A Design Improvement Strategy for Axial Blood Pumps Using Computational Fluid Dynamics

GREG W. BURGREEN, JAMES F. ANTAKI, AND BARTLEY P. GRIFFITH

During the initial stages of concept development of non-traditional axial flow pumps, numeric simulation offers an attractive advantage. Computational fluid dynamics (CFD) provides the rationale to evolve the design numerically such that undesirable flow features may be significantly mitigated before a physical prototype is fabricated. The initial design of a novel axial flow blood pump is shown through CFD analysis to exhibit large regions of reverse flow. Such fluid dynamic behavior not only decreases the pump's hydrodynamic efficiency, but, more significantly, increases its overall potential for blood trauma and thrombogenesis. The design improvement strategy consists of creating a geometric model of the blood wetted surfaces and changing the associated geometric parameters such that more desirable fluid dynamic behavior is systematically attained with each incremental modification. The fluid flow through each new pump design is analyzed by numerically solving the incompressible Navier-Stokes equations in rotating coordinates. Marked improvements in the major fluid dynamic aspects of the axial flow pump were observed over an evolutionary sequence of four generations of pump design. ASAIO Journal 1996;42:M354–M360.

The advent of the information age brings with it a reconsideration of the established or traditional process of design. A popular trend of the 1990s is the inclusion of some level of numeric modeling or simulation early in a product development cycle. Furthermore, this practice is rapidly becoming an increasingly effective means to guide design changes. The design cycles of the aerospace, automotive, civil engineering, and biomedical industries are beginning to incorporate elements of computer aided design guidance. Computational solid modeling of complex geometries reduces the time from drawingboard to end-product and also simplifies the integration of system development and manufacturing process control. Computer based rapid prototyping, meanwhile, is revolutionizing the early stages of the design process by enabling the physical realization of conceptual designs. Numeric simulations involving structural statics and dynamics, fluid dynamics, electromagnetics, and heat transfer provide valuable physical insights during the preliminary design stage of product development. Such insights often directly lead to immediate design refinements and improvements.

Within the bioengineering research field, computational fluid dynamics (CFD) is increasingly being used to augment fluid dynamic understanding when experimental diagnostics or flow visualization is not possible or lacks sufficient detail. The analysis capabilities of CFD are also beginning to be recognized by the biomedical community as a means to effect design improvements. Specifically, CFD is being used as a tool to help designers suggest improvements in the fluid dynamic behavior within prosthetic devices or pertaining to a surgical procedure. Some flow problems that have been analyzed with this intent in mind include blood pumps of rotary and positive displacement types and vascular grafts and stents. However, no examples of a rigorous, biomedical CFD-based design improvement application are found in the literature.

The intent of this report is to demonstrate a practical design improvement strategy that integrates CFD in the design development of a novel rotary blood pump. Our design strategy uses a "cut and try" approach, but totally within a computational framework. In this context, CFD provides the rationale to evolve the design numerically during the earliest stages of its development cycle. It will be shown that the fluid dynamic behavior of this rotary blood pump is significantly improved by our design improvement strategy.

**The Design Problem**

We are interested in improving the fluid dynamic behavior of a second generation cardiac assist device (artificial heart...
or blood pump). Our design improvement strategy is applied to a blood pump that takes the form of a small axial flow turbopump. Recently, several blood pump concepts that are based on axial flow turbomachinery principles have been introduced. However, the novel aspect of the present design is that the rotor body and its attached impeller blades will be magnetically supported about a stationary hub, hence forming an annular flow passage or gap between the two components. Consequently, in theory there will be no physical contact between the rotor assembly and any other component of the pump. A patent is pending on this blood pump design. Figure 1 shows a schematic outline of the defining profiles of the pump, and Figure 2 shows the resulting pump surface geometry. (Although not modeled in this work, a set of stator blade assemblies in the actual geometry will dually serve to straighten the flow and structurally support the stationary hub and flow guides.) The operating point of the four bladed blood pump consists of a rotational speed of 10,000 rpm and a volumetric flow rate of 5 L/min. In this study, we seek to improve the overall fluid dynamic behavior of the pump by refining its entire shape (excluding the housing inlet diameter of fixed 13 mm dimension).

Typically, pump performance is defined in terms of pressure head and brake horsepower, which in turn depend on flow rate, impeller diameter, rotational speed, and the working fluid properties of density and viscosity. However, the most crucial performance criteria in blood pump design are those that are physiologically based—in particular, criteria that involve the mitigation of blood damage due to deleterious fluid dynamic disturbances, direct mechanical forces, and time exposures to non-biocompatible surfaces. It has been established that direct or delayed destruction of blood cells (hemolysis) is closely correlated to high shear stress levels, cavitation, and turbulent stresses. In addition, the fluid dynamics couples with biologic activation mechanisms that initiate the deposition of blood elements on non-biocompatible blood contacting surfaces, and also encourage the formation of platelet aggregates. The potential for thrombus formation is increased proportionally to the local residence times of blood elements within the pump. Hence, local thrombogenesis is intensified by the presence of stagnant or recirculating regions in the flow path of the pump.

For the present design problem, the main objective is the improvement of the thrombogenic character of the blood pump. We propose to accomplish this by eliminating or reducing the regions of reverse and stagnant flow throughout the pump at its operating point. The defining profile shapes of the pump (Figure 1) are refined to this end.

The Design Improvement Strategy

The design improvement strategy embodies the traditional heuristic ("cut and try") approach in which a design configuration is systematically varied through incremental shape changes that are derived from human intuition and experience. Each design iteration involves 1) a CFD analysis, 2) a critical evaluation of the pump fluid dynamics, 3) an incremental shape change, and 4) a computational remeshing of the new configuration. Each of these major components of the design strategy is further discussed later as it pertains to this work.

**Computational Fluid Dynamics Analysis**

The physical domain about the blood pump geometry is spatially discretized by an unstructured tetrahedral mesh generated with an advancing front method. Figure 3 shows the unstructured surface mesh of the blood pump. Some features of the computational mesh include periodic midblade surfaces, a finite clearance between the blade tip and outer housing, and the presence of axisymmetric flow guide structures. Although rotor–stator interactions will exist in the ac-
tual geometry, this aspect is not modeled or analyzed in the present study. The incompressible laminar Navier-Stokes equations written in a rotating coordinate system are solved using RAMPANT (Fluent, Inc., Lebanon, NH). The numeric algorithm is a pseudocompressibility finite volume method using an explicit Runge-Kutta time integration scheme with multigrid acceleration. A Newtonian constitutive model is selected with viscosity equal to the asymptotic value for human blood (3.5 cp). A typical CFD analysis requires approximately 8 central processing unit hours on a Sun Sparc 10 (Sun Microsystems, Inc., Mountain View, CA) workstation for 170 multigrid cycles, resulting in a two order drop in the residual L2-norm.

Geometry Representation

Surface representation is at best a tedious exercise. With commercial computer aided drafting (CAD) packages, a particular design configuration may be satisfactorily generated, but any subsequent modifications to the configuration may require one to “scrap” the model and reinvest a substantial effort in generating the new modified geometry. In a situation requiring repetitive design changes, CAD software often lacks the required degree of parametric generality and user interaction to automate this.

A vital component of our design strategy is a mathematical model of the pump surface that is specifically customized to describe a wide range of axial flow pump geometries. This parameterized surface model is driven by a set of design variables that are directly linked to the key defining profiles of the pump geometry. The current model defines the axisymmetric profile shapes of the stationary hub, rotor body, outer housing, fore and aft flow guide cross-sections, and the impeller blade shape (Figure 1). The impeller blade shape requires the specification of profiles for the blade tip shape (to define the blade tip clearance), leading and trailing edge shapes, and the specification of axial distributions of the blade thickness, wrap angle, lean angle, and twist angle. In addition, each profile shape and axial distribution may be defined as either a linear function, a Bezier-Bernstein curve, one of various conic functions, or a point list. The final pump surface is generated by geometrically manipulating the defined profiles and axial distributions.

Fluid Dynamic Evaluation Criteria

The primary fluid dynamic feature that we desire to control is the presence of retarded or reversed flow within the pump. By minimizing such flow patterns that are favorable for deposition of blood elements, the thrombogenic character of the pump is improved. The regions of poor quality flow are identified through a qualitative evaluation of the interior and near-surface velocity vector field through graphic post processing of the CFD analysis results.

Design Modification

If the design is regarded as fluid dynamically deficient, corrective design changes are then prescribed based on an understanding of the fluid dynamic phenomena, intuition, and the collective pump design experience as brought to bear on the problem. In this study, the geometry modifications are restricted to the axisymmetric and impeller blade profile shapes; the axial distributions associated with the impeller blade shape are not modified.

Computational Domain Remeshing

Each design change results in a new pump geometry that requires a computational remeshing of the physical domain. Application of a simple automatic remeshing technique based on an elastic spring analogy for unstructured meshes consistently produced degenerate elements for this internal flow geometry. Consequently, each new pump geometry was remeshed from scratch using the previously mentioned mesh generation software.

Results

Design 1 Configuration

The first conceptual embodiment of the proposed axial flow pump is shown in Figures 1–3 and is referred to as Design 1. The slight axial flare of the stationary hub is intended to subject the rotor–hub annular gap flow to centrifugal forces and hence encourage a positive direction throughflow. Both fore and aft flow guide structures are included in the initial design. The fore structure is intended to help guide or funnel inlet flow into the annular gap region, whereas the aft structure is meant to entrain a portion of the energetic exit flow of the impeller to prevent flow separation on the aft hub section. The remainder of the pump design is specified based on a simple meanline analysis and also rather arbitrary aesthetic appeal.

Design 1 Analysis

The CFD flow analysis of the Design 1 configuration (Figure 4) indicates that very strong reverse flows exist in the hub, annular gap, and rotor body regions, as well through the interior portion of both flow guide structures. In fact, a two layered flow structure forms within the pump, with the more radial forward moving flow being recirculated back through the pump in a reverse direction through the flow guides and near-hub regions. Such massive amounts of reverse and recirculating flow are unacceptable because this behavior not
Design 2 Analysis

The bell shaped hub is found significantly to reduce the severity of reverse flow in the annular gap region. However, reverse flow still persists inside the aft flow guide structure and also near the blade root at the exit region of the impeller (Figure 5C). In fact, the aft guide structure appears to be instrumental in promoting the observed reverse flow pattern. The highly rotational forward moving flow exiting from the impeller tips remains radially above the aft flow guide structure.

Design 3 Configuration

The main physical characteristic of the third generation design is that a slightly mixed flow type of impeller is adopted (Figure 6A,B). This geometry modification is chosen to shorten the relative height of the impeller blades and, it is hoped, mitigate the degree of blade-to-blade reverse flow

only adversely affects hydrodynamic efficiency, but most certainly would result in thrombus formation over the entire pump.

Design 2 Configuration

Design 2 represents the second generation pump design (Figure 5A,B) resulting from our geometric design modifications to Design 1. To impart larger centrifugal forces to the annular gap flow, the hub axial flare is significantly increased to resemble a bell shape. The fore flow guide structure failed to funnel the low momentum inlet flow and is therefore eliminated. The aft flow guide structure is retained but repositioned to capture most of the exiting annular gap flow and other centrifugally induced flow. This is done to decrease the amount of separated flow expected at the aft portion of the bell shaped hub. To augment this anticipated momentum

Figure 5. The Design 2 blood pump. (A) Defining profiles; (B) surface geometry; (C) near-surface axial velocity contours (m/sec).

Figure 6. The Design 3 blood pump. (A) Defining profiles; (B) surface geometry; (C) near-surface axial velocity contours (m/sec).
observed in Design 2. The outer housing shape is altered to accommodate a larger diameter hub and rotor assembly. To minimize downstream flow separation, the use of an aft flow guide structure is abandoned, and instead, the housing-to-hub flow passage aft of the impeller blade exit is reduced. It is noted that the aft housing shape is a completely artificial construction to minimize flow separation in our numeric calculations. In actuality, a set of stator blades would be positioned in this vicinity to effect flow straightening and to provide hub support, but, as stated earlier, we do not account for the stator blade assembly. Finally, with intentions of increasing the fluid rotational speed and hence centrifugal forces within the annular region, the annular gap exit width is tightened to foster increased viscous effects.

Design 3 Analysis

The elimination of the aft flow guide combined with the modified housing shape leads to an overall reduction of flow reversal and separation in the pump (Figure 6C). However, a strong pocket of reverse flow still persists near the blade root at the exit region of the impeller. Moderate flow separation is observed downstream as the hub diameter decreases. The smaller annular gap exit width yields a slight improvement in the annular gap flow.

Because most of the flowfield is well behaved in Design 3, we focus our attention on the inflow blockage due to a ring vortex residing near the impeller blade tip inlet region. This flow structure is found in both of the previous designs, but was not addressed there because of our larger concerns of suppressing reverse flows elsewhere. The ring vortex stems from a combination of the leakage flow through the blade tip clearance and the cylindrically shaped fore housing section. In this design, the size of the vortex is rather large, extending far upstream and occupying the outer 50% of the housing radius (which corresponds to 75% blockage of the inflow area). This blockage significantly diverts the inflow, as evidenced by the unusually high velocities at the fore hub tip. The main concern with allowing the vortex to remain intact is the increased blood surface interaction, and hence increased possibility of thrombus formation.

Design 4 Configuration

In this configuration (Figure 7A,B), the geometry changes are specifically tailored to eliminate the three predominant flow disturbances of Design 3. To address the problem of reverse flow in the impeller exit region, the relative heights of the impeller blades are further decreased by increasing the taper of the rotor body. This modification significantly decreases the cross sectional areas of the flow passages in the impeller, hence increasing the mean axial velocities there. To suppress the ring vortex, the fore portion of the outer housing is modified such that a throat is formed upstream of the impeller blade inlet. The rationale behind this modification is geometrically to contain the growth of the vortex as well as to increase the momentum of the inlet flow to counteract directly the blade tip leakage. Finally, the housing outlet diameter is reduced to promote artificially flow attachment at the aft hub region.

Figure 7. The Design 4 blood pump. (A) Defining profiles; (B) surface geometry; (C) near-surface axial velocity contours (m/sec).

Design 4 Analysis

With these last geometry modifications, almost all undesirable fluid dynamic behavior is significantly mitigated or eliminated (Figure 7C). This fourth generation pump design is remarkably free of reverse flow. The main sources of reverse flow include the unavoidable mass leakage through the blade tip clearances and a small flow separation on the suction side of the impeller. The ring vortex is much diminished in strength and is completely damped out midway between the housing throat and the impeller inlet. Overall, the blade-to-blade impeller flow is directed in a predominantly forward direction. A small region of stagnant flow is located at the aft pressure side blade root of the impeller.

Discussion

Marked improvements in the major fluid dynamic aspects of an axial flow blood pump are observed over an evolutionary sequence of four generations of pump design. The effectiveness of the present heuristic design improvement strategy critically depends on 1) an accurate qualitative interpretation of the CFD analyses, and 2) an intuitive feel for the fluid dynamic system response to corrective geometry changes. The pump surface model and its parameterization
are found to be invaluable in the iterative refinement of the pump shape.

It is entirely feasible that a similar final pump design may have been obtained if an analogous experimentally based design approach had been performed. However, it is unlikely that certain flow phenomena within the pump would have been as evident as in the present computationally based approach. For example, the internal structures of the blade tip ring vortex or the flow disturbances within the impeller blade-to-blade and annular gap regions may not be readily identified or interrogated when using conventional flow visualization techniques.

For decades, the aerospace community has invested enormous resources into designing near-optimal shapes for external compressible flow configurations. However, over the same time period, pump design has remained resistant to major advancements because of its more complex internal flow interactions. Traditionally, pump design has been performed relying on human expertise and experience aided by simple meanline analyses. More recently, quasi-three-dimensional numeric flow methods have infiltrated into the pump design process. Ultimately, however, the final stages of pump design rely to some extent on an experimentally based heuristic design improvement cycle.

In this work, we bypassed the initial steps of traditional pump design and moved directly to a numerically based heuristic design improvement stage. Although it is prudent to incorporate the full extent of traditional pump design knowledge into an initial design, we have shown that it is not absolutely necessary to do so. We are not advocating a disregard of human expertise and experience, for these are essential to design properly for such exigencies as off-design conditions and cavitation phenomena. However, pumps with a working fluid of human blood impose critically vital flow constraints in pump design that are not traditionally considered, namely, blood damage criteria. In this situation, an extremely detailed analysis and understanding of the fluid dynamics is required for effective blood pump design. We believe that CFD substantially provides both the qualitative and quantitative information necessary for this situation. Furthermore, we are confident that our numerically based heuristic design improvement strategy provides a promising means to resolve some of the complicated issues associated with blood pump design that are otherwise totally left to human intuition and experimental testing.

Ongoing progress towards the completion of the rotary blood pump design described herein includes additional design optimization and experimental testing and validation. The improved preliminary pump design of this work is amenable to further refinement using a CFD based design optimization strategy. This more advanced design improvement strategy directly couples CFD with numeric optimization techniques to extract some quantitative index of performance. We have successfully applied such a design strategy toward the shape optimization of artificial heart components. The applicability of CFD based design optimization to three dimensional geometries has been demonstrated for aerospace applications, and recently has been applied to the present blood pump geometry. A physical prototype of the present geometry is currently being constructed using stereolithography techniques and is scheduled to be tested experimentally. With the availability of such testing, validation of the CFD analyses and optimization results can be performed, enabling a realistic evaluation of the potency of the numeric design improvement strategies described earlier.

Future work will involve upgrading the numerics and mathematical physics of the CFD analysis and also incorporating mathematical models for predicting blood trauma. The inclusion of non Newtonian viscosity effects will be especially important in the stagnant flow regions, and a turbulence model is necessary to predict better the flow separation characteristics of the pump.

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References

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Progressive Pressure Expansion in Skeletal Muscle Ventriole Conditioning

KENNETH J. GUSTAFSON, JAMES D. SWENEY, JOHN GIBNEY, AND TEDD A. BRANDON

Skeletal muscle ventricle (SMV) conditioning typically results in reduced muscle performance. This study investigated the effects of progressive SMV resting pressure expansion and dynamic muscle training on SMV pumping capability. SMVs were formed from latissimus dorsi muscle in five goats. Three experimental SMVs were conditioned against a compliant pneumatic implant system. SMV resting pressure was progressively increased as the SMV adapted to each increment. Resting pressure rose from 40 to 100–120 mmHg over an 8 week period of time. Two control SMVs were conditioned against a non expanded incompressible implant. Both experimental and control SMVs were electrically burst stimulated for at least 6 weeks after an initial 2 week vascular delay interval. Results demonstrate that 1) experimental SMVs increased in volume; 2) SMV passive and active (evoked isovolumetric pressure) pressure-volume curves adapted to the increasing or static resting volume; and 3) two of three experimental SMVs generated greater stroke volumes than control SMVs across a range of counterpulsation pressures and electrical stimulation parameters. Progressive pressure expansion using a compliant implant system improved final SMV pumping performance and merits further investigation. ASAIO Journal 1996;42:M360–M364.

Skeletal muscle ventricles (SMVs) can be created by wrapping a muscle, typically the left latissimus dorsi, into or around a ventricle shaped chamber. Electrical stimulation of peripheral nerve fibers innervating the latissimus dorsi is used to condition and control contraction of the relocated skeletal muscle. Prolonged, continuous electrical conditioning can result in the extremely high fatigue resistance required for cardiac assistance. However, as Salmons and colleagues have pointed out, the fiber atrophy that may occur results in a loss of power output and an overly slow velocity of contraction that is poorly matched to cardiac applications. Mobilization of skeletal muscle for cardiac assistance, and especially loss of resting tension, also have damaging effects on muscle performance. Conversely, Goldspink and colleagues, as well as others, have found that "mechanical conditioning" of skeletal muscles (through immobilization in a stretched state) can produce anabolic effects that include muscle lengthening and prevention of connective tissue accumulation. Mechanical loading and chronic electrical stimulation have been used in combination in rabbit hind limb skeletal muscles to produce a dramatic, rapid conversion of fast to slow fibers, while simultaneously inducing muscle lengthening and hypertrophy. Leighton and colleagues have shown that mechanical conditioning through tissue expansion can produce increased muscle mass and blood flow.

The primary objective of the research described in this paper has therefore been to determine whether combined electrical and mechanical conditioning methods can be used to produce highly fatigue resistant, yet large volume and powerful, SMVs. Mechanical conditioning through progressive increases of SMV resting pressure to just beyond that needed for peak performance (instead of immediate exposure to a high physiologic pressure) may allow the muscle to adapt incrementally without overstretch and resulting damage. It is known that SMV pressure generation performance adapts to a static resting pressure and to increases in SMV resting pressure in a compliant implant system. A compliant, pneumatic implant system also allows control of SMV resting pressure and allows dynamic muscle shortening during conditioning. SMV conditioning against a highly compliant workload can act to preserve muscle contractility and power generation capability. We have sought to determine if an experimental SMV conditioning protocol combining progressive SMV pressure expansion and electrically stimulated dynamic muscle shortening results in greater SMV pumping performance than an electrically stimulated control protocol using an incompressible implant. An improved conditioning protocol that maintains muscle mass and contractile speed should maintain muscle performance and thereby increase the future potential for cardiac assistance with SMVs. A brief report on early results of this study has appeared previously.