Internal Flow Analysis of a Shaftless Double-Suction Blood Pump (Postprint)

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Date: 2017-11-07T00:00:00+00:00

Abstract

A novel shaftless double-suction blood pump has been designed based on the concept of a double-suction impeller with impeller rotor. To investigate the hydraulic and hemolytic performance of the blood pump, three-dimensional unsteady numerical simulations were performed under design operating conditions. Additionally, considering pulsatile flow conditions for human cardiac applications, the velocity, pressure, and wall shear stress distributions within the internal flow field under pulsatile flow conditions were computed and analyzed using these as boundary conditions, along with their effects on pump operational stability. The numerical results indicate that under pulsatile flow conditions, the pump achieves fluctuating output of flow rate and head, with comparatively poorer pressure stability relative to non-pulsatile flow. These results provide a valuable reference for the application of blood pumps in modern mechanical assist systems such as artificial hearts and artificial dialysis.

Full Text

Internal Flow Analysis of a Shaft-less Double-suction Blood Pump

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Abstract: Based on the concept of a double-suction impeller combined with a motor rotor, a novel shaft-less blood pump has been designed. To investigate the hydraulic performance and hemocompatibility of the pump, unsteady numerical simulations of three-dimensional flows in the entire pump passage were conducted near the design condition. The pulsatile flow characteristics of a natural artery were applied as boundary conditions at the pump inlet to observe

flow features such as velocity, pressure, and wall shear stress distributions in the flow passage, as well as their effects on pump operating stability. Based on the numerical simulation results, the internal flow characteristics are further discussed.

Key words: blood pump; double-suction; unsteady; numerical simulation

1. Structural Design of the Blood Pump

The overall structure of the blood pump consists of two main components: the drive unit and the flow passage components, as shown in [Figure 1: see original paper]. The drive unit comprises the motor and controller, which include armature windings, silicon steel sheets, and permanent magnets. The flow passage components consist of the impeller, pump casing, and outlet pipe.

A liquid film gap is designed between the impeller rotor and the volute. When fluid flows through the pump, this gap functions as a hydrodynamic bearing that supports and lubricates the impeller rotor. After passing through the impeller, the fluid pressure at the impeller outlet becomes higher than at the inlet. Under this pressure differential, liquid flows back through the gap from the impeller outlet to the inlet, flushing the clearance passage and preventing the formation of flow stagnation zones. This design effectively prevents blood coagulation caused by high temperatures and prolonged blood stagnation, while the hydrodynamic suspension support avoids hemolysis associated with mechanical bearing-induced red blood cell damage.

2. Computational Model and Methods

2.1 Computational Model and Mesh Generation

The full flow path model comprises four components: the inlet pipe, impeller, volute, and outlet pipe. The impeller region and rotating suction section (middle portion of the inlet pipe) are treated within a rotating coordinate system, while the inlet pipe, volute, and outlet pipe regions are handled as stationary coordinate systems. The SST k-ω turbulence model was selected for the numerical calculations.

Considering both numerical accuracy and computational economy, a structured mesh with 860,000 nodes was adopted. The mesh generation results are shown in [Figure 2: see original paper]. To more accurately calculate the pressure and shear stress distributions on the blade surfaces, local mesh refinement was applied near the walls, with a minimum grid spacing of 30 m at the wall-adjacent region.

2.2 Pulsatile Flow Conditions and Simulation Method

Based on the physiological pulsatile flow variation curve of the human heart, the cardiac pulsation period was determined. The pump inlet was prescribed with a time-varying flow rate boundary condition. [Figure 3: see original paper] illustrates the mass flow rate variation at the pump inlet (single side) over two periods. To observe the influence of blade rotation on internal flow patterns, one period was divided into five time instances corresponding to different blade positions, with the black outline representing the same blade. The moments t = 0 and t = T correspond to the blade' s initial position and its position after one complete rotation, respectively. For convenience in analyzing the internal flow within the impeller and volute, the impeller blades and volute cross-sections are numbered as shown in [Figure 4: see original paper].

The mass flow rate variation curve over one pulsation period can be expressed by the following function: (n=1,2···) (1). The parameter values are listed in .

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3. Numerical Results and Analysis

3.1 Unsteady Flow Simulation Under Non-Pulsatile Conditions

Under the design rotational speed of 5000 r/min, unsteady numerical simulations were performed for the design flow condition (Q = Qd) to analyze the velocity, pressure, and wall shear stress distributions within the impeller and volute. The following dimensionless static pressure coefficient Cp and total pressure coefficient Ctp are defined:

$$C_p = \frac{p - p_0}{\frac{1}{2}\rho u_2^2}, \quad C_{tp} = \frac{p_t - p_0}{\frac{1}{2}\rho u_2^2}$$

where p is static pressure (Pa), pt is total pressure (Pa), is fluid density, and p0 is reference pressure (taken as the average pressure at the impeller inlet).

[Figure 5: see original paper] presents the relative velocity distribution and streamlines at the mid-span plane of the impeller under design flow conditions. As flow rate increases, the relative velocity within the impeller rises, reaching a maximum value of 6.7 m/s near the blade pressure surface outlet along the impeller rotation direction, close to the volute tongue. The relative velocity distribution varies at different time instances as the blade position relative to the tongue changes, with relative streamlines deflecting in local flow passages.

The absolute velocity distribution and streamlines in the volute under design flow conditions are shown in [Figure 6: see original paper]. The absolute velocity distribution in the volute is relatively uniform, with a maximum absolute velocity of 3.7 m/s and no significant high-velocity regions. This primarily occurs because the absolute velocity at the blade outlet is similar to that at the volute inlet under design conditions, preventing large velocity gradients and associated flow impingement. Due to the stationary tongue effect, low-velocity regions appear in the volute passage between sections 0 and II, while relatively high-velocity zones occur near the volute outlet.

The static pressure distribution in the impeller and volute at design flow conditions is illustrated in [Figure 7: see original paper]. The static pressure varies at different time instances. Overall, the pressure distribution in the volute and outlet section is relatively uniform, though localized high-pressure regions appear due to the interaction between rotating blades and the stationary tongue. Compared with low-flow conditions, the static pressure at the outlet section is relatively low at only 1.1 kPa.

3.2 Pulsatile Flow Simulation Results

3.2.1 Energy Performance Characteristics The pump head variation over two pulsation periods is shown in [Figure 8: see original paper]. As the inlet flow rate changes periodically, the pump head also pulsates accordingly, achieving pulsatile flow output. The overall head variation trend is opposite to that of the mass flow rate, with lower head occurring at higher flow moments. The head ranges between 1.37–2.58 m.

3.2.2 Internal Flow Field Characteristics The mass flow rate variation curve over one pulsation period and three typical monitoring points are shown in [Figure 9: see original paper]. Point t1 = 0.1094 s corresponds to the peak of the pulsation cycle, where the flow rate approaches the design condition. Point t2 = 0.19 s represents the average mass flow moment, while point t3 = 0.2674 s corresponds to the trough moment when the mass flow rate approaches zero.

The relative velocity distributions in the impeller region at three different moments (t = t1, t2, t3) are presented in [Figure 10: see original paper]. At moment t1, when the inlet mass flow rate is near the design condition, the relative velocity distribution within the impeller passages is relatively uniform with smooth streamlines. At moment t2, low-velocity zones appear in the passages with evident turbulent phenomena, chaotic streamline distribution, and vortices in some passages. At moment t3, large-area low-velocity zones appear in the passages, with significant vortices in each passage that severely block the flow path and inevitably cause substantial losses.

[Figure 11: see original paper] shows the absolute velocity distribution in the volute at three different moments under pulsatile flow conditions. At moment t1 (near design flow), the absolute velocity distribution in the volute is uniform with smooth, stable flow and minimal flow losses. At moment t2, the velocity distribution near the tongue is non-uniform, with locally high velocities at the volute inlet and low-velocity zones at the outlet section where streamlines deflect. At moment t3, the velocity distribution in the volute is highly non-uniform, with large low-velocity zones at the outlet and significantly deflected streamlines containing large vortices.

The static pressure distributions in the impeller and volute at three different flow moments (t = t1, t2, t3) are shown in [Figure 12: see original paper]. The results indicate substantial differences in static pressure distribution at different moments. At t1, static pressure in the impeller gradually increases from inlet to outlet, with uniform pressure distribution in the volute. At t2, the pressure distribution at the impeller outlet is non-uniform, with localized low-pressure zones near the volute tongue. At t3, the pressure distribution in both the impeller and volute is highly non-uniform, which severely affects pump performance.

3.2.3 Wall Shear Stress Analysis The wall shear stress distribution on impeller and volute surfaces at three different moments is presented in [Figure 13: see original paper]. The calculated maximum wall shear stresses in the volute at moments t1, t2, and t3 are 278 Pa, 377 Pa, and 360 Pa, respectively. Similar to the non-pulsatile condition, the maximum wall shear stress occurs at the tongue and at blade inlet and outlet locations. The maximum wall shear stress in the volute is higher than that in the impeller, while the average wall shear stress shows the opposite trend. The average wall shear stress follows the same variation trend as the mass flow rate, being larger at high-flow moments. Although the maximum shear stress in the volute is relatively high, the average wall shear stress remains low, indicating that the overall wall shear stress is small. During volute optimization, reducing the localized maximum wall shear stress should be considered.

[Figure 14: see original paper] and [Figure 15: see original paper] illustrate the temporal variation of maximum wall shear stress τmax and average wall shear stress τavg on impeller, shroud, and volute surfaces over one pulsation period under pulsatile flow conditions.

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4. Conclusions

1) This study applies human cardiac pulsatile flow conditions to a blood pump to analyze the effects of pulsatile flow on the operational stability of a shaft-less double-suction blood pump, providing valuable references for applications in modern mechanical assist systems such as artificial hearts and dialysis.

2) Under pulsatile flow conditions, the pump achieves fluctuating output of flow rate and head. Within one pulsation period, the pressure variation trend at each monitoring point correlates with the pumped mass flow rate, with pressure fluctuations in the impeller being more significant than those in the volute.

3) Over one pulsation period, the maximum wall shear stress in the volute

exceeds that in the impeller, but the average wall shear stress is lower. During volute optimization, reducing localized maximum wall shear stress values should be considered.

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