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FSI SIMULATION OF BIOPROSTHETIC AORTIC VALVES: COMPARISON BETWEEN IB AND ALE FOR THE MESH REPRESENTATION

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SUMMARY

To the best of our knowledge, all the reported 3D heart valve FSI simulations are based on the IB technique, allowing for large deformations of the structures without compromising the quality of the fluid domain [1-4]. Nevertheless, it is well-known that the ALE-FSI approach guarantees more accurate results at the interface. We present a feasibility investigation by comparing FSI-ALE and FSI-IB simulations for a 2D and 3D aortic valve model. If in the 2D case the differences are unsubstantial, in our experience the performance of a full-3D ALE-FSI simulation is significantly limited by technical problems related to the ALE formulation.

Key words: *heart valve simulations, IB-FSI simulation, ALE-FSI simulation*

1 INTRODUCTION

When setting up a FSI simulation, several choices have to be made to select the most suitable approach for the case of interest. To simulate flexible leaflet cardiac valves, the most important decision is the type of discretization of the fluid domain, which can be described with an ALE (arbitrary lagrangian-eulerian) or an IB (immersed boundary) formulation. In the immersed boundary approach (IB), introduced by Peskin in 1972 [5], the fluid domain is discretized with an eulerian grid, while the structure is modeled with a lagrangian mesh, free to move through the fluid domain. The effect of the presence of the solid bodies immersed in the fixed grid is taken into account by the introduction of an external body force term in the Navier-Stokes equation: it accounts for the effect that the structure has on the underlying fluid by means of an interpolating function, therefore no real interface exists in the fluid domain. On the contrary, if the ALE approach is used, the fluid grid is built as the negative of the solid, and it is allowed to move accordingly. Imposing the no-slip condition at the interface guarantees the equivalence of the fluid and solid grid velocity at the interface. Within each iteration, the grid velocity is extended to the entire fluid domain with extension functions or by solving a system of equations resulting from e.g. a spring model or a pseudo-elasticity model. If the movement of the structure is large, in an ALE-FSI simulation remeshing of the fluid domain is usually needed.

The main advantage of the IB over the ALE-FSI approach is that only the structural grid deforms, the calculation of the variables is less expensive and there are no issues related to a highly deformed fluid grid. Thanks to the fixed fluid grid, this approach is commonly used to solve problems where the structural domain undergoes large displacements, as in case of heart valve dynamics studies [6]. On the other hand, this technique results in a less accurate description of the fluid-structure interface, which is normally the area of interest in the cardiovascular field. Quantities such as the wall shear stress (WSS) and pressure on the leaflets can be an indicator for the occurrence of pathologies e.g. the calcification of the valve [3]. Also, a very refined fluid mesh in the area where the movement is expected is required. These are some of the reasons why, theoretically, it would be preferable to perform heart valve simulations using the ALE technique [7]: in this case, in fact, the interface is sharply defined, and the variables are actually calculated on the surface and not obtained from interpolation, as it happens in the IB instead. In this case, the large deformation in the fluid domain introduces the need for remeshing, increasing the computational time of the solution.

2 METHODOLOGY

To perform all the described IB-FSI simulations, the module Abaqus/CEL of the commercial software Abaqus (Dassault system) has been used. It is an extended version of Abaqus/Explicit, which includes the capability of combining the structural (lagrangian) calculation with the eulerian calculation of a fluid material behavior. Abaqus/CEL results in an explicit, monolithic FSI algorithm, with the eulerian (i.e. IB) description of the fluid domain. To perform the ALE-FSI simulations the in house-written algorithm Tango has been used [7]. It allows for a strongly coupled, partitioned FSI simulation, by coupling any fluid (Fluent Ansys, in this work) and structural (Abaqus/Standard) solver. The interaction between the two is ensured by the coupling algorithm, which iterates the solution of the two segregated solvers until convergence is reached. By using Fluent Ansys as a CFD solver, the fluid mesh can be described with the ALE approach.
In this work, we compare the results obtained for a IB-FSI and ALE-FSI simulation on a 2D and 3D aortic valve model. A Carpentier-Edwards PERIMOUNT Aortic Heart Valve (Edwards Lifesciences LLC, Irvine, California) was scanned with a μCT scan. The images were segmented with the commercial software Mimics (Materialise, Leuven, Belgium) to obtain the desired geometry. To realize the 3D geometry (Figure 1.b), the reconstructed valve was placed into a straight rigid tube with three hemispherical enlargements to mimic the sinuses of Valsalva [8]. In the 2D case a section of this geometry is considered, including two symmetric leaflets placed in a straight rigid tube with two enlargements (Figure 1.a). The dimensions of the domains are consistent in the IB and FSI models, and have been chosen according to literature data.

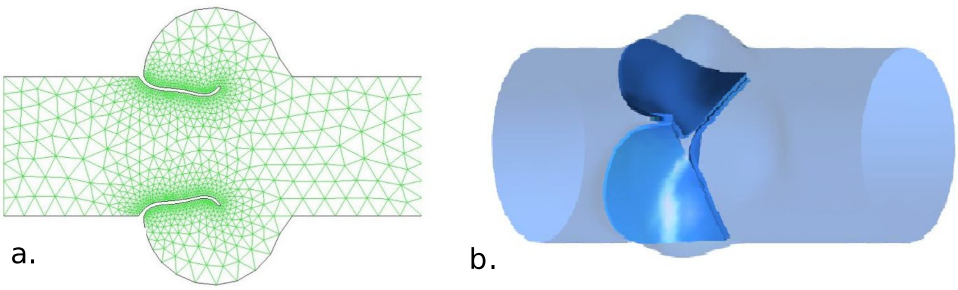

Figure 1: computational domain; 2D model (1.a), 3D model (1.b)

The solid and the fluid meshes have been realized, the dimensions are listed in table 1.

		SOLID	FLUID
2D	ALE-FSI	492	2418 (initial)
	IB-FSI	328	3506
3D	ALE-FSI	5184	150000 (initial)
	IB-FSI	1248	1147392

Table 1: dimension of the computational meshes.

The leaflets tissue is assumed to be linear elastic (Young modulus 1 MPa, poisson ratio 0.45) , and the blood is modelled as an incompressible newtonian fluid (density 1060 Kg/m^3, viscosity 0.003Pas). A negligible compressibility factor is added to the fluid to enhance the convergence of the solution, in particular for the IB-FSI [3]. In all the simulations, two physiological pressure curves have been applied on the inlet and outlet surfaces (ventricular and aortic surface, respectively). Before the loading cycle begins, in all the simulations a preconditioning cycle has been performed, to ensure the removal of initialization artefacts, to provide an initial developed inlet flow, and to limit the influence of the minimal compressibility of the flow. The contact between the leaflets is managed differently between the ALE and the IB simulations, due to the intrinsic differences of the solving codes. The solid-solid contact in the IB-FSI is managed within the Abaqus/CEL software, which includes the interaction in the calculation. The contact in the

ALE-FSI is not available automatically, therefore on the fluid software a condition on the displacement has to be imposed: when coming into contact, the motion of the valve is hampered and a two-layers-cell gap is preserved between the leaflets, with an arbitrarily high artificial porosity value imposed in this gap to limit the backflow [9]. This method replicates the contact between the structures without splitting the fluid domain, which would cause the failure of the ALE simulation.

3 RESULTS AND CONCLUSIONS

3.1 2D comparison

Despite the differences between the two approaches, the comparison of the IB-FSI and ALE-FSI results is satisfactory in the two dimensional case. In the figure below, the flow field velocity (2.a) in three significant time-points, as well as the displacement of the extremity of the leaflet (2.b) are reported.

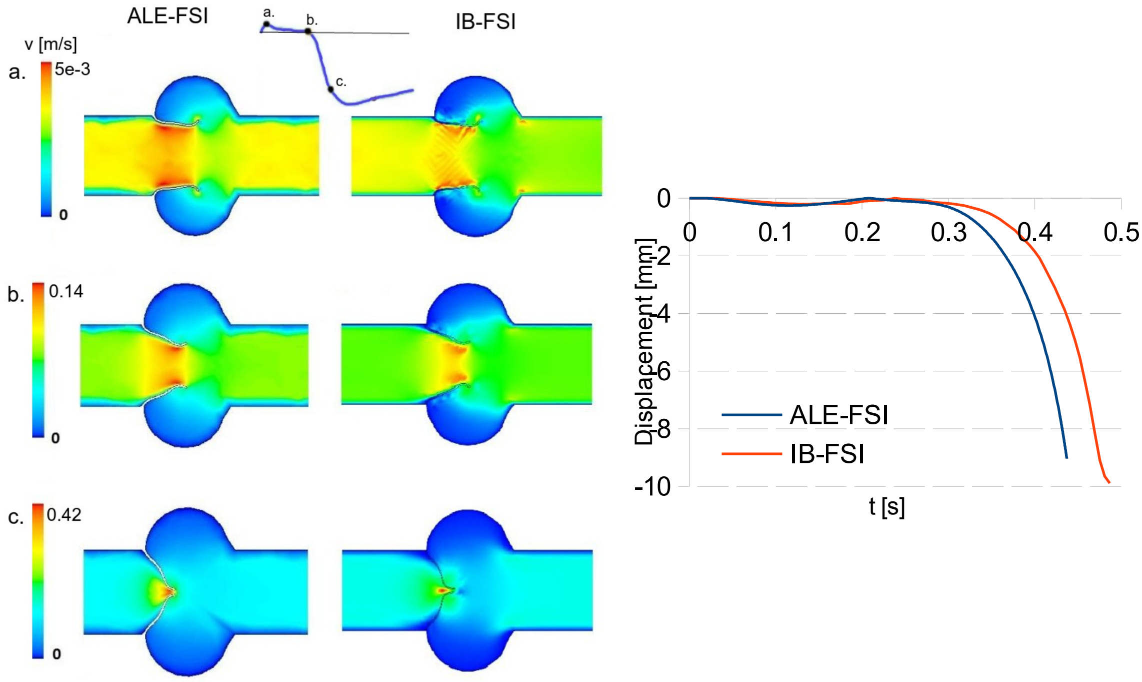

Figure 2: Velocity field (left panel) during systole (a), closing phase (b), and closed position (c). In the right panel the displacement of the distal portion of the leaflet.

Although a small time delay is noticeable in the results of the IB simulation (Figure 2, right panel), the obtained kinematics of the leaflets and velocity in the domain are overall comparable among the 2 models and in agreement with the available literature studies, both based on the IB-FSI [10] and on the ALE technique [11]. The 2D configuration, due to its features, cannot bear the ΔP during diastole, and the leaflets reverse into the ventricle in both cases (results not shown).

3.2 3D comparison

Due to the high complexity of the simulated design and the great computational challenge of the ALE-FSI simulation (i.e. the large deformation of the fluid grid, the need of extensive remeshing, and the computational restriction on the allowed time-step size), the 3D ALE-FSI simulation failed and no full comparison is available for the 3D case. Considering the closed position as the initial configuration for this model, the ALE-FSI simulation stops in the early opening phase, as shown in figure 3.a, due to the generation of negative volume cells in the fluid domain. On the contrary, the FSI-IB simulation reaches the open configuration (figure 3.c). The velocity and the leaflets position in the early opening phase are comparable between the two different approaches (figure 3.a, 3.b). In figure 3.c the velocity is reported for the fully open configuration of the IB-FSI valve.

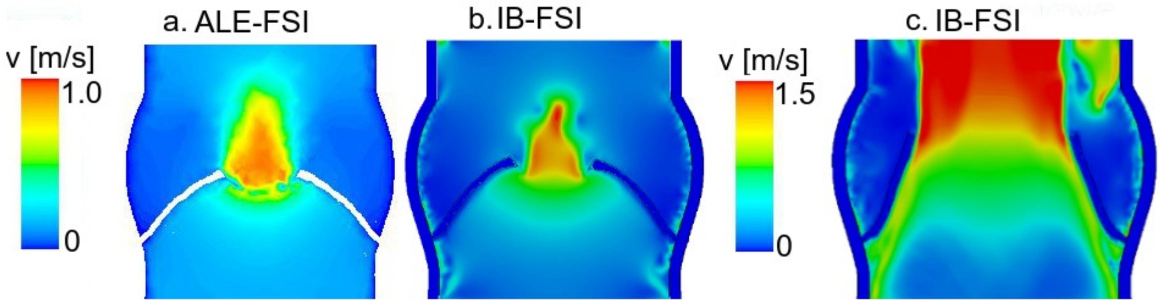

Figure 3: 3.a maximum open configuration for the ALE-FSI 3D simulation; 3.b IB-FSI in the same configuration of 3.a; 3.c open configuration for the IB-FSI 3D simulation. All the IB-FSI images include the solid domain in the visualization (leaflets and rigid wall in blue).

3.3 Computational time

The computational time required is significantly different, as table 2 shows. All the simulations are run on the same number of CPUs.

	IB-FSI	ALE-FSI
2D	1h	48h
3D	3 days (entire opening)	> 3 weeks (early opening)

Table 2: computational cost

3.4 Conclusion

To our experience, due to the technical limitation of the ALE formulation in case of large displacements, the simulation of a heart valve with a fluid-structure approach seems to be infeasible, even though preferable, in theory. Beside this aspect, the IB-FSI offers a significant advantage in terms of computational costs, even though the number of elements is much higher in the 3D IB-FSI, compared to the corresponding ALE-FSI. In alignment with the previously published works, the present study verifies the feasibility of both the ALE and IB FSI simulations in the 2D case, but agrees on the necessity of an IB approach when simulating structures which undergo severe deformation and displacement.

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