Simulation of Mechanical Heart Valves

Riju Chatterjee, Om Prakash, Nirmal Kumar, Arun Sharma

Dept. of Mechanical Engineering, Birla Institute of Technology, Mesra, Ranchi

Author Emails
[a)]btech10015.18@bitmesra.ac.in
[b)]omprakash@bitmesra.ac.in
[c)]nirmal@bitmesra.ac.in
[d)]sarun@bitmesra.ac.in

Abstract. Various designs exist for Mechanical Heart Valves, but each of them has significant differences in operation from a natural heart valve. It is unavoidable, therefore, that the blood flow patterns through and near MHVs be to some degree unnatural, with medical implications for the patient. Most notably, MHVs are associated with increased formation of blood clots and higher risk of thrombosis. This paper considers representative examples of a few popular classes of MHV, and presents FEM simulations of the flow through them in vitro. A comparison is made between the simulation results for each type of valve with a focus on factors contributing to clotting - including flow velocity, shear stress and stagnation - as well as on leakage or regurgitation. A simulation-based insight is therefore developed for the relative strengths and weaknesses of each type of valve, and this insight is discussed in the context of common practices in the medical field.

Keywords: Mechanical Heart Valve, CFD, Ansys

1 INTRODUCTION

The human heart, like a reciprocating pump, works by forcing fluid to flow in and out of a chamber by periodically increasing and decreasing the internal volume. It is then ensured, through the use of directional flow valves or one-way check valves, that fluid may only enter the chamber through the inlet and may only leave it through the outlet, thereby setting up a net flow in the desired direction. Clearly, valves are a vital part of this process and the performance of the heart depends largely on the performance of heart valves.

In patients with heart valve disease, one or more heart valves may not open or close properly, leading to disruption of blood flow, inefficiencies in pumping, or other problems. For several decades it has been possible to address such issues through the surgical implantation of Artificial Heart Valves to replace damaged or otherwise functionally deficient natural heart valves. Two broad categories of Artificial Heart Valve have been used. These are Mechanical Heart Valves and Bioprosthetic Heart Valves. Bioprosthetic Heart Valves are the more recent of the two, and are meant to emulate natural heart valves as closely as possible. They are made of biomaterials such as animal tissue, human tissue, or synthetic or lab-grown tissue substitutes. Mechanical heart valves, on the other hand, are made of biocompatible engineering materials such as titanium alloys, graphite, and silicone. They are generally designed to emulate the general function, but not the form, of natural heart valves [5].

One of the major drawbacks of Mechanical Heart Valves compared to Bioprosthetic Heart Valves is an increased chance of formation of blood clots, which can thereafter hinder the proper functioning of the valve or can lead to thromboembolism and possibly stroke. Patients who receive Mechanical Heart Valve Implants are generally required to use anticoagulant medication indefinitely, to reduce the risk of such potentially lethal side-effects. The increase in clotting, itself, is widely attributed to the flow patterns through and near Mechanical Heart Valves being unnatural, due to the valves' material properties and geometry being significantly different from those of natural heart valves. Since Bioprosthetic Heart Valves are made of tissue or tissue-like materials and are similar in shape and function to natural heart valves, the flow patterns are much closer to those in natural heart valves, and thromboembolism is not generally a concern [4].

In spite of Bioprosthetic Heart Valves' superior performance in terms of thromboembolic rates in the absence of blood thinners, they are unable to render Mechanical Heart Valves obsolete simply because the life of a Bioprosthetic Heart Valve is far shorter than that of a Mechanical Heart Valve. Where the life of a Mechanical Heart Valve is generally expected to be comfortably greater than a human lifespan, Bioprosthetic Heart Valves typically only last for 10-15 years, after which a patient would be required a second time to undergo heart valve replacement, along with all the risks associated with open-heart surgery [6]. As such, Bioprosthetic Heart Valves are not recommended for patients under the age of 50, and Mechanical Heart Valves continue to be commonly implanted, and remain important subjects of research [7].

Various designs and models of Mechanical Heart Valve have been implanted in patients over the decades. Broadly, these can be categorized into caged ball valves, single disc valves and two-disc or bi-leaflet valves [5]. Caged ball valves were the first type to be implanted, with early experiments taking place in the 1940s and further development through the 1950s leading to the Starr-Edwards valve, which has been implanted in hundreds of thousands of patients since. While caged ball valves are no longer recommended for implantation, they remain an important subject of research. This is due to the large number of patients alive today with such prostheses, whose condition must continue to be monitored and maintained. Over time, many new types of valve have been conceived, created and implanted. The most prominent of these have been valves with one or two tilting discs, i.e., single-disc and bi-leaflet valves [7].

The relative performance of different types of valves has been investigated in several medical studies. Bloomfield [6], looking at multiple such studies, made recommendations for choice of heart valve replacement based on durability and observed rates of thromboembolism. Gott et al. [7], along with an overview of the history of various heart valves' development and implantation, presented conclusions drawn from their performance and post-operative statistics. Goldstone et al. [13] compared the performance of mechanical heart valves with bioprosthetic heart valves by performing a follow-up study of post-operative mortality rates in people who received prostheses in California. The focus was on 50 to 69-year old patients, an age range that had been found to be of particular interest given the gap here in standard age-based recommendations for choice of heart valve prosthesis. Cannegeiter et al. [4] Investigated the risks associated with thromboembolic and bleeding complications in recipients of mechanical heart valves, focusing on the effect of short interruption of anticoagulant medication.

While medical studies generally form the basis for medical practitioners' decision-making at the individual patient level, looking at such statistical performance data alone is unhelpful in fringe cases and cannot meaningfully inform the design of new heart valves. Physics-based investigation of heart valve performance is therefore required in order to gain a fuller understanding of the various factors at play in deciding how the design of a heart valve is likely to affect its performance over its lifetime. Many publications have looked at the physics of heart valves through CFD simulations or other computational means. Cheng et al. [8] extended previous 2-Dimensional simulations to 3D and compared the results for a bileaflet valve, taking into account the fluid-structure interactions responsible for valve operation. De Tulio et al. [9] performed direct numerical simulation of pulsatile flow, also through a bileaflet valve and including the effects of coupled blood-valve dynamics. Nguyen et al. [12] used a blood analog of a sodium iodide and glycerin solution to experimentally validate their 3D simulations of a 29 mm ATS Medical Open Pivot Heart Valve, considering any non-Newtonian flow characteristics to be negligible. Borazjani et al. [10] performed high-resolution CFD simulations of a bileaflet valve in anatomically accurate aorta geometry. Simon et al. [11] simulated the flow fields in a bileaflet aortic valve, focusing on pulsatile flow in the geometrically complex hinge regions.

One of the most important areas for practical research into the physics of heart valve prostheses has been the investigation of the properties of blood, and in particular the identification of flow characteristics that are likely to contribute to clotting. Such research in recent times has allowed the results of CFD simulations of flow in heart valves to be usefully interpreted to inform or evaluate the design of heart valves. Zakaria et al. [14] reviewed the various numerical methods used to simulate mechanical heart valves with respect to the potential for blood clotting, and described the effect of local flow on clotting. It was pointed out that although high shear stress in the flow has generally been accepted as the most important cause of clotting, likely areas for clotting can also be identified by the vorticity of the flow, and that in addition to shear stress, stagnation has been found to contribute to clotting.

A review of the literature reveals that although many CFD simulations of flow in heart valves have been published, as have comparative studies between different designs of heart valve, it is unusual for simulations to be used as the basis for comparison. There is, therefore, a lack of physics-driven insight into the relative strengths and weaknesses of the various common heart valve designs. This paper presents Ansys simulations of flow through three common types of Mechanical Heart Valve, and contrasts the findings, in order, hopefully, to highlight the peculiarities of the flow patterns in each valve design that may contribute to clotting. Forward flow simulations through the open valves are performed by specifying the flow velocity, for which representative values are assumed, based on the work of Mowat et al. [3]. Backward flow simulations are performed by specifying pressure, for which representative values are taken from the work of Mathur et al. [1]. In all cases, the valve geometry used is based on descriptions from the work of Witowski [5], and the non-Newtonian nature of blood is captured using the power-law viscosity model in Ansys, where the coefficient and the exponent are derived by fitting a power law curve to data from experimental work by Rand et al. [2]. It is found that, in accordance with common medical practices, the caged ball valve performs the worst, and that the bileaflet valve appears to perform the best by most, but not all, metrics.

2 METHOD

This paper presents a total of twelve CFD simulations for three valve designs, performed in Ansys 2021 R1. Simulated conditions include forward flow through valves that are fully open and backward flow through valves that are 90% closed. Visualizations are generated of flow variables accepted to be indicative of clotting tendency, and the performance of the different valve designs are contrasted with reference to each such flow variable considered.

2.1 Valve Designs and Geometry

Although many different designs of Mechanical Heart Valve have been developed by various pharmaceutical companies and have been implanted in patients over the decades, the most popular designs can be categorized into three major classes. These are the caged ball type, the tilting disc or single disc type, and the bileaflet type [7]. One valve design of each of these types has been considered in this paper, based on the valves' reputation and the number of patients to have received implants.

Valve geometries used in the simulations in this paper are based on drawings of the Starr-Edwards caged ball valve, the Medtronic-Hall single disc valve, and the St. Jude Medical bileaflet valve from a review of Artificial Heart Valve designs by Witowski [5]. A representative diameter of 22 mm was used for all the valves. 3D renders of the CAD geometry used in simulations of the caged ball valve, the single disc valve, and the bileaflet valve, both in the open positions and the 90% closed positions, can be seen in Figure 1, Figure 2, and Figure 3.

Figure 1: CAD geometry used in simulations of the caged ball type valve

Figure 2: CAD geometry used in simulations of the single disc type valve

Figure 3: CAD geometry used in simulations of the bileaflet type valve

Each valve geometry seen above was subtracted using a Boolean Difference tool in CAD software from a cylindrical channel of diameter 22 mm to generate a fluid domain for CFD simulation.

2.2 Simulation Parameters

The following representative values of blood pressure and aortic blood velocity were extracted from experimental data published in work by Mathur et al. [1] and Mowat et al. [3]:

- The systolic blood pressure averaged for men and women aged 41 to 50 years old was found to be 128.3 mmHg.
- The average diastolic blood pressure in men and women aged 41 to 50 years old was found to be 81.9 mmHg.
- The aortic blood flow velocity averaged for men and women aged 45 to 54 years old was found to be 1.02 m/s.

The choice of age range from among the available data was based on the fact that Mechanical Heart Valve implants are generally recommended for patients under 50 years old [6].

In forward flow simulations through open valves, the aortic flow velocity was used as a domain inlet velocity boundary condition and the systolic blood pressure was used as a domain outlet pressure boundary condition. In backward flow simulations through 90% closed valves, the diastolic blood pressure was used as a domain inlet boundary condition. In all cases, a no-slip boundary condition was imposed on all domain boundaries except for the inlet and outlet faces.

2.2.1 Non-Newtonian Viscosity of Blood

One of the interesting properties of blood from a fluid dynamic standpoint is its non-Newtonian viscous behaviour. Ansys allows simulation of laminar non-Newtonian flow using a power law viscosity model. In order to obtain the appropriate coefficient and exponent for simulation of blood, a power law was fit to experimental viscosity data measured at 37°C, i.e., body temperature, by Rand et al. [2]. The fit is shown in Figure 4.

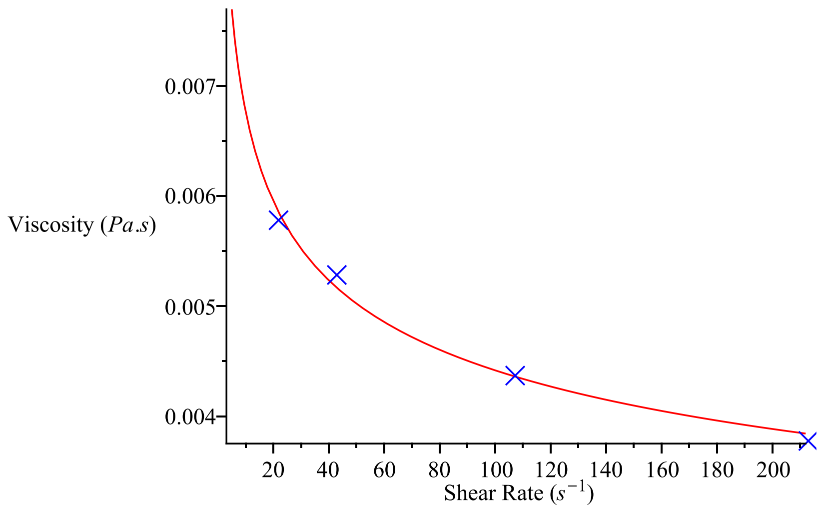

Figure 4: Power law fit to experimental viscosity data for blood at body temperature

The expression

$$\nu = 0.01037\dot{\gamma}^{-0.1854} \tag{1}$$

was obtained, where ν is the viscosity of blood and $\dot{\gamma}$ is the shear rate. The coefficient and exponent from (1) were used in all simulations in this paper.

2.3 Quantities Visualized

It is generally accepted that blood clotting in and near Mechanical Heart Valves is caused primarily by platelet activation due to high shear stress in the flow, and recent research has made it clear that clotting can also occur due to stagnation [4].

Flow velocity is one of the most important quantities to visualize in a heart valve flow simulation, since low flow velocities are indicative of stagnation, and high flow velocities can lead to turbulence and thereby to high shear stresses. Sharp velocity gradients, moreover, are indicative of high laminar shear stress. In backward flow simulations, the velocity fields are expected to be relatively simple, with any flow at all being due to leakage through the small gaps left by incomplete closing of the valve. A volume-rendered visualisation of flow velocity magnitude is therefore generated in backward flow simulations in this paper, affording a complete view of the flow velocities in all parts of the domain at the cost of some level of visual ambiguity in the spatial distribution of rendered velocities. In forward flow simulations, the velocity fields are expected to be more complex, making volume rendered visualisations less useful. Velocity magnitude is therefore displayed along two planes, one normal to the net direction of the flow and one parallel to it.

As with velocity gradients, sharp pressure gradients are a good indicator of likely regions of high shear stress. Pressure is therefore visualised on all domain walls in forward flow simulations in this paper.

Lastly, since shear stresses in laminar flows are often the greatest in boundary layers, wall shear is visualized in forward flow simulations on the entirety of the valve surface itself.

3 RESULTS

Results generated from the simulations described in Section 2 now follow.

3.1 Forward Flow Velocity

Figure 5, Figure 6, and Figure 7 show the simulated magnitudes of the forward flow velocities in the fully open valve positions through the caged ball valve, the single disc valve and the bileaflet valve respectively.

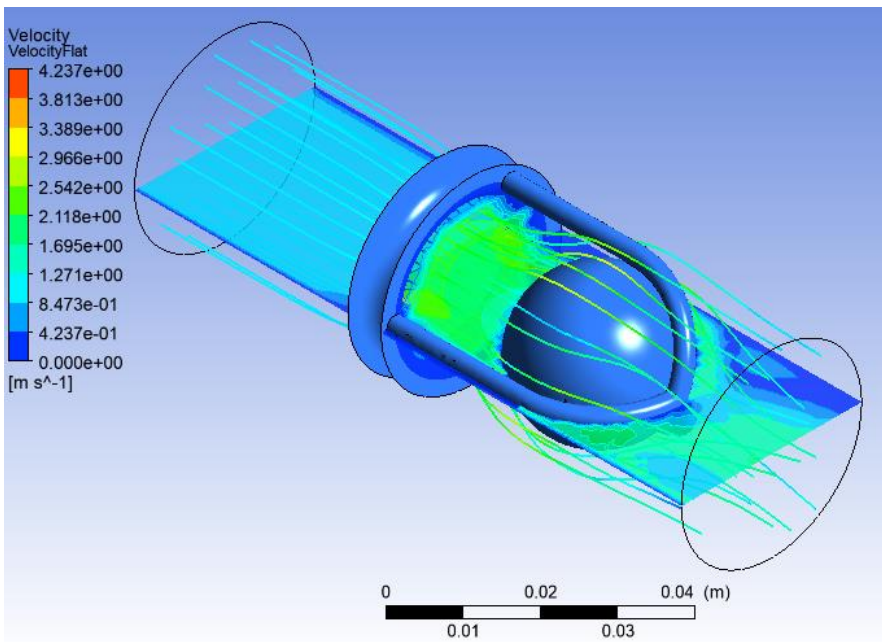

Figure 5: Forward flow velocity through an open caged ball valve

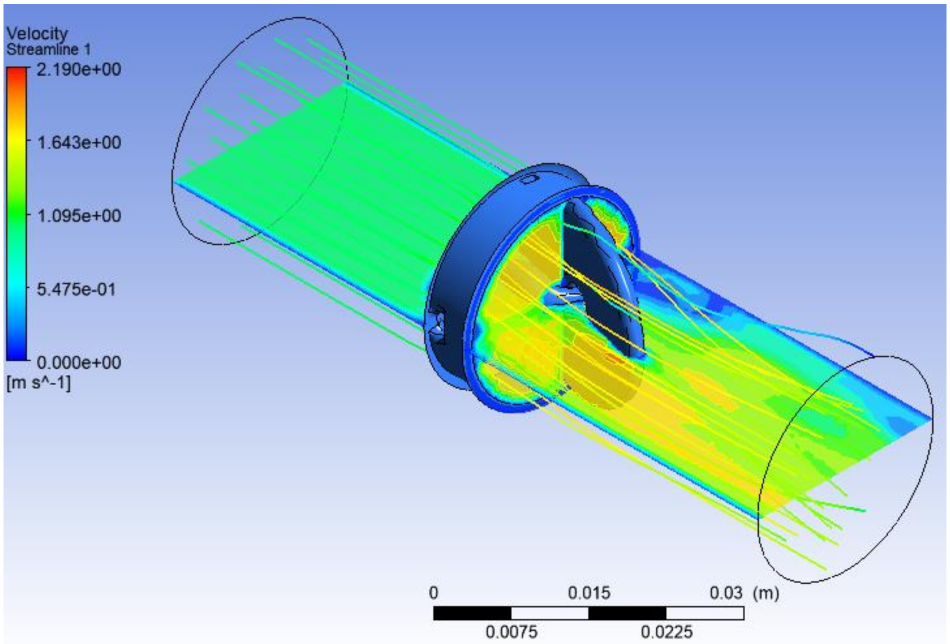

Figure 6: Forward flow velocity through an open single disc valve

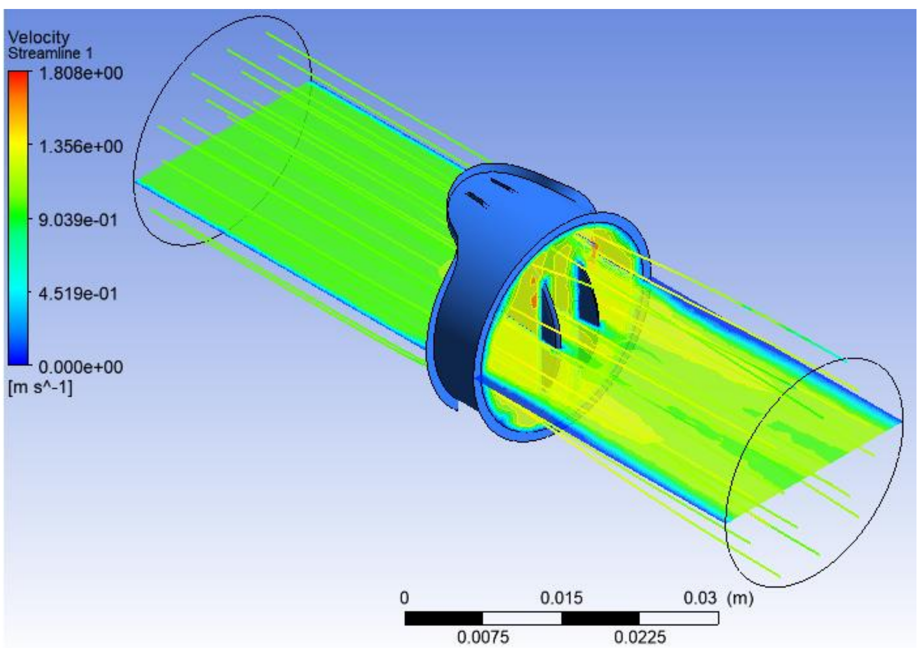

Figure 7: Forward flow velocity through an open bileaflet valve

The widest range of flow velocities is seen in the case of the caged ball valve, and the narrowest range with the bileaflet valve. There are distinct stagnation regions due to flow separation along the smooth trailing surface of the ball in the caged ball valve and behind the strut in the single disc valve. There are no such stagnation regions seen with the bileaflet valve.

3.2 Forward Flow Pressure

Figure 8, Figure 9, and Figure 10 show the simulated pressure on the domain walls in forward flow simulations through the fully open caged ball valve, single disc valve and bileaflet valve respectively.

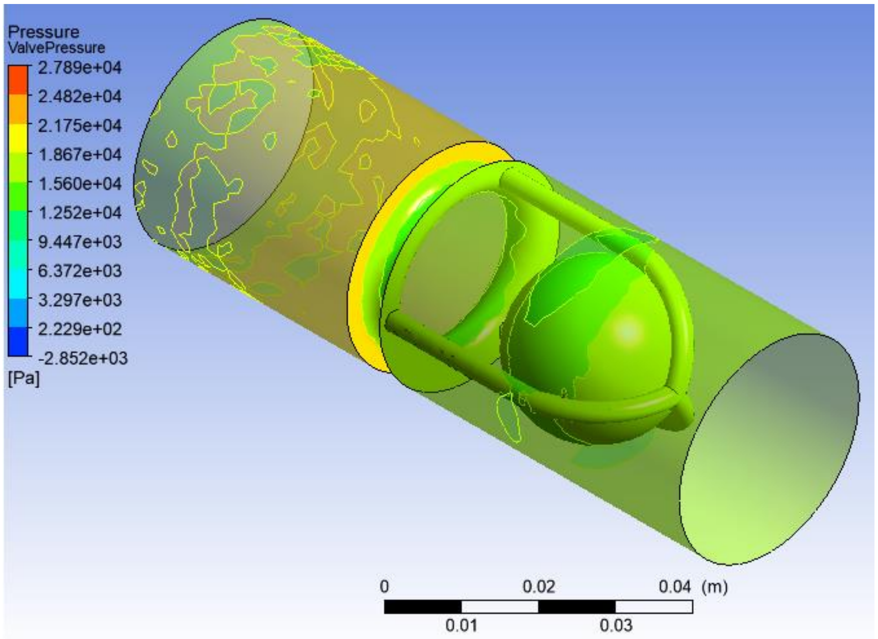

Figure 8: Pressure on domain walls in forward flow simulation through caged ball valve

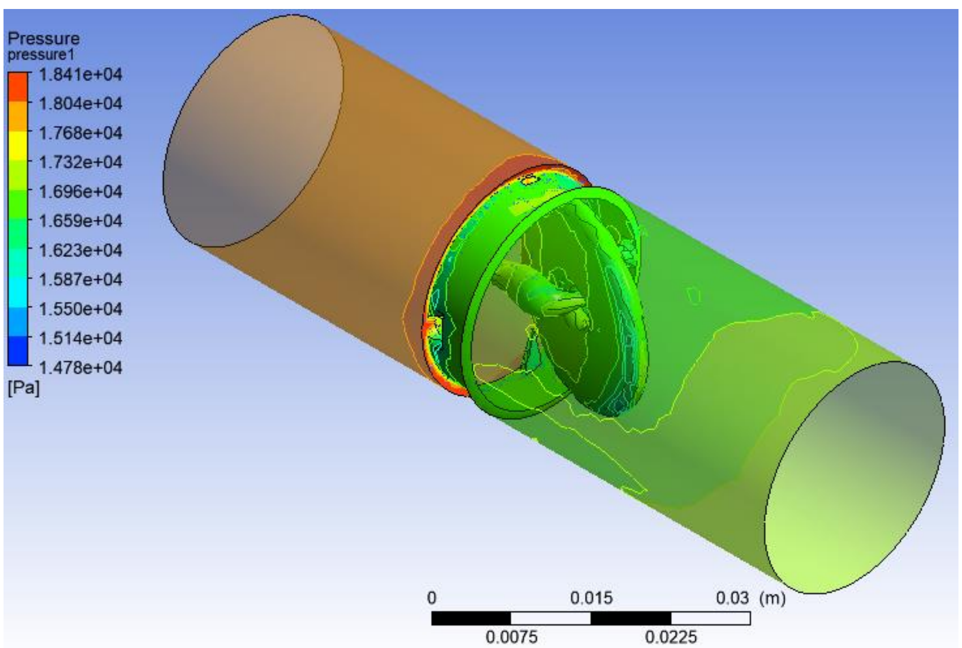

Figure 9: Pressure on domain walls in forward flow simulation through single disc valve

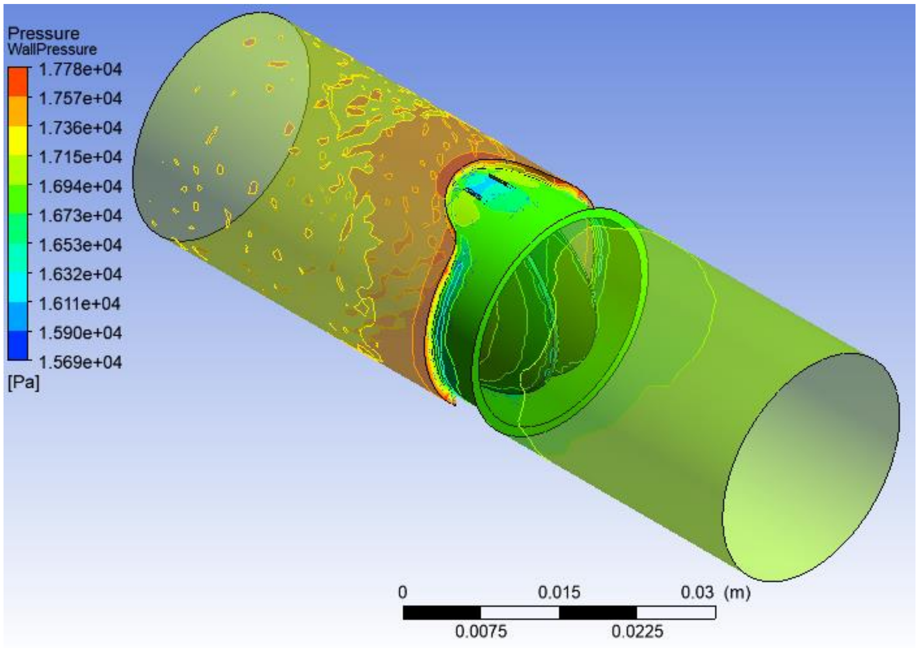

Figure 10: Pressure on domain walls in forward flow simulation through bileaflet valve

As with flow velocity, the widest range of pressures is seen in the case of the caged ball valve, and the lowest in the case of the bileaflet valve. Sharp local pressure gradients are present in all three cases, but it can be seen that the net pressure drop across the valve is smallest in the case of the bileaflet valve.

3.3 Valve Wall Shear

Figure 11, Figure 12, and Figure 13 show the simulated wall shear on the caged ball valve, the single disc valve, and the bileaflet valve respectively.

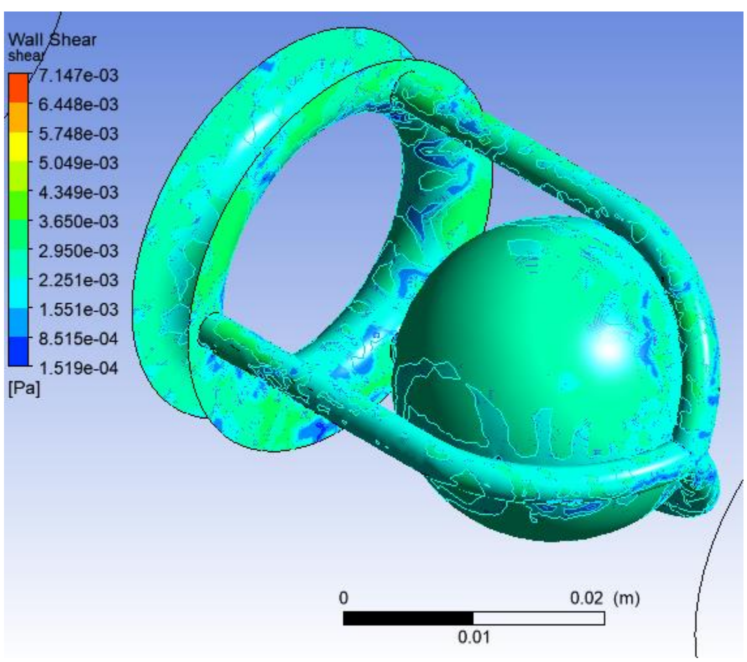

Figure 11: Wall shear on caged ball valve

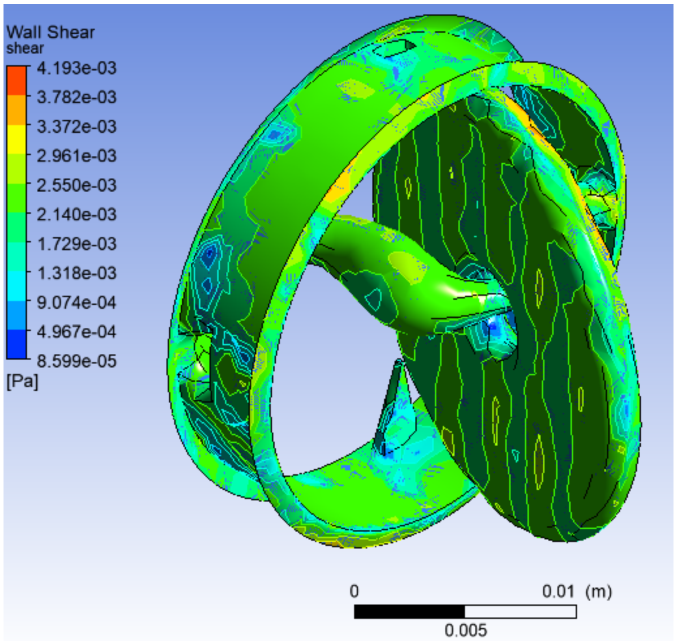

Figure 12: Wall shear on single disc valve

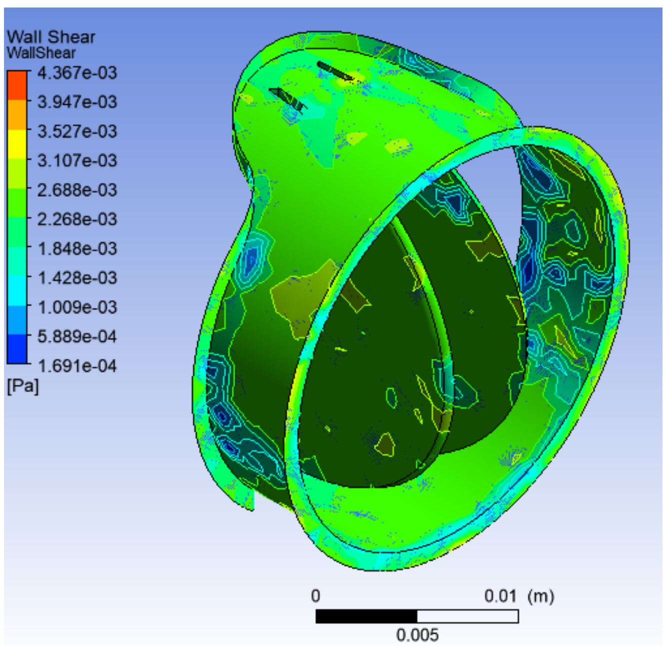

Figure 13: Wall shear on bileaflet valve

The greatest wall shear is seen in the case of the caged ball valve, and the lowest in the case of the single disc valve.

3.4 Backward Flow Velocity

Figure 14, Figure 15 and Figure 16 show the simulated backward flow velocity through the 90% closed caged ball valve, single disc valve, and bileaflet valve respectively.

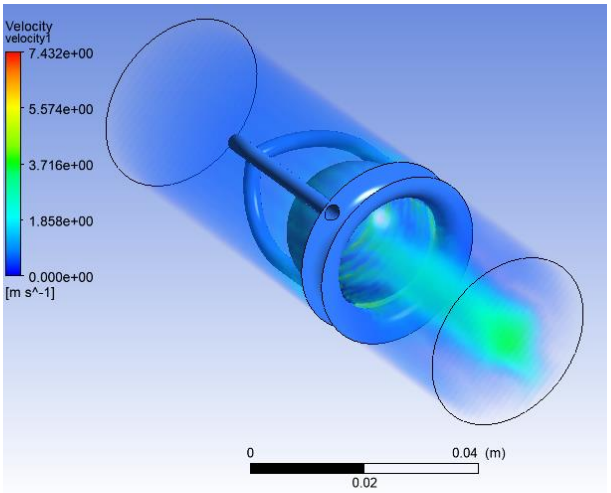

Figure 14: Backward flow velocity through 90% closed caged ball valve

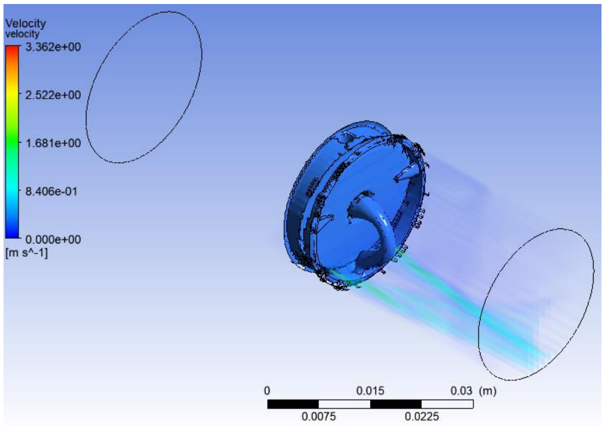

Figure 15: Backward flow velocity through 90% closed single disc valve

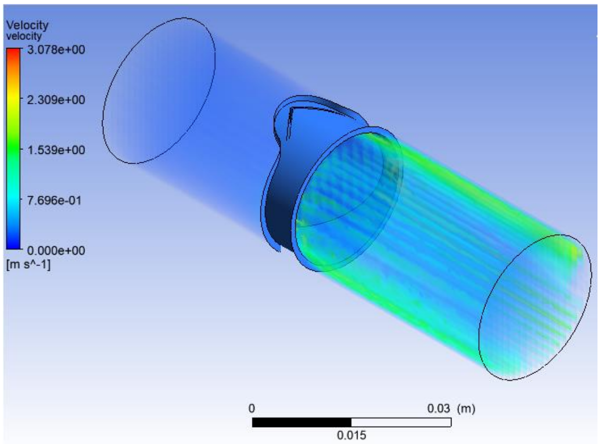

Figure 16: Backward flow velocity through 90% closed bileaflet valve

From superficial inspection, it can be seen that the most leakage occurs through the caged ball valve, and the least through the single disc valve.

4 CONCLUSIONS

This work suggests that among the caged ball valve, the single disc valve, and the bileaflet valve, the caged ball valve is most likely to cause clotting and allows the most backward leakage. The bileaflet valve seems least likely to cause clotting, but allows marginally more leakage than the single disc valve.

These physics-driven findings are in complete alignment with common medical practices, given that caged ball type valves are no longer implanted in patients and bileaflet valves are the most commonly implanted type of Mechanical Heart Valve. It may be noted that, subject to appropriate medical studies, this work might indicate that single disc type valves could potentially be preferable in some patients with reduced cardiovascular function, in case backward leakage flow is of particular concern in such patients.

5 SCOPE FOR FURTHER RESEARCH

This paper has made a comparison between the performance of Mechanical Heart Valve types based on static fluid flow simulations. As has been demonstrated through the agreement of the findings with common medical practices, such simulations can effectively capture much of the individuality of each valve type. However, given that heart valves are designed to regulate blood flow while simultaneously being controlled by the same blood flow, the coupled fluid-structure dynamics involved is of non-negligible significance. Future work that performs simulation-based heart valve comparison while incorporating the effects of fluid-structure coupling will therefore be very effective in further improving physics-based understanding of the relative strengths and weaknesses of Mechanical Heart Valve types.

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