HEMODYNAMIC ASPECTS OF THE BERLIN VENTRICLE ASSIST DEVICE

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Abstract- **A New Ventricle Assist Device (VAD), with an improved energy converter unit, was investigated both numerically and experimentally. An experimental Continuous Digital Particle Imagining Velocimetry (CDPIV) was combined with a computational fluid dynamics (CFD) analysis. These tools complement each other to result into a comprehensive description of the complex 3D, viscous and time-dependent flow field inside the artificial heart ventricle. A 3D numerical model was constructed to simulate the VAD pump and a time-dependent CFD analysis with moving walls was performed to predict the flow field inside the VAD during the cardiac cycle. A commercial finite element package (FIDAP, Fluent Inc., Evanston) was used to solve the Navier-Stokes equations. In the experimental analysis, an optically clear elastic model of the VAD was placed inside a 2D CDPIV system. Continuous flow visualization and CDPIV calculations of the flow were used for validating the CFD simulations. Once validated, the CFD results provide a detailed 3D and time dependent description of the flow field, allowing the identification of stagnation or high shear stress regions.**

***Keywords* - Artificial hearts, PIV, CFD, cardiovascular flow.**

I. INTRODUCTION

Ventricular assist devices (VAD) are mechanical devices that assist or replace a failing heart ventricle (left or right) of end-stage patients. Currently there are a few clinically approved blood pumps for short-term use of bridge to transplantation (BTT) or bridge to recovery (BTR), but none of them is for chronic use or is fully implantable. Improvement of blood pump designs is essential for the extension of their application to permanent use.

Thrombosis is one of the primary causes of death in cardiac prosthesis patients and it could be avoided by improving the blood pump hemodynamics. Hemolysis and calcification could be reduced as well by proper design. High shear stresses, turbulence, flow separations and stagnant flow regions have to be minimized. Better understanding of the flow field inside the blood chamber and through the valves is crucial for better design of blood pumps and for long-term applications of these devices.

<u>*Experimental Flow investigation*</u>

Several experimental methods have been used to investigate the flow in blood pumps, usually in mock circulatory systems. Invasive methods such as hot film anemometry required the placement of probes in the fluid path, with the potential disturbance of flow at the probe site. Laser Doppler velocimetry has been used to measure the point-wise velocity (e.g. [1]). However, this method is limited when a comprehensive description is needed. Methods that rely on flow visualization using seeded particles in the flow overcome some of these limitations and can be utilized to evaluate flow patterns. It can reveal high shear locations and identify areas where fluid is stagnant. Therefore, in most studies dedicated to designing and improving AHs, flow visualization is a major tool for locating possible hemolisys and thrombus formation sites. Recently, the *particle image velocimetry* (PIV) technique is increasingly being used in the investigation of flow in blood pumps, since it can also provide quantitative data, such as flow velocity or stresses magnitudes [2]-[6].

In all these studies, visualization and PIV have been proven to be useful tools for describing the global features of the flow in terms of stagnation or separation regions and flow washout. However, even though they are very often used, flow visualization techniques have several limitations. The major limitation is locating extreme flow characteristics. The flow in blood pumps is complex and is characterized by relatively high velocities, especially near the valves. This results in a required frame rate of more than 4000 frame/sec ([4]; [7]; [8]). The frame rate exceeded 500 frames/sec only in very few studies [6]; [7]; [9], and therefore images taken from critical areas have often poor quality and the apparent flow velocity might be underestimated. One way to overcome this limitation is to use enlarged models (while keeping constant the Reynolds, Archimedes, Stiffness and Strouhal numbers), but it might be complex to manufacture the scaled-up components of the blood pump [10].

The second limitation of visualization techniques is the necessity of using transparent models. Reference [11] suggested the nuclear scintigraphy as an alternative method that can investigate flow in opaque models. However, the results are of very poor resolution and only global qualitative flow behavior could be obtained.

Another limitation of visualization methods is the description of 3-D flow. According to [2], the flow disturbances in the ventricles might be in planes other than the one containing the inflow and outflow valves. Although there are a few visualization techniques to measure the third dimension [12], the flow inside blood pumps is much too complex to employ these techniques.

Visualization techniques have also failed to estimate shear stresses on the walls. There are several other experimental techniques to measure the wall shear stress. For example, [9] combined PIV with paint erosion method to estimate wall

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Abstract

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shear stress and evaluate the function of valve orientation on the flow pattern and washout in an electrohydraulic TAH. However, this technique is useless for pulsating flow. Reference [13] suggested using visualization techniques on a fluid mixed with a dye and illuminated with diffused light, but this technique required very complex procedures and was not proved to be completely satisfactory.

In summary, flow visualization is a very powerful and useful technique for the investigation of blood pumps hemodynamics, but in order to evaluate the detailed 3D flow characteristics in the complex continuously changing geometry of the blood chamber, it has to be combined with an additional complimentary technique to obtain a comprehensive flow description.

Numerical Flow investigation

Computational fluid dynamics (CFD) simulations can provide a detailed description of the unsteady flow field (including in regions which are difficult to access experimentally) and may be extensively used for optimization or for isolating factors which in many cases cannot be separated experimentally. Moreover, after building and validating a suitable numerical model, changes in the numerical model for evaluation of the various parameters such as geometry or operating conditions can be easily preformed. However, very few numerical studies of the flow in artificial hearts have been done, probably due to the complexity of such analyses. The numerical simulation of the full problem is beyond the routine capabilities of existing computational resources, since the flow in the blood chamber is very complex. The flow is at relatively high Reynolds number, time-dependent, three-dimensional, and occurs within flexible moving walls and through passively moving valves.

Reference [4] simulated the flow in simplified 2-D models of pulsating blood pumps, using steady analysis. Both laminar and turbulent (using k-e model) flow conditions were considered. Models with tri-leaflet or flap polyurethane inflow valves were examined. Their results suggest that flap valve (or tilting disk valve) result in better flow dynamics compared to trileaflet valve.

A more complex analysis was done by [14]. They simulated the 3-D, time-dependent flow in the pulsating Penn State TAH pusher-plate chamber during the cardiac cycle. The simulations included the motion of the pusher-plate and the two tilting-disk valves. The motion of the valves was specified from experimental data. A zonal method (division to sub domains) and an overlapped-grid embedding scheme (for the large displacements) were employed. The results were compared qualitatively to flow visualization results and showed good agreement in the global flow patterns.

In constant flow blood pumps the numerical analysis is considerably simple because steady analysis is adequate, and the geometry is fixed. Reference [15] simulated the 3D flow in Nikkiso HPM-15 centrifugal pump. They preformed a steady flow analysis using k-ε turbulence model for three different pump models, and validated their results with visualization results.

Reference [16] performed numerical analyses to optimize their non-pulsating VAD geometry. The numerical model was validated using experiments, and the spacing and geometry of the clearance regions were optimized according to flow velocity and pressure in various conditions of blood flow rate and pump speed.

A few numerical studies of the flow inside the native heart were also performed. Reference [17] studied the effect of time-varying left ventricular ejection using 3D moving wall realistic model of the left ventricle. Reference [18] simulated the ventricular ejection mechanism with studying the fluid-structure interaction for the transient intravalvular pressure gradients in an axisymetric model of the ventricle. Reference [19] investigated the transient flow during skeletal muscle ventricle filling. Particle path lines from a 2-D moving wall numerical simulation were compared with corresponding flow-visualization pictures.

However, the most notable works on native heart are the fascinating simulations based on the immersed-boundary method developed by the group of Prof. Peskin, (e.g. [20], [21]). They simulated the flow in the natural heart, incorporating the elasticity of the wall fibers and the interaction of the tissue and the blood. However, these fully three-dimensional calculations required very large computational resources and the Cartesian meshes did not allow the resolution of the finer viscous details of the flow.

In summary, CFD can be used as a complementary technique to evaluate the detailed flow within blood pumps. However, to make the problem traceable, the numerical model has to be simplified.

II. METHODOLOGY

The present research employs both experimental and numerical methods to study the flow field inside an innovated VAD that is based on a new improved energy converter – the Berlin Left Heart Assist Device [10]. For this purpose, an experimental Continuous Digital Particle Imagining Velocimetry (CDPIV) was combined with a computational fluid dynamics (CFD) analysis. These tools complemented each other to result into a comprehensive description of the complex 3D, viscous and time-dependent flow field inside the artificial heart ventricle.

The Berlin VAD

The Berlin VAD uses a new energy converter that was developed in the laboratory of Prof. Klaus Affeld of Humboldt University, Berlin. It is still in its laboratory phase. This fully implantable heart assist device is a pulsating pump that uses only one moving part, which is the rotor of an electrical motor with an attached impeller of a centrifugal pump (see figure 1).

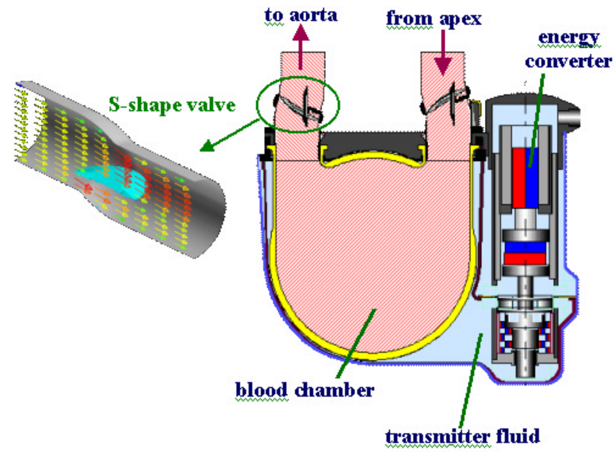

Figure 1 – Berlin VAD

It rotates and delivers the transmitter fluid that compresses a blood chamber. By an electromagnetically axial shift of a few millimeters, the rotor reverses the flow of the impeller to another channel and thus the blood chamber is expanded (figure 1). In this way, a pulsating action of the blood chamber is achieved, while a total separation between the moving elements and the blood is kept [22], [8]. The Berlin VAD includes Bjork-Shiley tilting disk valves placed in an S-shape centerline duct thus the leading edge of the occluder is in line with the incoming fluid to avoid flow separation.

The CDPIV Method

The experimental investigation method uses the *continuous digital particles image velocimetry* (CDPIV) method. DPIV is a powerful, non-intrusive full field flow measuring optical technique that combines flow visualization with quantitative measurements and provides the instantaneous 2-D velocity vector field of the flow over an extended plane. In this method, the fluid is seeded with particles and single exposure images of the particles illuminated by a pulsed laser light sheet are recorded using a CCD video camera. The displacements of the particles are obtained by locally cross-correlating sequential images, and the velocity vector fields are computed from the known timing of the laser pulses. The cross-correlation function of the two samples is calculated using FFT techniques. The displacement of the cross-correlation peak provides the average spatial shift of the particles in each subsample pair [23]. In the present study, the continuous laser light sheet (Coherent Innova 70, 5W, water-cooled CW argon Ion laser) was strobed using an electronic light shutter, triggered by a NTSC CCD video camera (Pulnix TM-9701, 29.97fps, 8Bit, grayscale, 480x768). A combination of mirrors and cylindrical lenses spread the laser beam to a 1 mm thick sheet and area that covers the entire cross section of the flow field in the chamber. To obtain the time-dependent flow, the DPIV system was developed to operate continuously in real-time. The CDPIV system is capable of sampling 30 images per second, resulting in 15 velocity vector plots per second. Detailed description of the system and method can be found in [24,].

The Experimental Setup

The blood chamber was placed inside a pressure chamber. Contraction of the blood chamber was achieved by applying external pressure that was produced by an attached piston pump.

The VAD chamber is a flexible sac composed of two symmetric Polyurethane parts and a transparent Perspex profile between them. These parts were assembled using stainless screws and hermetized by special sticking plasticine. The mitral and aortic Bjork-Shiley valves were placed in the S-shaped inlet and outlet ducts. The pressure chamber is a transparent box made of Perspex with glass windows on each side. The VAD chamber was connected to a mock flow loop that simulates cardio-vascular conditions. The loop contains a tube with a compliance chamber (that represents arterial compliance), flow regulator (that represents afterload conditions) and a reservoir (that represents preload conditions). Blood was simulated by a 40% water-glycerin mixture seeded with 10μ silver coated glass microspheres. The same fluid without particles was used as a transmitter fluid in the pressure chamber. An electromagnetic flowmeter (MAGFLO MAG 3000, Danfoss), located downstream the VAD aortic valve, measured instantaneous and average flow rate (Cardiac Output).

The pump generated quasi-physiological volume time curves as described in [25]. The VAD was investigated at physiological working conditions of 2 liters/min flow rate and 50 pulses/min.

The numerical method

The analysis of the flow inside the VAD was based on a 3D realistic model of the Berlin VAD blood chamber. A time-dependent CFD analysis was performed to predict the flow behavior in the changing geometry of the VAD during the cardiac cycle. A laminar, incompressible and Newtonian flow was assumed. A commercial finite element package (FIDAP, Fluent Inc., Evanston) was used to solve the Navier-Stokes equations. The mesh consisted of 44,000 hexahedral elements. The motion of the flexible walls was simulated by normal flow through rigid walls. This simplification was based on the assumption that the motion of the walls does not affect the basic flow. The normal flow waveform imposed on the walls corresponds to physiological conditions of 50 beats/min and CO of 2 liter/min. The motion of the valves was not modeled. They were held fixed in their fully open or fully closed positions, depending on the phase of the cardiac cycle. During systole, the aortic valve was opened and the mitral valve was closed and during diastole, the mitral valve was opened and the aortic valve was closed. An Euler backward implicit time-advancing algorithm was employed together with a projection method for solving the pressure field. A segregated iterative solver using the GMRES method was used for solving the discrete equations. Streamline upwinding was activated to smooth the solution at the higher *Re* number cases.

III. RESULTS and DISCUSSION

Visualization images of the particles in the illuminated plane were taken along the cardiac cycle. Velocity fields of the visualized flow were calculated using the DPIV algorithm to obtain the time-dependent velocity field. A time-dependent numerical analysis of the flow in the 3D model was also performed. Figure 2 shows the experimental and numerical results for three-time instants: at the beginning of the diastole, at the end of diastole and at peak systole.

The experimental results show a fast velocity blood jet across the mitral valve at the beginning of the diastole, while during the diastole a large rotating flow was formed in the entire blood chamber. At the systole, the large rotating flow was completely washed out through the aortic valve. No significant stagnant regions were found in the flow during the cardiac cycle.

The numerical results were in agreement with the experimental results. A strong jet flow through the mitral valve was found at the beginning of the diastole, and a large rotating flow was filling the blood chamber during diastole. At the systole, the large rotation was washed out completely through the aortic valve, and as with experiments, no stagnant regions were found in the flow.

The three-dimensional component of the flow during diastole and the shear stress on the wall during systole (figure 3) are examples of flow parameters that can be easily extracted from the numerical simulations, but cannot be obtained from the experimental methods employed in the present work.

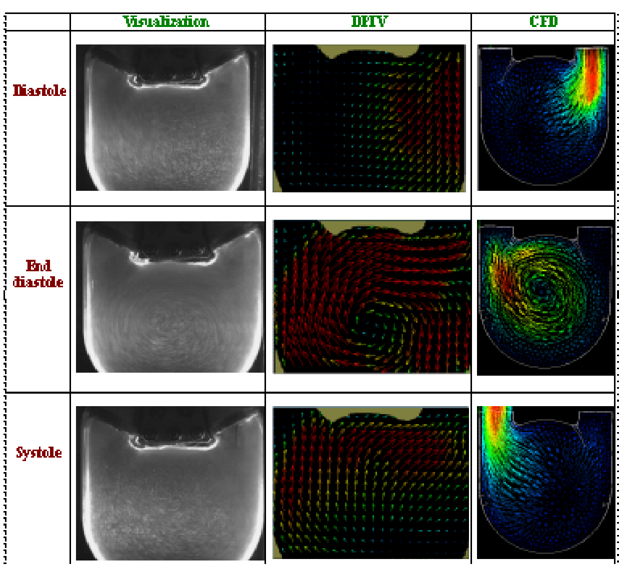

Figure 2 - visualization, PIV and CFD results

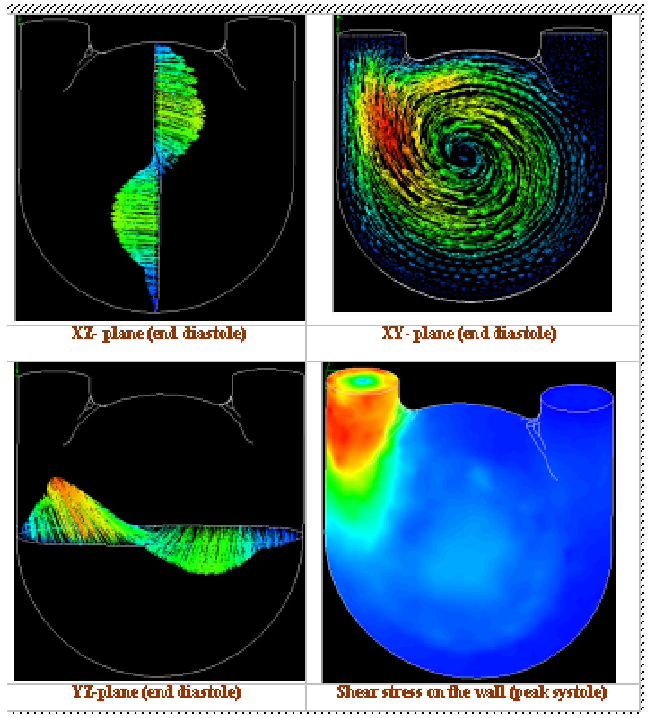

Figure 3 – 3D velocity at end diastole and shear stress on the wall at peak systole

IV. CONCLUSION

The study of the hemodynamics of the blood chamber is important in the design of VAD. To evaluate the 3D flow field in the VAD, visualization techniques have to be combined with an additional complimentary technique. In the present study, numerical techniques were combined with DPIV to obtain the flow across the blood chamber of the Berlin VAD. CFD is a powerful tool for fast investigation of the flow field, for prediction of clinical cases and optimization.

The numerical results were in good agreement with the experiments. A large rotating flow was found in the blood chamber during diastole. It was washed out during systole. Regions of low and high shear stresses could be identified in the CFD simulations. The present study results may assist in improving the design of the Berlin VADs.

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