



DESIGN AND DEVELOPMENT OF MINIATURE SLOTTED IMPELLER FOR SAI SPANDAN TOTAL ARTIFICIAL HEART FOR DESTINATION THERAPY

Cardiology

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ABSTRACT

In recent years, the use of rotary blood pumps (RBPs) as continuous flow VADs has surged ahead, and virtually eliminated the use of pulsatile-flow or volume-displacement pumps for implantable, chronic mechanical circulatory support (MCS). Circuit Design modifications like that in Saispandan has imparted pulsatility into RBP. Impeller designs are a significant factor when designing centrifugal pumps as mechanical circulatory assist devices as smaller diameter impellers with higher rotational speeds to achieve target outputs would cause more blood component trauma compared to larger diameter impellers. Hydraulic performance and hemolysis tests in the same pump housing with different prototypes is needed. Ventricular assist parameters for efficient circulatory support would include an output of 5 L/min against 100 mmHg at speeds of 2500-3500 rpm. Vein height does not contribute significantly to evaluation metric in most studies.

KEYWORDS

TAH total artificial heart, DT Destination Therapy, Mechanical circulatory assist MCA, CFD Computational Fluid Dynamics, Particle Image Velocimetry (PIV), ANN artificial neural network, LAP left atrial pressure

INTRODUCTION

Problem Statement:

Design of Miniature slotted impeller centrifugal pump for blood flow capacity of 5lit/min, Pressure Head of 0.135bar and rated speed of 2500rpm. The design data concept of impeller of Saispandan Total Artificial Heart for destination therapy is outlined below.

Design Data:

- 1) Q = flow rate in $m^3/hr = 5lit/min = 0.3m^3/hr$
- 2) N = Rotational speed of Impeller in r.p.m. = 2500rpm
- 3) D_1 = Impeller Inlet Diameter = 30mm
- 4) D_2 = Impeller Outlet Diameter = 40mm
- 5) α_1 = Blade inlet Angle = 16°
- 6) α_2 = Blade Outlet Angle = 21°
- 7) Z = number of blades = 4
- 8) b_1 = Blade width at the Inlet = 4mm
- 9) b_2 = Blade width at the Outlet = 5mm
- 10) h = Blade height
- 11) Impeller outlet width = 3mm
- 12) Volute Angle = 9°
- 13) Throat Area = $253mm^2$

- 14) Blood specific gravity = 1.0621

$$15) \text{ Specific speed } N_s = \frac{N\sqrt{Q}}{H^{3/4}} = \frac{2500\sqrt{0.3}}{1.295^{3/4}} =$$

- 16) H = Pressure head of blood developed in the pump
- 17) Pressure Head of blood developed in the pump ' H ' = $\frac{p}{\rho g} = \frac{0.135 \times 10^5}{1.062 \times 9.81}$

$$18) \text{ Shaft Power } P' = \frac{\rho g Q H}{\eta_o}$$

- 19) Take $\eta_o = 85\%$

$$20) \text{ Shaft Power } P' = \frac{1062 + 0.3 \times 9.81 \times 1.295}{0.85 \times 3600} = 4.76kW/3600 = 1.32 \text{ Watts}$$

- 21) The minimum power output of the electromagnet actuated motor should be 1.32Watts.

$$22) \text{ Shaft diameter } d_s = \sqrt[3]{\frac{360000 P (hp)}{N \tau_t}} = \sqrt[3]{\frac{360000 \times 1.776 \times 10^{-5}}{2500 \times 450}} = 0.08283mm$$

- 23) Velocity of flow at inlet $C_{m1} = ?$

$$24) x_i = \tan^{-1} \frac{C_{m1}}{V_1} = \tan^{-1} \frac{C_{m1}}{x D 1 N}$$

$$25)_{\text{lam}} 16 \frac{C_{m1}}{x D 1 N} \frac{C_{m1} \times 60}{x \times 0.025 \times 2500} 0.983 \text{ m/s}$$

$$26) \text{ Radial Thrust } T_R = \frac{K \times H \times D_2 \times b_2}{2.31}$$

where $K=0.21$, $H=1.295$, $D_2=60\text{mm}$, b_2 =Outlet width of the blade

$$27) \text{ Axial Thrust } T_A = (A_1 - A_2) P_1$$

Where A_1 =Area of the inlet including shroud cm^2

A_2 =Area of shaft including sleeve cm^2

Blade Profile Design:

$$= \rho \frac{R_2^2 \sin^2 \alpha_2 - R_1^2 \sin^2 \alpha_1}{2(R_2 \cos \alpha_2 - R_1 \cos \alpha_1)}$$

R_1 = Impeller Inlet Radius

R_2 = Impeller Outlet Radius

α_1 = Blade inlet angle

α_2 = Blade Outlet angle

Take a random point 'A' on circle D_1 . Draw a line with a α_1 angle and a length ' r ' in negative direction of the X-axis. Then take a point 'E' as the center and ' r ' as the radius to make the arc intersect the inlet and outlet circles at 'A' & 'B'. This 'AB' is the arc profile.

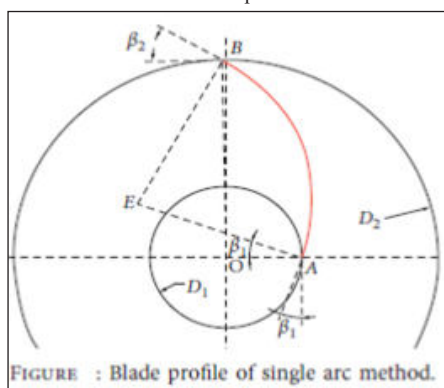


FIGURE : Blade profile of single arc method.

DISCUSSION

Fully implantable continuous flow pumps have opened the flood gates of long term mechanical support using these devices (1,2). Centrifugal pumps provide non pulsatile flows but technological modifications could induce pulsatility to output as with the Saispandan Total Artificial Heart. Many different impeller shapes have been analyzed (5) for efficiency and hemolysis patterns. Impeller internal diameter (ID) does not greatly change the hydraulic performance of a small centrifugal blood pump. The proper combination of the impeller ID and inlet port offset obtains less hemolysis.

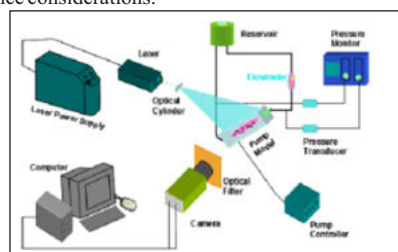
The larger impeller ID is considered to have less thrombogenicity around the top bearing(6). Evaluation is done on mock circulatory loops (figure 3). Advantages of continuous flow devices includes 1) no inflow and outflow valves are required, thus reducing the chance of thromboembolic complications 2) devices can be small and compact, that allows easier implantation and removal and 3) devices are less invasive, requiring smaller priming volume than the pulsatile devices. These favorable features lead to easier implantation and removal, simpler flow control to cover a wide range of patients, and estimation of pump flow through noninvasive monitoring of motor power consumption and rotational speed. Reverse flow through the pump at low pump speeds can be detrimental to the heart. Mechanical durability, biocompatibility of the shaft seal, power source recharging cycles, thermal issues of the motor are areas that would have to be attended to long-term durability. There are distinct advantages of inducing pulsatility too which would augur well for pediatric patients also(3,4) for recovery of cardiac function. For the pediatric population priming volume and conduit size are also critical factors. Volume displacement pumps require higher priming volume. Centrifugal- or axial-flow devices, are better suited for smaller subjects. The impeller and housing were fabricated with acrylic resin to yield a see-through view of the pump inside, whereas the hydrodynamic bearing's parts were made with titanium and polyethylene material.

External diameter of pump is 40mm, height of 20mm and impeller diameter is 30mm. The pump head weighs only 50 g (150 g when the motor driver is included). It met our design criteria by providing flows

of 0.5–5.0 l/min against a head pressure of 100–120 mm Hg at rotational speeds of 2,500–3500 rpm. Because of the wide flow range from 0.5 to 4.0 l/min, the device may be suited to support the circulation of pediatric population. The normal cardiac output requirement of 80 ml/kg–1 min–1 and 4.0 l/min, pump output is sufficient to support circulation of 50-kg patients. A small priming volume of 5 ml was also achieved, which is favorable for application to pediatric population. The inflow and outflow conduit size of 1/4" is for the pediatric prototype. The electrical-to-hydraulic efficiency ranges from 5% to 22.5%, requiring extremely low power from 1 to 6.0 W. Hemolysis testing is usually conducted at a flow rate of 1.7 l/min against a head pressure of 70 mm Hg and normalized index of hemolysis is calculated. CFD is the standard tool when designing RBPs, as it can calculate pressure-flow characteristics for a given impeller speed. Deriving shear forces helps in prediction of hemocompatibility by constitutive equations derived from empirical data. Particle image velocimetry (PIV) (figure 1) rapidly obtains velocity in fluids, which is employed experimentally to verify CFD-based predictions of flow field. Mock loop circuit is set with water solution with 37% of volume filled with glycerin to simulate the blood viscosity and density Figure 3. One of the key technologies required for rotary blood pumps is sealing of the motor shaft. Seal can be formed by injecting ferrofluid into the gap between the pole and the motor shaft and to minimize the possibility of ferrofluid making contact with blood, a shield with a small cavity should be provided. ANN can be used to estimate the LAP based on pump speed, pump flow, and aortic pressure, obtained from a mock circulation. Important reasons for hemolysis: 1) the drift of the pump output (including the volume and efficiency) from the design point; 2) impeller vane angles, i.e., the radial logarithmic spiral angle and the axial helical spiral angle; 3) roughness of vane surface and other blood contacting surfaces of the sealing box and pump housing; 4) vibration of the rotor resulting from dynamic disequilibrium; and 5) prerotation swirl at the inlet of the pump. The blood pressure to be pumped has been shown to have no influence on pump hemolysis. After eliminating the hemolysis factors, the blood trauma of the impeller heart has been reduced remarkably. The index of hemolysis of the nonpulsatile pump is 0.015.

3 D printing allows complexity and single unit fabrication without seams. It is less time consuming and requires less surveillance. Material selection depends on performance characteristics of the impeller(7,8).

Selecting the most appropriate motor to drive these miniature pumps requires the design team to reiterate the key parameters of pressure and flow, and conflicting requirements of lifetime, noise, temperature, current draw, feedback and weight and balancing all of these performance considerations.



PIV set up Figure 1



Figure 2, 3D Printing of Impellers

a- Single Arc, b- Double arc, c- Logarithmic spiral d B spline curve

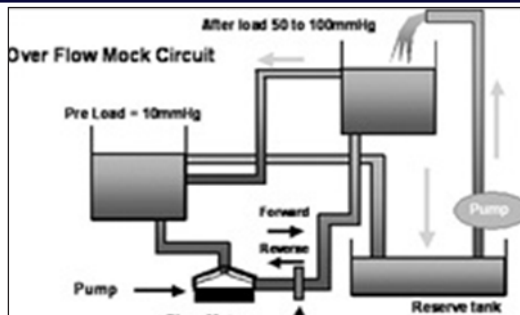


Figure 3: Evaluation by Mock Circulatory Loop

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