Investigation of the influence of centrifugal pump wet part geometry on hemolysis index

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Abstract: The present study focuses on the influence of the circulatory support pump wet part geometry on the level of damage experienced by blood cells in it. The object of the study is a centrifugal pump: axial input, impeller and volute. The values of density and viscosity are constant and are close to the physiological parameters of blood of a healthy human. In order to estimate hemolysis blood cells are considered as spherical particles, which do not interact with each other. At each point of the trajectory of the particle equivalent shear stress is calculated according to Huber-Mises criterion. Hemolysis is then calculated in accordance with Lagrangian approach to particle movement in a flow. As an experimental hemolysis curve a power law of equivalent shear stress depending on the time the particle has experienced this stress was chosen. For a wet part optimization criterion of maximum efficiency and minimum hemolysis are considered.

1. Introduction
Interest in wearable circulatory systems has increased significantly over the last ten years. Every year the number of publications and patent applications in the world on this subject is steadily increasing (according to lens.org$^1$ the number of patents and patent applications on this subject in 2014 reached 5 784). Based on the large amount of works, it can be concluded that in the medium term, miniature assist devices will be the main means to save patients’ lives at the terminal stages of heart failure, in case the heart transplantation is impossible. However, during the period of circulatory support, wearable assist pumps have complications [1–5]: bleeding, thrombosis, stroke, etc.

Auxiliary circulation pumps are divided into two types: rotary and volumetric. Currently, according to statistics from [1] rotor pumps of two types prevail in medical practice: axial and radial. The presented schemes differ structurally from each other and suggest different approaches to implantation, but in general there is now little data that suggests the absolute advantages of one type

$^1$ Request: “(Blood pump) AND implantable”
over the other. However, in a number of works similar comparisons are made. The article [6] presents data on the implantation of 105 axial pumps HeartMate 2 (Thoratec Inc.) and 34 centrifugal HVAD (HeartWare Inc.). Groups of patients were selected to ensure comparability of obtained results. As a result of the study, no statistically significant difference in medical indications was found for the case of implantation of an axial or radial pump. The study [7] (the sample consists of 194 HeartMate 2 devices and 209 HVAD) also notes that the final patient parameters are nearly the same, but pump replacement (due to breakage, unit wear, scheduled replacement) is more often required for axial HeartMate 2. We note here the study [8], in which 68 patients participated. The patients had radial pumps Duraheart \((n = 15)\), Evaheart \((n = 23)\) and axial pumps HeartMate 2 \((n = 22)\), Jarvik 2000 \((n = 8)\) installed. As a result of the study, the survival rates in two groups of patients (with an installed radial or axial pump) had similar survival rates and the frequency of infection of the supply cable, the researchers also noted that patients with radial pumps more often have a violation of the cerebral circulation. Some additional data on the results of clinical practice can be found in [9–11].

The interest of this work is focused on the analysis of design flaws in the radial HVAD. Based on the analysis, a modification of the centrifugal pump is proposed and calculations of hydraulic parameters are given, theoretical hemolysis in the wet part of the pump is also calculated.

2. Geometric model
The simulation object is a centrifugal pump: axial input, impeller and volute. A pump HVAD [12] (HeartWare Inc., USA) is considered as a prototype. Figure 1 shows that this pump has an unusual for centrifugal pumps geometry – there are no curved blades in impeller, interblade channel is narrowed.

![Figure 1. Geometry of HVAD wet part.](image)

3. Mathematical simulation methods
Hydrodynamics is simulated using STAR-CCM+ software with additional C++ macro providing assessment of the accumulated damage. In order to solve differential equations a finite volume method is used. The flow region is discretized into small cells, in this case there are 1.5 million of them, in each cell such parameters as pressure, velocities, etc. are calculated. Reynolds number inside the pump can reach the value of 72 000 (at the interblade channels), at the same time there are areas with laminar flow there.

Shear Stress Transform Method (SST \(k–\omega[13–16]\)) is used to describe the flow. This model has already showed its feasibility during other simulations [13]. Since the flow cannot be considered as fully turbulent more common \(k–\varepsilon\) and \(k–\omega\) models can lead to computational errors [17–19]. Hybrid SST \(k–\omega\) model is a combination of the models mentioned above. It uses \(k–\omega\) model at the areas by the
wall (where laminar flows are concentrated) and \( k - \varepsilon \) model for turbulent areas; transient zones are simulated with special transition algorithms.

In order to save machine time as a first approximation all models are calculated at steady state, then a few models with best parameters are simulated at unsteady state. Mass flow of 5 l/min is assigned as an inlet boundary condition, 0 Pa pressure is assigned to the outlet boundary, the impeller rotational speed is 2 900 rpm. Characteristics of the fluid under consideration are close to those of blood: Newtonian fluid with dynamic viscosity of 0.0035 kg/m-s, which corresponds to a hematocrit level of 33% (the most probable class of patients to have such a device installed), density \( \rho = 1050 \text{ kg/m}^3 \). Control particles are non-interacting, their trajectories do not affect the flow.

4. Optimization parameters
Simulation of flow in the base model (original HVAD geometry) resulted in an unsatisfying flow pattern – a pressure drop can be seen in the channels and intensive vorticity is obtained (figure 2). This configuration of the wet part also has relatively low values of hydrodynamic parameters (head – 64 mm Hg, efficiency – 51.8%, HI – 0.0285).

**Figure 2.** Pressure (left) and turbulent kinetic energy (right) distribution at the pump cross section.

For further analysis geometry with curved blades was chosen (figure 3). It helps to obtain more uniform flow with better hydraulic characteristics, which in its case leads to decreased blood cells damage.

**Figure 3.** Geometry of the suggested pump wet part.
The following parameters were changing during optimization process (figure 4): number of impeller blades – \( z \) (varying from 3 to 6), impeller outlet width – \( b_2 \) (varying from 3.5 mm to 6 mm), impeller outer diameter – \( d_2 \) (varying from 24 mm to 46 mm), blade angle at impeller outlet – \( \beta_2 \) (varying from 20º to 40º), blade wrap angle – \( \varphi \) (varying from 90º to 170º).

Figure 4. Optimization parameters of the suggested pump.

5. Optimization algorithm
A huge amount of machine time required for the process of pump optimization with CFD methods imposes constraints on optimization algorithms, number of designs considered and parameters, that can be chosen. In this case optimization is based on LP\( _T \) sequences [20] which finds global minimum of the function relatively fast and is flexible in terms of chosen number of designs to be considered.

During the optimization 2 criteria were considered – maximum of efficiency and minimum of hemolysis.

Efficiency is calculated according to the following formula:

\[
\eta = \frac{Q(p_2 - p_1)}{\omega M},
\]

where
\( Q \) – volumetric flow, m\(^3\)/s;
\( p_2 \) – total pressure at the outlet of the pump, Pa;
\( p_1 \) – total pressure at the inlet of the pump, Pa;
\( M \) – torque, N\( \cdot \)m;
\( \omega \) – impeller rotational speed, rad/s.

Hemolysis. The value of shear stress experienced by blood cell during a period of time is usually considered as a source of cell damage. At the moment in order to obtain the level of hemolysis the following power law is used:

\[
HI(\%) = \frac{\Delta Hb}{HB} \cdot 100 = C\tau^{\alpha},
\]

where
\( HI \) – hemolysis index for one particle, %;
\( \Delta Hb \) – change of extracellular hemoglobin concentration in plasma;
$HB$ – total hemoglobin concentration;
$C$– empirical constant;
$\tau$ – equivalent shear stress;
$t$ – time interval of the cell experiencing the stress.

Formula (2) is one of the most common one (excluding more complex models, where there is still a power law with some additional features to obtain special blood cells behavior [21]). To apply formula (2) it is required to describe stress tensor in a scalar way. The following calculations use Huber—Mises criterion

$$\tau = \sqrt{\frac{1}{2} \left( (\sigma_1 - \sigma_2)^2 + (\sigma_2 - \sigma_3)^2 + (\sigma_1 - \sigma_3)^2 \right)}$$

where
$\sigma_i$– primary stress components.

Values of constants for (2) are based on [22].

To estimate the level of cells damage the index of hemolysis is averaged over all control particles [23,24]:

$$HI_{total} = \frac{1}{N} \sum_{p=1}^{N} HI_p$$

where
$HI_p$ shows the amount of cell $p$ damage accumulated on its way through the pump;
$N$ – number of control particles.

Simulation was carried for 128 design variants. In order to decrease the amount of numerical error each model was calculated using the same mesh parameters and boundary conditions.

6. Optimization results
The results of steady state simulations for the most representative designs are shown in table 1, the distributions of physical values are shown on figures 5–9.

Table 1. Some of simulation results.

| No. | $z$ | $b_2$ | $d_2$ | $\beta_2$ | $\phi$ | Efficiency, % | $H,m$ | $H,\text{mm Hg}$ | $HI$ |
|-----|-----|-------|-------|-----------|-------|---------------|-------|-----------------|------|
| 13  | 5   | 4.8   | 33.7  | 22.1      | 135   | 64.5          | 1.41  | 108.94          | 0.0451 |
| 14  | 4   | 6.3   | 44.2  | 31.1      | 96    | 57.7          | 2.73  | 210.92          | 0.249  |
| 20  | 4   | 4.5   | 38.3  | 21.6      | 142   | 67.0          | 1.73  | 133.66          | 0.0617 |
| 31  | 5   | 5.5   | 24.8  | 38.2      | 153   | 22.2          | 0.18  | 13.91           | 0.0263 |
| 32  | 4   | 4.0   | 35.3  | 29.2      | 114   | 68.4          | 1.43  | 110.48          | 0.0297 |
| 40  | 4   | 5.0   | 36.6  | 34.8      | 148   | 66.2          | 1.85  | 142.93          | 0.0495 |
| 48  | 4   | 3.7   | 36.0  | 23.0      | 151   | 67.4          | 1.39  | 107.39          | 0.0257 |
| 93  | 5   | 5.5   | 45.0  | 29.9      | 106   | 47.1          | 2.74  | 211.69          | 0.444  |
| 109 | 4   | 4.5   | 44.0  | 25.1      | 95    | 52.3          | 2.69  | 207.83          | 0.351  |
Figure 5. Pressure distribution at the cross sections of models no. 14 (left) and no. 13 (right).

Figure 6. Velocity distribution at the cross sections of models no. 14 (left) and no. 13 (right).

Figure 7. Velocity distribution at the cross sections of models no. 14 (left) and no. 13 (right) diffusors.
Figure 8. Equivalent shear stress distribution at the cross sections of models no. 14 (left) and no. 13 (right).

Figure 9. Equivalent shear stress distribution at the cross sections of models no. 14 (left) and no. 13 (right) diffusors.

In accordance with the data obtained the following dependences of the criteria under consideration on the parameters described above were pictured (figure 10).
A few of the designs with the best criteria values were calculated then at an unsteady state. The results of these simulations are represented in table 2.
Figure 10. The effect of optimization parameters on efficiency (left) and hemolysis index (right), approximation line (red) — polynomial of the third degree.

Table 2. Results of unsteady simulation.

| No. | z  | $b_2$ | $d_2$ | $\phi$ | Efficiency, % | $H_m$ | $H_{mm}$ Hg | $HI$ |
|-----|----|-------|-------|--------|---------------|-------|-------------|------|
| 133 | 5  | 4.8   | 33.7  | 4.8    | 65.2          | 1.44  | 111.25      | 0.0252 |
| 20  | 4  | 4.5   | 38.3  | 21.6   | 67.7          | 1.84  | 142.16      | 0.0345 |
| 32  | 4  | 4.0   | 35.3  | 29.2   | 69.1          | 1.5   | 115.89      | 0.0166 |
| 40  | 4  | 5.0   | 36.6  | 34.8   | 68.7          | 1.88  | 145.25      | 0.0277 |
| 48  | 4  | 3.7   | 36.0  | 23.0   | 68.2          | 1.47  | 113.57      | 0.0143 |
In accordance with optimization results the wet part geometry with the best criteria combination was chosen (design no. 32, table 2) and is shown on figure 11.

![Optimized wet part geometry of the pump under consideration.](image)

**Figure 11.** Optimized wet part geometry of the pump under consideration.

7. **Comparison of wet part geometry configurations in a wide range of mass flow values**

Requirements imposed on such pumps can change during the day, depending on patient’s kinesis and the medicine he is taking [25]. In particular, it will affect the mass flow the pump should work on. Therefore, in order to compare two wet part configurations described above, it is necessary to compare them in a wide range of mass flow values.

Both variants were calculated in the same conditions: unsteady flow, mesh and solver parameters were kept the same; volumetric flow was varying from 20% to 120% from BEP (from 1 l/min to 6 l/min). Some of the results to compare are shown on figures 12–14.

![Simulation results at \( Q = 0.4Q_{\text{BEP}} \).](image)

**Figure 12.** Simulation results at \( Q = 0.4Q_{\text{BEP}} \).
Figure 13. Simulation results at $Q = 0.8Q_{\text{BEP}}$.

Figure 14. Pump characteristics in a wide range of mass flow value: a) comparison of efficiency curves; b) comparison of hemolysis index curves.
8. Discussion and Conclusions
Conducted calculations allowed to conclude, that the suggested pump configuration with curved blades demonstrates better behavior than HVAD geometry at BEP and near it (at volumetric flow from 3 l/min to 6 l/min). High vorticity is obtained in the geometry with narrow channels. Under given conditions the suggested geometry efficiency is 14% higher and hemolysis index is nearly twice lower at BEP. Moreover, the mass of the device is also decreased in this case, what make it more comfortable to use. Nevertheless, in the zone of relatively low mass flows (from 1 l/min to 3 l/min) configuration with curved blades behaves significantly worse than the one with narrow channels (for volumetric flow 1 l/min its efficiency is 30% lower and hemolysis index is nearly 3-times higher). Such details should be taken into account when choosing frequency control system.

Obtained criteria dependences on parameters allow to conclude on the recommended intervals of the values to choose from when designing such devices: impeller width at the outlet ($b_2$) is recommended to vary between 3.5…4 mm, impeller diameter ($d_2$) between 32…37 mm, blade angle at the outlet ($\beta_2$) between 220°…270°, blade wrap angle ($\phi$) between 120°…140°. The data calculated is also a good base for the following analysis and additional optimization to be carried.

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References
[1] Kirklin J K et al 2017 Eighth annual INTERMACS report: Special focus on framing the impact of adverse events J. Hear. Lung Transplant. 36(10) pp 1080–86
[2] Gouskov A M, Sorokin F D and Banin E P 2016 Simulation of an Inlet Structure of an Implantable Axial Blood Pump Biomed. Eng. (NY)
[3] Su B, Chua L P and Zhong L 2013 Numerical Studies of an Axial Flow Blood Pump With Different Diffuser Designs J. Mech. Med. Biol. 13 (3) pp 1350029
[4] Wu J et al 2012 Computational Fluid Dynamics-Based Design Optimization for an Implantable Miniature Maglev Pediatric Ventricular Assist Device J. Fluids Eng. 134 (4) pp 041101
[5] Karimov J H et al 2017 The axial continuous-flow blood pump: Bench evaluation of changes in flow associated with changes of inflow cannula angle J. Hear. Lung Transplant. 36 (1) pp 106–112
[6] Gaffey A C et al 2018 Is there a difference in bleeding after left ventricular assist device implant: centrifugal versus axial? J. Cardiotorac. Surg. 13(1) pp 22
[7] F.H. S. et al 2017 Characteristics and outcomes of LVAD patients who undergo pump exchange: Axial flow vs. centrifugal flow therapy J. Hear. Lung Transplant. 36 (4) pp S99–S100
[8] Kimura M et al 2017 Cerebrovascular Accident Rate Is Different Between Centrifugal and Axial-Flow Pumps, but Survival and Driveline Infection Rates Are Similar Transplant. Proc. 49 (1) pp 121–124
[9] Kimura M et al 2015 Midterm outcome of implantable left ventricular assist devices as a bridge to transplantation: Single-center experience in Japan J. Cardiol. 65 (5) pp 383–389
[10] Lalonde S D et al 2013 Clinical Differences Between Continuous Flow Ventricular Assist Devices: A Comparison Between HeartMate II and HeartWare HVAD J. Card. Surg. 28 (5) pp 604–610
[11] Coffin S T et al 2015 Adverse neurologic events in patients bridged with long-term mechanical circulatory support: A device-specific comparative analysis J. Hear. Lung Transplant. 34 (12) pp 1578–1585
[12] Bell P E 1992 United States Patent. 2 (12)
[13] Versteg H and Malalasekera W 2007 An introduction to Computational Fluid Dynamics: The
Finite Volume Method. 2nd ed. (Pearson Education Limited, Harlow)

[14] Bui V et al 2017 Numerical study of the influence of flow blockage on the aerodynamic coefficients of models in low-speed wind tunnels (Springer) 24 (6) pp 857–866

[15] Isaev S A et al 2010 Influence of the Reynolds number and the spherical dimple depth on turbulent heat transfer and hydraulic loss in a narrow channel Int. J. Heat Mass Transf. 53 (1–3) pp 178–197

[16] Isaev S A et al 2016 Numerical simulation of the turbulent air flow in the narrow channel with a heated wall and a spherical dimple placed on it for vortex heat transfer enhancement depending on the dimple depth Int. J. Heat Mass Transf. 94 pp 426–448

[17] Fraser K H et al 2011 The use of computational fluid dynamics in the development of ventricular assist devices Medical Engineering and Physics 33 (3) pp 263–280

[18] Devisilov V A, Sharai E Y 2016 Numerical study of the flow structure in a hydrodynamic filter Theor. Found. Chem. Eng. 50 (2) pp 209–216

[19] Kulik V V, Parkin AN and Navasardyan E S 2016 Numerical Modeling Procedure for Micromachined Cryogenic Cooler Elements Using ANSYS Fluent Software and Viscous Flow in a Small-Diameter Channel with Heat Transfer as an Example Chem. Pet. Eng. 52 (7–8) pp 531–538

[20] Chaburko P S, Lomakin V O and Kuleshova M S B.M.N. 2016 Comprehensive optimization of the hermetic pump’s flow part using LP-TAU search method Pumps. Turbines. System 1 (18)

[21] Yu H et al 2017 A Review of Hemolysis Prediction Models for Computational Fluid Dynamics Artif. Organs. 41 (7) pp 603–621

[22] Giersiepen M et al 1990 Estimation of Shear Stress-Related Blood Trauma in Heart Valve Prostheses in vivo Comparison of 25 Aortic Valves Artif. Organs. 12 (5) pp 130–136

[23] Gouskov A M et al 2017 Minimization of Hemolysis and Improvement of the Hydrodynamic Efficiency of a Circulatory Support Pump by Optimizing the Pump Flowpath Biomed. Eng. (NY)

[24] Gouskov A M et al 2016 Assessment of Hemolysis in a Ventricular Assist Axial Flow Blood Pump Biomed. Eng. (NY)

[25] Demidova M and Tikhonenko V 2001 Circadian rhythm of heart rate variability in healthy subjects Journal of Arrhythmology 23 pp 52–58