Design and development of caged ball heart valve using solid works

MIGCE j 0Xkf {cuci ct Department of Biomedical engineering, D.B.N.C.O.E.T, Yavatmal, India Tcxk'Xcto c'P0 Department of Biomedical engineering, D.B.N.C.O.E.T, Yavatmal, India

Abstract--Currently, over 2, 90,000 heart valve surgeries are performed worldwide annually and that number is estimated to triple by 2050. Even though patients with prosthetic valves lead a life relatively free from symptoms, problems like physiological complications and valve failure are significant. To date, all mechanical heart valves are plagued with complications associated with hemolysis and coagulation. These complications are believed to be associated with non-physiological blood flow patterns in the vicinity of the artificial heart valves. The geometry of the valve prosthesis with respect to the mitral valve annulus may significantly affect the flow dynamics in the human heart. It is thus essential to assess the hemodynamics of mitral prosthetic caged ball valve to improve the design of the device.

This research work presents a 3D model of the left human heart with optimized mitral caged ball valve engineered by a computational tool (SolidWorks 2009). The performance of the valve like hemodynamics across the valve, stress analysis and physical properties like mass, surface area, etc. is assessed virtually. This limits the need to perform extensive, costly and timeconsuming in-vitro and animal tests. Thus optimization of the caged ball heart valve design facilitates reduction of flow-induced thrombogenicity and reduces the need for post-implant anticoagulants.

Index Terms: Mitral valve, left ventricle, modeling, turbulence, hemolysis, stress

I. INTRODUCTION

The heart has four chambers. Blood is pumped through the chambers, aided by four heart valvesmitral, tricuspid, aortic and pulmonary. The valves open and close to let the blood flow in only one direction. Each valve has a set of flaps. When working properly, the heart valves open and close fully.

An artificial heart valve is a device implanted in the heart of a patient with heart valvular disease. Several different mechanical valves are currently available. Many of them have good bulk forward flow hemodynamics, lower transvalvular pressure drops, larger effective orifice areas, and fewer regions of forward flow stasis. The design of the first artificial ball-and-cage valve consists of a ball within a metal cage. The ball occludes the valve orifice and thus passively prevents backflow. The movement of the ball is pressure driven across the two chambers. Since 1962, the Starr–Edwards valve has undergone many modifications to improve its performance; however, the changes have focused on materials and construction techniques and have not altered the overall concept of the valve design. Caged ball valve implants generate flow patterns that are significantly different from those present in normal cardio vascular system (CVS). These flow patterns include regions of high velocity and high shear stress that potentially damage the blood cells [2]. Therefore, thromboembolic complications are being measured by turbulence shear stress experienced by blood cells. The threshold for haemolysis is at 800 N/m² [3].

II. PROBLEM DEFINITION

Patients who receive prosthetic heart valve (PHV) implants require mandatory anticoagulation medication after implantation due to the thrombogenic potential of the valve [1]. Optimization of the caged ball heart valve design facilitates reduction of flowinduced thrombogenicity and thus reduces the need for post-implant anticoagulants.

III. METHODOLOGY

A. Designing

Solid Works is a 3D mechanical Computer Aided Design (CAD) tool useful for modeling and analysis of various structural elements. This tool has been employed in human physiology to model and analyze the various human organs. The left human heart is modeled according to human anatomy [4]. It is semielliptical in shape with major and minor axes as 80 mm and 58 mm respectively (Fig.1). The volume of modeled left ventricle is 126 ml and surface area is 28598.71 mm². The conventional caged ball valve is designed in accordance with standard values using SolidWorks. The mitral annulus diameter is 30mm, orifice diameter is 19mm and ball diameter 23.75mm (Fig.2). SolidWorks enables to specify the materials the ball is of silicone rubber, cage is of titanium alloy and the sewing ring is of Teflon (PTFE).



 $Fig.1-3D \ model \ of \ left \ human \ heart$

The caged ball valve is then assembled in the mitral valve position in the left ventricle to analyze the hemodynamics across the prosthetic valve using FloXpress tool in SolidWorks.

The Boundary Conditions given as input in SolidWorks to analyze hemodynamics are:-

Inlet

Mass flow rate	=	0.126126 kg/sec
Volume flow rate	=	$1.26126 \text{ e-}04 \text{ m}^3/\text{sec}$
Temperature	=	310K
Pressure	=	15996 Pascal

Outlet

Pressure = 799.80 Pascal

Hemodynamic analysis with FloXpress tool shows a maximum velocity of 4.89 m/sec across the conventional caged ball valve (Fig.3).



Fig. 2- 3D model of caged ball valve



Fig.3- Hemodynamic analysis with FloXpress tool

However velocity profile in mitral blood flow in normal human heart is only 38.2 cm/sec - 123.9 cm/sec. High velocity of 4.89 m/sec across the prosthetic valve reults in hemolysis. Hence this research work deals with design optimization of caged ball valve to obtain the hemodynamics within the physiological limit.

B. Optimization

The strut of the caged ball valve is optimized so as to obtain the hemodynamics within the physiological limit. Four different designs for cross section of struts are considered – (A) circle (B) square (C) ellipse and (D) spline (Fig. 4)



Fig. 4 Strut design (A) circle (B) square (C) ellipse and (D) spline

Caged ball valve with different dimensions of each strut design ranging from 0.55 mm to 0.35 mm is modeled. The design of the model is evaluated on 3 parameters-

- (a) Hemodynamics within human physiological limit.
- (b) Maximum pressure bearing capacity- i.e. least stress over a range of pressure (10mmHg to 160 mmHg)
- (c) Minimum surface area

IV. RESULT

The first parameter - hemodynamic analysis of each design is evaluated using FloXpress tool of SolidWorks. The second parameter - pressure bearing capacity, is evaluated by performing Finite Element Analysis (FEA) over a range of 10mmHg to 160 mmHg for each design of the valve. The third parameter- surface area is measured for each design.

Design of strut	Dimension (in mm)	Blood Flow Velocity (m/sec)	Stress (N/m ²) at 10 mmHg	Surface Area (mm ²)
Circle	0.40	0.623	1324.2	1447.74
Square	0.45x0.45	0.625	1174	1363.21
Ellipse	0.45x0.35	0.624	1112.7	1455.43
Spline	0.40x0.40	0.622	4072	1398.82

Spline design of the strut is eliminated because of maximum stress development at 10mmHg. This indicates spline design of the strut has the least pressure bearing capacity. The above optimized dimensions of the three remaining strut designs, circle, square and ellipse, are then evaluated for pressure bearing capacity over a range of 10mmHg to 160mmHg

Table 2-Pressure Bearing Capacity

Pressure (mmHg)	Circular Strut Stress (N/m ²)	Square Strut Stress (N/m ²)	Elliptical Strut Stress (N/m ²)
10	1324.2	1245.7	1112.7
40	5296.7	4982.6	4450.7
80	10593.4	9965.2	8901.4
120	15890.3	14947.9	13352.1
160	21186.9	19930.5	17802.8





The graph indicates that the pressure bearing capacity of the elliptical strut caged ball valve is superior (least stress) than the circular strut caged ball valve with almost similar blood flow velocity and surface area.

V. CONCLUSION

The result of hemodynamic analysis of the left human heart with caged ball mitral valve and FEA of the valve indicates that elliptical strut of the valve is a superior design compared to conventional circular strut. Hence this optimization of the structure of the caged ball valve would greatly reduce flow induced thrombogenicity.

VI. FUTURE WORK

This model provides with a prospect to further optimize the caged ball valve. The pressure-stress analysis for different biomaterials can also be performed to increase the durability of the valve.

REFERENCES

[1] Ajit P. Yoganathan, K.B. Chandran, and Fotis Sotiropoulos, Flow in Prosthetic Heart Valves: State-of-the-Art and Future Directions, Annals of Biomedical Engineering, December 2005, Vol. 33. No. 12, pp. 1689–1694

[2] Lakshmi P Dasi, Helene A Simon, Philippe Sucosky, and Ajit P Yoganathan, *Fluid Mechanics of Artificial Heart Valves*, Clin Exp Pharmacol Physiol. *February 2009, Vol. 36. No. 2, pp. 225–237*

[3] P.C. Lu, H.C. Lai and J.S. Liu, A reevaluation and discussion on the threshold limit for hemolysis in a turbulent shear flow, Journal of Biomechanics, 2001; Vol.34, pp. 1361–1364

[4] Richard T. Schoephoerster and Krishnan B. Chandran, Velocity and turbulence measurements past mitral valve prostheses in a model left ventricle, Journal of Biomechanics, 1991, Vol. 24. No. 7, pp. 549-562

Creative Commons Attribution License 4.0 (Attribution 4.0 International, CC BY 4.0)

This article is published under the terms of the Creative Commons Attribution License 4.0 <u>https://creativecommons.org/licenses/by/4.0/deed.en_US</u>