Numerical analysis of the internal flow field in screw centrifugal blood pump based on CFD

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Abstract. As to the impeller blood pump, the high speed of the impeller, the local high shear force of the flow field and the flow dead region are the main reasons for blood damage. The screw centrifugal pump can effectively alleviate the problems of the high speed and the high shear stress for the impeller. The softness and non-destructiveness during the transfer process can effectively reduce the extent of the damage. By using CFD software, the characteristics of internal flow are analyzed in the screw centrifugal pump by exploring the distribution rules of the velocity, pressure and shear deformation rate of the blood when it flows through the impeller and the destructive effects of spiral blades on blood. The results show that: the design of magnetic levitation solves the sealing problems; the design of regurgitation holes solves the problem of the flow dead zone; the magnetic levitated microcirculation screw centrifugal pump can effectively avoid the vortex, turbulence and high shear forces generated while the blood is flowing through the pump. Since the distribution rules in the velocity field, pressure field and shear deformation rate of the blood in the blood pump are comparatively uniform and the gradient change is comparatively small, the blood damage is effectively reduced.

1. Introduction
Heart disease has become a major hazard to human health. As a signal of the worsening of heart disease, the failure of cardiac pump function directly threatens the patient's life. According to statistics, the number of heart failure patients increases alarmingly by 50 million per year worldwide and 10 million of these patients die every year. People eagerly want to create a kind of device which can help the heart to pump blood and thus save more lives. Therefore the study for artificial heart and mechanical heart-assisting device becomes the focus of researches in both the field of medicine and engineering [1, 2]. Some foreign research in situations began to develop the blood pump since the 1950s. United States, Britain, Germany, France, Japan, Australia and some other countries already achieved some positive results in the research and development of blood pump [3].

According to its working principle, blood pump can be divided into volumetric blood pump and impeller pump. Volumetric blood pump consists of the total artificial heart and circulatory assisting device developed in early stage. Its main function is to simulate the pulsating of natural heart to achieve blood supply. But it is bulk as well as difficult to implant, and prone to bring physiological exclusion. Impeller blood pump belongs to third-generation blood pump and is well known by its simple structure,
small size and high efficiency. It is very likely to become long-term implantable heart assisting devices. Impeller blood pump can be divided in three types: centrifugal, axial and mixed flow type. These three kinds of impeller blood pump gain great leaps in respect of size, weight, energy and control. However, there are also many problems including, most commonly, blood loss, seals and so on [4].

There are two blood loss situations: thrombosis and hemolysis. Thrombosis represents the phenomenon that activated platelet aggregates and precipitates on the blood pump contact surface with blood. Thrombosis affects the circulation of blood and appears mostly in the dead zone of blood flow and sealing places. Hemolysis represents the situation that red blood cells are destructed and then destructed hemoglobin dissociates in plasma, thereby reducing the quality of the blood and affecting the blood supply [5]. From studies carried out by domestic scholars, it is clear that the shear failure of blood is one of the main causes of hemolysis. When the shear force is greater than 200Pa, the hemolysis starts [6] and when the vertical impact speed reaches 6m/s or more, the red blood cells may rupture, causing hemolysis [7].

In order to reduce blood loss, the study of flow field inside the pump is needed. Nowadays, the main research method is taking experiments the costs of which are extremely high. However, the arising of Computational Fluid Dynamics (CFD) alleviated this thorny question. In this paper, by taking advantage of CFD software, the numerical analysis of inside flow field of microcirculation spiral magnetic levitation centrifugal blood pump is analyzed and the studies of the distribution of the speed, pressure, shear deformation rate and other parameters together with the spiral blades’ damaging effects on the blood are carried out. Therefore the capacities of the microcirculation spiral magnetic levitation centrifugal blood pump to not only effectively solve problems of dead zone of blood flow and sealing, but also alleviate the problem of high shear stress caused by high rotating speed of impellers, and thus reduce the extent of the damage to the blood.

2. Model Establishing and Grid Dividing

2.1. The establishment of the screw centrifugal blood pump channel solid model
Using a design theory design method distribution on the impeller, extrusion chamber, the suction chamber hydraulic design, then according to the obtained parameters of the hydraulic design to establish the impeller, extrusion chamber, the suction chamber channel entity model respectively. Assembly, the solid model of screw centrifugal blood pump channel is shown in Figure 1.

![Figure 1. Three-dimensional solid model](image-url)

2.2. The grid dividing of the screw centrifugal blood pump channel solid model
Importing the solid model into the ICEM software, using unstructured tetrahedral grid to achieve grid dividing, to optimize grid number, in calculation, 1481363 computational units and 312463 compute nodes are divided. Grid model is shown in Figure 2.
3. The Control Equations and Boundary Conditions

3.1. The control equations

The continuity equation:
\[
\frac{\partial \rho}{\partial t} + \nabla \cdot \rho \mathbf{u} = 0
\]  
(1)

The momentum conservation equation:
\[
\frac{\partial}{\partial x_i} (\rho \mathbf{u}_i) = f_i - \frac{1}{\rho} \frac{\partial p}{\partial x_i} + \nu \frac{\partial^2 \mathbf{u}_i}{\partial x_j \partial x_j}
\]  
(2)

The energy conservation equation:
\[
\frac{\partial (\rho T)}{\partial t} + \text{div}(\rho \mathbf{u} T) = \text{div} \left( \frac{k}{c_p} \text{grad} T \right) + S_T
\]  
(3)

Where \( f_i \) is the mass force inside the impeller flow region, \( \nu \) is a viscous stress factor, \( \rho \) is the density of blood, the \( k \) is the heat transfer coefficient, \( \rho \) is the temperature of blood and \( c_p \) is the constant-specific-heat.

3.2. Numerical simulation method

The method of Multiple Reference Frame (MRF) is selected for the computation, and the impeller region is in the rotating reference frame. Use standard wall function near the field of solid wall. Establish the Reynolds-averaging continuity equation, Navier-Stokes averaging equation and the other control equations, use standard k–\( \varepsilon \) turbulence model to close equations. To save compute time under a high accuracy condition, use first order upstream scheme on convection term and centre difference scheme on dissipation term, set the convergence precision as \( 10^{-5} \), and introduce the SIMPLEC method to iterate the pressure-velocity.

3.3. The boundary conditions

The internal flow medium of the screw centrifugal blood pump is blood, assuming blood is incompressible Newtonian fluid, density is \( 1.055 \times 10^3 \text{ kg/m}^3 \), viscosity is \( 3.5 \times 10^{-3} \text{ Pas} \), blood flow is \( 0.3 \text{ m}^3/\text{h} \), the import and export of pressure about the design conditions are 100mmHg column (13332Pa), and the impeller speed is 2980rpm.

1) The entrance boundary condition. The entrance is velocity-inlet, the speed of size is 0.44m/s.
2) The export boundary. The export is pressure-outlet, the pressure value is 110mmHg column (14665.22Pa).
3) The solid wall boundary. Defining the wall formed by the screw centrifugal blood pump impeller as rotating boundary, its speed is equal to the speed of rotation of the impeller is 2980rpm, and other wall is defined as the no-slip solid wall boundaries.
4. Simulation Results and Analysis

4.1 The internal velocity distribution of the screw centrifugal blood pump

Figure 3 is the internal velocity distribution of the screw centrifugal blood pump, from the velocity distribution figure, we can observe the whole flow trend within the internal pump is good, and no obvious flow separation; the velocity of the fluid from the entrance to the export increases gradually, the maximum velocity is at the export of the impeller while the minimum velocity is at the entrance; the velocity flow field is well-distributed, the mechanical damage to the blood is small, reflecting the gentle conveying characteristics of the screw centrifugal blood pump.

![Figure 3. Internal velocity field of blood pump (m/s)](image)

4.2 The internal pressure distribution of the screw centrifugal blood pump

Figure 4 is the internal pressure, distribution of the screw centrifugal blood pump, from the pressure distribution figure, pressure along the flow channel increases gradually, the minimum pressure is at the flow of entrance while the maximum pressure is at the flow of export. From the figure of the internal pressure field of blade (Figure 5), we can observe that pressure along the direction of blade rotation is gradually increasing, uniform increase, which is conductive to the safety of the blood.

![Figure 4. Internal pressure field of blood pump (Pa)](image)

![Figure 5. Internal pressure field of blade (Pa)](image)

The internal shear stress distribution of the screw centrifugal blood pump Niimi H[8] studies have shown that: the reasons of hemolysis are not only related to the size of the shear stress $t$, but also related to the shear stress exposure time $t$. There exists a critical shear stress value of about 1000Pa, when the shear stress exceeds this value, even the exposure time is very short, the red blood cells will be damaged; when the shear stress is in the range of 150–1000 Pa, if the exposure time is too long, the red blood cells also can be damaged. Therefore, by analyzing the internal shear stress distribution of the screw
centrifugal blood pump, we can acquire an intuitive understanding of the pump hemolysis. Figure 6 reflects the internal shear stress distribution of the screw centrifugal blood pump, we can directly read out the range of the shear stress of the pump. Most region is less than 150 Pa, which shows that the screw centrifugal blood pump in pumping simultaneously can effectively protect the red blood cells, and the loss of the blood is relatively small.

![Image](image)

**Figure 6.** Distribution of internal shear stress in blood pump (Pa).y.

**Figure 7.** Relationship between shear stress and red blood cell exposure time

5. Conclusions

1) The screw centrifugal blood pump’ rotational speed is low, the speed is well-distributed, and flow gently, which can effectively decrease the blood cells hemolysis phenomenon due to high-speed impact breakage triggered.

2) The screw centrifugal blood pump internal pressure distribution is evenly distributed, pressurized stability, which can satisfy peoples’ physiological requirements.

3) The screw centrifugal blood pump internal shear stress’ area which is higher than the critical value (150Pa) is minimum, and the damage to the blood is relatively small.

4) The opening of the backflow hole can lead the back impeller generate the microcirculation, avoiding flow dead zone, thereby reducing the thrombotic probability.

5) The maglev designs can solve sealing problems, thereby improving the blood pump on blood contamination.

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