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Flow Visualization of Three-Dimensionality Inside the 12 cc Penn State Pulsatile Pediatric Ventricular Assist Device

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Abstract

In order to aid the ongoing concern of limited organ availability for pediatric heart transplants, Penn State has continued development of a pulsatile Pediatric Ventricular Assist Device (PVAD). Initial studies of the PVAD observed an increase in thrombus formation due to differences in flow field physics when compared to adult sized devices, which included a higher degree of three-dimensionality. This unique flow field brings into question the use of 2D planar particle image velocimetry (PIV) as a flow visualization technique, however the small size and high curvature of the PVAD make other tools such as stereoscopic PIV impractical. In order to test the reliability of the 2D results, we perform a pseudo-3D PIV study using planes both parallel and normal to the diaphragm employing a mock circulatory loop containing a viscoelastic fluid that mimics 40% hematocrit blood. We find that while the third component of velocity is extremely helpful to a physical understanding of the flow, particularly of the diastolic jet and the development of a desired rotational pattern, the flow data taken parallel to the diaphragm is sufficient to describe the wall shear rates, a critical aspect to the study of thrombosis and design of such pumps.

Keywords

Pediatric ventricular assist device; Thrombosis; Fluid dynamics; Particle image velocimetry

Introduction

A quarter of the over 35,000 babies born in the United States with congenital heart disease will require invasive surgery within the first year of life.¹ Although there have been medical advancements in corrective procedures, most notably transplantation, organ availability is limited. Neonates under one year of age have the lowest organ availability and in 2005, 40% of infants died while awaiting transplant.¹² Therefore, there is a need to aid these patients during the “bridge-to-transplant” period. Some current options for pediatric patients include extracorporeal membrane oxygenation and adult sized ventricular assist devices (VADs),^{5,16} however neither of these fulfills the need for long term circulatory support in neonates and small pediatrics.^{3,6}

To help combat this issue, Penn State began developing a 15 cc pediatric sized VAD in the late 1980s that was based on the successful adult 70 cc Pierce-Donachy device (now the Thoratec®). Modifications from the original Pierce-Donachy design included placing the ports at an angle to the body of the device in order to account for the smaller volume. Because of the limited number of commercially available pediatric sized valves at the time, specially made

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ball-and-cage valves were used. This design was tested clinically using animal models and the initial studies showed thrombi, not found in the adult-sized device, on areas of the blood sac, with fibrin and platelet deposits throughout the sac and with embolization in the kidneys of the animals.⁹ Thrombus formation is a function of the coagulability of blood, the properties of the blood-contacting surface and the flow field.⁹ Because the same animal models and blood-contacting surfaces were used in both the pediatric and adult devices, it was reasonable to assume that the increase in thrombus formation was associated with the flow in the smaller pump.

The fluid mechanics of pulsatile VADs has been studied in depth.¹⁰ Because the mechanical heart valves used in VADs are known to cause hemolysis and platelet activation, the fluid mechanics within the VAD can either promote or reduce the chance of thrombus deposition. Extensive studies of pulsatile VAD flows have shown that reducing the chance of deposition requires a strong flow rotation during diastole, which reduces the blood residence time and limits the areas of stasis within the device.^{4,13}

Bachmann *et al.* performed a study of flow visualization in the 15 cc Penn State PVAD. Using laser Doppler velocimetry (LDV), they found that the PVAD had shear rates along the walls low enough (below 500 s^{-1}) to cause the increase in thrombogenicity. They hypothesized that one reason for the low shear rates were the ball-and-cage valves used. When compared to a bileaflet valve, the ball-and-cage valve had a weaker inlet jet, a poorer penetration to the bottom of the pump and a later start to the diastolic rotation. In addition the small, 6 mm, ball-and-cage valves have a much larger pressure drop than the bileaflet valves, which may have contributed to an increase in blood damage. Bachmann *et al.* observed that the Strouhal number, which together with the Reynolds number provides a measure of geometric similarity, was much higher in the PVAD than the parent device (45.3 vs. 8.3) and they suspected that this large difference may be a factor in the difference between the adult and pediatric device fluid mechanics.²

In 2004, the development of the Penn State PVAD became part of The National Heart, Lung and Blood Institute's (NHLBI) Pediatric Circulatory Support Program. This program awarded five contracts for the purpose of developing a circulatory support system for pediatric patients weighing from 2 to 25 kg.³ As part of this program, Penn State modified the 15 cc scaled Pierce-Donachy device to develop 12 cc and 25 cc devices. Changes included a redesign of the inlet and outlet ports to accommodate larger valves.

Particle image velocimetry (PIV) in fields parallel to the diaphragm of the 12 cc Penn State PVAD have been performed to look at the flow patterns, to consider the effect of beat rate reduction and to contribute to the valve selection process.^{8,21,24} Each of these studies has indicated that the flow inside the PVAD has a higher degree of flow normal to the diaphragm than that found in larger devices. This three-dimensionality occurs more within the PVAD because of its angled ports, higher curvature and less uniform diaphragm motion. To illustrate these design differences, we show the PVAD and a 50 cc adult device in Fig. 1. Using planar PIV to observe the three-dimensional flow in the PVAD, we can potentially miss important flow characteristics that involve out of plane fluid movement. However three-dimensional PIV methods are not practical for the PVAD because of the high calibration errors associated with these methods and the small volume that is being studied.¹⁸ Therefore, the purpose of this study is to observe the flow patterns in fields normal to the diaphragm, and by combining and comparing this information with previous parallel flow data, to assess the reliability of the parallel data and to look at the effects of three-dimensionality on the fluid mechanics, as these may influence the thrombogenicity of the device.

Methods

The PVAD was constructed of optically clear acrylic to allow for flow visualization. For these studies two separate sets of valves were used. One pump contained 17 mm Bjork-Shiley Monostrut (BSM) valves, which have an effective orifice area (EOA) of 1.1 cm^2 , and the other, 16 mm CarboMedics (CM) bi-leaflet valves, with an EOA of 1.06 cm^2 .²⁷ The BSM valves were oriented so that the major orifice opened to the outside wall of the device and the strut was parallel to the diaphragm. The CM valves were oriented so the B-datum line, which separates the two leaflets, was perpendicular to the diaphragm.

To mimic the systemic circulation, the PVAD was placed in a mock circulatory loop first developed by Rosenberg *et al.* The loop was modified slightly for pediatric application by reducing the volume and increasing the resistance.²³ A viscoelastic fluid with the properties of 40% hematocrit pediatric blood was used in the loop. This fluid also matches the refractive index of acrylic, 1.488. The analog was made from Xanthan gum (0.03%), glycerin (16%), sodium iodide (50%), and water (35%) by weight. Long *et al.* describe the fluid in detail.¹⁹ The fluid was seeded with $10 \mu\text{m}$ beads (Potters Industries Inc., Valley Forge, PA) for the PIV measurements. The pump was attached to a pneumatic driver that allowed for variation of beat rate, systolic duration, drive pressure and vacuum. Atrial and aortic pressures were measured using Maxxim Medical pressure transducers (Maxxim Medical, Athens, TX). Signals were converted and displayed using a WaveBook acquisition system and WaveView software (IOtech, Inc., Cleveland, OH).

The flow rates both into and out of the PVAD were measured using ultrasonic flow probes (Transonic Systems, Inc., Millis, MA), in combination with the WaveBook acquisition system. We measured both average and instantaneous flow rates and monitored the flow waveforms. These waveforms are used to ensure complete filling and ejection of the pump, which is necessary clinically to reduce the chance of blood damage by valve cavitation, as shown by Lukic *et al.* and further discussed in Cooper *et al.*^{8,20} A normal operating beat rate of 75 bpm with a systolic duration of 340 ms and an average flow rate of 1.47 L/min was employed.

PIV data was acquired in 12 planes in the device. Three were parallel to the diaphragm at 7, 8.2, and 11 mm from the edge of the ports as shown in Fig. 2a. The first two of these planes allows for a view of the entire flow field, including the area near the ports. The 11 mm plane allows for visualization of the body as deep into the device as possible before the diaphragm blocks the laser light plane for a significant amount of the cycle. The other nine planes were taken normal to the diaphragm as shown and described in Fig. 2b. A 6 mm diameter beam was used as the light source and was produced by dual Nd:YAG lasers, part of the Gemini PIV 15 system (New Wave Research, Inc., Fremont, CA). A -25 mm cylindrical lens coupled with a 25 mm diameter high-energy mirror and a 500 mm spherical lens was used to form the beam into a $200 \mu\text{m}$ thick light sheet.

The images were taken with a one megapixel charge-coupled device (CCD) camera (TSI, Inc., Shoreview, MN) with a Micro-Nikkor 60 mm f/2.8D lens (Nikon Corporation, Tokyo, Japan). The CCD chip size was 1000×1016 pixels. The frame straddling technique used was controlled through a LaserPulse Synchronizer (TSI, Inc., Shoreview, MN) and a personal computer (Dell, Inc., Round Rock, TX). InsightTM 6 software (TSI, Inc., Shoreview, MN) was used to control the laser and camera. The images were captured in time by incorporating a trigger signal from the start of device inflow and a trigger delay box. Two-hundred images were taken at 50 ms increments through the cardiac cycle except for early diastole and late systole where diaphragm interference with the light sheet precluded measurement. The magnification was designed so that the measurement planes included the entire planar flow field, which is desirable for the

interpretation of flow field patterns. Planes parallel and normal to the diaphragm have a resolution of 55 $\mu\text{m}/\text{pixel}$ and a resolution of 45 $\mu\text{m}/\text{pixel}$, respectively.

The 200 images from each time step were masked using an in-house Matlab[®] 7.1 (The MathWorks, Inc., Natick, MA) program that allows the fluid boundary of the image to be located and background noise to be removed from the image. The images were then processed using Insight[™] 3G software (TSI, Inc., Shoreview, MN) which cross-correlates the two images taken at a known time step apart. This time step is defined as the pulse delay and can be changed to accommodate differing magnifications and velocities. Here the pulse delay was set at 200 μs . A deformation grid was used. The starting interrogation region size was 32×32 pixels and the final region was 16×16 pixels. Because of its efficient use of extra information gathered from correlation algorithms that overlap interrogation regions, the Hart Correlation was used for the cross correlation.¹¹ The particle (or fluid) velocities were found by finding the average particle displacement of each region and dividing by the known pulse delay. An average flow map for each time step was found by averaging the 200 image pairs taken.

The PIV derived velocities were then used to find the wall shear rates along the surface of the device. This process includes finding the walls of the device and using a centroid shifting technique to find velocities very near the wall. A full description of this process is given in Hochareon *et al.*¹⁴ A study by Hubbell and McIntire found that wall shear rates above 500 s^{-1} on polyurethane materials, such as that used in the sac of the Penn State pediatric device, is sufficient to prevent thrombus formation.¹⁵ Although the process of thrombus formation is quite complex, previous *in vitro* and *in vivo* comparison in a 50 cc version of a pulsatile VAD have shown 500 s^{-1} to be a reasonable threshold value.¹³ Because of this, our design goals include producing a shear rate $> 500 \text{ s}^{-1}$ on the walls of the PVAD for some part of the cycle.

Results

The results presented include two-dimensional velocity flow maps, where colors indicate magnitude and arrows show direction, and wall shear maps derived from the PIV data. A detailed explanation and an error estimate for the application of planar PIV to three-dimensional flow fields can be found in Cooper *et al.*⁸ As the flow fields parallel to the diaphragm have been discussed previously in Cooper *et al.* and Roszelle *et al.*, the results presented here focus on integrating the normal with the parallel plane data to improve the interpretation of the entire data set.^{8,24} Because of the large amount of data, the results presented are those important to thrombogenicity.

The first area of interest is the inlet jet and its behavior for the BSM valve during diastole as shown in Figs. 3 and 4, which present the three inlet port normal planes and the 8.2 mm parallel plane at 250 and 300 ms, respectively. The inlet jet forms early in diastole with maximum velocities of 1.4, 1.1, and 0.9 m/s in the 3.75, 7.5, and 11.25 mm normal planes, respectively. As highlighted in both Figs. 3 and 4, the jet formation in the 7.5 and 11.25 mm normal planes shows the influence of the leaflet and its resulting wake, as indicated by the reduced velocity magnitudes in the planes near the valve. These planes show larger velocity magnitudes below the occluder and wake as the jet spreads and penetrates into the device. Figure 4 shows how this spreading correlates well to the movement of the inlet jet originally presented in the parallel planes. Notice that the jet penetrates to the bottom of the device and sets up the desired rotational flow within the body, as highlighted in Fig. 4a. The formation of the rotational flow early in diastole is a consequence of the concentration of the inlet jet of the BSM valve through the major orifice and nearest the outer perimeter of the pump as is clear from Figs. 3 and 4b–4d. In comparison, in the 3.75 mm normal plane through which the flow from the major orifice of the valve is concentrated, the magnitude of the jet is much greater and converges as it enters the device. As shown in Figs. 3 and 4, this convergence comes from the movement of the jet

into a volume still contained by the diaphragm. The magnitude of the inlet jet in the parallel plane reaches almost 1.7 m/s at mid-diastole (250 ms) while the maximum in the 3.75 mm plane at the same time is 1.3 m/s. Clearly, the 3.75 mm plane does not cross through the strongest part of the jet.

Figures 5 and 6 highlight the behavior of the CM inlet jet by presenting all three inlet normal planes and the 8.2 mm parallel plane at 250 and 400 ms. In contrast to the BSM device, the CM valve has three distinct jets at mid diastole (250 ms), where each plane contains a jet with maximum velocities near 0.6 m/s, as presented in Fig. 5. As diastole continues the three jets combine near the outer device wall to form a single jet, similar to that for the BSM valve, as first shown by Cooper *et al.*⁸ The magnitude in the 3.25 mm normal plane near the wall increases to 1.5 m/s at late diastole (400 ms, Fig. 6) while the 11.25 mm normal plane flow shows a decrease in magnitude to 0.4 m/s at the top of the plane and to 0.7 m/s at the bottom as presented in Fig. 6. Higher magnitudes lower into the device is an indication of the effect of the wake from the valve leaflet, and is similar to that found in the 11.25 and 7.5 mm inlet normal planes in the BSM device. The jet in the 3.25 mm normal plane at 400 ms also shows two peaks, labeled as jet 1 and jet 2 in Fig. 6. The peak labeled jet 2 represents the movement of the center and inner jets toward the outer wall.

The behavior of the inlet jets of both devices influences the penetration and rotation during diastole. In the parallel flow fields it was noted that the inlet jet of the BSM device penetrated to the bottom of the device sooner than did that of the CM device. This allowed for the rotation to set up more quickly and provide better fluid movement and more adequate wall washing.^{8,24} Wall shear data at the bottom of the device further confirms the earlier penetration by the BSM device. Figure 7 provides the wall shear maps of the two surfaces in the bottom of the 11.25 mm normal plane in both the BSM and CM devices. The x -axis is the position along the surface and the y -axis is the time point from the start of diastole. Shear in the counterclockwise direction is designated as positive, while shear in the clockwise direction is designated as negative. The maps are normalized by the threshold shear rate of 500 s^{-1} , therefore the low shear we want to avoid is between 1 and -1 on the color legend and is shown in green. The shear rate in the BSM device goes above the 500 s^{-1} threshold sooner, in some areas on the walls as soon as 200 ms into the cycle, and lasts throughout diastole and into mid systole (650 ms), the shear rate for the CM device does not exceed the threshold until late diastole, 350 ms. Once found the high shear region, which is similar to that of the BSM pump, lasts until 650 ms into the cardiac cycle.

Another characteristic of interest found in the inlet side was an area of stagnation near the port in the BSM device. This area of stagnation is found only in the 11.25 mm inlet normal plane of the BSM device and is highlighted in Fig. 8. Beginning in late diastole the area contains low velocities, below 0.3 m/s, and becomes an area of stagnation during systole. Because of its location near the diaphragm side of the device, it was compared to flow in the 11 mm parallel planes. As presented in Fig. 8, the corresponding parallel plane shows no signs of this stagnation during the transition from diastole to systole. To further investigate this characteristic wall shear maps of this location in the 11.25 mm normal plane were derived. Figure 9 shows the wall shear along this surface, which shows that overall the wall shear remains between 500 and -500 s^{-1} .

Both the BSM and CM pumps show a reduction in flow velocities on the inlet side of the pump during systole. This is to be expected due to the movement of the fluid toward the outlet valve. The CM device however shows a larger drop in velocity and has larger areas of stagnation during systole on the inlet side than does the BSM pump as shown in Fig. 10. In the 3.75 mm normal plane of the CM device the velocity drops from 1.4 m/s at 450 ms to 0.6 m/s at 500 ms. Because diastole ends at 460 ms, this period shows the transition from diastole to systole. In

comparison, the BSM has a more gradual transition. The velocity magnitude in the 3.75 mm normal plane drops from 1.3 to 0.9 to 0.5 m/s at 400, 450, and 500 ms, respectively. This more gradual transition seems to provide for a more coherent rotational flow in the parallel flow fields, as presented in Fig. 10 for the BSM device when compared to the same plane in the CM device. Recall that each of the plots in Fig. 10 represents an average of 200 snapshots. Therefore, the definition of the center of rotation provides a useful indication of the beat-to-beat variation of the flow field, and this is clearly less well defined for the CM valve than for the BSM valve.

A negative characteristic of both valves during diastole is outlet port regurgitation. For the BSM device, this regurgitation was not observed in the parallel planes data, but is apparent in the 11.25 mm normal plane. The regurgitation from the outlet valve sets up an area of recirculation in the outlet port, however as indicated from Fig. 11, this area of recirculation does not affect the rotational flow in the parallel planes. Areas of slight recirculation like this may in fact be beneficial because they keep the fluid in the port moving and reduce the chances of thrombus deposition.

In contrast the CM device shows much stronger regurgitation in both the normal and parallel planes, which are presented together in Fig. 12. This regurgitation can be found as early as mid-diastole (250 ms) and has velocity magnitudes up to 0.4 m/s in late-diastole (400 and 450 ms). In comparison the velocities in the BSM recirculation do not exceed 0.25 m/s. While the regurgitation in the CM device sets up similar areas of recirculation in the outlet port as that observed in the BSM device, the higher velocities also affect the rotational flow in the parallel body planes as presented in Fig. 12.

One final characteristic of the PVAD outlet not observed in the larger pumps is an area of flow separation upstream of the outlet valve during mid to late systole in the BSM device, as noted in earlier work by Cooper *et al.*⁸ To pinpoint this separation, a higher concentration of PIV data was taken (at planes every 2 mm throughout the port, see Fig. 13). The separation was found in the 9 and 11 mm normal planes, as shown in Fig. 13. This area is of concern as the separation can lead to a larger than normal pressure drop across the valve, which may in turn cause more blood damage.

Discussion

PIV provides an accurate way to view whole flow fields and wall shear rates as a function of cardiac cycle in VADs.^{13,14} In contrast to other measurements tools of similar accuracy, such as LDV, the necessary data may be taken quickly enough to impact the design process. The adult sized devices previously studied with planar PIV show that fluid movement is contained to planes parallel to the diaphragm for all but very early diastole and very late systole, where diaphragm blockage precludes optimal interpretation. In contrast the flow field of the PVAD contains a higher degree of fluid motion normal to the diaphragm, a characteristic that brings into question the ability of planar PIV to accurately capture the fluid mechanics of the smaller device. By taking PIV data in planes both parallel and normal to the diaphragm, we have more completely observed flow characteristics that may affect thrombogenicity. Through this study we were also able to examine the ability of planer PIV to suitably present the physics of the flow field and the correspondence between the two orthogonal sets of data.

The three-dimensional flow of the PVAD was observed in planes normal to the diaphragm throughout the cardiac cycle. A good example of this is the cross flow observed in the body of the device as presented in Fig. 14. The fluid moves normal to the diaphragm during the initial filling of the device, early to mid diastole, and as the rotational flow becomes stronger the cross flow decreases and the flow velocities drop below 0.25 m/s, an indication that the parallel flow

dominates during late diastole and early systole in this part of the device. This behavior is different from what is observed in the adult devices and highlights the differences in fluid mechanics predicted by previous studies.^{2,9}

The greatest concern of this increased three-dimensionality observed in the PVAD is the effect on the overall thrombogenicity of the device. By taking data both parallel and normal to the diaphragm, this study allowed us to discover flow characteristics that may aid or abet thrombus formation. An example of this is the areas of recirculation that appear during late diastole and early systole in the outlet ports of the devices. As shown in Fig. 11, the 11 mm parallel plane of the BSM device, as well as the 7 and 8.2 mm parallel planes not shown, do not indicate that the flow in the outlet port hamper the development of a desired rotational flow parallel with the diaphragm. In contrast, data from the 8.2 mm parallel plane of the CM device shows some interference with the desired rotational flow at the outlet port around the same point in the cycle, as shown in Fig. 7. In this case all three normal planes contained areas of recirculation with velocities almost double of what was observed in the BSM device. This highlights the importance of viewing the PVAD in both parallel and normal planes, as the difference in flow could affect the thrombogenicity of the device. By observing the flow in both parallel and normal planes, we were able to understand the physics of the disruption of rotational flow for the CM valve more completely.

In general, the data set indicates that the flow field physics in both the parallel and normal planes correlate well with each other and that the valve selection results originally based on the parallel plane data by Cooper *et al.* was sound.⁸ This is clearly shown in the behavior of the inlet jet in both devices. Original parallel plane data comparing the BSM and CM devices found that the BSM device set up rotational flow more quickly and had greater wall shear rates in the bottom of the device than did the CM device.⁸ The normal plane data agreed with these findings as the flow maps showed penetration to the bottom of the device 100 ms earlier in the BSM device. The wall shear maps in Fig. 7 also confirm wall shear rates above the 500 s^{-1} threshold earlier in the cycle for the BSM device. Another reason for the BSM selection by Cooper *et al.* was the regurgitation found at the outlet valve by the CM device, as discussed earlier.

While the interpretation of multiple plane data is difficult and other flow visualization methods for direct 3D interpretation are available, they have severe limitations such as complex calibration methods and greatly increased time requirements. Stereoscopic PIV, for example, is highly sensitive to calibration and would be nearly impossible to use currently in the small volume of the PVAD containing high curvature and an air to liquid interface.^{18,22}

Summary

While the combination of the normal and parallel planes allowed for a greater comprehensive view of the flow physics of the PVAD, the conclusions drawn by Cooper *et al.* and Roszelle *et al.* using 2D PIV are sound. Multiple plane data, however, should be acquired when feasible.

The use of PIV measurements to help choose the BSM valve for the PVAD highlight the importance of flow measurements to pump design. With this in mind, studies are currently being performed to look at the orientation of the valve in the device. Previous work has shown that orienting the inlet valve at a 30° angle to the diaphragm can have positive effects on the flow in the larger pumps.¹⁷ In the PVAD, which the smaller volume and angled valves, this orientation may be even more important.

Acknowledgments

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FIGURE 1.

Model drawings of a 50 cc (a, b) adult VAD and the 12 cc (c, d) pediatric VAD. As shown in the side views (b, d) the pediatric size has a greater port angle as well as a more severely rounded body. These elements increase the observed three-dimensionality.

FIGURE 2.

The PIV planes where data was required. All of the distances are in mm. (a) 7, 8.2, and 11 mm planes parallel to the diaphragm of the device. (b) 9 planes taken normal to the diaphragm. Three planes each were taken in the outlet and inlet ports, each 3.75 mm apart. Three planes were also taken 2.5 mm apart in the body of the device.

FIGURE 3.

The BSM 8.2 mm parallel plane (a) and 3.75 (b), 7.5 (c) and 11.25 mm (d) normal planes at 250 ms. Gray areas are disturbances in the visual flow field from either the diaphragm or valve motion or shadowing. The cross-locations of the normal and parallel planes are shown by red lines. As indicated by the red arrow the jet in the 3.25 mm plane shows convergence as it enters the pump. In contrast the 7.5 and 11.25 mm planes show reduced flow at the top of the ports (as indicated by red circles), effects from the wake of the inlet jet. The normal planes also correspond well to the 8.2 mm parallel plane which indicates the highest jet magnitudes along the outer wall and spreading of the jet as it moves into the bottom of the device.

FIGURE 4.

The BSM 8.2 mm parallel plane (a) and 3.75 (b), 7.5 (c) and 11.25 mm (d) normal planes at 300 ms. Gray areas are disturbances in the visual flow field from either the diaphragm or valve motion or shadowing. The cross-locations of the normal and parallel planes are shown by red lines. As in Fig. 3 the highest magnitudes are in the 3.75 mm normal plane and the spreading of the jet continues toward the bottom of the device as highlighted by the red circles. The desired rotational flow in the body of the device has begun as observed in the 8.2 mm parallel plane.

FIGURE 5.

The CM 8.2 mm parallel plane (a) and 3.75 (b), 7.5 (c) and 11.25 mm (d) normal planes at 250 ms. Gray areas are disturbances in the visual flow field from either the diaphragm or valve motion or shadowing. The cross-locations of the normal and parallel planes are shown by red lines. The CM valve produces three inlet jets, which can be observed in the 8.2 mm parallel plane. Each of the normal planes also contains velocities of similar magnitude, a large difference from the BSM device.

FIGURE 6.

The CM 8.2 mm parallel plane (a) and 3.75 (b), 7.5 (c) and 11.25 mm (d) normal planes at 400 ms. Gray areas are disturbances in the visual flow field from either the diaphragm or valve motion or shadowing. The cross-locations of the normal and parallel planes are shown by red lines. As the CM device continues through diastole the three jets found in Fig. 5 combine into a single jet along the outside wall, similar to what is presented from the BSM device in Figs. 3 and 4. This combination is observed in the 3.75 mm plane where the jet from the middle crosses to the outer wall as highlighted by the arrows labeled jet 1 and jet 2.

FIGURE 7.

The wall shear of surfaces 3 and 4 of the 11.25 mm inlet normal plane, which are highlighted on the left, for the BSM device (a, b) and CM device (c, d). The x -axis of the wall shear maps indicates the location along the surface. The y -axis of the map is the time location during the cardiac cycle, where 0 ms is the start of diastole. Positive numbers on the legend indicate shear in the clockwise direction while negative numbers indicate shear in the counter-clockwise direction. The legend has been normalized by the 500 s^{-1} threshold, therefore the low shear area is between 1 and -1 (indicated here by green tones). High shear (above 2000 s^{-1}) is indicated by shades of red or dark blue. The shear rate increases above the 500 s^{-1} threshold in the BSM device around 300 ms but not until 400 ms for the CM device. Both devices show a decrease in shear around 600 ms.

FIGURE 8.

The 11.25 mm normal and 11 mm parallel planes of the BSM device at 400 (a, b), 450 (c, d), 500 (e, f), and 550 ms (g, h). Gray areas are disturbances in the visual flow field from either the diaphragm or valve motion or shadowing. The cross-locations of the normal and parallel planes are shown by red lines. The area of stagnation in the 11.25 mm normal plane, highlighted by red circles, is found at the top of the inlet port. This area does not appear in the 11 mm parallel plane, indicated by the red circles, and does not affect the desired rotational flow in the body of the device. There is a slight regurgitation at 450 ms, indicated by the dashed red circle, which occurs just before the end of diastole and is a characteristic of the inlet valve closing.

FIGURE 9.

The wall shear of surface 7 of the BSM 11.25 inlet normal plane as highlighted on the left hand side. The x -axis of the wall shear maps indicates the location along the surface. The y -axis of the map is the time location during the cardiac cycle, where 0 ms is the start of diastole. Positive numbers on the legend indicate shear in the clockwise direction while negative numbers indicate shear in the counter-clockwise direction. The legend has been normalized by the 500 s^{-1} threshold, therefore the low shear area is between 1 and -1 (indicated by shades of green). Shear between this 1 to -1 threshold is found throughout the surface 7 of the plane, indicating that this could be a possible area for thrombus formation.

FIGURE 10.

The 3.75 mm inlet normal plane and 11 mm parallel plane for the BSM and CM devices at 400, 450, 500, and 550 ms. The cross-locations of the normal and parallel planes are shown by red lines. During this transitional period between diastole and systole both devices show a drop in velocity magnitude in the 3.75 mm plane. This drop is markedly larger in the CM device which leads to a less uniform rotational flow in the 11 mm parallel plane including areas of stagnation, highlighted by red circles.

FIGURE 11.

The 8.2 mm parallel and 11.25 mm outlet normal planes of the BSM device at 350 (a, b), 450 (c, d) and 550 ms (e, f). The cross-locations of the normal and parallel planes are shown by red lines. The 11.25 mm plane contains an area of regurgitation (a) that continues into an area of recirculation (d, e) during systole (highlighted by red circles). The recirculation does appear in the 8.2 mm parallel plane and the desired rotational flow continues unaffected during systole.

FIGURE 12.

The 8.2 mm parallel plane and 11.25, 7.5, and 3.75 mm normal outlet planes of the CM device at 350 (a–d) and 450 ms (e–h). The cross-locations of the normal and parallel planes are shown by red lines. Areas of recirculation appear in all three normal planes and contain higher velocity magnitudes than those presented earlier in the BSM device (Fig. 11). This regurgitation appears in the parallel planes (a, e) and affects the rotational flow as highlighted by the red circles.

FIGURE 13.

PIV planes were taken every 2 mm throughout the entire outlet port of the BSM device and are shown in the upper right corner. There were used to locate an area of separation discovered in previous parallel planes. The separation was found in the 9 and 11 mm planes, which are highlighted with dotted lines. The 8.2 mm parallel plane (a) and 9 and 11 mm outlet normal planes (b, c) at 700 ms. Gray areas are disturbances in the visual flow field from the diaphragm motion or shadowing. The cross-locations of the normal and parallel planes are shown by red lines. The stagnation observed previously in the parallel planes but not found in the initial normal planes was captured in new normal planes and highlighted in red.

FIGURE 14.

The 8.2 mm parallel plane and 2.5, 5.0, and 7.5 mm normal body planes of the BSM devices at 250 (a–d) and 450 ms (e–h). The normal body planes show some cross flow in mid diastole (b–d), which is highlighted with red circles. As the device finishes diastole this cross flow is eliminated and the flow in the normal body planes remains below 0.25 m/s (f–h). This is an indication of the dominate flow in the parallel planes as the desired rotational flow continues.