Coupled Fluid-Structure Interaction Simulation of Prosthetic Heart Valves

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Abstract

Artificial heart valves are medical devices that are implanted in patients to replace a diseased native heart valve. They could be classified according to their shape and materials used to manufacture them into mechanical, biological, tissue-engineered and polymeric valves. Approximately 2% of the US population suffer from valvular heart disease (VHD) with the most common causes being aortic stenosis (AS) mostly due to calcification of the aortic valve and aortic valve insufficiency. This paper deals with the numerical simulation of a biological prosthetic aortic valve (AV). This type of valves is composed of three leaflets configured in a complex hemispherical geometry. The leaflets have a variable thickness distribution being thicker at the attachments and free edges and thinner at the belly of the leaflet. Important design parameters for PHVs include effective orifice area, jet velocity, pressure gradient, regurgitation and thrombogenic potential. The objective is to showcase a framework within LS-DYNA® to perform a coupled Fluid Structure Interaction simulation (FSI) of a prosthetic valve and the possible different procedures used to evaluate the design parameters which can be used for a later optimization procedure.

Introduction

Before we begin it is important to clarify that the idea of this paper is not to provide a medical solution in the field of cardiovascular simulation and specifically prosthetic heart valves (PHV) but to give an explanation of how LS-DYNA ICFD can be a tool to support this kind of applications. In particular this paper is aimed at LS-DYNA users that would like to understand the basics of how we are modeling heart valves. Indeed the subject is not new to LS-DYNA with some initial results in mitral heart valve simulation dating back to 2004 [1]. Since then the number of academic developments in the field of cardiovascular simulations and in particular heart valves has grown significantly. Nevertheless the penetration of this work into the biomedical industry has been small with the industry still relying heavily on experimental results. On one side this is due to regulatory constrains since animal testing and clinical trials are a key part of the approval process for medical devices. But on the other side this is because the numerical formulation of the problem is very challenging and the tools available require training to non-numerical experts that need to spend too much time creating meshes for analysis instead of optimizing heart valve prototypes.

In this paper we will provide a short reference of what is available in LS-DYNA to assist in the solution of prosthetic heart valves with a focus in the ICFD solver and the implicit mechanics solver. We hope to inspire users to not be afraid of tackling this kind of applications by providing a guideline of current best practices. It is important to highlight that for now the analysis are aimed at the design cycle of the valve and not the operational cycle. At a later stage in the evolution of this simulation process we will focus on assessment of malfunctioning valves, diagnosis, implantation, etc.

In the first part of the paper we will review the goals and objectives expected as outcomes from a numerical simulation together with the components involved in PHV modeling. In the second part the meshing process will be detailed with a special focus on the needs of the ICFD solver for fluid structure interaction (FSI) coupled simulations. In the third section a simplified problem will be presented and the solution will be explored.

Scope and objectives

It goes with no saying that a successful prosthetic heart valve design should replicate the functionality of a healthy native valve in terms of durability, hemodynamics and resistance to the generation of thrombus. In this work we will focus on the hemodynamics of the valve and how we can replicate in-vitro results numerically. To study the performance of the valve during the design the valves are placed in machines that provide pressure pulses similar to those present in the human heart (see Fig. 1) and through sensors and measurements the engineers asses the capabilities of the PHV. These machines can also take high resolution pictures and slow motion movies for a visual study and measurements of orifice aperture and closure shape. The typical values obtained from in-vitro testing are: ventricular and aortic pressure, aortic flow and mitral pressure drop.

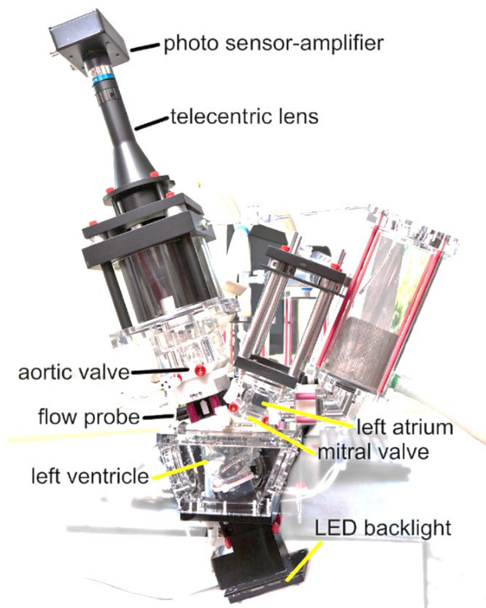


Fig. 1: Sample illustration of pulsatile flow loop with a camera, flow sensor and valve locations [2].

Using these measurements and imaging techniques valvular leakage or regurgitation can be computed together with geometric orifice area (GOA). These measurements and derived values are used in the evaluation of a valve performance. It is the objective of our solution to assist in obtaining these values through simulation.

Valve geometry and pre-processing requirements

A simplified geometry of a PHV can be divided in four parts: the leaflets (cusps), the skirt, the frame that holds everything in place and a pipe or tube that defines the internal domain. In Fig. 2 there are pictures of the type of valves that will be discussed in this work. They are biological valves being either sutured in to the anatomy of the patient to perform an aortic valve replacement or percutaneous which means that they are deployed on top of the native symptomatic valve. In modelling terms the leaflets will be composed of *shell elements* meaning that they will have no thickness during the pre-processing. During the analysis shell elements will be assigned a thickness equal to the real thickness of the leaflet.

Fig. 2: the type of prosthetic valves that will be considered in this study are biological valves. The left valve is used for aortic valve replacement while on the right both valves are percutaneous.

The ICFD solver uses body fitted meshing techniques meaning that all surfaces need to conform to the walls geometry and match exactly. This is very convenient in term of numerical accuracy but it requires some extra work from the pre-processing step. One of the main advantages of body fitted meshes is that in the presence of walls exact boundary conditions can be applied. In the case of valves it means that on the leaflet *non-slip* conditions are satisfied exactly and that pressure discontinuities are allowed across the leaflet even when they are represented geometrically by walls with no thickness. Both are pre-requisites to accurately assess the performance of any valve by providing predictions of pressure drop and flow rates.
We will present two different geometries to illustrate the process. In Fig. 3 there are two valves that use slightly different approaches to define the computational domain. We will refer to them as valve 1 for the left valve and valve 2 for the right valve.

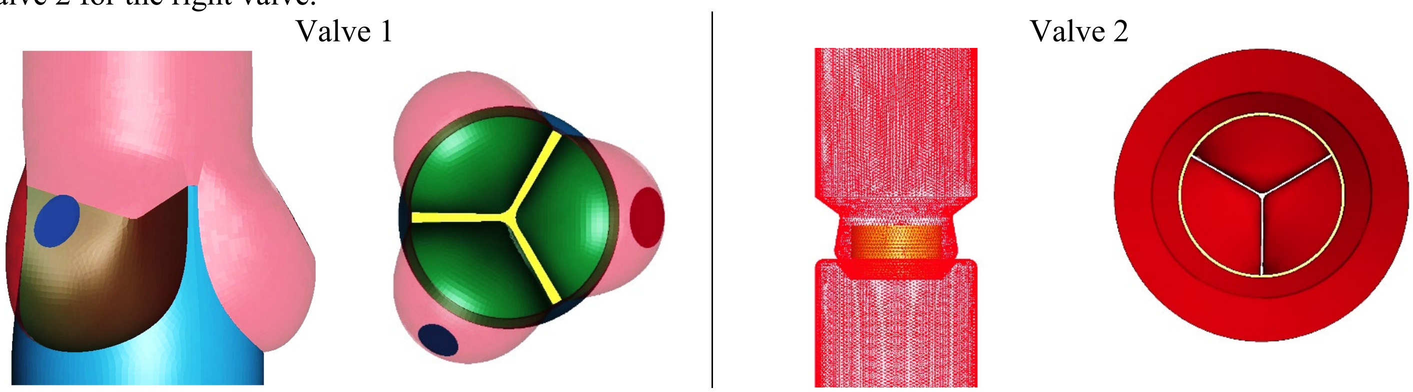


Fig. 3: two different domains for simulation. The one on the left (valve 1) is a short domain with the aortic sinus more similar to the anatomic valve. The one the right (valve 2) shows a long pipe with a valve inserted for testing.

Now let us *dissect* valve 1 into its different components which is shown in Fig. 4. And now let us do the same for valve 2 which is shown in Fig. 5. The main differences for valve 1 and valve 2 are shown in Table 1.

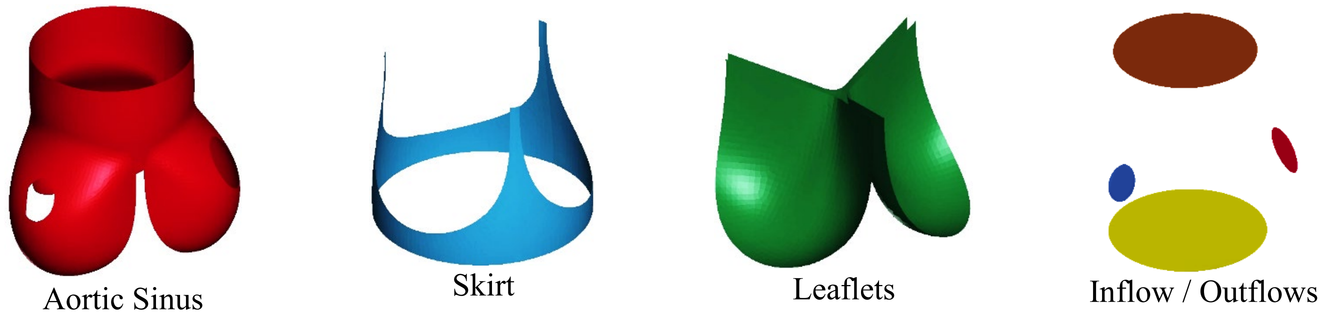


Fig. 4: different components that generate the valve domain for valve 1.

Now that all the components of the valves have been defined, they can be assembled together to define the computational domain. The next step in the process is meshing. It is not the intent of this work to give a detailed explanation about the meshing process since all tool are different. What is important to remember is that the final goal is to perform a FSI analysis and then some parts of the domain will belong to the fluid solution and other parts to the structural solution. One important aspect is that even though some parts the fluid and structural domain coincide they both need to have their own copy for the structural and fluid solver separately. The reason being that the mesh resolution requirements for the fluid and the structural problem may have to be very different in some applications. For instance in aero elastic simulations the fluid mesh has to be very fine to capture all the turbulent effects of the wind while the structural mesh may be much coarser since the displacements are small. This is the methodology for generality. In the particular case of hemodynamics the mesh resolution requirements for the fluid and the structure are very similar and most of the time both meshes coincide but even in this case the solver requires two meshes: one for the fluid and one for the solid problem. In the case of valve 1 every wall is considered flexible so that the skirt, the leaflets and the aortic sinus can deform. In the case of valve 2 only the leaflets are flexible so the structural solver needs a mesh only for the leaflets. For a graphic explanation see Fig. 6. Observe that in the case of valve 1 since the aortic root and sinus are both considered flexible then the structural domain covers the fluid domain except for the boundary conditions. For the case of valve 2 only de leaflets are deformable so the structural solver needs only a mesh at the leaflets. For both problems the structural modeling is done using shell elements exclusively.

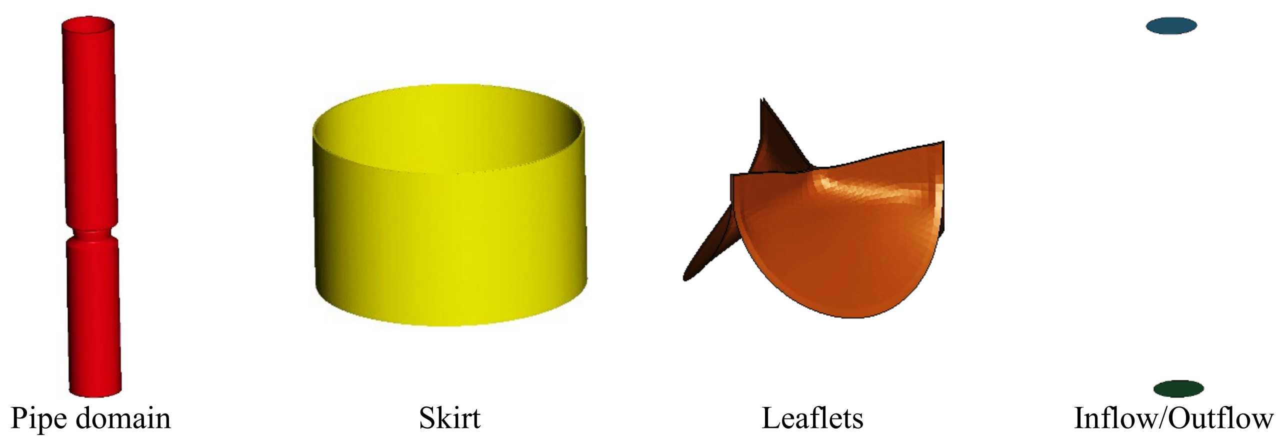


Fig. 5: different components that generate the valve domain for valve 2.

	Domain	Skirt	Leaflets
Valve 1	Composed by aortic sinus + skirt	It is part of the boundary and attached to the leaflets.	Attached to the walls
Valve 2	An external pipe	Internal surface not attached to leaflets	"Floating" inside the domain. Close proximity to skirt to avoid leakage

Table 1: highlight of the differences for valve 1 and valve 2.

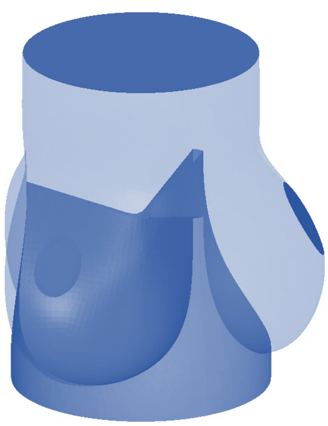
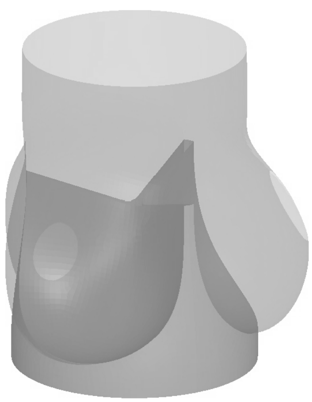
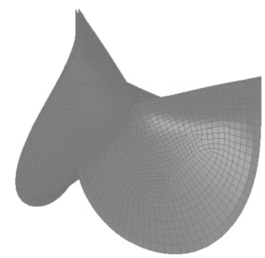
Valve 1		Valve 2	
			
Fluid domain	Structural domain	Fluid Domain	Structural domain

Fig. 6: the fluid and structural domain side by side for valve 1 and valve 2.

There are two more differences between the models that are worth highlighting. In valve 1 the leaflets surface and the skirt / aortic sinus surfaces intersect and the meshes need to match at the intersection while in valve 2 the leaflets and the skirt do not intersect and there is a small gap (see Fig. 7).

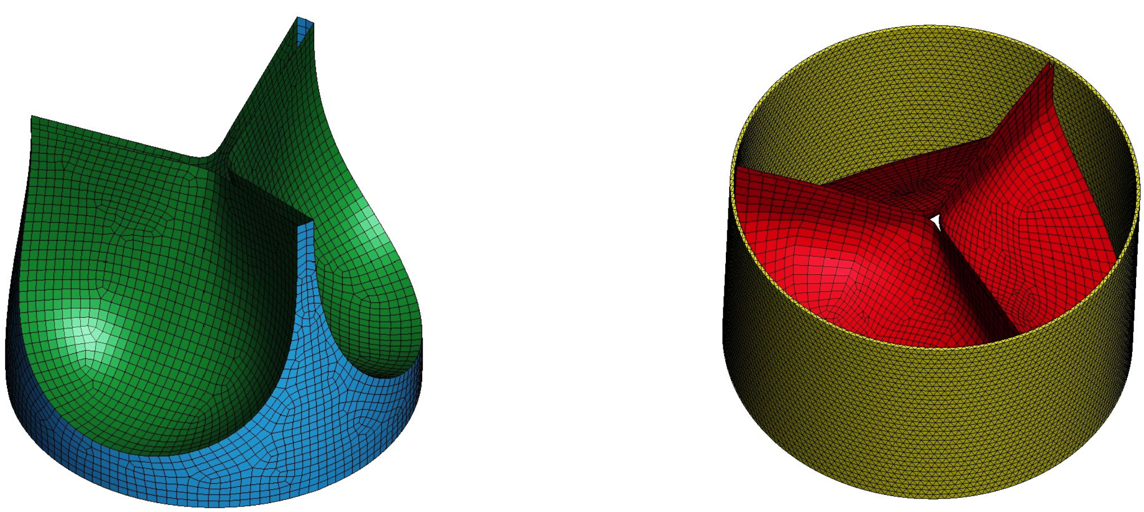


Fig. 7: close view of the mesh in the proximity of the leaflets and walls. Valve 1 intersects the walls while Valve 2 is in close proximity.

In Fig. 3 the top view of the valves is shown. A small initial gap is observed between the leaflets. The initial position of the leaflets has to be such that they do not intersect or overlap. The body fitted meshing strategy does not allow for surface elements to intersect each other. During the analysis the solver will perform several re-meshing steps and the contact algorithm in LS-DYNA will take care of keeping the surfaces from intersecting so that the re-meshing process is successful. This is perhaps a limitation of the body fitted approach when compared to immersed interfaces. Nevertheless the increased accuracy of body fitted finite elements compared to unfitted techniques may counterbalance this limitation.

Results

Let us take valve 1 and using the models created in the previous section show some results and comparisons. The best tool for post-processing this kind of FSI problems is LS-PrePost® (version 4.7 or above is recommended) since both CFD and structural solutions can be explored simultaneously. The boundary conditions for this model prescribes a pressure pulse represented by a pressure differential as shown in Fig. 8. From the image we can see that the maximum aperture orifice of the valve is expected to happen right before $t = 0.2sec$. A sudden drop in pressure will close the valve and it will remain closed from approximately $t = 0.3sec$ until the end of the simulation.

After opening the model one possible action could be to explore the flow structures present in critical parts of the domain. For example one could track vortices in key location areas like at the trailing edge of the leaflets during valve opening or inside the sinus during valve closures (see Fig. 9). These metrics could provide insight into the thrombogenicity of the design.

We can go further now and change some design parameter like the skirt stiffness and evaluate how that affects the solution. After running the model and inspecting the solution we observe that one of the most obvious effects happens at the closure of the leaflets where the shape at the center twists for the flexible case (see Fig. 10) while it remains more “straight” for the stiffer case. The next step could be a comparison between the two designs for the flow rate and leaflet displacement (see Fig. 11). We can see that the case with the flexible skirt shows a lower maximum flow ejection than the rigid skirt. The flow rate for the flexible case presents some oscillations that follow the response of the leaflet structural dynamics showing a strong correlation in terms of frequency of the oscillations. Needless to say that this is a “toy” case to exemplify the capabilities of the solver, a realistic valve will not show this kind of flow variations. More realistic material models, material properties, geometry and surrounding domain are also an important component for a real industrial model that are not considered here nor are the objective of this paper.

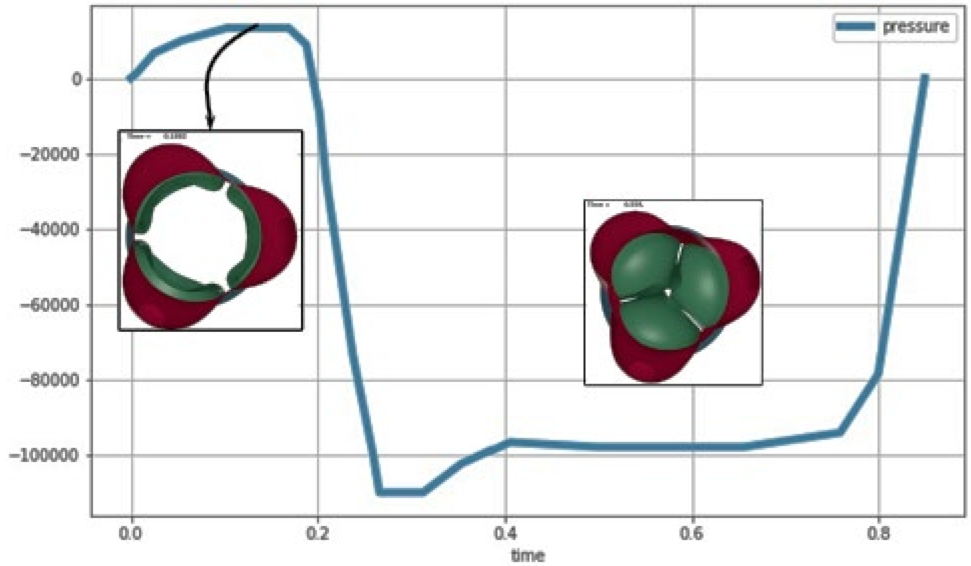


Fig. 8: Pressure differential boundary condition for the FSI simulation in valve 1 and respective valve positions.

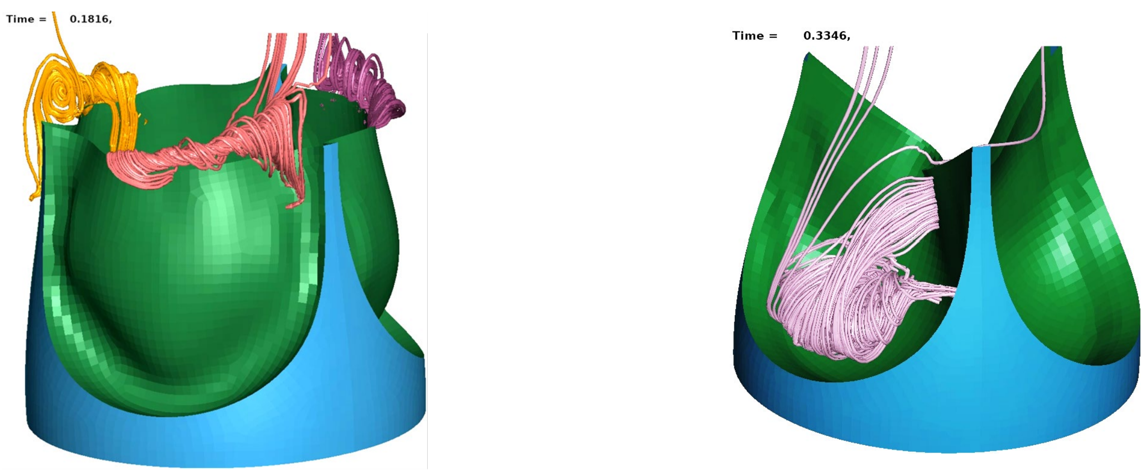


Fig. 9: vortex core visualization to predict areas of recirculation.

Some simple statistics can show the performance of both cases in Table 2. Doing a simplistic analysis solely based on flow rates one could conclude that the more rigid skirt has better properties since it provides a larger maximum ejection, mean ejection and a lowest minimum. A next step should analyze and compare pressure drop between the two valves.

Conclusions

It was shown in this paper the basic principles for setting up a heart model problem using LS-DYNA implicit and ICFD. The specifics in terms of keywords, material models and properties may follow in a different paper. The objective was to present the workflow used for heart valve modeling in our current framework. We mentioned that leaflets are modeled using shell elements and the reasons why we need a body fitted mesh at this point. We also explained that a model needs to have a fluid part and a structural part and that both models need their own domains. For the sake of completeness results on valve model were presented and using LS-PrePost we ventured some conclusions.

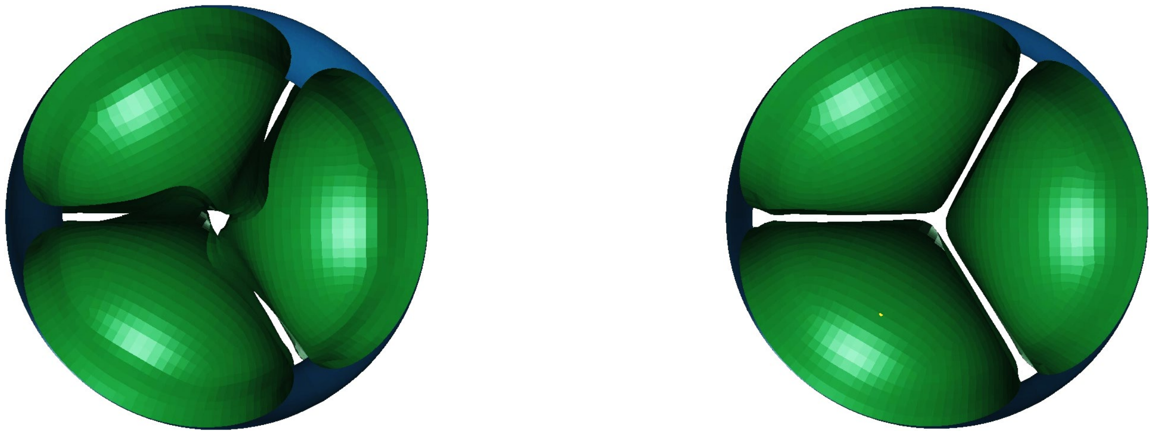


Fig. 10: leaflet closure for a flexible skirt (left) and rigid skirt (right).

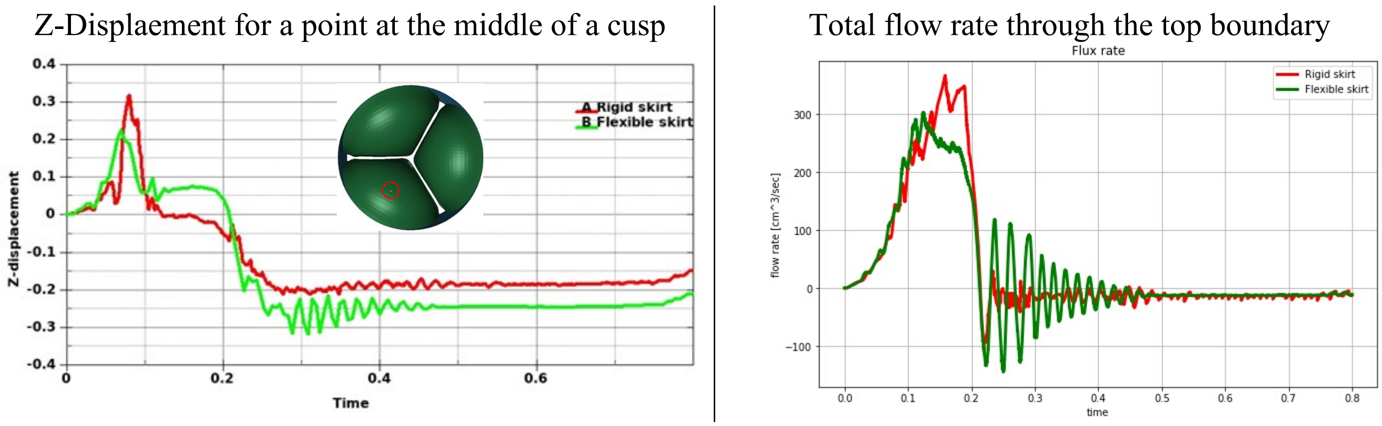


Fig. 11: time dependent plots of a point displacement on the leaflet (left) and total flux through the top boundary (left).

	Flexible skirt	Rigid skirt
Mean flow	32.19	35.25
Min. flow	-144.6	-105.5
Max. flow	302.7	366.6

Table 2: simple statistics comparing the flexible and rigid skirt cases.

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