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Effect of Arched Leaflets and Stent Profile on the Hemodynamics of Tri-leaflet Flexible Polymeric Heart Valves

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Abstract

Polymeric heart valves (PHV) can be engineered to serve as alternatives for existing prosthetic valves due to higher durability and hemodynamics similar to bioprosthetic valves. The purpose of this study is to evaluate the effect of geometry on PHVs coaptation and hemodynamic performance. The two geometric factors considered are stent profile and leaflet arch length, which were varied across six valve configurations. Three models were created with height to diameter ratio of 0.6, 0.7, and 0.88. The other three models were designed by altering arch height to stent diameter ratio, to be 0, 0.081, and 0.116. Particle image velocimetry (PIV) experiments were conducted on each PHV to characterize velocity, vorticity, turbulent characteristics, effective orifice area (EOA), and regurgitant fraction. This study revealed that the presence of arches as well as higher stent profile reduced regurgitant flow down to 5%, while peak systole downstream velocity reduced to 58% and Reynolds Shear Stress values reduced 40%. Further, earlier reattachment of the forward flow jet was observed in PHVs with leaflet arches. These findings indicate that although both geometric factors help diminish the commissural gap during diastole, leaflet arches induce a larger jet opening, yielding to earlier flow reattachment and lower energy dissipation.

Keywords

Particle Image Velocimetry; Fluid Mechanics; Commissure Coaptation; Reynolds Shear Stress; Turbulent Kinetic Energy; Prosthetic Heart Valve; Polymeric Heart Valve

Introduction

Cardiovascular disease is the leading cause of death in the United States [23] and heart valve (HV) replacement, usually for aortic valve disease, is the second most common cardiac operation performed in the US [16, 31]. Aortic valve disease is prevalent in 2.5% of the overall US population, and 10.4% of its elderly population[24, 35]. Currently, the most

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common types of prosthetic heart valves used in aortic HV replacement procedures are either mechanical or bioprosthetic. However, these valves introduce significant risk of thromboembolism (which requires the patient to undergo lifelong anticoagulant therapy), or the patients become more prone to valve degeneration and tissue failure (which would require reoperation) respectively [3, 7, 27, 30]. Despite decades of research, there remains a need for a prosthetic valve that would be durable and does not necessitate anticoagulation therapy[15].

Synthetic material heart valves, if designed right and made of the right materials, can potentially serve as a superior alternative to existing valve designs [17]. Based on this premise, multiple studies focused on developing polymeric heart valves[4, 15, 39] (PHVs) made out of homogeneous materials or more recently as inhomogeneous textile heart valves (THVs) [1]. In another study we demonstrated that a linear low density polyurethane (LLDPE) PHV enhanced with hyaluronic acid would provide good hemocompatibility for prosthetic heart valves [26]. While the material and its microscopic morphology are important from the standpoint of platelet adhesion, immune response, and the flexibility of the leaflets itself, little is known about the macroscopic design features of tri-leaflet heart valves. For instance what are the fluid dynamic benefits changing the valve profile height or the leaflet shape?

From a fluid mechanics perspective, the efficiency of the prosthetic valve can impact its blood damage characteristics. A valve with a poor performance index would dissipate more energy due to the losses from turbulence in the central jet emanating from the valve orifice[22], which in turn increase hemolysis and platelet activation. Now, the large scale features of the central jet, including its point of re-attachment must be largely dictated by the valve's orifice geometry and the geometry of the aorta distal to the orifice itself. So the question that naturally arises is: what design features of the valve can help maximize energy efficiency? Observations on the structure of the native aortic valve leaflet indicate the presence of an arch at free-edge which may contribute to proper sealing without requiring the whole valve apparatus to extend far into the aortic sinus. These arched leaflet designs are however not utilized in bioprosthetic pericardial HVs. The objective of the present study is to understand the flow-control relevance of leaflet height, as well as the arch in trileaflet heart valve designs on the valve function and efficient transport from the ventricle into the aorta with minimal turbulence losses.

Previous studies have shown polymeric valve prosthesis provide beneficial aerodynamic design [17]; at the same time these studies indicate that design features such as commissural design and the high shear stress along the edge of the central jet of the PHVs are potential causes of blood clot formation [17]. Additionally, no detailed hemodynamic study has been conducted to address flow recirculation regions in PHVs, which have been strongly correlated to hemolysis and platelet activation [2]. Such a study will provide insight into effect of delayed reattachment of the flow and the resulting regions of recirculation, to evaluate the potential of platelets aggregation and platelet activating factors. The aim of the current study is to evaluate the effect of geometric parameters to improve hemodynamic characteristics of PHVs by alleviating reattachment and reducing shear stress. This is a

crucial step toward reducing turbulent losses and flow leakage before designing the next generation of heart valve prostheses.

Materials and Methods

Manufacturing Process

To address the questions regarding the relevance of leaflet arches and its potential benefit towards optimizing polymeric HVs, we designed a modular fixture. This fixture will assist us in evaluating the effect of geometric parameters on the commissure coaptation and fluid dynamics in the valve and ascending aorta. As shown in Fig. 1, six different HV models have been designed and assembled, each of which consists of three sections: the bottom stent part, the top stent part, and the leaflets. The aspect ratio between stent height and diameter, hereby referred to as H/D , was varied for three different valves with ratios listed in Table 1. The leaflets for each valve were made of linear low density polyethylene (LLDPE). A rectangular strip of LLDPE was used as the primary leaflet design. A secondary design involved additional arch shapes for each leaflet. The arch height to diameter aspect ratio, hereby referred to as h/D , was varied (Table 2) to evaluate its effect on coaptation. Alteration of H/D and h/D results in nine different valve configuration, however, the aim of adding arches to leaflets is to improve commissure coaptation, while at the same time longer leaflets may cause warping and delayed leaflet opening. Therefore, we have sufficed to studying the configurations that provide full commissure coaptation. This aim is accomplished by the addition of short arches to the valve with medium aspect ratio, and no arches to the valve with highest aspect ratio. Hence, six out of nine configurations are further studied. Each of the six valves has an inner diameter of 21.5 mm. The two stent parts were 3D printed from acrylonitrile butadiene styrene (ABS) using uPrint SE (Stratasys, Eden Prairie, MN, USA).

The bottom stent part has a cylindrical end, which serves to keep the leaflets in place and three stent posts to form the three cusps of the valve. The rectangular piece of LLDPE is cut and wrapped around the bottom stent part, creating a cylinder. This piece is accurately fixed to the three stent posts to ensure a symmetric geometry in closed position of the leaflets. Fixing is performed by connecting the bottom stent part to an air vacuum and fixing the LLDPE leaflet by matching the three cusps. In order to mimic the native aortic heart valve (AHV), the top stent part is placed on top of the leaflets in a way that creates three cusps resembling the tri-leaflet AHV in the normally closed position. The top and bottom stent parts are secured together using three orthodontic rubber bands. Fig. 1 shows the assembling process for each of these six AHV models LPNA has low profile no arch, LPSA has low profile short arch, LPLA has low profile long arch, MPNA has medium profile no arch, MPSA has medium profile short arch and HPNA has high profile no arch.

Flow Loop Setup

Each valve was inserted into a transparent acrylic aortic chamber machined to mimic the outer walls of aorta. The tube is designed analogous to the chamber used by Leo *et al.* [17]. The chamber was then placed in the aortic position of a left heart simulator (Fig. 2) which was controlled by an in-house LabVIEW program. Parameters of the left heart simulator were adjusted to simulate the physiological flow and pressure conditions of the aortic valve

in the *in vitro* setup, which were measured using ultrasonic flow probes (Transonic Inc., Ithaca, NY), and the pressure upstream and downstream of the valve was measured with Validyne pressure transducers (Validyne Engineering Corp., Northridge, CA). An ensemble average of flow curves was obtained over 20 cycles at average resting condition (Heart Rate (HR) = 60 bpm, Mean Aortic Pressure (MAP) = 100 mmHg, Cardiac Output (CO) = 5 L/min) using the aforementioned LabVIEW program. The working fluid for the flow loop was a mixture of water/glycerin with 38% glycerin concentration to produce a Newtonian blood analog of similar kinematic viscosity and density ($\nu = 3.5 \text{ cSt}$, $\rho = 1080 \text{ kg/m}^3$).

High Speed Imaging

The valves' dynamic models were evaluated using high speed imaging and particle image velocimetry (PIV). Both of these were done using the same aortic chamber, obtaining data from front viewing window and lateral side. High speed videos were captured to analyze the overall performance of the designed PHVs and to evaluate leaflet closure. To do this, an acrylic dog-leg chamber was added to the downstream of aortic chamber and a high-speed CMOS camera (FASTCAM SA3, 60kfps, 1024×1024 px, Photron, Tokyo, Japan) was positioned in front of the dog-leg window. Images were acquired at 1000 fps.

Particle Image Velocimetry

To visualize particle movement during the PIV process, the flow was seeded with PMMA-Rhodamine B seeding particles (microParticles GmbH, Berlin, Germany) with particle sizes ranging from 1 to 20 μm . A laser sheet cut through the center plane of one leaflet and illuminated the particles. The laser sheet was created using a Nd:YLF single-cavity diode-pumped solid-state, high-repetition-rate laser (Photonic Industries, Bohemia, NY) coupled with external spherical and cylindrical lenses and an orange filter. The high-speed camera was placed on the side of aortic chamber to view the laser sheet. Measurements were phase-locked with acquisition of 250 double-frame images at each time point [11]. Commercial PIV software, DaVis (LaVision, Germany), was used for data acquisition and processing. Velocity vectors were calculated using an advanced PIV cross-correlation method with a 50% overlap multi-pass approach with an initial interrogation window of 64×64 pixels which progressively reduced to 8×8 pixels interrogation window passes. No pre-processing was done, but post-processing was performed using a median filter that rejected vectors outside 2 standard deviations of the neighbor vector. The effective spatial resolution was $27 \mu\text{m/pixel}$ and the temporal resolution was 1000 Hz. Particle seeding density was approximately 0.02 particles per pixel [36] and the particle displacement was around 0–10 pixels per frame (average 5 pixels). The number of particles per interrogation window averaged 20.5 which are acceptable for high-quality PIV [20]. Velocity vectors and vorticity contours were ensemble averaged across 5 trials.

Definition of the Calculated Parameters

The following parameters are used to characterize fluid dynamics in the valve and the flow chamber used as model of the ascending aorta:

Regurgitation fraction—In order to make a comparison between regurgitant fractions that occur in each valve, flow diagrams obtained from LabVIEW program assigned to the

flow loop were analyzed. Regurgitation percentage is defined as the ratio between reverse flow to the total amount of flow in the loop [29].

Effective Orifice Area (EOA)—The effective orifice area (EOA) is another parameter to evaluate valve's hemodynamic performance. EOA was calculated using Gorlin equation [13], in which it is a function of both flow rate and transvalvular pressure gradient.

$$EOA = \frac{Q_{rms}}{51.6 \sqrt{\Delta P_{mean}}} \quad (1)$$

Calculations were acquired over 10 cardiac cycles and then averaged to provide the representative values in table 3 [40].

The coordinate system used to calculate different parameters are defined so that X direction is along the center plane of the heart valve and Z- direction is perpendicular to our interrogation window. U and V are velocity components respectively along X and Y directions with u' and v' as the velocity fluctuations in each direction.

Vorticity (ω_z)—Regions of high vorticity indicate high tendency of fluid elements swirling and rotating along one axis. Even though flow field in aorta is three-dimensional, the vorticity along z-direction is dominant. 2D PIV results along X-Y plane provide vorticity field along Z axis which is defined as follows:

$$\omega_z = \left(\frac{\partial V}{\partial x} - \frac{\partial U}{\partial y} \right) \quad (2)$$

Principal Reynolds Shear Stress (RSS)—It is known that high shear stress results in destruction of red blood cells (hemolysis) [12, 18, 22, 25]. One method to evaluate the performance of designed HVs is to calculate RSS to obtain the optimal design for minimal blood damage. RSS is a component of total shear stress calculated from Reynolds decomposition of Navier-Stokes equation. RSS is defined as:

$$RSS = \rho \sqrt{\left(\frac{\overline{u'u'} - \overline{v'v'}}{2} \right)^2 + (\overline{u'v'})^2} \quad (3)$$

Turbulent Kinetic Energy (TKE)—This parameter represents the kinetic energy per unit mass for eddies in a turbulent flow. The physiological effect of high TKE is extended exposure of cells to high shear stress and will result in cell membrane rupture, platelet activation and thrombosis. For a 2D PIV, TKE is expressed as:

$$TKE = \frac{1}{2} (\overline{u'^2} + \overline{v'^2}) \quad (4)$$

Results

Hemodynamics results

Pressure and flow data were acquired for all six models from LabVIEW. Fig. 3(a) shows the pressure and flow curve for the MPSA HV. The flow loop was adjusted to match physiological conditions. The pressure curves for the other four valves other than LPNA have the same general shape. However, LPNA failed to maintain the desired 120/80 mmHg pressures. Flow has been normalized by the maximum flow rate for each valve model and the normalized flow curves for all six valves are presented in Fig. 3(b). The comparison between flow curves obtained for each valve indicates an improvement in commissure coaptation based on the reduction in back flow as profile length increased or when leaflet arches were added. The regurgitation fraction is further quantified for both H/D and h/D aspect ratios. Fig 4(a) and 4(b) show a decrease in regurgitation fraction with the increment in H/D and h/D respectively. The error bars in Fig. 4(a) and 4(b) represent the calculated standard deviation for each value.

Leaflet kinematics

Coaptation kinematics were further studied using high-speed camera images of each valve in the flow loop. Images were acquired during four time points throughout the cardiac cycle: Early Systole (ES), Peak Systole (PS), Late Systole (LS) and Diastole. The first sets of experiments were carried out using the three valve models without arches to evaluate the effect of H/D (Fig. 5). The second sets were for the medium and short profile valves with short and long arch added to their leaflets to study the effect of h/D (Fig. 6). As observed in Fig. 5 and 6 the large gap at the center of commissures in the LPNA valve is diminished as either the stent profile or the arch length increase. This observation indicates the increment in commissural contact as the aspect ratio and arch height increase. The LPLA valve in Fig 6 shows warping of commissures at diastole and small amount of asymmetry at PS. Fig. 5 also indicates that obtaining a symmetrical leaflet opening is easier when the leaflets have no arch added. MPSA HV had an asymmetric leaflet opening with a lag time for one leaflet to open all the way which can result in undesired fluid motion during early systolic phase.

Quantitative Flow Results

High resolution PIV was used to capture turbulent characteristics through the aortic valve and ascending aorta. Ensemble averaged velocity vectors superimposed by vorticity contours are shown at two time points throughout the cardiac cycle for each HV in Fig. 7 and Fig. 8 to show the effect of stent height and leaflet arch respectively. At early systole (ES) the HVs with an arch result in an asymmetric velocity profile skewed toward the top along the centerline of the chamber. No significant difference was observed in the values for velocity for each HV. The maximum velocity during ES was ranged from 1.3 m/s for MPNA to 1.8 m/s for LPLA. The flow streamlines for LPLA and MPSA were curved, showing a downward motion towards lower part of the flow chamber and then upward motion to obtain a straight streamline through the rest of the measurement plane. The central orifice jet in short profile valves initiated perfectly symmetric during ES, however during PS the central jet for LPNA showed an upward motion at the trailing edge of the leaflet and resulted in larger separation region in the lower portion of the chamber. Except for LPSA, the other

HVs produced separation zone only on the lower part of the flow chamber, corresponding to the side of the valve with the stent post, as opposed to a free leaflet edge. No reattachment point was observed for the HPNA HV in the measurement plane; while the rest of the HVs showed reattachment points in 1–2 diameters downstream of the annulus, as observed in Fig. 9. Maximum velocity during PS is assigned to LPNA HV and was measured 3.1 m/s. Maximum peak velocity for the other HVs ranged between 1.8 m/s to 2.2 m/s. The velocities downstream of the trailing edge of the leaflet during diastole were typically less than 0.1 m/s for all the HVs except for LPNA and MPNA which were measured 0.35 m/s and 0.17 respectively. The reason for higher diastolic velocity in these two HVs was the presence of a leakage jet at the center of the commissural region. The velocity for this leakage jet was 0.7 m/s and 0.2 m/s for LPNA and MPNA models respectively.

RSS and TKE distribution

Reynolds shear stress (RSS) is a critical parameter in predicting blood damage and calcification in heart valves [12, 22]. Contours of normalized RSS near HV leaflets and inside the ascending aorta are provided in Fig. 7 and Fig. 8 respectively for varying stent profile and arch length to study the propensity of blood cell damage in each of the six designed HVs. Normalized values were calculated to take the difference in peak velocity for each HV into account. The Highest value for RSS is observed in the LPNA HV during PS, which is 1940 dyne/cm² with normalized RSS value of 0.14 ± 0.008 . Adding leaflet arches as well as increasing aspect ratio decreased RSS one order of magnitude. However, MPNA is showed higher normalized RSS values in comparison to the other four valve, and was measured to be 0.04 ± 0.004 . LPLA valve configuration resulted in the smallest value of normalized RSS, which was 0.016 ± 0.0014 . The measured valued for normalized RSS is presented in table 3. Generally high regions of RSS were observed in the shear layer region between the central orifice jet and at the trailing edge of the leaflets. As shown in Fig. 7 and 8, regions of high RSS values were diminished when arches were added to the leaflets.

The normalized TKE for all six valve models at different time points during the cardiac cycle is demonstrated in the supplementary materials (Fig. S1). Contours of TKE bear a resemblance to RSS plots shown in Fig 7 and 8. LPNA had highest magnitudes of TKE, corresponding to higher levels of kinetic energy for turbulent eddies. Similar to RSS, TKE was also decreased with the addition of leaflet arches and increment in aspect ratio.

Discussion

The results of this study are discussed in the following sequence: (1) the effect of heart valve profile; (2) the effect of leaflet arch; and (3) the combined effects and interactions.

Effect of Heart Valve Profile

Hemodynamics and Kinematics of Leaflets—One major parameter that needs to be taken into account for the AV hemodynamics is to ensure a negligible regurgitation fraction during diastole to minimize the work of the heart and avoid a high shear stress jet that can induce platelet activation and hemolysis [10, 19, 21, 37]. Appropriate sealing of the leaflets is obtained through a proper aspect ratio or a modified leaflet design. Nonetheless it has

been previously studied that the short profile heart valves are of more interest due to the reduced dead space by diminishing the blockage of coronary ostium and aortic sinuses [32]. The dead space created in higher profile HVs supports coagulation and will increase the risk for thromboembolic complications [34, 38].

The high-speed camera snapshots shown in Fig. 5 indicate that the aspect ratio of LPNA and MPNA models was not large enough to cover the central gap, which leads to flow regurgitation throughout diastole. The regurgitant flow in these two valve models can be detected in the diastolic phase of the cardiac cycle (see video S1).

All designed models were able to withstand the physiological flow and pressure condition except for the LPNA model which showed higher pressure drops across the HV during systolic phase. This observation is due to the increment in the peak flow rate to maintain an average flow rate compensating for regurgitation.

Based on the EOA values provided in table 3, we observed that the highest value for EOA was achieved in the MPNA valve. A decrement in EOA was observed in the HPNA valve, which we believe is due to the fact that the larger leaflets take more time to fully open. The EOA for LPNA valve is shown to be relatively high, and we believe this is due to the fact that LPNA valve configuration was not able to withstand normal pressure gradient of 80–120 mmHg.

Velocity and Vorticity Results—The Velocity and vorticity pattern observed in the center orifice jet shows a great dependence to the HV geometry. LPNA and MPNA models created jets with higher systolic peak velocity throughout the ascending aorta. The higher downstream velocity present in these models is to compensate for the regurgitating flow during diastole and keep the average flow rate at the desired value of 5 L/s. No reattachment points were observed for the lower part of the HPNA model in our view plane. The reason for delayed reattachment appears to be the presence of longer stent posts act as a barrier and create a larger turbulent shear zone which delays reattachment (Fig 9(a)). This is an important observation that implies leaflet coaptation does not necessarily correlate to an optimum HV geometry.

Comparing all three valves together shows that the central orifice jet for the MPNA valve models achieved unidirectional flow sooner than the other two models by creating reattachment points in the 1D downstream of the trailing edge of the leaflets.

Effect of RSS and TKE—High values of RSS are strongly correlated to hemolysis and blood damage. The RSS values reduced dramatically as a result of an enhanced leaflet coaptation. The high values of RSS and TKE were more dominantly observed in the LPNA and MPNA valve models, which shows the contribution of the leakage jet in increasing the risk of blood damage. The HPNA model also resulted in larger values of RSS and TKE throughout the ascending aorta. The values measured for RSS in these valve models is well below the presented results in previous studies [17], however they still are above the reported values for RSS threshold ($\sim 400 \text{ dyne/cm}^2$) suggested by previous studies which can cause hemolysis and platelet activation [8, 9, 14].

Effect of Leaflet Arch

Leaflet Kinematics—The results indicate that leaflet arch dramatically improves commissure coaptation. The leakage flow percentage reduced from 150% in the LPNA model to 5.6% in the LPLA configuration. As a result the maximum PS downstream velocity reduced to 58% of its value in LPNA to 1.8 m/s in LPSA valve model. Additionally, maximum RSS value reduced 40% from the LPNA to LPLA model. Increments in leaflet arch length resulted in warping in the leaflet tips during diastole, which can increase the chance of fatigue. Warping was observed in the LPLA and became more noticeable for the MPSA, which also resulted in reduced EOA. An asymmetric leaflet opening and closing seen for MPSA model (Fig 6(h), 6(i)) is associated with the warping of the leaflets in their closed position. Long leaflet arches resulted in one leaflet being pushed underneath the other two which is another reason of the asymmetric opening and closing of the MPSA model.

Additionally, the EOA value for the MPSA is lower than LPSA and LPLA and this can be due to the larger size of leaflets and the lag in the opening time compared to the other two.

Velocity and Vorticity Results—Earlier reattachment points were observed in both top and bottom sections of the ascending aorta for LPLA and MPSA models. We speculate that the flexible arches create a wider opening for the jet of fluid exiting through the HV. As shown in the schematic drawing in Fig 9(b), due to Coanda effect [33] the fluid jet stays attached to the arch surface and this attachment deviates flow from the straight path guiding the flow to a higher point in the sinus area. As shown by Dabiri *et al.* increasing the exit diameter as the starting flow develops results in higher-energy vortex ring structures with peak vorticity located further from the axis of symmetry relative to a static nozzle case [5]. The flexible arches facilitate transfer of impulse to a greater volume of fluid and result in an enhanced turbulent mixing as well as helping the flow to slow down. This reveals the critical role that leaflet arches play in damping velocity fluctuations and initiates sooner reattachment. Hence, reduces the residence time in recirculation zones and decreases the chance of platelet activation and blood damage.

Even though arches provided earlier reattachment to the ascending aorta, the added arch increases leaflet area, requiring more pressure to open the leaflet all the way. The delayed opening of flaps during early systolic pushes the fluid towards the center of the chamber, resulting in vortex rings in the trailing edge of the leaflets (Fig 8 c, e)(See video S2). Viscous Shear Stress (VSS) data presents much lower values for all six valve models in comparison to mechanical heart valves [22], indicating a much lower blood damage potential. Video S3 shows VSS contours for all six valves during one cardiac cycle, which appear to be similar to RSS contours.

Effect of RSS and TKE—The RSS values reduced dramatically, since the reduced regurgitation associated with coaptation resulted in reduced peak flow to maintain an average cardiac output. Additionally the presence of arches at leaflet tips could act as damping mechanism by providing a variable exit diameter for the orifice jet and reduce RSS and TKE values.

Combined Effects and interactions

Comparison of coaptation in MPNA and LPSA indicates commissural contact is more easily accomplished by the addition of leaflet arches. The combination of leaflet arch and increased profile height observed in MPSA design led to better overall hemodynamic performance. This is due to the fact that longer arches increase asymmetric leaflet opening due to warping; therefore, making short arches preferable.

Limitations

Several other parameters can play role in improving the design of PHVs, such as leaflet and stent elasticity. In this study we merely focused on stent height and leaflet arch, however, future studies with the inclusion of flexible stent posts will be beneficial in optimizing trileaflet heart valve geometry. Another major limitation of this work was the use of a rigid, straight, chamber to resemble the aortic sinuses and the ascending aorta. While this assumption is not physiological, we used a rigid chamber analogous to the chamber used by Leo *et al.* and general flow features should be similar for a compliant curved chamber [17]. Additionally, there is a three dimensional flow through the HVs and the ascending aorta, further computational modeling or the use of 3D PIV can help address this issue. Another constraint of this study was that only six heart valve models were studied. Further work with inclusion of a greater number of HV designs can improve our understanding of the effects of HV geometry on the hemodynamics and kinematics of the valves. Lastly, even though 250 ensembles may not be enough to obtain statistical convergence on Reynolds Shear Stress calculations, this number is equal to or larger than what previous PIV studies used to study polymeric [17], tissue [28] and mechanical HVs [6].

Conclusion

In conclusion, in this study a modular fixture was created to characterize the impact of the stent and leaflet geometry on the hemodynamics of polymeric tri-leaflet heart valves. It is demonstrated that leaflet arches and higher HV profile provide several advantages for such as: 1) resulting in a dramatic decrease in the RSS, 2) yielding to better leaflet coaptation, and 3) minimizing regurgitation percentage. However, high stent profile may delay reattachment of flow in the aorta and slightly increases RSS. This increment can potentially escalate hemolysis and blood damage. Leaflet arches result in great enhancement of leaflet kinematics and HV hemodynamics by optimizing the low profile design of the HV. These findings will help optimize future design of polymeric heart valves by reducing the stent height and diminishing the blockage of coronary arteries while providing an optimized level of shear stress in the ascending aorta.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1. Assembling leaflet and stents for manufacturing each heart valve (left), Top view of each six HVs: (a)LPNA, (b) LPSA, (c) LPLA, (d) MPNA, (e) MPSA, (f) HPNA.

Figure 2.
Schematic of left heart flow loop, used to impose physiological pressures and flow rates across the aortic valve.

Figure 3. Flow and pressure curved obtained from LabVIEW data for each HV. (a) Flow and pressure for MPSA valve model. (b) Flow and pressure curves for LPNA valve model.

Figure 4. Regurgitant flow fraction during each cardiac cycle, error bars are standard deviation for 10 cycles. (a) Reverse flow fraction for HVs without arch and varying length. (b) Reverse flow fraction for short profile HVs with varying arch length.

Figure 5.
Snapshots of HV models without arch at four phases during the cardiac cycles.

Figure 6.
Snapshots of HV models with arch at four phases during the cardiac cycles.

Figure 7. Effect of stent height on left: ensemble averaged velocity vectors and vorticity contours; and right: normalized Reynolds Shear Stress contours, at selected time points during the cardiac cycle.

Figure 8. Effect of leaflet arch on left: Ensemble averaged velocity vectors and vorticity contours; and right: Normalized Reynolds Shear Stress contours, at selected time points during the cardiac cycle.

Figure 9. Schematic drawing of the effect of left: stent profile; and right: leaflet arch, on the hemodynamics of the prosthetic aortic heart valve.

Table 1

Values assigned to leaflet height.

Height to Diameter Aspect Ratio			
	Low profile	Medium profile	High profile
H/D	0.6	0.7	0.88

Table 2

Values assigned to leaflet arch length.

Parameters used to design leaflets			
	No arch	Short arch	Long arch
h/D	0	0.081	0.116

Table 3

Calculated values for the Effective Orifice Area and normalized RSS of each valve model.

	LPNA	LPSA	LPLA	MPNA	MPSA	HPNA
EOA	2.57±0.087	1.99±0.007	2.01±0.415	2.74±0.02	1.72±0.031	1.89±0.013
Normalized RSS	0.14±.008	0.023±0.0029	0.016±0.0014	0.040±0.005	0.017±0.0001	0.029±0.002