COMPUTATIONAL FLOW ANALYSIS OF A CONTINUOUS FLOW PEDIATRIC VENTRICULAR ASSIST DEVICE

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MOTIVATION

Recent statistics suggest that approximately 10,000 to 20,000 pediatric patients each year may benefit from long-term circulatory support [1]. These pediatric patients suffer from debilitating congenital heart defects, which ultimately may require a heart transplant. While these patients wait for a donor heart to become available, they normally receive supplemental mechanical circulatory support. With a limited number of donor organs (~400 each year), the need for long-term mechanical circulatory support for this population continues to grow [1,2].

BACKGROUND

To date, most clinical experience with pediatric mechanical assist systems has only been for short support durations [1,2]. The Berlin Medos-HIA system, Berlin Heart, Medtronic Biomedicus pump, Abiomed BVS 5000, Hemopump, ECMO, and the IABP system have shown promise in reversing cardiac failure and providing short-term circulatory support to patients with congenital heart defects [1]. Many of these systems, however, encompass a large number of instruments, elaborate plumbing, and a finite capacity or stroke volume capability. These extracorporeal assist devices further require larger quantities of anticoagulants, continuous monitoring, and patient immobility for effective operation. Despite these drawbacks, the assist systems proved successful for short-term support and encourage the belief that effective, implantable long-term support is possible [1-2].

PEDIATRIC VAD DESIGN

We seek to develop a reliable and effective pediatric circulatory assist system to meet the growing need for long-term support. This research work details the initial design of an implantable pediatric ventricular assist device (VAD) as illustrated in Figure 1. The pediatric VAD, a centrifugal pump, consists of an inlet volute, exit volute, four-bladed 35 *mm* diameter impeller, and a clearance region between the impeller and internal housing. As a geometrically similar and scaled down version of the Virginia Artificial Heart Institute's adult left ventricular assist device (LVAD), the pediatric pump's impeller is fully suspended by magnetic bearings. Based on a demographic study reported by Dr. Thomas Spray at the 48th Annual ASAIO Conference in June of 2001, the pediatric VAD is designed to support patients from 2 to 12 years of age. For resting conditions, this age range would require a cardiac output of 2 to 5 *lpm* with an average aortic pressure of 70 to 85 *mmHg*, which correspond to approximate pump operating conditions [3]. The pump's fluid regions measure 22 *mm* in height by 54 *mm* in diameter.

COMPUTATIONAL FLOW MODEL

A computational flow model of the continuous flow pediatric VAD was created with approximately 360,700 grid elements. Employing computational fluid dynamics (CFD) software, the bulk performance parameters (head-flow curves, pump efficiency, fluid force exerted on the impeller, and shear stress levels) were determined for steady state flow conditions. The simulations involved flow rates of 2 to 5 *lpm* for 2750 to 3250 *RPM* and incorporated a k-ε fluid turbulence model, which has been applied in the design of several heart pumps [4]. Blood, an incompressible fluid, exhibits Newtonian behavior for shear rates greater than 100 s^{-1}, as predominantly observed in this study [5]. Therefore, the whole blood density and viscosity were taken to be constant values of 1,050 kg/m^3 and 0.0035 $kg/m\text{-}s$, respectively [6].

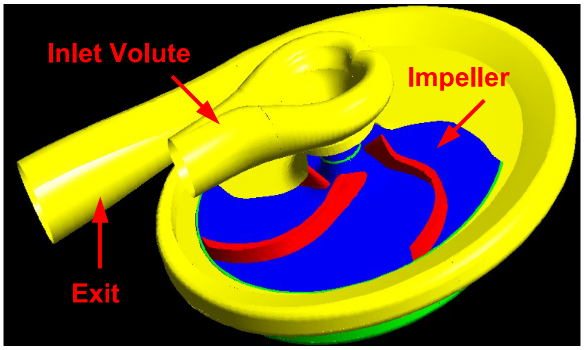

Figure 1. Full Computational Model of the Pediatric VAD

PERFORMANCE RESULTS

Figure 2 illustrates the head-flow performance curves for three rotational speeds. Each data point corresponds to a steady state CFD simulation. These results demonstrate the pump's ability to deliver 2 to 5 *lpm* at a pressure of 70 to 95 *mmHg*, depending on the operating rotational speed. The CFD results further indicated best efficiency points ranging from 25% to 28%, which correlate well with typical efficiencies for blood pumps.[4,6]

Since the impeller is fully suspended by magnetic bearings, a blood-filled clearance exists between the impeller and housing due to the magnetically levitated design. The pressure gradient between the exit volute and inlet volute causes retrograde flow through this clearance. Therefore, a small proportion (approximately 10%) of blood recirculates through the pump. The thickness of the back clearance gap is designed to create a balance between minimizing exposure time and preventing excessive reguritant flow. The fluid in this clearance exerts a force on the impeller in the positive axial or y-direction, which the suspension system must counteract to maintain the rotor's optimal position and performance. Hence, the force's magnitude is critical to the magnetic suspension design. Using a macro in the CFD program, the axial fluid force magnitudes were determined and ranged from 0.5 to 2 *N*, depending on pump operating conditions.

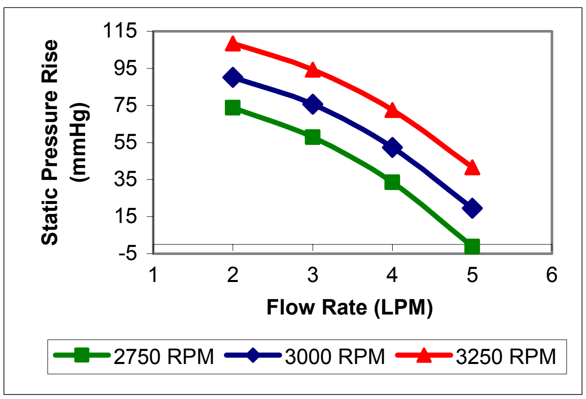

Figure 2. Head / Flow Performance Curves

In addition to calculating Reynolds' averaged velocity flow fields, pressure gradients, and pump efficiencies, CFD includes the ability to determine the shear stress at any nodal location in the flow field. Qualifying and quantifying the shear stress within the pump allows designers to estimate whether hemolysis or thrombosis may occur and adjust the design parameters accordingly to reduce the likelihood of occurrence. To account for the three-dimensional nature of the shear field, we adopted scalar shear stresses (τ) as originally introduced by Bludszuweit [7]:

$$\tau = (\frac{1}{6}\sum(\tau_{ii} - \tau_{jj})^2 + \sum{\tau_{ij}}^2)^{\frac{1}{2}}$$

Figure 3 displays the scalar stress values for the plane along the tip of the impeller blades. This surface shows the highest magnitude of shear stresses in the pump. In this plane, the higher shear stresses exist along the trailing edge of the impeller blade prior to entering the exit volute, on the suction of the blades, and directly along the blade tip surfaces, particularly at the trailing edge. Maximum shear stress values of approximately 300 *Pa* are found in this plane; the fluid velocities are large enough to produce a residence time of only a few milliseconds in the impeller.

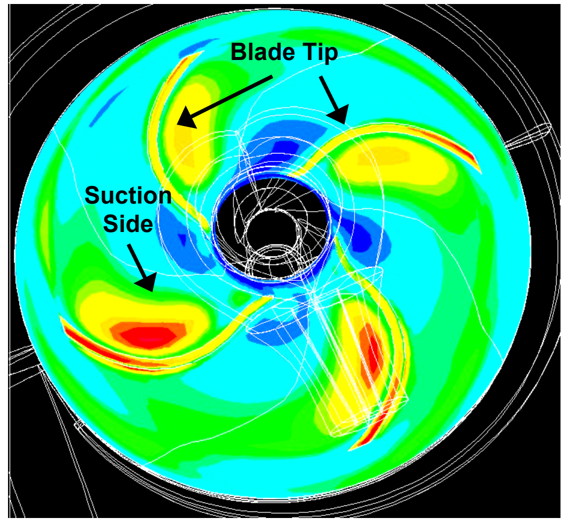

Figure 3. Shear Stress Along Blade Tip

DISCUSSION

This first generation design demonstrates the potential of this system to support pediatric patients. The desired output flow, pressures, and efficiencies were achieved for a fully implantable pump design. Shear stress levels throughout the VAD remained below 300 Pa, thereby signifying low levels of hemolysis. A closer examination of the velocity profiles in the inlet volute, exit diffuser, and blade regions, however, revealed signs of possible irregular flow patterns in need of optimization. Nevertheless, this design represents a good starting point for future work initiatives and model enhancements.

ACKNOWLEGDMENTS

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