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PROXIMAL ISOVELOCITY SURFACE FOR DIFFERENT MITRAL VALVE HOLE GEOMETRIES

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Abstract. *The mitral valve can be subject to several pathologies resulting in blood regurgitation from the ventricle to the atrium. The proximal isovelocity surface area (PISA) method is widely used to assess the severity of the regurgitation. However, it relies on several assumptions on both fluid motion and morphology. We designed numerical simulations to assess the influence of breaking those assumptions on the PISA model. A simplified heart geometry was created and three different defects were introduced in the mitral valve. Four heart cycles were simulated using a fluid/structure interaction method. The simulated results were then compared with the estimation provided by the PISA method. In the simulation presented, the mitral defect shapes strongly affects the shape of the isovelocity surface and this results in important estimation error of clinically relevant quantities. Moreover, it was found that the aortic outflow influenced the regurgitation. These observations would justify the use of patient-specific models for clinical evaluations.*

1 Introduction and previous work

The mitral complex is located between the left atrium and the left ventricle. It is composed of the mitral annulus, the mitral leaflets, the chordae tendinae and the papillary muscles [13]. The physiological functionning of the mitral valve is a result of complex biomechanical interactions. Its role is to prevent blood from going back in the atrium during the ventricular systole. Due to various pathologies such as annulus calcification, damaged chordae or ventricle hypertrophy,for example, the mitral valve can behave unproperly thus allowing blood to leak back into the atrium at high velocities (up to several meters per seconds [4]). This behavior is called mitral valve regurgitation (**MR**). When the blood leak is too important, physicians have to choose an appropriate course of action for the patient treatment. Alternatives include for example the replacement of the valve, the placement of a mitral clip, or the repair of the chordae.

To assess the severity of the mitral regurgitation, quantitative measurements have been designed, namely the Effective Regurgitant Orifice Area (**EROA** , the area that is subject to regurgitation on the mitral valve), the Vena Contracta Width/Area (**VCW/VCA**, ie the width/area of the narrowest portion of the regurgitant jet of blood in the atrium) or the Regurgitant Fraction (**RF**, ie the percentage of blood that went to the ventricle and that was regurgitated). Guidelines have been given to physicians to assess the severity of the regurgitation depending on those measurements. For example, a regurgitation with an EROA of more than 0.40 cm^2 is associated with a severe MR [3] and it is recommended to perform a surgical intervention in this case [8].

Depending on the image modality, several techniques can be used to assess MR. One of the most commonly used technique is based on color-Doppler echocardiographic images and is called the proximal isovelocity surface area (**PISA**) method [15]. In particular, it allows the estimation of the EROA and RF based on several hydrodynamic principles and morphological simplifications. Assuming that the mitral valve defect is small, circular and lies on an infinite planar surface, the velocity field of an inviscid fluid forms hemispheric isovelocity concentric shells of increasing value and decreasing area when converging to the hole. The PISA method requires the physician to 1) locate the center of the holes in the mitral valve, 2) select a velocity $|\mathbf{u}_a|$ and 3) find the distance between the center of the hole and the isovelocity surfaces of the selected velocity. Using those parameters, the physician can then calculate the isovelocity surface area ($2\pi r^2$) and the instantaneous regurgitant flow ($2\pi r^2|\mathbf{u}_a|$). The purpose of this work is to evaluate the validity of the PISA model when the assumptions are violated by assuming more realistic scenarios such as a non circular hole or the presence of a neighboring aortic flow.

1.1 Previous work : CFD analysis of the MR

In 1992, [12] used a simple 2D CFD model to assess the validity of the PISA model when the regurgitant hole in the mitral valve is not infinitesimally small, breaking one of the assumptions of the PISA model. They observed that the expected hemispheric isovelocities were flattened in the vicinity of the hole and proposed a correction factor to take that deformation into account.

Yap et al. [17] presented a novel method to quantify mitral regurgitation. Using 3D echography, they proposed an automatic algorithm claimed to reduce the variability of the measures. The authors used a similar simple 2D CFD model to validate the flow velocity scheme assumed by the PISA model. They modeled a regurgitation with a finite sized hole and several valve leaflets configurations. They concluded that the flow velocity scheme of the PISA model was generally valid for the tested configurations although there is a strong error in the orientation of the flow in the vicinity of the hole.

3D CFD of simplified mitral valve geometries was performed by [14] to assess the possibility

of using CFD to help characterize mitral regurgitation and improve treatments. In vitro experiments were performed and a CFD model with equivalent geometry and boundary conditions was used to validate the CFD model. It was shown that realistic three dimensional data could be obtained. They also stressed that the use of CFD allows the investigation of data that are not obtainable using traditional echography.

A similar study was performed by Quaini et al. [11]. On top of validating a 3D CFD model with in vitro experiments, the PISA model was compared with the simulated data. It was shown that the reliability of the PISA measure is highly dependent of the shape of the holes. They also proposed the use of CFD to evaluate the limitations of the ultrasound (**US**) acquisition on the quantitative measures of mitral regurgitation. The same authors went further by introducing a 3D Fluide-Structure interaction (**FSI**) model where the mitral valves were considered as an elastic solid [10]. The model was validated using in vitro experiments. The goal of this last paper was solely focused on the validation of the model.

To summarize the current state of the art, it has been shown that reliable results can be obtained to perform flow analysis in the context of the mitral valve regurgitation using all kind of model complexity. Although PISA method is well known and has been used for decades, it is known that it leads to rough estimations. CFD simulations have helped quantify the drawbacks of the PISA method and provided useful data to build better estimation techniques for realistic holes and mitral valve configuration.

2 Materials and methods

A FSI simulation has been developed to simulate blood flow behavior in patients affected with MR. The heart model used consisted of three cylinders. The ventricle and the atrium were represented by two cylinders sharing the same diameter and were connected along their longitudinal direction. The connection (circular) between them represents the mitral valve. The third cylinder modeled the aorta and was intersecting the ventricle with an angle of 50°, which was similar to clinical measurements [7]. An internal circle in the aorta was placed to mimic the aortic valve. Figure 1 shows the simplified heart model that was used in this study. Diameters and length of each cylinder are presented in the table 1.

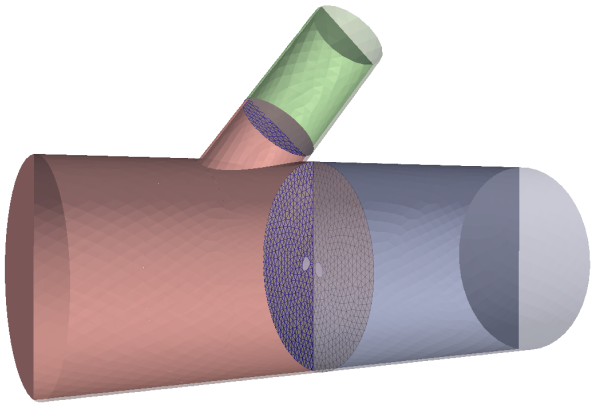

Figure 1: Heart model. The atrium, ventricle and aorta are represented by the blue, red and green cylinder respectively.

To model the impact of the hole shapes on the PISA estimation error, three different mitral valve were modeled (see figure 2). Sim #1 contains a round hole in the middle of the valve, Sim #2 contains an ellipsoid hole. Sim #3 contains a valve defect that was taken from segmented

ultrasound data of a human mitral valve. It contains two holes, which total area was coherent with reported values (≈ 0.3 cm^2). The holes of Sim #1 and Sim #2 holes have been designed so their total area was equal to the defect area of Sim #3 to facilitate the comparison among simulations.

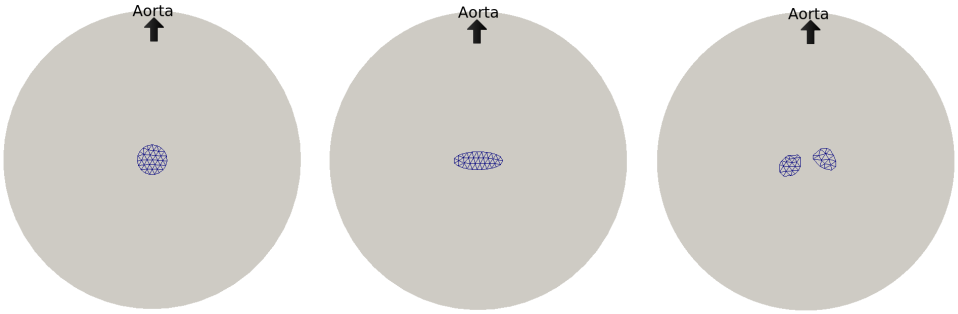

Figure 2: Hole configuration used in the presented work. The rightmost configuration is a mitral defect segmented from ultrasound images.

The heart geometry was then meshed with 3D tetrahedra using Salome (Version 7.7.1 : www.salome-platform.org) with specific refinements in the vicinity of the mitral valve defect. The fluid simulation was performed using the open-source software Openfoam (Version 2.4 : http://www.openfoam.com).

The kinematic viscosity ν was set to 3.5×10^{-6}m^2 s^{-1} [5]. It has been decided to use a Newtonian fluid model since blood can be modeled realistically with constant viscosity if the vessel diameter is greater than 0.5mm [2].

Physiological pressures were imposed as dirichlet boundary conditions for both the atrium and aortic outlet (11.5 mmHg and 100 mmHg respectively [9, 16]). The contraction and the relaxation of the ventricle was simulated by deforming the mesh and this solid boundary displacement was coupled with the CFD solver. To keep the deformation simple, the farmost wall was compressing the ventricle like a piston. The wall velocity was designed so that the ventricular volume evolution was coherent with physiological behaviors. As the amount of contraction was constrained by the aorta positioning, the fraction of blood ejected during systole was below physiological values (40% of the initial ventricular volume).

The aortic and mitral valves were opened and closed concurrently using user-defined condition. The aortic valves opens when the contraction of the ventricle starts and it closes when fluid is coming back into the ventricle. The mitral valve acted in the reverse way.

Simulations were initialized with zero velocity and pressure fields. To alleviate this erroneous initialization, four heart cycles were computed and only the results of the last cycle were considered for further analysis. The isovolumetric contraction and relaxation were not simulated. To ensure the independency of the results with respect to the mesh, four different meshes

Denomination	Reported value
Ventricle cylinder radius	3 cm
Aortic cylinder radius	1.2 cm
Atrium cylinder length	8 cm
End-diastolic ventricle length	8 cm
US-segmented defect area	0.3 cm^2

Table 1: Table of the geometrical dimensions of the model

of increased refinement (120k, 200k, 320k and 1300k elements) were investigated and the isovelocity surface area of each mesh was compared to the one of the most refined mesh. The third mesh was used to compute the results for all the cases presented in the next section as it showed convergence of the quantities of interest and was a good compromise as it was computationally accessible.

2.1 Assumption evaluation

For each simulations, two area measurements were made. First, the PISA estimation of the isovelocity surface area was obtained. A velocity magnitude $|\mathbf{u}_a| = 0.25$ m.s^{-1} was chosen to locate the isovelocity surfaces in all the PISA measurements, which was alined with clinical recommendations [3]. The second measurement was the isovelocity surface area that was provided by the CFD simulations for the same velocity magnitude. The chosen velocity $|\mathbf{u}_a|$ allows the simulated isovelocity surface to be in the region where the mesh was refined, thus allowing a more accurate estimation of this surface area.

3 Results

3.1 Simulation assessment and observations

The CFD simulations were able to reproduce physiological haemodynamics in the location of interests. During systole, blood convergence to the valve defects and blood jets in the atrium were observed. Key caracteristics of the mitral jet, such as the vena contracta, the region where the jet is the narrowest, were observed. As expected, the vena contracta occured just behind the mitral hole due to the fact that the fluid cannot abruptly change direction. The blood velocity at the mitral defect for the three simulations was approximately 3 m.s^{-1} which is comparable to reported clinical measurements [4]. In the case of the round mitral hole, the blood velocity is increasing as it approaches the hole and form hemispheric isovelocity surfaces, resulting in a good correlation between the CFD results and the hemispherical isovelocity surface assumption of the PISA methods. Nevertheless, several discrepancies between the PISA technique and the CFD results were found in the other defect configurations (Sim #2 and Sim #3).

3.2 Comparison of the simulation results with the pisa measurements

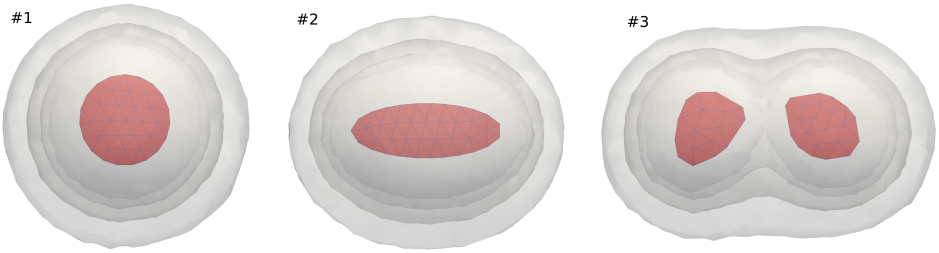

Figure 3: Isovelocity surfaces for the three hole configuration as a top view. Red surfaces represent the holes. The isovelocity area were hemispheric for Sim #1 but not for the two other cases.

Figure 3 depicts proximal isovelocity surfaces for the different hole configurations. Sim #1 exhibits successive hemispheric isovelocity surfaces. The PISA measurements was straightforward as the center of the round hole was not difficult to locate and the radius to perform the PISA measurement was easily picked. A good agreement between the simulations and the assumptions of the PISA method was found (less than 3% difference). For the sim #2, the

isovelocities were no longer hemispherical. As the distance between the center of the ellipsoid and the isovelocity surface are no longer constant, it is not possible to provide a single PISA estimation of the surface. It has been decided to bound the PISA measurement between its maximum and minimum possible values by taking the extremum distances between the center of the ellipsoid and the isovelocity surface. Finally, sim #3 involves two distinct holes in the mitral valve. In this case, the isovelocity surfaces separates in two distinct surfaces when approaching the holes. However, at the chosen isovelocity, a unique isovelocity surface is found for the two holes. As in the ellipse case, this isolvelocity surface is not hemispherical. Moreover, the choice of the center of the defect is harder because there are two holes. To perform the PISA estimation, a point centered between the two hole was taken. Table 2 reports the variation between the estimated PISA and the actual surface area.

Simulation	Surface area (cm^2)		Percentage difference
	Simulated	Pisa method	
#1	2.65	2.58	-3%
#2	2.61	2.18	-16%
		3.45	32%
#3	2.99	2.24	-25%
		4.47	49%

Table 2: Surface area and variation for a velocity $|\mathbf{u}_a| = 0.25$ cm.s^{-1}. The two reported PISA estimations for the sim #2 and #3 are values obtained by selecting the smallest and largest distance to the iso-surface. Percentage difference was calculated as $100 \times (1 - \frac{S_{Pisa}}{S_{Sim}})$

3.3 Aortic outflow influence

During the systole of an healthy heart, the blood is directed to the aorta. However, for a patient with mitral regurgitation, part of this blood is regurgitating back in the atrium through the mitral valve. It has been decided to used the simple round hole case to investigate the influence of the aortic outflow on the haemodynamics near the mitral valve. In particular, the PISA methods assumes that the blood is directed to the center of the hole. To verify this assumptions, the angle deviation between this assumption and the simulation has been computed on the simulated isovelocity surface at $|\mathbf{u}_a|$. As seen on figure 4, the distribution of this angle deviation is not uniform on the surface. At $|\mathbf{u}_a|$, a deviation of approximately 10° is exhibited while the distal part is closer to the PISA assumption. In the PISA method, this assumption is important because it is used to compute the flow through the isovelocity surface. If the velocity is not perpendicular to the surface, the relationship used to compute the flow through the surface doesn't hold anymore. In the specific case of a deviation $\theta = 10^\circ$, the error is less than 2% (the corrected flow was calculated as $2\pi r^2 |\mathbf{u}_a| \cos(\theta)$). Nevertheless, this effect should not put aside. If the defect was closer to the aorta, or if the isovelocity chosen was higher, this angle deviation could rise and thus impacting the PISA estimation even more.

4 Discussions and future work

In the presented cases, a significative difference between the isovelocity surface area of the holes and their respective PISA estimation were found. More precisely, it has been shown that the isovelocity surface area could be under or overestimated using the PISA method depending on the valve defects. Specifically, using an US segmented defect from a real mitral regurgita-

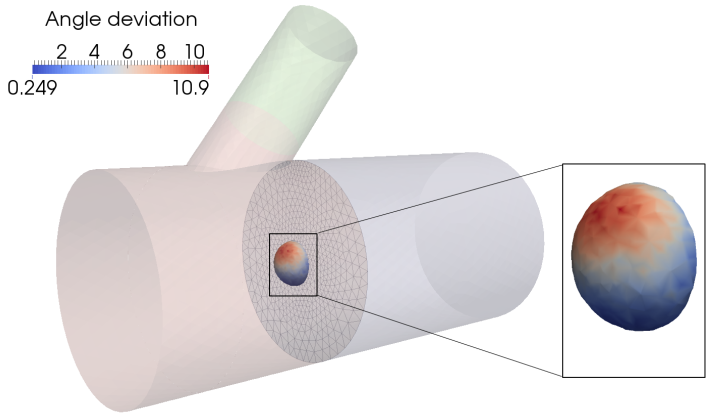

Figure 4: Angle deviation between the flow velocity and the PISA assumption (convergence toward the center of the hole). A deviation of the flow on the side of the aortic output can be observed.

tion, the estimation could have been under/overestimated by 25% to 50% respectively using the traditional PISA method. The variation of estimations of the isovelocity surface area is a result of the difficulty to choose the two parameters of the PISA estimation method. The patient-specific defect, for example, does not have a clear "center" definition. Moreover, choosing a distance between the *so-called* center and the isovelocity surface is not straightforward when the isovelocity surface is not hemispheric. These results corroborate previous observations [11].

In this work, it has been chosen to include a case where the mitral defect was extracted from clinical echocardiographic images. Using a more complex patient-specific model could be beneficial in the future as it could allow a more in-depth exploration of the mitral regurgitation pathology. This is important as a better understanding of the pathology and of haemodynamics can allow us to propose tools that provide more reliable quantitative assessment of the MR severity.

We found that the aortic outflow had an influence on the PISA estimation. Additional investigation needs to be conducted to understand its real influence. For example, it could be possible that the influence of aortic outflow varies depending on the aliasing velocity chosen as it means that the blood is closer or farther to the aortic root. This has not, to our knowledge, been highlighted in previous studies. As this haemodynamic behavior influence the evaluation of the mitral regurgitation, it is of interest to continue research further in this direction.

5 Conclusion

A one-way FSI model has been used to simulate three different mitral valve defects. It was exposed that differences in mitral holes geometries can have a great impact on the estimation of clinically relevant quantities (e.g. isovelocity surfaces) using traditional estimation methods. In particular, the hypothesis of hemispheric isovelocity surfaces close to the holes, which is the basis for the PISA methods, was shown to not hold in every cases. Moreover, thanks to the inclusion of an aortic output in the simulation, it was found that the aortic flow can also influence the estimations. The presented work is an example of the use of numerical techniques to improve the current clinical evaluation tools by improving our understanding of the haemodynamics and of the pathology.

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