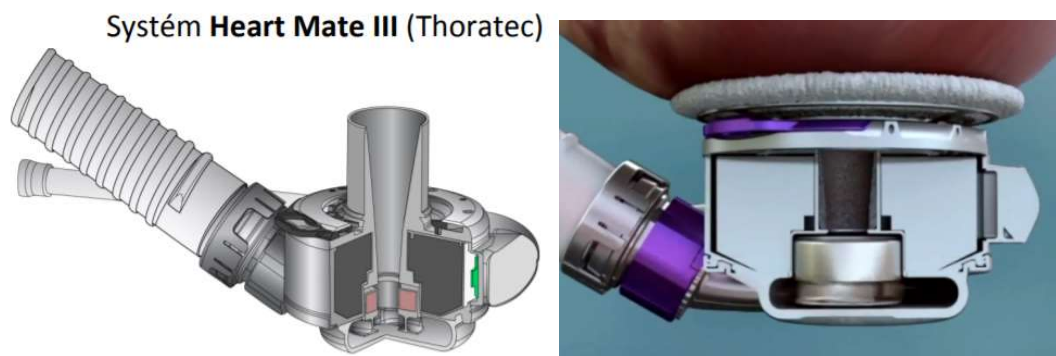


Fig. 1.HeartMateIII LVAD

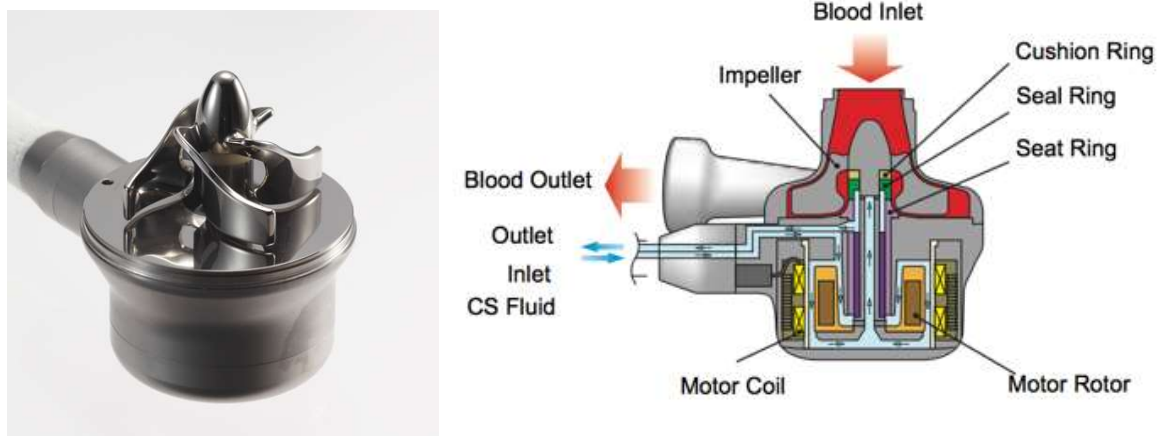


Fig. 2.Evaheart LVAD.

On the basis of the above-mentioned prototypes, initial models of two LVADs were created (Fig.3, 4).

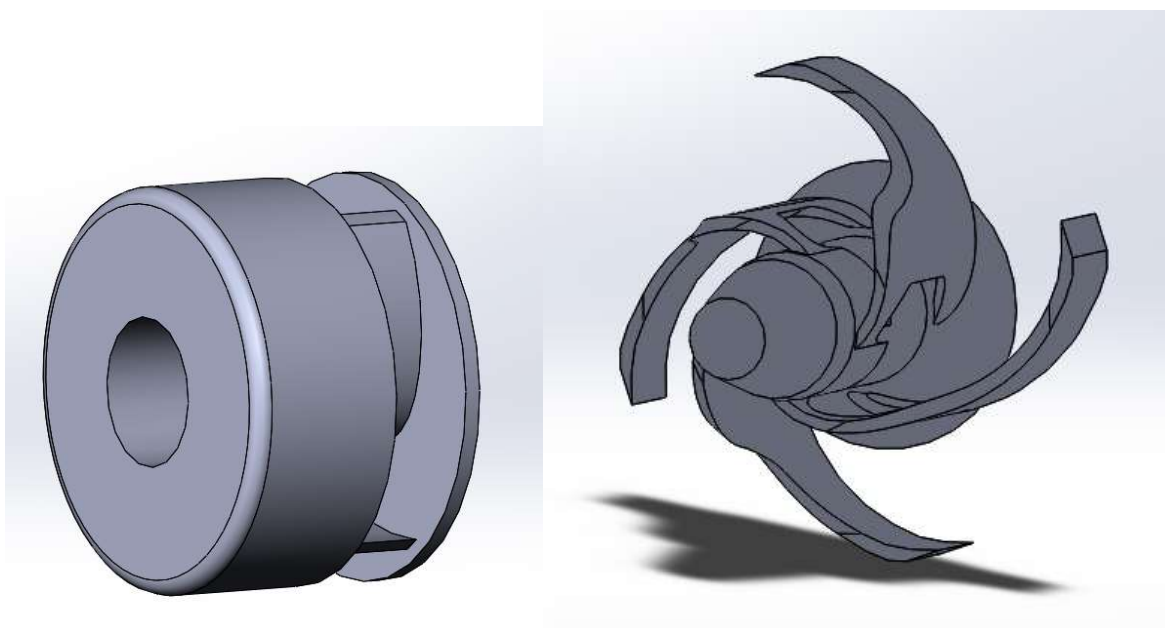


Fig. 3.Impellers built in SolidWorks

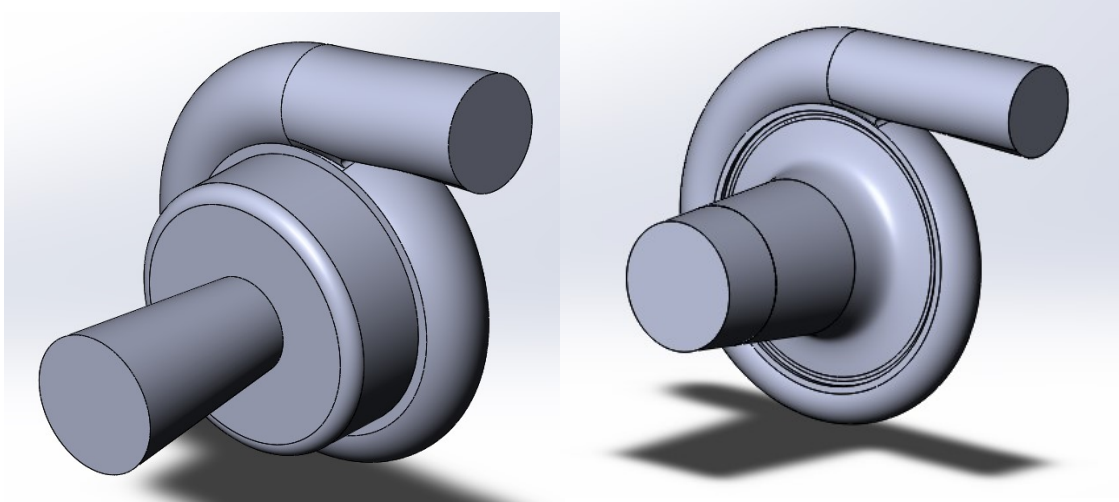


Fig.4.Flow parts built in SolidWorks

In the case of calculating the flow of incompressible fluid inside the pump, the following equations are applied:

The continuity equation for incompressible fluid.

$$\frac{\partial V_i}{\partial x_i} = 0$$

The equation of conservation of momentum averaged over time (Navier – Stokes equation averaged over Reynolds).

$$\rho \left(\frac{\partial V_i}{\partial t} + V_j \frac{\partial V_i}{\partial x_j} \right) = -\frac{\partial p}{\partial x_i} + \frac{\partial}{\partial x_i} \left(\mu \left(\frac{\partial V_i}{\partial x_j} + \frac{\partial V_j}{\partial x_i} \right) - \rho V_i' V_j' \right) + F_{mi}$$

Terms of the form $\rho \langle V_i' V_j' \rangle$ are called Reynolds stresses. They appear in the equations as a result of the formal averaging procedure and make the system of equations open. As a result, it becomes necessary to introduce additional assumptions and relationships in the original system of equations.

One of the main assumptions is the adoption of the Boussinesq hypothesis:

$$-\rho V_i' V_j' = \mu_t \left(\frac{\partial V_i}{\partial x_j} + \frac{\partial V_j}{\partial x_i} \right) - \frac{2}{3} \rho k \delta_{ij}$$

The combination of new relationships and assumptions forms a model of turbulence.

The model is complemented by two additional equations for the transfer of the kinetic energy of turbulence and the relative dissipation rate of this energy.

$$\frac{\partial k}{\partial t} + V_j \frac{\partial k}{\partial x_j} = P_k - \beta^* k \omega + \frac{\partial}{\partial x_j} \left((\nu + \sigma_k \nu_T) \frac{\partial k}{\partial x_j} \right)$$

$$\frac{\partial \omega}{\partial t} + V_j \frac{\partial \omega}{\partial x_j} = \alpha S^2 - \beta \omega^2 + \frac{\partial}{\partial x_j} \left((\nu + \sigma_\omega \nu_T) \frac{\partial \omega}{\partial x_j} \right) + 2(1 - F_1) \sigma_{\omega 2} \frac{1}{\omega} \frac{\partial k}{\partial x_i} \frac{\partial \omega}{\partial x_i}$$

The simulation was carried out in the software package STAR-CCM + 13.06.012-R8.

In order to simulate the flow of blood in the devices, a computational grid and the most illustrative sections of the computational domain were created (Fig.5). The grid captures the entire computational area of the prototypes, which makes it possible to calculate not only the hydraulic component of the efficiency, but also the volume component. On the second model, 2 extruders were created at the pump inlet and at the outlet. This is done to smooth the inlet vortices and the unevenness of the velocity distribution field at the outlet.[6]-[9]

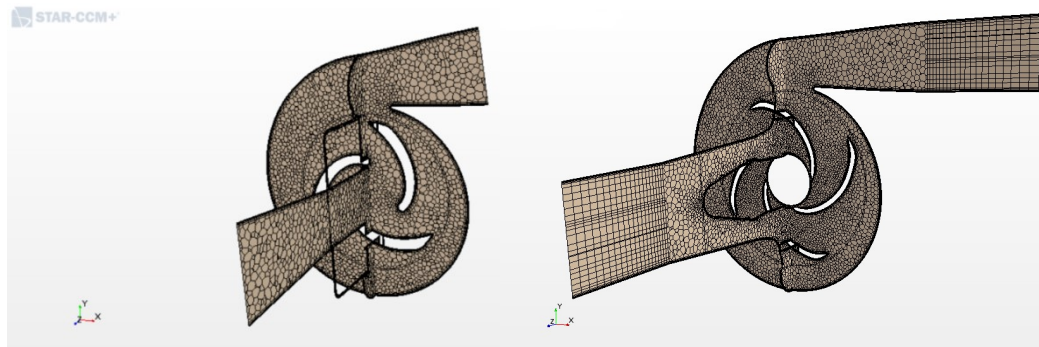


Fig. 5. Cross sections of the grid.

The input speed is set based on the flow rate values, the cross-sectional area and the rotor speed. It is 1.18 m/s and 5000 rpm for the first prototype and 0.52 m/s and 2600 rpm for the second.

Simulation results

After the flow stabilized in the models, the following graphs and flow patterns in the flow parts were obtained[10] (Fig.6,7,8,9):

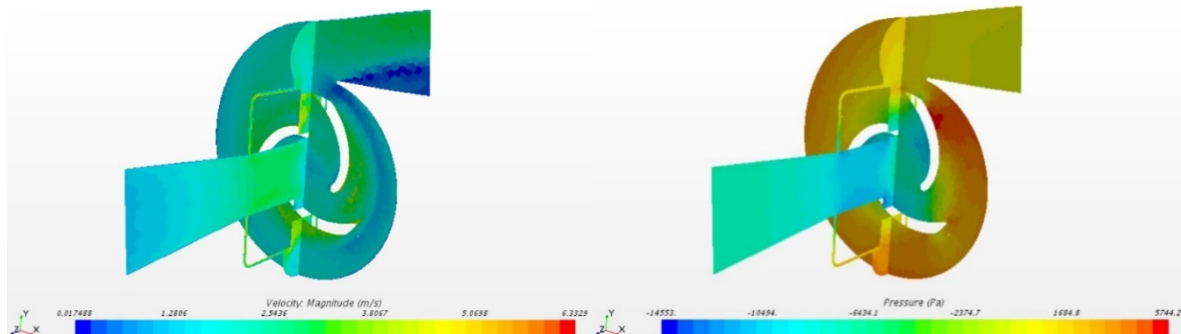


Fig.6. Velocity and pressure distribution scenes in the flow part of the first model.

In the pictures of the velocity and pressure distributions there are areas of fairly strong irregularity near the “cut-water” of the outlet (Fig.6). However, the stresses arising in these areas are permissible, and, presumably, will not have a large effect on hemolysis.

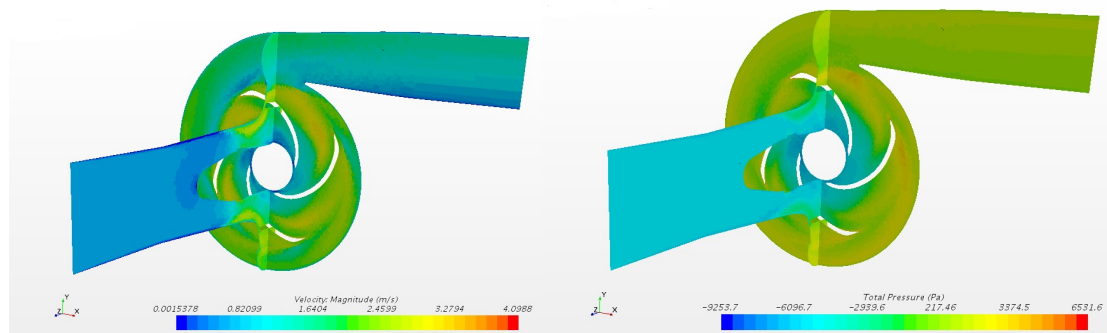


Fig.7.Velocity and pressure distribution scenes in the flow part of the second model

Despite the quite complicated shape of the impeller and the flow part, the fluid moves evenly and practically without turbulence, which indicates a successful shape of the flow part. The pressure in the flow part also varies uniformly, without obvious deviations (Fig. 7). All this suggests low damage to blood cells in the process of passing through the impeller and the possibility of testing this model.[11]-[15]

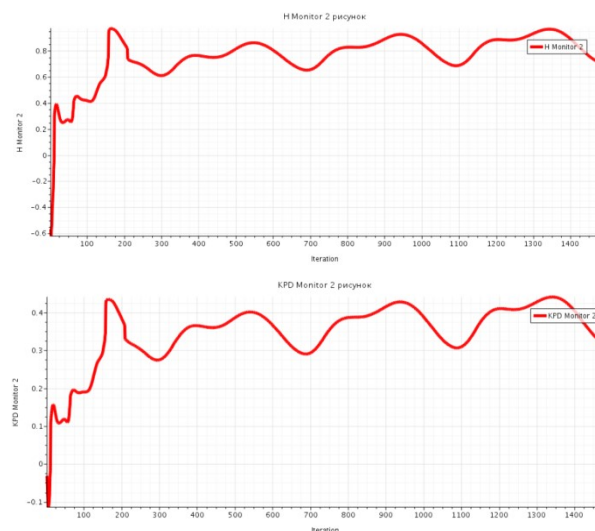
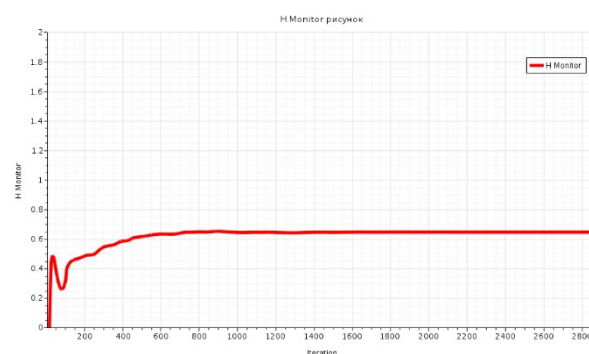


Fig.8.Head and efficiency graphs of the first model

During the calculation, the pump impeller turned about 1.3 times. The graphs show regularly repeated pulsations during the passage of each blade of a certain position (Fig.8). The results of calculations of the first prototype indicate some irregularity of flow in the pump [16]. The obtained average pressure value (0.75 m) can provide sufficient blood circulation during quiet life, the efficiency value of 35% is a good indicator for blood pumps.[17]-[18]



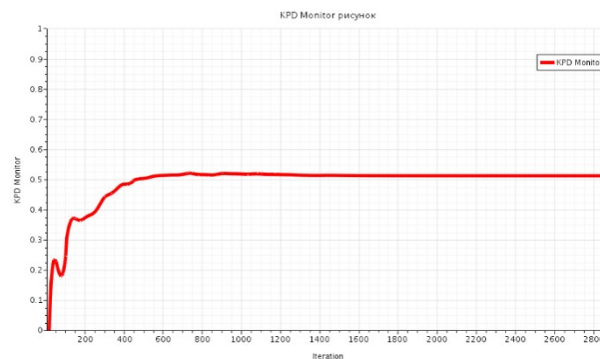


Fig.9.Head and efficiency graphs of the second model

The results of calculations of the second prototype showed that a well-received head of 0.65 m can ensure normal blood circulation in conditions of complete calmness of a person. Regarding the energy consumption of this model, a hydraulic efficiency of 51.5% was obtained, which is a good indicator, given the size of the device and the complicated shape of the flow part. The smoothness of the graphs shows that the prototype ensures continuous operation in this mode and maintains the necessary head for a long time (Fig.9).[19]-[20]

Conclusion

Based on the obtained simulation results, it can be concluded that the developed prototypes are quite promising, which allows us to continue working on their optimization. Already at this design stage, we can say that the first model outperforms the second in the head, but is inferior to it in efficiency. These models allow further calculation of hemolysis, which will give a more complete picture of the flow of blood in the flow parts.

Reference

- [1] FrancisD. Pagani. Continuous-FlowRotaryLeftVentricular Assist Devices with “3rd Generation” Design. *SeminThoracCardiovascSurg* 2008, 20: 255-263.
- [2] Steven M. Parnis, Mehmet HakanAkay, O.H.Frazier. Newer-Generation Rotary Blood Pumps. *Springer Science+Business Media New York* 2013.
- [3] SvetankAgarway, Kane M. High. Newer-genertion ventricular assist devices. *Best Practice & Clinical Anesthesiology* 26 (2012) 117-130.
- [4] Hideo Hoshi, TadahikoShinshi, SetsuoTakatani. Third-generation Blood Pumps With Mechanical Noncontact Magnetic Bearings. *Artificial Organs* 30(5):324–338, *Blackwell Publishing, Inc.*
- [5] Kenji Yamazaki, Shinichiro Kihara, Takehide Akimoto, Osamu Tagusari, Akihiko Kawai, MitsuoUmez, Jun Tomioka, Robert L. Kormos, Bartley P. Griffin, Hiromi Kurosawa. EVAHEART: An Implantable Centrifugal Blood Pump for Long-Term Circulatory Support. *Jpn J ThoracCardiovascSurg* 2002, 50: 461-465.
- [6] Lomakin, V.O., Kuleshova, M.S., Bozh’eva, S.M. Numerical Modeling of Liquid Flow in a Pump Station (2016)*Power Technology and Engineering*, 49 (5), pp. 324-327.
- [7] JeerasitBumrungetch, Andy Chit Tan, Shu-Hong Liu, Xian-Wu Luo, Qing-Yu Wu, Jian-Ping Yuan, and Ming-Kui Zhang. Flow Evaluation and Hemolysis Analysis of BVAD Centrifugal Blood Pump by Computational Fluids Dynamics. *International Journal of Fluid Machinery and Systems Vol. 7, No. 1, January-March 2014.*
- [8] M. ErtanTaskin, Tao Zhang, Katharine H. Fraser, Bartley P. Griffith, Zhongjun J. Wu. Design Optimization of a Wearable Artificial Pump-Lung Device With Computational Modeling. *Journal of Medical Devices* SEPTEMBER 2012, Vol. 6 / 031009-1.

- [9] Valeria Marinova, ImanKerroumi, Andreas Lintermann, Jens Henrik Gobbert, Charles Moulinec, Sebastian Rible, Yvan Fournier, Mehdi Behbahani, Numerical Analysis of the FDA Centrifugal Blood Pump. *NIC Series 2016, Vol. 48, ISBN 978-3-95806-109-5, pp. 355.*
- [10] Loktionov, E.Y., Mikhaylova, M.G., Sitnikov, D.S. Experimental setup for studying dynamics of the calcium interaction in cells (2016)*Instruments and Experimental Techniques*, 59 (4), pp. 601-606.
- [11] Christoph Gross, Christiane Marko, Johann Mikl, Johann Altenberger, Thomas Schlöglhofer, Heinrich Schima, Daniel Zimpfer, Francesco Moscato. LVAD Pump Flow Does Not Adequately Increase with Exercise. *Artificial Organs* 2018, 0(0):1–7.
- [12] Gousskov, A.M., Lomakin, V.O., Banin, E.P., Kuleshova, M.S. Minimization of Hemolysis and Improvement of the Hydrodynamic Efficiency of a Circulatory Support Pump by Optimizing the Pump Flowpath (2017) *Biomedical Engineering*, 51 (4), pp. 229-233.
- [13] BenteThamsen, Bastian Blümel, Jens Schaller, Christian O. Paschereit, Klaus Affeld, Leonid Goubergrits, Ulrich Kertzscher. Numerical Analysis of Blood Damage Potential of the HeartMate II and HeartWare HVAD Rotary Blood Pumps. *Artificial Organs* 2015, 39(8):651–659.
- [14] Vladimirov, I.Y., Korchagin, N.N., Savin, A.S. Effects of the Influence of a Wave on an Underwater Obstacle by a Stratified Flow of Finite Depth (2018) *Doklady Earth Sciences*, 478 (1), pp. 132-136.
- [15] Gousskov, A.M., Lomakin, V.O., Banin, E.P., Kuleshova, M.S. Assessment of Hemolysis in a Ventricular Assist Axial Flow Blood Pump (2016) *Biomedical Engineering*, 50 (4), pp. 233-236.
- [16] M. Behbahani, M. Behr, M. Hormes, U. Steinseifer, D. Arora, O. Coronado and M. Pasquali. A review of computational fluid dynamics analysis of blood pumps. *Euro. Jnl of Applied Mathematics* (2009), vol. 20, pp. 363–397.
- [17] Gordon Paul, Amin Rezaenia, EldadAvital, TheodosiosKorakianitis. Machinability and optimisation of shrouded centrifugal impellers for implantable blood pumps. *Journal of Medical Devices*, March 17, 2017.
- [18] GousskovA.M.,SorokinF.D., BaninE.P. Simulation of the input design of an implantable axial blood pump. *Medical Equipment* 2016.- № 1.- pp. 11 – 14.
- [19] Yurechko, V.N., Omel'yanchuk, A.M., Yurechko, R.V., Kuznetsova, E.I., Fadeev, A.A., Radkevich, F.A. Study of the Functional Characteristics of Artificial Cardiac Valves (2003)*Biomedical Engineering*, 37 (2), pp. 101-110.
- [20] BaninE.P.,GousskovA.M., SorokinF.D. Analysis of modern approaches to the design of artificial ventricles of the heart of the rotor type. *Science and education: a scientific publication of BMSTU* 2015.- № 02.