4D Flow MR Imaging for Blood Velocity Field Validation: A Preliminary Study in Atrial Fibrillation Patients

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Abstract

Computational fluid dynamics (CFD) represents a unique tool to simulate fluid dynamic conditions on a complex system, such as the cardiac blood flow in the heart chamber, in a non-invasive, fully controllable, and reproducible way. Unfortunately, validation of the simulation results is still an open issue.
The purpose of this preliminary study was to obtain an initial validation of a patient-specific computational model comparing the simulated blood velocity field in left atrium (LA) from CFD, with the velocity field provided by 4D flow MRI data, to select the simulation context best suited to the specific clinical question investigated. Simulations were run on patient-specific dynamic LA models derived from MRI data and CT data on three AF patients. The focus of the validation was to compare the flow rate simulated at the mitral valve (MV) with the one measured from the 4D flow MRI.
Our preliminary analysis suggests that the flow rate simulated with the anatomical model derived from MRI has a better time synchronization with the measured flow rate, compared with the simulation result with CT model; on the other hand, the CT model was more realistic in reproducing flow rate amplitudes. A large-scale testing and validation of CFD results are required to increase the accuracy of simulations.

1. Introduction

Computational fluid dynamics (CFD) represents a valuable non-invasive approach to determining and assessing physically meaningful parameters and indicators in a complex fluid dynamics system, such as the left atrium (LA) in atrial fibrillation (AF).

CFD modelling of LA blood flows has reached an appropriate level of complexity by using dynamic wall motion [1,2], patient-specific boundary conditions [3,4] and has provided valuable insight correlating flows and stresses with AF state. However, there is no consensus on how to model atrial flows to find the best simulation set-up suitable for the clinical question to be investigated and to reduce the inaccuracies introduced by the assumptions applied in the model. The only way we can validate the results of the simulations is through the availability of 4D flow MRI data, which unfortunately are very difficult to acquire and are not part of clinical practice.

In this scenario, a quantitative comparison between blood flow simulated under specific assumptions and blood flow derived from 4D flow magnetic resonance analysis could highlight differences and help spread more informed use of CFD simulation tools.

In this preliminary study, we analysed the impact of different anatomical models derived from MRI and CT on blood flow simulations in the LA in patients with AF.

2. Materials and Methods

2.1. Patients data

The data utilized in this investigation were obtained thanks to the collaboration with the Hôpital de la Pitié et Salpetriére, Laboratoire d'Imagerie Biomédicale, Université Pierre et Marie Curie, Sorbonne Universités, Paris. To carry out this preliminary study, 3 patients with AF were selected.

4D flow MRI images were acquired using a Siemens 1.5T MRI scanner applying a T1-weighted sequence. Twenty-five volumes (pixel size 2.375 mm, slice thickness 3 mm), spanning one cardiac cycle from the end of ventricular diastole, were reconstructed by retrospective ECG and respiratory gating.

CT data for the reconstruction of the anatomical model of LA of the same patients were acquired from a Siemens scanner. Eleven volumes (pixel size 0.5 mm, slice thickness 0.6 mm), spanning one cardiac cycle from the end of ventricular diastole as the 4D flow MRI data, were reconstructed by retrospective ECG and respiratory gating.

2.2. Data analysis

The process to analyse 4D flow MRI and CT data is depicted in Figure 1 and includes several steps that are described in the next section.

4D flow MRI consists of an anatomical reference volumetric acquisition and three acquisitions in different views that contain the velocity values throughout the cardiac cycle. The anatomical reference acquisition was processed to extract the 3D LA anatomical model; the velocity field within the LA was

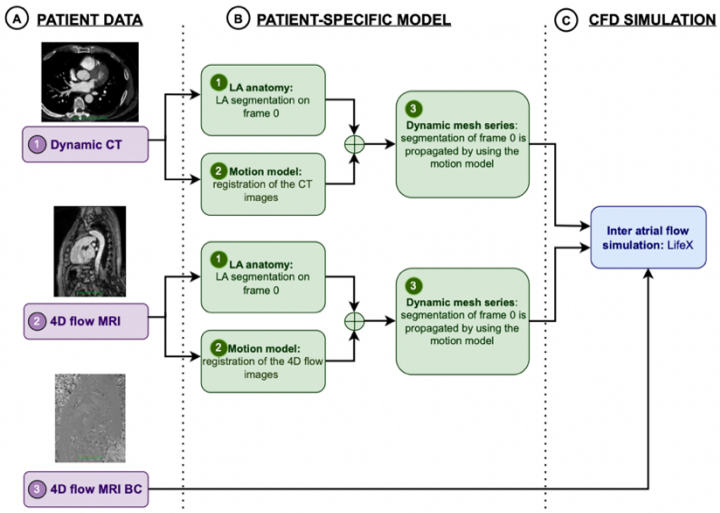


Figure 1: Workflow of the study: patient-specific CT (A1) and 4D flow MRI (A2) data are processed to derive the LA anatomical and displacement models (B1-2) from which a dynamic model (B3) is computed; the latter represents the computational domain for the personalized CFD model (C) using PVs flow rate from 4D flow MRI as boundary conditions (A3).

extrapolated from the three time-series acquisition. The first volume of 4D flow data acquired for each patient during the cardiac cycle was manually segmented using ITK-SNAP (version 3.8.0, Kitware Inc.). The output of the segmentation was saved as a 3D binary volume.

CT data were also processed to extract 3D LA anatomical model. The first volume of the CT data was segmented in MATLAB environment. First, a region of interest (ROI) including the LA chamber was defined.

Then, for 2D segmentation, the histogram of the image was analysed and the peak corresponding to the grey intensity of the LA was identified. Based on this value, two thresholds were applied to perform a preliminary segmentation of the atrium. To improve the initial segmentation, its contour was regularized and smoothed using morphological operators and a Laplacian filter, respectively.

Once all the steps described above were applied, the 3D LA surface was reconstructed using data resolution and the 2D LA contours in each slice both from 4D flow MRI and CT data.

To obtain the patient-specific displacement model, 4D flow MRI and CT volume registration and initial mesh propagation were performed using Elastix [5]. Together with Transformix, this library allows to evaluate and then apply the transformation matrix to register the reference volume. A B-spline model was utilized to execute a non-rigid registration and the mean squared difference was used to optimize registration performance. Following the physiological periodicity of the heart motion, Fourier series interpolation was applied, which recovers a continuous and periodic function from the discrete data provided.

Finally, to run fluid-dynamic simulations in SR conditions, the dynamic model was supplied as input to the customized CFD model [6]. For each patient, we run two simulations, one using the 4D flow MRI-derived dynamic model and one using the CT-derived dynamic model.

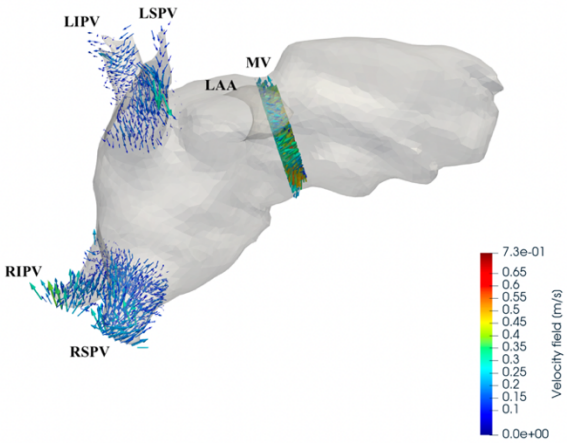


Figure 2: 3D left atrio-ventricular anatomical model with superimposition of the velocity field isolated at PVs and MV, LIPV: left inferior pulmonary vein, LSPV: left superior pulmonary vein, RIPV: right inferior pulmonary vein, RSPV: right superior pulmonary vein, LAA: left atrial appendage, MV: mitral valve.

The use of the incompressible Navier-Stokes equations in the Arbitrary Lagrangian Eulerian [7] frame of reference allowed for the replication of blood flow as a fluid.

Patient-specific boundary conditions (BCs), in terms of PVs flow rate, were extracted from 4D flow MRI velocity field and applied to the CFD simulations. For each PV, the velocity field was evaluated exclusively inside a VOI by multiplying the 3D velocity volume, containing the intensity of three x, y, and z velocity components for each voxel, by the corresponding LA 3D binary mask derived from the segmentation step. In Figure 2, through the use of Paraview (version 5.10.1, Kitware Inc.), we visualize the 3D velocity field superimposed to the LA mesh. For each PV and for each time frame the mean velocity value inside the VOI was computed and used as BC for the corresponding PV.

In addition, to take into account the size of each PV, we applied the mass balance equation as follows:

$$Q_{LIPV} + Q_{LSPV} + Q_{RIPV} + Q_{RSPV} + Q_0 + Q_{wall} = 0$$

$$Q_{wall} = \frac{dV}{dt}$$

where Q_{LIPV}, Q_{LSPV}, Q_{RIPV}, Q_{RSPV} are the flow rate at the left inferior, left superior, right inferior and right superior PV respectively, Q_0 is the flow rate at the MV and Q_{wall} is the flux associated to the volume variation. Each flow rate was calculated as $Q_{PV} = v_{PV}A_{PV}$, where v_{PV} indicates the mean velocity and A_{PV} indicates the area of the PVs section from both 4D flow MRI and CT model.
Since the blood was assumed as a Newtonian fluid, the viscosity μ was considered as constant.

The dynamic viscosity was adjusted to 0.035 poise, the density to 1.06 g/cm^3, and the time step to 0.001 seconds. The LifeX package was used to run the simulations [8].

To mitigate the impact of the initial fluid velocity conditions on the results, we conducted a simulation with three heartbeats and solely examined the final one.

To validate the blood velocity fields obtained by applying the CFD model in the two settings we evaluated the velocity field at the MV. As previously described for the PVs we extracted the average velocity at the MV from the 4D flow MRI, throughout the cardiac cycle. This average MV velocity-time curve was compared with the two simulated velocity-time curves obtained by applying the MRI and CT-derived anatomical models. The measured flow rate at the MV was calculated as $Q_{MV} = v_{MV}A_{MV}$, where v_{MV} indicates the mean velocity and A_{MV} indicates the area of the MV section, measured from the 4D flow anatomical model and the CT model.

3. Results and Discussion

In all study subjects, data processing and simulations were feasible. Two simulations were run for each of the three AF patients, one with the CT model and one with the 4D flow model.

For each patient, the MV flow rate, the LA volume curve, and the anatomical models were carefully analysed. The evaluation of MV flow rate considered different phases of the cardiac cycle, and results in one patient are shown in Figure 3A.

During the first phase (atrial diastole), up to the MV opening, in all simulations, the velocity near the MV was zero because it was imposed with the no-slip BCs. On the other hand, the velocities measured at the MV from the 4D flow MRI revealed that the velocity in the proximity of the valve is never equal to 0, but for reasons of continuity with the simulated flow rate it was set to zero until the opening of the MV.

Overall, in all curves, it was possible to distinguish the E wave (first velocity peak of trans-mitral blood flow in early LV diastole) and A wave (second velocity peak due to atrial contraction).

Some differences emerged between the flow rate measured from the 4D flow data (Figure 3A, blue curve) and the simulated ones (Figure 3A, orange curves). Indeed, in all the three patients analysed we recognized the repetition of two main patterns mainly focused on time synchronization and flow rate amplitude.

One difference between the results of the simulation with 4D flow model and CT model is in terms of time synchronization of the E wave and A wave peaks between the measured and simulated flow. In fact, in Figure 3A, looking at the CT model results (Figure A.2), both peaks are out of phase by a temporal shift. In the curve derived from the 4D flow MRI model (Figure A.1), we observe that the peaks are almost totally synchronized. It is well known that CT acquisitions are characterized by low temporal resolution (up to 20 frames per cardiac cycle) and this might be the cause of such shifting. We cannot exclude a different heart rate during the CT and 4D flow MRI exam, which were acquired at different times. This could also have caused such shifting. In the case of the MRI model, it was directly derived from 4D flow MRI.

The second difference that emerges between the results of the two simulations concerns the flow rate amplitude between the MRI model and the CT model. In the MRI model, the difference between measured and simulated flow rates is particularly pronounced (Figure A.1) compared to the result from CT model (Figure A.2), where the discrepancy between simulated and measured flow rates is drastically reduced. This may be related to the differences in the anatomical models represented in Figure 3B. CT data spatial resolution is higher, thus the derived anatomical model is more accurate and allows for no loss of details in small anatomical structures. Due to this difference, a larger sectional area at both PVs and MV was found, resulting in a greater flow rate value used as BC.

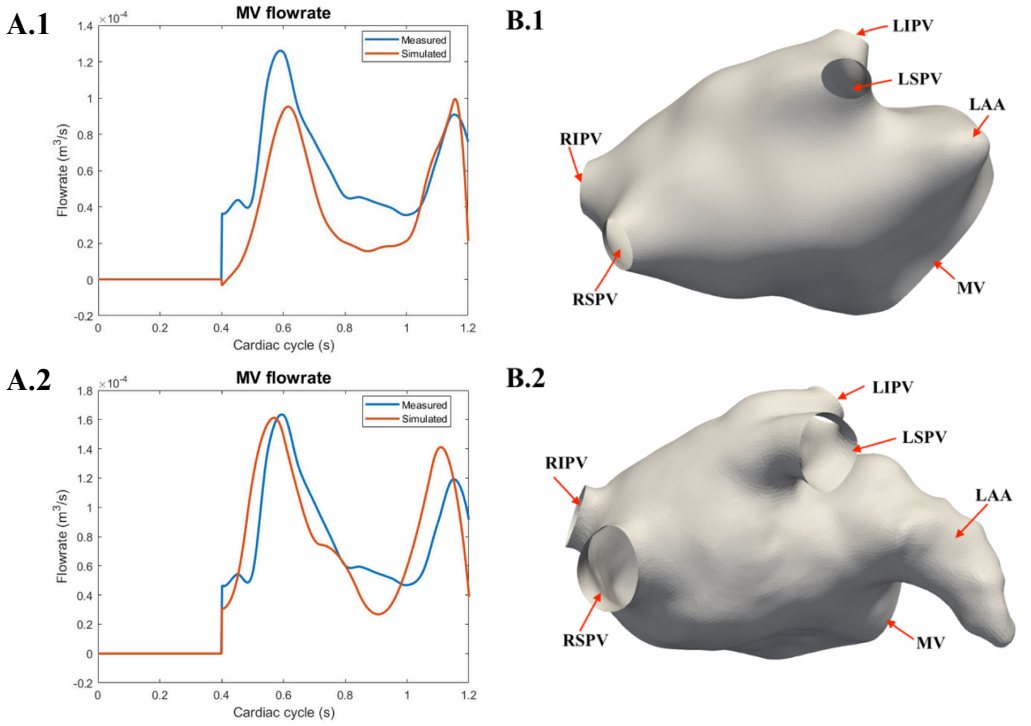


Figure 3: A: comparison between measured and simulated MV flow rate from 4D flow MRI anatomical model (A.1) and CT anatomical model (A.2). B: representation of the LA anatomical model from 4D flow MRI data (B.1); representation of the LA anatomical model from CT data (B.2).

4. Conclusions

To conclude, these preliminary results suggest that the use of anatomical models derived from different imaging modalities impacts CFD simulation results. Spatial and temporal resolution of the imaging data used to derive anatomical models affect the capability of the CFD model to accurately assess blood flow in terms of time and amplitude, whilst the shape of the flow rate curve is perfectly matched for both simulations.

The obtained results can be considered a good basis for further investigating CFD model validation on a larger sample population on which we are testing additional factors that might impact simulation results.

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