NUMERICAL STUDY FOR FLOW IN PHOENIX LEFT VENTRICULAR ASSIST DEVICE AT PEAK SYSTOLE

CHEUNG-HWA HSU

Department of Mechanical Engineering, Southern Taiwan University of Technology, Taiwan

ABSTRACT

The Phoenix left ventricular assist device is driven by pneumatic pump outside a human body to provide air flow in-and-out of a pneumatic diaphragm inside the blood chamber. The inflation and deflation of the diaphragm drives the blood flows. At peak systole, blood flows into aorta with a maximum flow rate both from natural ventricle and artificial ventricle. To assess the critical condition, blood flow pattern of the device at peak systole is studied. We apply the finite volume method with $\kappa - \epsilon$ turbulent model to simulate the steady flow at peak systole. Results provide a comprehension of fluid dynamic effect not only the region inside the ventricle but also that in the vicinity of the inflow channel and outflow valves of flow field. The numerical results show that maximum velocities were smaller within the ventricular chamber with bi-leaflet outflow valve than with tilting disk outflow valve. Analysis of the numerical solutions suggests that geometry similar to the bi-leaflet outflow valve results in a better flow dynamics within the ventricle chamber compared to a tilting disk valve.

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Keywords: left ventricular assist device, peak systole, bi-leaflet valve, tilting disk valve

1. INTRODUCTION

During the last decade, a variety of left ventricular assist devices (LVADs) have been developed and demonstrated clinical efficacy in salvaging patients with heart failure following either open heart surgery or acute myocardial infarction [22]. Thrombosis remains to be one of the major obstacles in long-term implantation of the LVADs. Because the blood flow cannot completely wash out the inner walls of the ventricle, stagnation of blood flow results in thrombosis. Thrombosis is also caused by high turbulence shear stresses near the valves that damage

red blood cell and platelets. Under the influence of turbulence shear stresses, platelets will adhere to and aggregate on the device [6]. As indicated by Lamson's studies [7], the shear stress, if controlled appropriately, can both reduce the amount of platelet aggregation and wash the inner walls of the ventricle. However, the excessive increase in shear stress will deform the platelets and increase their adhesion, thus activating the mechanism of the platelets resulting in thrombosis.

Thrombosis formation is a very complicated process. Smith et al. [15] and Stein and Sabbah [16] pointed out in their experiments that thrombosis formation is related to turbulence. Goldsmith and Karino [3] indicated the separation of flow and presence of stagnation regions cause platelets to aggregate. According to Hellums and Brown [4], hemolysis occurs when Reynolds stresses reach 1500 dyn/cm² for an exposure time of 0.01 sec. According to Tillman et al. [20], the upper limit of hemolysis is 2000-4000 dyn/cm² for an exposure time of 1-10 ms. Sutura et al. [17] pointed out that hemolysis occurs at stresses of 1000-2000 dyn/cm² for an exposure time of 10⁻³ sec. However, the limit for platelet damage drops

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Correspondence: Cheung-Hwa Hsu, Professor
Department of Mechanical Engineering,
Southern Taiwan University of Technology,
1 Nan-Tai Street, Yung-Kang City,
Tainan County, Taiwan
E-mail: chhsu@mail.stut.edu.tw

to 100-500 dyn/cm² [6]. Once platelets are damaged, they release adenosine diphosphate, which causes thrombosis.

Numerous pathological findings [1] have shown strong correlation of retarded flow with the deposition of blood elements. In the flow field where elevated shear followed by separated flow, platelets are activated by shear exposure and subsequently introduced to a low shear environment where they are free to aggregate or attach to absorbed protein [19], so as to form thrombosis.

The phoenix LVAD consists of a ventricle with an outflow valve. The ventricle is orthotopically positioned and attached to the patient's ventricle and aorta with atrial sewing cuffs and vascular graft material. The device is driven by compressed air by a console that adjusts pumping rate, driveline inflation pressures and percent systole. A flexing diaphragm is found in ventricle to pump the blood from the ventricle when the diaphragms are inflated. In order to obtain detailed information of flows, visualization technique and laser Doppler anemometer were used to observe the flow field over the past decades. However, due to a practical limitation of measurements within LVADs especially in the vicinity of the outflow valves, data obtained from the experimental measurements were limited.

Computational fluid dynamics provides an alternative technique in which more detailed analysis of fluid dynamic effects of flow within a LVAD can be evaluated. Extensive numerical analysis on natural heart flow dynamics has been reported by Peskin [12]. They initially solved the Navier-Stokes equations for a two dimensional model of the left ventricle and mitral valve by a finite difference scheme. The heart was considered to be submerged in an infinite pool of blood, with the mitral valve and ventricular walls moving at the local fluid velocity and acting as regions in space where extra forces are applied to the fluid. They reported atrial and ventricular pressure and flow patterns which showed reasonable agreement with the experimental data for the case of the natural mitral valve. They extended the studies to perform three dimensional numerical calculations of blood flow in the human heart. Rogers et al. [14] presented numerical solution of incompressible Navier-Stokes equations for steady-state and time-dependent flow through a model of a Penn state TAH. The solution was obtained for a Reynolds number of 100 based on unit length and velocity. In this paper, we present a numerical method to evaluate Phoenix LVAD's performance. We use the finite volume method to solve the fully incompressible three-dimensional Navier-Stokes equations plus $k-\varepsilon$ turbulent model. The analysis provides a comprehension for the physical phenomena not only for the region inside the ventricle but also the region in

the vicinity of the outflow valve.

2. METHOD

The Phoenix LVAD is driven by air pump outside a human body to provide air to flow in-and-out of a pneumatic diaphragm inside the blood chamber. At systolic phase the diaphragm inflates and drives blood flows. The LVAD drains blood from the left ventricular apex via inlet cannulae (with no inlet valve) and ejects into the aortic root via an outlet conduit along with an outlet valve to avoid reverse flow at diastolic phase. Blood flows into aorta with a maximum flow rate at peak systole both from natural ventricle and artificial ventricle.

The governing equations used for flow analysis at peak systole are Reynolds-averaged equations of Cartesian rectangular coordinate. The tensor forms of the Reynolds-averaged continuity and momentum are given in the following.

$$\frac{\partial}{\partial x_j}(\rho u_j) = 0 \quad (1)$$

$$\frac{\partial}{\partial t}(\rho u_i) + \frac{\partial}{\partial x_j}(\rho u_j u_i) = -\frac{\partial p}{\partial x_i} + \frac{\partial}{\partial x_j} \left(\mu_{eff} \left(\frac{\partial u_i}{\partial x_j} \right) \right) \quad (2)$$

where u_i is velocity component in the x_i direction, p is static pressure, ρ is fluid density, μ_{eff} is viscosity ($\mu_{eff} = \mu + \mu_t$, μ is the laminar viscosity, μ_t is the turbulent viscosity). The $k-\varepsilon$ turbulent model along with wall function is adopted in the present study to solve important parameters including Reynolds stress, turbulent energy flux, and fluctuating viscous work.

Taking the volume integral on Eqs. (1), (2) and transforming equations into the surface integral forms using Gauss's theorem, the governing equations are transformed to

$$\int_s \rho u_j dn_j = 0 \quad (3)$$

$$\begin{aligned} \frac{\partial}{\partial t} \left(\int_v \rho u_i dv \right) + \int_s \rho u_j u_i dn_j = \\ - \int_s p dn_i + \int_s \mu_{eff} \left(\frac{\partial u_i}{\partial x_j} + \frac{\partial u_j}{\partial x_i} \right) dn_j \end{aligned} \quad (4)$$

Where subscriptions v and s represent the volume and surface integrations respectively, n_j is the outward normal surface vector. Eqs. (3), (4) are then discretized in the following.

$$\sum_{ip} (u_j \Delta n_j) = 0 \quad (5)$$

$$\rho \Delta V \left(\frac{u_i - u_i^o}{\Delta t} \right) + \sum_{ip} (\rho u_j \Delta n_j)_{ip} (u_i) = - \sum_{ip} (p \Delta n_i) + \sum_{ip} \left(\mu_{eff} \left(\frac{\partial u_i}{\partial x_j} + \frac{\partial u_j}{\partial x_i} \right) \Delta n_j \right)_{ip} \quad (6)$$

where ΔV is the volume of the control volume, the subscript "ip" denotes an integration point, the summation is over all the integration points of the surface, Δn_j is the discrete outward surface vector, the superscript "o" means "at the previous time level", and the over-bar on the source terms indicate an average value for the control volume.

Critical state of peak systole is simulated and the model is simplified to be steady. The diaphragm inflates completely with the outflow valve open. The outflow of blood continues so long as pressure in the ventricle is higher than that in the aorta. For flow simulation during peak ventricular systole, a relatively long channel is chosen for the outflow channel where the flow is fully developed. The inflow at peak diastole is proposed to be uniform. No slip and no-flux boundary conditions are imposed on all solid surfaces.

3. RESULTS AND DISCUSSION

In this initial attempt to design the turbulent flow within the LVAD chamber, a three-dimensional geometry of the blood chamber was established. A tilting disk valve was simulated in the outflow orifice of Model I while a bi-leaflet valve was simulated in the outflow orifice of Model II. Fig.1 shows the flow pattern of the flow chamber at peak systole. The analysis is conducted of outflow rate of 3 l/min with $Re=600$ based on the diameter of the inflow channel and the mean velocity in the inlet. The diaphragm expands during the systolic phase. At peak systole, the diaphragm is absolutely stationary as the diaphragm volume has reached a maximum. The movement of the diaphragm provides a dynamic source for the blood to flow through the outlet and into the aorta. Blood inflow is mainly pushed through the lower rigid wall to the outlet of the ventricle, where the flow has high velocity. Maximum velocity reaches 38.5 cm/s near the pivoting mechanism of sewing ring. There is greater velocity in the lower cavity compared to the upper region, especially for both upstream and downstream the outflow valve.

Cases with various outflow rates of $Re=250$, 600,

2100 and 5600 are presented in Fig. 2~5. They demonstrate a similar flow pattern but different velocity scales. As shown in Fig.2 (a), a high velocity with a maximum magnitude of 5.5 cm/s is observed distal to the inflow. After the flow passes through the inflow orifice it rapidly decelerates toward the diaphragm wall. Majority of the blood flows though

Fig 1. Streak lines at $Re=600$ for the model with tilting disk valve

Fig 2. Velocity distribution at $Re=250$ (a) for whole flow field (b) for the vicinity of outflow valve

Fig 3. Velocity distribution at Re=600 (a) for whole flow field (b) for the vicinity of outflow valve

the lower cavity of the chamber and toward the outlet orifice. As the blood flows through the outlet valve, two jet-like flows emanate as shown in Fig.2 (b). One is induced on the right-side orifice and the other on the left-side. Flow rate is higher through the right-side orifice with a maximum velocity of outflow jet of 5.3 cm/s. Maximum velocity of the flow domain is discovered somewhere near the S-type pivoting mechanism. The mechanism provides a gap between the leaflet edge and the sewing ring. Velocity increases extremely on the gap. It reaches 12.8 cm/s when Re=250 (Fig.2), 38.5 cm/s when Re=600 (Fig.3), 115.5 cm/s when Re=2100 (Fig.4), 308 cm/s when Re=5600 (Fig.5).

Flow pattern is rather different, especially for the outflow orifice, compared to Mode I when the outlet valve is replaced by a bi-leaflet valve of Mode II. Fig.6 shows the flow pattern of the flow chamber at peak systole when outflow rate equals 3 l/min with Re=600. The movement of the diaphragm pushes the blood to flow through the outlet and into the aorta. A velocity jet with a maximum magnitude of 13.8 cm/s is observed distal to the inflow channel. After the flow passes through the inflow channel it flows toward the

Fig 4. Velocity distribution at Re=2100 (a) for whole flow field (b) for the vicinity of outflow valve

lower cavity of the diaphragm. Maximum velocity of the flow domain takes place somewhere downstream the inflow channel with a value of 27.4 cm/s. Majority of the flow is pushed through the lower cavity and toward the outflow channel of the ventricle. Downstream the leading edge of the outlet valve, the maximum blood velocity reaches 19.5 cm/s. Cases with various outflow rates of Re=250, 600, 2100 and 5600 are displayed in Fig. 7~10. They display a similar flow pattern but different velocity magnitudes. As shown in Fig.7 (a), a velocity with a specific magnitude of 5.5 cm/s downstream the inflow. After the flow passes through the inflow orifice, majority of the blood flows through the lower region of the chamber and toward the outlet orifice. There exists a maximum flow velocity somewhere downstream the inflow. As the blood flows through the outlet orifice, three jet-like flows emanate as shown in Fig.7 (b). They are on the right-side orifice, the left-side orifice and the central orifice. Flow velocity through the central orifice is much higher than those through the side orifices as a rather narrow flow passage in the region and thus inducing a higher speed. The maximum velocity of outflow in the central orifice is

(a)

(b)

Fig 5. Velocity distribution at $Re=5600$ (a) for whole flow field (b) for the vicinity of outflow valve

Fig 6. Streak lines at $Re=600$ for the model with outflow bi-leaflet valve

6.5 cm/s specifically. Maximum velocity of the flow domain is discovered somewhere downstream inflow. It reaches 9.15 cm/s when $Re=250$ (Fig.7), 27.4 cm/s when $Re=600$ (Fig.8), 82.4 cm/s when $Re=2100$ (Fig.9), 220 cm/s when $Re=5600$ (Fig.10).

Several conclusions can be drawn from the results

(a)

(b)

Fig 7. Velocity distribution at $Re=250$ (a) for whole flow field (b) for the vicinity of outflow bi-leaflet valve

of this simulation.

1. At the inflow channel, the egg-shaped diaphragm results in a flow dynamics with a reduction in regions of relative stasis. Flow interacts with the ventricular chamber and wall motion of the diaphragm and wash out of blood in this region is predicted. In the outflow channel, however, large velocity gradient takes place between the leaflets and the rigid wall. At peak systole, as the diaphragm of the chamber pushes the fluid towards the outflow orifice, the fluid in the region behind the leaflet is trapped and a smooth washout may not be achieved. It appears that, in the case of LVAD chamber, a bi-leaflet valve opening towards the outer wall will induce a smaller velocity field along with a smaller velocity gradient compared to that of tilting disk valve will induce a smoother washout of fluid in the outflow chamber.

2. Magnitudes of shear stress are predicted larger distal to the outflow valves compared to other regions within the chamber. On the other hand, the magnitudes of shear stress to the bi-leaflet valve at the outflow channel are expected smaller than that of the tilting disk valve in the same position.

3. The numerical solution restricts the solution for

Fig 8. Velocity distribution at $Re=600$ (a) for whole flow field (b) for the vicinity of outflow bi-leaflet valve

Fig 9. Velocity distribution at $Re=2100$ (a) for whole flow field (b) for the vicinity of outflow bi-leaflet valve

Fig 10. Velocity distribution at $Re=5600$ (a) for whole flow field (b) for the vicinity of outflow bi-leaflet valve

steady, the valves are assumed to be in the fully open position. In extending the solution to the unsteady flow case similar to the in vitro situation, the motion of the leaflets as well as the diaphragm and the dynamic

interaction between the fluid and solid interface need to be taken into account. The work is essential for analyzing flow performance of the device of complete heart beat cycle.

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