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Original Research Paper

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OPTIMIZED CENTRIFUGAL PUMP DIMENSIONS OF SAISPANDAN: HYBRID BSRM TOTAL ARTIFICIAL HEART FOR DESTINATION THERAPY

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ABSTRACT Optimization of pump characteristics when supporting the mechanical functions of heart as destination therapy is essential because of limitations imposed on design on size and power. Computational fluid dynamics assessment with fluid bearing similar properties is done. Knowledge of previously designed similar pumps is an added advantage. Pumps supporting heart operate in Reynolds numbers of 10^5.Limited data is available on pumps that operate in this range. It is useful to present data using traditional Cordier diagram of nondimensional speed vs diameter. Flow, pressure and rotational speed are the main parametric analytic points. Similitude helps in efficient design concept.

KEYWORDS : CFD Computational Fluid Dynamics, FEA finite element analysis, VAD ventricular assist device, TAH-total artificial heart, MATLAB, ANSYS

Introduction

Blood for pumps acting as heart reaches from the inflow cannula, enters the casing and through of the eye of impeller flows out into the outflow cannula. Variations in speed can impart pulsatiliy to flow.

Multiple impeller designs are available for total artificial heart design. Design of centrifugal pumps should minimize shear stress that is cause of hemolysis and thrombosis. The design chosen for impeller of Saispandan is backwardly curved four bladed designs. The impellers are powered by twin 12/14 bearing less switched reluctance motor powering its core. It is a twin motor double impeller assembly with proposed separate control systems for the right and left side. The development envisages in addition to a unique impeller design an automatic speed control system, automatic right/left balance, minimal hemolysis achieved with elegant designing of its elements and analysis using computational fluid dynamics. No thrombogenic zones were noted on analysis.

An open type backwardly curved four bladed impeller made in titanium was designed with maximum efficiency and stable hemodynamics. Backward blades within the normal working range of power avoid overloading at higher flow rates. Elaborating on concepts of similitude for efficient design of blood pumps machines with identical geometry, velocity triangle at congruous points in flow path, ratio of gravitational to inertia forces in flow path and handle the fluids with similar thermodynamic characteristics have equal efficiencies. Since pumps powering VADS and TAH are smaller speed vs diameter graphs applicable to industrial pumps may not result in similar performing characteristics. Cordiers concept that data points plotted on a specific speed diameter diagram could be fitted into curves based on their efficiency [1].Modification of this concept by Balje [2] by relating highest achievable efficiency relating to size, speed, pressure rise and flow rate has been used extensively by designers for turbo machines resulting in up to 80% efficiency. Estimate of size speed performance of variety of blood pumps can be done employing Smiths technique [3]. Present evaluation of pump was based on CFD, speed specific diameter diagrams and single loop rig test for validation.

Material and methods

Buckingham's Pi theorem for parametric grouping involving physical process is expressed [4].Non dimensional pump parameters are enlisted in Table 1. Parametric groups include Reynolds number, efficiency, Laval number, ratio of specific heats, specific speed and diameter. In the design process efficiency, Reynolds number, specific speed and diameter are most important. For impeller the pump size, number of blades, blade thickness, inlet an outlet width, diameters and angles need to be predetermined [5]. Meridional coordinate sytem was used to determine the optimum curvature. Cubic Bezier curve created profile from data points of meridional profile. Design procedure developed as MATLAB code that generated 3 D coordinates for blade and diffusion volute. Model computation as done using Solidworks and imported to ANSYS workbench for the impellers. Navier-Stokes equations predicted flow through blood pumps. Fluid volume around the rotor surface created the rotating fluid zone and volute zone was created by the fluid volume inside the volute section. Specific boundary conditions defined the rotational speed, inlet pressure and outlet flow rate (5L/min).Circulatory system included an impeller housing, valve to change loop resistance, flow meter and pressure transducers. Pressure variations were analyzed against steady flow rates. Cordier diagram for highest efficiency points of specific designs are plotted.

Result

Figure 1 shows the profile of open four bladed design with 3cm diameter, 60 degree outlet angle, 4mm outlet width and 1.25mm gap. Each experiment was repeated thrice for verification of repeatability [6].Numerical and experimental efficiency and head ratio versus flow ratio is shown in figure 2.Figure 3 plots numerical and experimental efficiency versus specific speed. Figure 4 shows the flow and head coefficients relation. Good data fit was obtained from second and third order polynomial. Blood analog solution of 65% water and 35% glycerol by volume is used for evaluation. Fluids with greater viscosity will result in higher shear stress, higher friction and slightly higher power input with lower efficiency.

Discussion

Shear stress of viscosity will lead to loss of pressure head. This phenomenon occurs in both tubes and impeller surface. Higher shear rates on centrifugal pumps impart a incompressible Newtonian behavior to blood. When shear rates exceed 100/s non Newtonian properties like shear thinning and viscoelastic nature are negligible. Shear stress on impeller surface is 20-40 times of that seen on tubes. For suspended impellers volute design that keeps a constant angular momentum assumes greater significance. At the design point efficiency is more for single volute designs. Q/Q design of 0.8 gives maximum difference between numerical and experimental efficiency. In addition to frictional loss, surface irregularities contribute to generation of this difference. At higher Reynolds numbers K- epsilon turbulence model improves accuracy. Lower number of blades lowers the slip factor [7].At higher speeds low frictional loss and lower pressure rise in system would be noted. Gap between housing and impeller design should be such that efficiency is maximum at an acceptable shear stress. When it exceeds 150Pa hemolysis is significant [8].Keeping non dimensionals in a selected range, parametric variation could help in detection of optimal hemolysis index. Hemodynamic compatibility is very crucial in blood pumps for use as destination therapy. Compared with mechanical seal pumps magnetic drive pumps have lower efficiency [9, 10].

Calculation domain for magnetic pump field and mesh assembly is given in figure 5. Much like the Starling response achieved with pulsatile pumps, centrifugal pump output can increase with a growing preload; however, unlike pulsatile pumps, centrifugal devices are also sensitive to the afterload [11]. This characteristic may allow pumps placed in series, as in the classic Saxton and Andrews experiment, to autoregulate their output in response to instantaneous variations in systemic and pulmonary pressures and return flows.

Conclusion

Geometric parameters have optimum value for each pump. Validation and experimental parametric optimization is the best available current method. Open impeller designs offer ability to operate at higher speeds with greater head. Backwardly directed impeller blades are in fact suited for all higher efficiency high pressure applications where system efficiency is the key indicator. Design optimization is yet another requirement of ultralow speed magnetic pumps. The factors affecting efficiency in the descending order include blade number, bias angle in peripheral direction, inlet diameter and deflection angle. Optimization of magnetic pumps should extend to noise and vibration reduction [Figure 6]. Magnetic bearings are both feasible and effective as blood pumps. Fully implantable devices with autonomous power generation or transcuteanous charging with biocompatible materials will drive the heart of artificial hearts of future. The patented design of impeller housing of Saispandan Total Artificial Heart for Destination Therapy is shown in Figure 7.

Table 1 Non dimensional pump parameters

Efficiency	η	(ideal power/actual power)
Reynolds number	Re	$\rho CD / \mu$
Specific speed	n_s	$N_s \cdot \sqrt{Q} / \left(\frac{\Delta p}{\rho}\right)^{0.75}$
Specific diameter	d_s	$D \cdot \left(\frac{\Delta p}{\rho}\right)^{0.25} / \sqrt{Q}$
Flow coefficient	ϕ	$Q / N_s \cdot r^3$
Head coefficient	ψ	$\Delta p / \rho \cdot N_s^2 \cdot r^2$

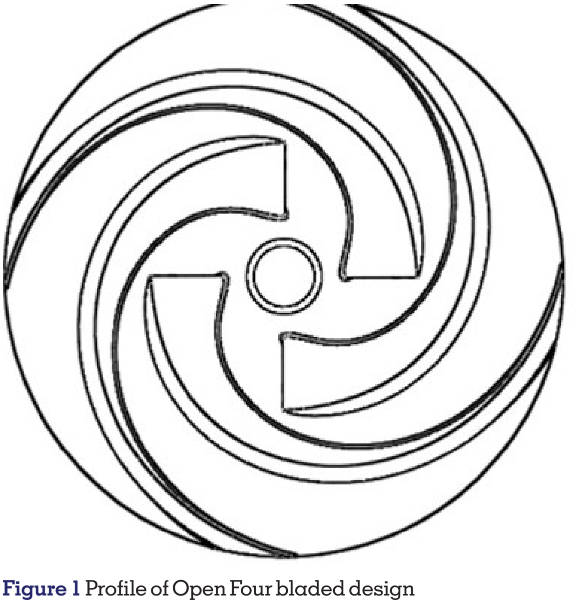

Figure 1 Profile of Open Four bladed design

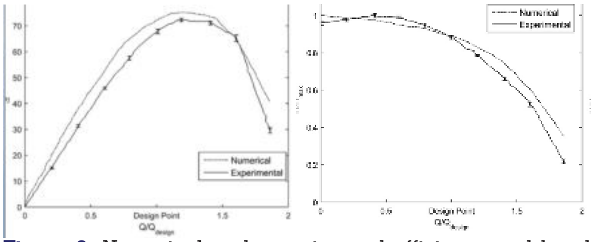

Figure 2: Numerical and experimental efficiency and head ratio versus flow ratio

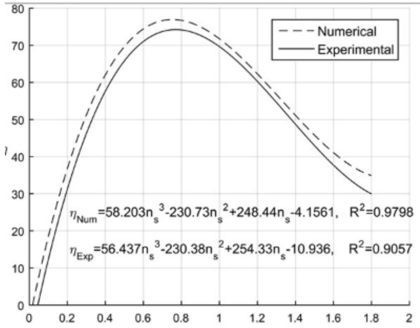

Figure 3: Plot of numerical and experimental efficiency versus specific speed.

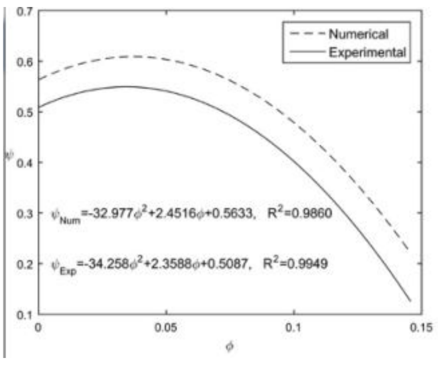

Figure 4: Flow and head coefficients relation.

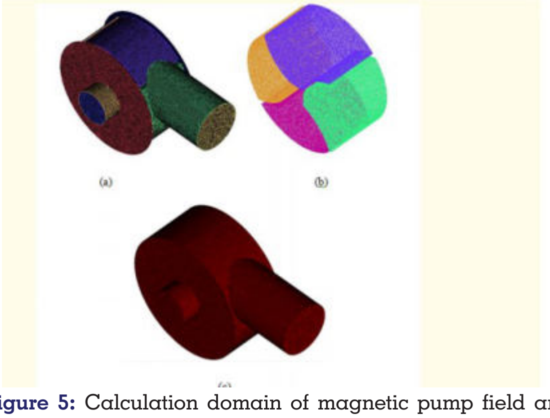

Figure 5: Calculation domain of magnetic pump field and mesh assembly.
A all parts, b four blades, c blood flow domain

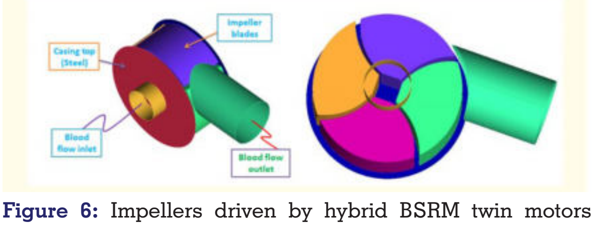

Figure 6: Impellers driven by hybrid BSRM twin motors Meshing of Saispandan Impeller Design 3 D geometry with four backwardly curved blades

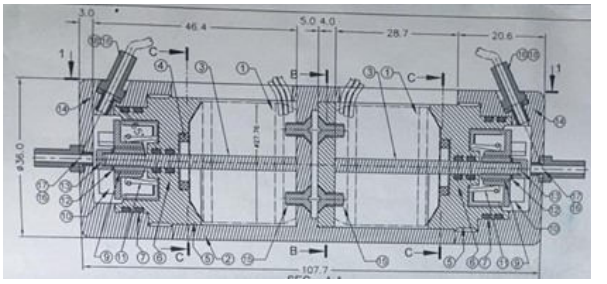

Figure 7: Design details of Saispandan Total Artificial Heart Impeller Housing and Design: Obtained with special permission from Division of Cardiothoracic Surgery - Saispandan Team Gitam University

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