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SUPPLEMENTARY DATA

Supplementary methods

3D model construction

For each selected patient, the left atria (LA) anatomy was extracted from the clinical computed tomography (CT) images with the open-source software Slicer 4.10.1 (https://www.slicer.org/) and a semi-automatic region-growing technique. A binary mask representing the LA was obtained, plus a surface mesh was created with the classical Marching Cubes algorithm followed by a smoothing process to correct the irregularities generated by the segmentation process. A Taubin smoothing filter was applied (λ =0.5, μ= -0.53),[1] followed by the manual removal of self-intersecting faces and non-manifold edges, when necessary. The software used to reconstruct the models was Meshlab 2016.12 (ISTI-CNR, Italy), and Meshmixer 3.5 (Autodesk, Inc., United States). The pulmonary veins (PV) could be reconstructed directly from the CT-based segmentation. To consistently define the inlets of the computational models for all 3D models we followed a common criterion to determine the PV length by defining a cut just before the first branch coming out from the LA.

To solve the fluid domain inside the LA, a volumetric mesh was generated. The Delaunay algorithm available in the Open-Source software Gmsh 4.0.4[2] was used creating volumetric meshes with 500k elements. The Netgen (https://ngsolve.org) software was also used to further optimize the resulting volumetric mesh.

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In-silico simulations set-up

The CFD simulations were performed with the ANSYS Fluent Solver 19.2 (Ansys, Inc., USA). Blood was modelled as an incompressible Newtonian fluid with a 1060 kg/m^3 density and a 0.0035 Pa/s viscosity. The Reynolds numbers, computed at the MV and PV, indicated that a laminar regime shall be considered. The flow was deemed to be Newtonian[3] since the shear rate values were > 100 s^{-1}, eg in a range where blood has been reported to behave as a Newtonian fluid when circulating in large vessels[4]. The time step was set to 0.01 s. Residuals for continuity equations were set as 0.005 for convergence criteria.

A dynamic mesh algorithm, implemented in the ANSYS Fluent 19.2, was used to add wall displacement to the left atria. From a diffusion-based method it turns out that it follows the Laplace equation:

$$\nabla \cdot (\gamma \nabla \vec{u}) = 0,$$

where $\vec{u}$ is the velocity field of mesh displacement. The diffusion coefficient, γ, controls propagation (eg, smoothing out) of the boundary displacements away from the application points:

$$\gamma = \frac{1}{d^{\alpha}}$$

where d is a normalized distance to the boundary, and α is a diffusion parameter introduced by the user. In this study, α was set equal to 0, yielding a uniform diffusion of the boundary motion throughout the mesh. We applied the dynamic mesh algorithm to simulate the displacement of the MV annulus based on a function described in the work conducted by Veronesi et al.[5] Patients with non-valvular AF presented similar longitudinal MV annular movement as the healthy ones, eg, from 10 mm

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to 8 mm. The pressure wave, wall displacement, and velocity all synchronized with the ECG extracted. Therefore, all fell into the same rhythm although not in the same patient. Synchronization was performed through linear interpolation using MATLAB R2018a (Mathworks, United States).

Supplementary references

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