Design optimization of flow channel and performance analysis for a new-type centrifugal blood pump

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Abstract. In this paper, a new-type centrifugal blood pump, whose impeller is suspended inside a pump chamber with hydraulic bearings, is presented. In order to improve the hydraulic performance of the pump, an internal flow simulation is conducted to compare the effects of different geometrical parameters of pump flow passage. Based on the numerical results, the pumps can satisfy the operation parameters and be free of hemolysis. It is noted that for the pump with a column-type supporter at its inlet, the pump head and hydraulic efficiency decrease compared to the pump with a step-type support structure. The performance drop is caused by the disturbed flow upstream impeller inlet. Further, the unfavorable flow features such as reverse flow and low velocity in the pump may increases the possibility of thrombus. It is also confirmed that the casing shape can little influence pump performance. Those results are helpful for design optimization in blood pump development.

1. Introduction
In the world today, heart disease has become the top health killer because of high mortality. However, the scarcity of donor hearts leads to prolonged periods on the device. Fortunately, it is confirmed that left ventricular assist device technology can be used successfully for heart failure patients. For examples, 74 Heart Mate II axial flow pumps and 41 Heart Ware centrifugal pumps were implanted in patients, and a 1-year survival of 73% and a 2-year survival of 69% were recorded [1].

Therefore, the development of sufficient and stable support devices for these patients is very important and urgent. Many researchers have contributed for blood pump technology [2]. However, it is believed that the rotor (impeller) supporting technology is crucial for a turbo-type blood pump. Akamatsu et al. [3] proposed a centrifugal blood pump with a magnetically suspended impeller. The impeller diameter is 50mm, and the pump has a complicated control system for the magnetic bearings. Recently, hydraulic dynamic bearings are used for blood pumps. Luo et al. [4] developed a shaft-less miniature turbo pump, whose rotor was suspended by a hydrodynamic bearing. The experimental test indicated that the pump can be operated steadily and silently. The pump made use of the double suction design to achieve the self-balancing axial force. But as a blood pump, it is not convenient when the ventricle is joint with double suction surgically [5].

Attention should also be paid on the internal flow inside a blood pump because the flow features decide the operation quality. For the case of Heart Ware centrifugal pump, it is noted that there are
scraps of meat and thrombus in the groove of hydrodynamic thrust bearings [6]. Besides, hydraulic performance is also need to be considered during blood pump development.

This paper presents a novel centrifugal blood pump, whose 25mm impeller is suspended by hydrodynamic bearings, including a thrust bearing and a radial bearing, and driven by a disc motor. To improve the operation reliability, the effects of support structure and casing shape on pump performance are analyzed by numerical methods. Based on the results, an optimized model is selected for blood pumps.

2. Pump Design and Geometries

The blood pump is designed with the following operation parameters: \(Q_d=6\) L/min, and \(H_d=2\) m at the rotational speed \(n=5000\) min\(^{-1}\). Thus, the specific speed \(n_s\) is \(108.5\) m\(^3\)s\(^{-1}\)min\(^{-1}\).

A semi-open impeller without a hub is selected. The velocity-coefficient method is used to design the 25mm diameter impeller with four blades. Figure 1 shows the meridional flow passage. The geometrical parameters of impeller and casing are listed in Table 1.

![Figure 1. Meridional configuration for pump impeller.](image)

| Parameter                        | Symbol | Value |
|----------------------------------|--------|-------|
| Inlet pipe diameter              | \(D_i\)/mm | 12    |
| Outlet pipe diameter             | \(D_o\)/mm | 9     |
| Impeller inlet diameter          | \(D_0\)/mm | 11    |
| Blade inlet diameter ratio at tip| \(D_{1t}\)/mm | 12.6  |
| Blade inlet diameter ratio at hub| \(D_{1h}\)/mm | 8     |
| Impeller exit diameter           | \(D_2\)/mm | 25    |
| Basic circle diameter of casing  | \(D_3\)/mm | 26    |
| Width of blade exit              | \(b_2\) | 3     |
| Exit angle of blade              | \(\beta_2\) / (°) | 34    |
| Blade number                     | \(Z\) | 4     |

Table 1. Geometrical parameters of the pump.

For a blood pump with hydrodynamic bearing, support structure plays a very important role at the starting period of operation while impeller hasn’t been suspended inside volute completely. At the same time, it has to come to terms with the invasion inside flow passage.

For comparison, models with two kinds of support structure are designed as demonstrated in figure 2. Note that the impeller shroud having a large width is applied to hold permanent magnets for a motor, and to form axial and radial hydrodynamic bearings during pump operation. Model 1 has a column-type supporter in pump inlet pipe and impeller inlet; Model 2 has a step at pump cover near impeller inlet. Both Model 1 and Model 2 have the same impeller blades and meridian flow passage.
The pump casing is a spiral one, whose sections are calculated according to identical velocity. In order to match the impeller, three different casing shapes are investigated as shown in figure 3. The basic casing shape is marked as “Casing 0”, Casing 2 has a smaller width near the casing centre, and Casing 1 has a larger width compared with the basic casing shape.

3. Numerical methods

3.1. Calculation domain and mesh
Because support structures locate inside the flow passage, the full flow passage including the inlet pipe and outlet pipe is considered as the calculation domain as shown in figure 4. Focusing on hydrodynamic performance, we ignore the clearance flow between the impeller and pump casing in this study. The unstructured tetra grid is used. About 1.76 million unstructured tetra mesh nodes have been discretized in each full flow passage.
3.2. Boundary conditions
The boundary conditions are defined as follows:
- The inlet of calculation domain is specified with averaged total pressure;
- At the outlet of the calculation domain, averaged mass flow-rate is given basing on mass equilibrium;
- For all solid walls, the non-slip wall condition is specified. The flow in all stationary components is considered in an absolute coordinate system. For the rotating impeller, a rotating coordinate system is set.
- Further, the sliding interfaces between two frames are treated as GGI method.

3.3. Operation conditions
Given different operation flow rate, such as $0.5Q_d$, $0.75Q_d$, $Q_d$, $1.25Q_d$, $1.5Q_d$ at the same rotational speed $n = 5000\text{ min}^{-1}$, a series of simulation have been completed respectively using water and blood as a flowing medium. It is noted that blood is treated as Newtonian fluid in this study though it is not in nature. The fluid properties of blood are defined as follows: density of $1055\text{Kg/m}^3$ and dynamic viscosity of $0.036\text{Pa.s}$.

The three-dimensional steady flows in the pump full passage were simulated based on RANS equations using the ANSYS CFX. The scalable k-epsilon turbulence model is applied for the turbulent flow analysis.

4. Results and analysis
4.1. Effects of support structure
Figure 5 and figure 6 are the hydraulic performance charts for the pump with different support structures. The horizontal axis stands for flow rate, and the vertical axis stands for pump head in Figure 5 and hydraulic efficiency in Figure 6.

It is noted that hydraulic performance of Model 2 is better than that of Model 1. With a column-type supporter near the impeller inlet, Model 1 has a lower head and efficiency than Model 2.

For both pump models, at the design condition, the pump head with blood of Model 1 is $2.07\text{m}$ and that of Model 2 is $2.14\text{m}$. That is to say, the output energy of blood pump can completely meet the need of systemic circulation for adults.
In order to make clear difference on the concrete details of internal flow, the velocity distributions at two sections i.e. Section A and Section B are shown in figure 7. Note that Section A is at the impeller inlet, and Section B is at the leading edge of impeller blade. It is obvious that the flow upstream impeller inlet is much more uniform for Model 2 compared with that of Model 1, which is disturbed by the supporter. For Model 1, there are zones with very small velocity, where there will be danger of thrombus.

Due to the influences of support structure, the flows inside the pump are also different. Figure 8 shows the streamlines in full flow passage and near blade pressure side for both models. It can be seen that the flow in Model 2 is smoother, while there seems to have reverse flows in impeller for Model 1. The unfavourable flow features such as reverse flow and non-uniform velocity distribution will result in performance drop for Model 1.

**Figure 5.** $H$-$Q$ Curves.

**Figure 6.** $\eta$-$Q$ Curves.

**Figure 7.** Velocity contour on Section A and B (blood flow).
In generally, pump with Model 2 has better hydraulic performance and less thrombus in main flow passage. However, the flow in clearance passage should be studied in future.

Figure 9 shows the wall shear stress value along the intersecting line of blade surface and the section at blade mid span. Basically, the wall shear stress on suction surface is large, whose value is less 60 Pa. The wall shear stress changes much at different area along the blade surface, and there is the largest wall shear stress at blade leading edge. On blade suction surface, the wall shear stress for Model 1 is larger than that for Model 2. According to Giersiepen et al. [7], the risk of hemolysis is negligible for both pump models based on the present results.

Figure 8. Streamlines of whole passage and blade pressure surface (blood flow).
4.2. Effects of casing shape
The comparison of pump performance with different casing shapes is listed in Table 2. Among these casings, Casing 0 is the best because it has the highest hydraulic efficiency and the smallest wall shear stress. It is clear that the effects of casing shape on hydraulic performance as well as hemolysis are limited. It is convenient to change the casing shape according to the needs of pump design optimization.

It is noted that the maximum wall shear stress for all pump models are within 255 Pa. Thus, those pump models are free of hemolysis. Due to the difference of viscosity, the maximum wall shear stress occurs at different location in the pump: blade trailing edge for water flow, and volute tongue for blood flow.

| Casing No. | Fluid: Water | Fluid: Blood |
|------------|--------------|--------------|
|            | 0            | 1            | 2            | 0            | 1            | 2            |
| Head (m)   | 2.101        | 2.12         | 2.118        | 2.07         | 2.095        | 2.09         |
| Efficiency (%) | 78.93     | 78.5         | 78.38        | 78.8         | 78.04        | 78.02        |
| Max wall shear stress (Pa) | 202.678 | 208.424     | 206.867      | 211.796      | 211.968      | 211.402      |
| Max wall shear stress value location | Trailing edge | Trailing edge | Trailing edge | Volute tongue | Volute tongue | Volute tongue |

The wall shear stress near casing surface is also shown in figure 10. It is noted that near the blade leading edge along suction side and near the impeller exit, the wall shear stress is bigger than other areas. The wall shear stress near the blade leading edge increases with the width near the casing centre.

Thus, the suitable geometry for the blood pump is Model 2 with Casing 0.

![Wall shear stress near casing surface](image)

**Figure 10.** Wall shear stress near casing surface.

5. Conclusion
In this paper, the internal flows in a new-type centrifugal blood pump with hydrodynamic bearings are treated to investigate the effects of different pump flow passage structures on the pump performance. Based on the numerical results, the pumps can satisfy the operation parameters and be free of hemolysis.

For the pump with a column-type supporter at its inlet, the pump head and hydraulic efficiency decrease compared to the pump with a step-type support structure. The performance drop is caused by the disturbed flow upstream impeller inlet. Further, the unfavorable flow features such as reverse flow and low velocity in the pump will increases the possibility of thrombus.

The casing shape can little influence pump performance.
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Nomenclature

| Symbol | Definition |
|--------|------------|
| $b_2$  | Blade width at impeller exit [m] |
| $D_2$  | Impeller exit diameter [m] |
| $H_d$  | Pump head at design point [m] |
| $n$    | Rotational speed of pump shaft [min$^{-1}$] |
| $n_s$  | Specific speed, $=3.65n \cdot Q^{0.5} \cdot H^{-0.75}$ |
| $Q_d$  | Flow discharge at design point [m$^3$s$^{-1}$] |
| $\beta_1$ | Blade inlet angle [$^\circ$] |
| $\beta_2$ | Blade angle at impeller exit [$^\circ$] |
| $\eta$ | Pump hydraulic efficiency |

References

[1] Anna L M, Doris M, Christoph B and Axel H and Martin S 2013 European journal of cardio-thoracic surgery 43 1233-6
[2] Malchesky P S 2006 J Artif Organs 30(3) 129-129
[3] Akamatsu T 2001 Turbomachinery 29 7-16
[4] Luo X W, Zhu L, Zhuang B T, Xu H Y, Yu W P and Nishi M 2010 Sci China Tech Sci 53 105-110
[5] Luo X W, Ji B, Zhuang B T, Zhu L, Lu L and Xu H Y 2012 Sci China Tech Sci 55 795-801
[6] Siddique A, Wrightson N, Macgowan GA and Schueler S 2013 Annalis of Thoracic Surgery 95 1508-1508
[7] Giersiepen M, Wurzinger L J, Opitz R, Reul H 1990 Int J Artif Organs 13 300-6