Image-Based Patient-Specific simulation, a computational modelling of the human left heart haemodynamics.

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Introduction

The demand for early diagnosis of heart disease and more generally for a better quantitative knowledge of the heart flow dynamics is a continuous source of motivation for the development of non-invasive exploration tools. Thanks to new advances in Computational Fluid Dynamics (CFD) techniques, an extremely detailed analysis of the haemodynamics and its interaction with the endocardium can be provided. Models taking into account the interactions between the blood flow and the wall motion (fluid-structure interaction) are promising [1] but require the complete knowledge of the complex tissue rheology, the electrical properties or even the external load due to the surrounding organs. These quantities are impossible to get in a non-invasive way today. A model with the geometry and geometry movement extracted from medical images seems to be better suited for patient-specific analysis and is compatible with clinical routine. Research teams tried to make CFD analysis of the flow in the human left ventricle based on medical imaging, but some limitations arise notably from the inflow direction and characterisation [5], the lack of realistic valves [2] or the time consuming segmentation of the medical images or even numerical issues [6].

Methodology

The objective of this study is to develop a methodology for patient-specific simulation of the blood flow within the whole left heart, including the ventricle, the atrium and the aortic and mitral valves. The input of the chain comes from non-invasive medical images that provide time-varying geometry. The process has been developed by R. Moreno et al [3] and is presently adapted to the heart blood flow. It consists in getting first a native geometry by an interactive segmentation of one cardiac phase using an image processing software (ScanIP ; Simpleware Ltd., Exeter, UK).

The high-fidelity three-dimensional geometric reconstruction covers all the space occupied by blood in the left heart cavity, including pulmonary veins and the aorta root. However, all geometrical details are not included. As a first step, the trabeculae carneae or the left atrial appendage are omitted for example, but investigation on how to include theses elements based on medical images is ongoing. The valves were not reconstructed by direct segmentation either; a particular reconstruction methodology has been developed and is described later. The realistic geometry resulting from the segmentation is discretized using a commercial mesh generator (Gambit, ANSYS) and a particular attention was given to grid quality.

This particular native mesh is then deformed by applying an automatic mesh deformation to fit each medical image [3]. After getting this couple mesh/image for each medical image, an interpolation is performed to represent heart geometry for each instant during the whole cardiac cycle.

Given the incompressible nature of the blood flow and the volume variations of the flow domain, the inflow and outflow boundary conditions in the simulations are prescribed in agreement with the physiological volume variations. An alternative to obtain inflow velocity values is to use MRI measurements. The phase contrast velocity data can be measured at the pulmonary veins and imposed directly in the simulation.

Simulations

This methodology was applied to a set of medical images taken at different instants during the heart cycle. Ten 3D images of the left heart were

acquired by a CT-scanner to generate time-varying volumetric data. Slice thickness and in-plane spatial resolution were set to 0.490 mm and 0.577 mm, respectively. Between each slice a space of 0.857 mm has been set. We obtained a one-million-cell mesh which allows us to compute a complete cycle in about one hour. The flow simulations were performed using finite-volume method as implemented in the YALES2 code [4] (CORIA, Rouen, France). An Arbitrary Lagrangian Eulerian formulation was implemented for our applications to solve the incompressible flow equations in a moving domain under large displacements.

A numerical method to account for the aortic and mitral valves was developed, based on the immersed boundaries formulation [7]. Valve rings are detected and valves are modelled by a simple plane of a typical thickness of four mesh cells. (see Fig. 1 for the example of the aortic valves).

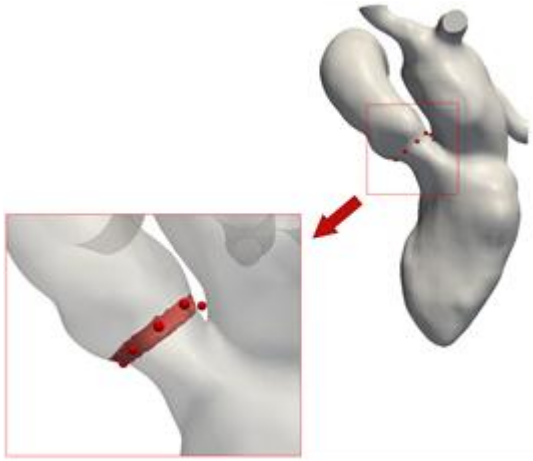

Figure 1: Aortic valve plane.

Valves are either closed or opened during the cardiac cycle, by either imposing or cancelling the immersed source term in the valve region. By imposing or cancelling the immersed source term, the open or close state of the valves can be reproduced. This simplistic approach is the first step in the valves modelling process. It enables to separate the different parts of the heart flow. Moreover, the effect of flow blockage by the mitral valve is reproduced by opening the valve only partially.

The simulations output is a functional imaging description of the left heart flow. First results for a healthy patient left heart will be presented during the meeting (see Fig. 2). Left heart flow haemodynamics will be described in details by discussing the evolution of velocity, wall shear stress and pressure gradients along the heart cycle.

The authors are grateful to GENCI/CINES for giving to super-computing facilities.

Figure 2: Left heart model during diastole phase. The vectors represent the velocity magnitude.

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