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# Wall shear-stress computation on heart valves

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## ABSTRACT

The flow through heart valves and the aortic valve in particular involves the strongly coupled interaction of blood flow ejected through the valve into the aortic arch. Many of the occurring phenomena are not fully understood such as e.g. the thrombus formation in mechanical heart valves or the localised calcification of bio-prosthetic valves. Fully understanding the flow and being able to improve replacement valve designs is of vital importance as every year 250000 aortic valves are replaced [9].

A number of numerical approaches to solving heart valve flows has been proposed. The earliest methods were immersed boundary or ficticious domain methods [5] where the moving valve mesh sweeps over the fixed fluid mesh and the valve boundary is represented through Lagrange multipliers. While this circumvents the need for a modification of the fluid mesh, the the valve surface is not represented in the mesh and accurate prediction of the flow field next to the leaflets is difficult leading to low accuracy of wall shear stresses and local bending.

As an alternative, the fluid mesh deforms along with the embedded valve mesh using an ALE formulation. This approach does allow to impose physical and accurate boundary conditions at the valve surface, however it does require a strategy to modify the mesh under large valve displacements. Global remeshing onto a new mesh [7] introduces excessive diffusion during the interpolation from old to new mesh. Regional remeshing of poor quality regions [2] improves this, however the most deformed regions tend to be the regions of interest near the moving valve surface which will still suffer from excessive diffusive errors.

This study is based on previous work [4] which proposes two novel approaches to simulating heart valves. The method uses a local remeshing where only a cell and its immediate edge-neighbours or face-neighbours are affected by a topological operation such as edge-deletion, node-insertion and edge- or face-swaps. A simple second-order non-conservative interpolation is used, owing to the locality of the topology change diffusive artefacts are minimised. Computational overheads for remeshing are negligible compared to the CFD solver time by the use of a combined node-to-face and node-to-boundary face data-structure.

The second novel aspect is the approach coupling of fluid and structure. Commonly, biological FSI simulations use a projection-type solver for the incompressible flow which requires the use of a strong

coupling method where the equations and compatibility conditions at the fluid-structure interface are converged using a fixed-point iteration, typically some 10 subiterations are needed per time-step [6]. The new method uses an artificial compressibility formulation for the incompressible Navier-Stokes equations, which allows to use the linearly stable midpoint rule algorithms [3]. In particular the ISS scheme is chosen which satisfies velocity continuity and the geometric conservation law by solving fluid and structure at staggered half time-steps. The method was validated for mechanical heart valves on a simplified experiment [8] and has been shown to provide very good accuracy for velocities and opening angle at very low mesh resolution.

The original work of [4] used a preconditioning approach based on the Lax-Wendroff scheme requiring only a single flow iteration per timestep to achieve second-order accuracy. However, while the preconditioning kept compressible effects small, they were still present. This paper presents an implementation method based on Chorin's original formulation [1] embedded in a dual-timestepping loop. The paper will compare the results of the preconditioning formulation of [4] with Chorin's [1] in terms of the effects of compressibility on the FSI, the wall shear stress and the numerical efficiency.

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