Structure-based modelling of human left ventricle using an extended FE-IB method

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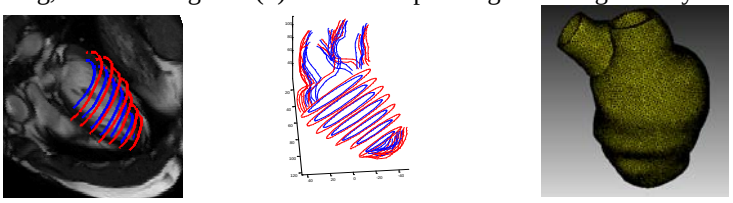
Summary Computational modelling of the human left ventricle (LV), especially the finite strain/stress analysis on imaged based reconstruction of fibre imbedded left ventricle with fluid-structure interaction and electrophysiology, provides a unique way for our understanding of myocardial energetic, blood flow, cardiac dysfunction. In this study, we aim to develop a platform to incorporate state-of-the-art magnetic resonance imaging techniques for patient specific modelling by using adaptive finite element-immersed boundary method. The model will include: (1) the fluid-structure interaction; (2) a structure-based finite strain constitutive law; (3) coupled electrophysiological simulation.

INTRODUCTION

Heart diseases, such as acute myocardial infarction (MI), remain a major public health burden all over the world. Paradoxically, since early survival is improving post-MI, more and more people are living with injured hearts and are at risk of heart failure and premature death. Improvements in risk assessments and treatments are urgently needed in order to identify high-risk patients and stratify therapy.Computational models, which can deal with sophisticated constitutive modelling as well as fluid structure interaction, such as the hybrid finite element/immersed boundary method (FE-IB)[1], can provide important tools for understanding heart diseases, and have potentials for guiding and optimizing the treatments.

METHODOLOGY

MRI Imaging and Geometry Reconstruction Cardiac MRI was performed on a healthy volunteer (male, age: 28) by a Siemens Magnetom Avanto (Erlangen, Germany) 3.0 Tesla scanner with an 8-element phased array cardiac surface coil. The study protocol was approved by the local ethics committee and written informed consent was given before the scan. Conventional cine MRI sequence was applied, parameters for short axial slices were: flip angle=50°, echo time=1.51ms, bandwidth=977Hz/pixel, field of view = 216x340mm with matrix size of 216x256, slice thickness= 10 mm. 4-chamber view long axial slice was also acquired to provide the apex structure. Endocardial and epicardial boundaries at end-diastole phase for the seven short axial slices covering from LV base to apex were manually delineated with an in-house Matlab code, shown in Figure 1(a), superimposed with the long axial slice. The apex location was identified with 4-chamber view long axial slices. Furthermore the boundaries for left atrium and aorta were also delineated with long axis views. After manual segmentation, LV boundary points were imported into SolidWorks™ for geometry reconstruction using a B-spline surface fitting, shown in Figure 1(b). The corresponding meshed geometry is shown in Figure 1(c)



(a) (b) (c)

Figure 1 (a): left ventricular boundary segmentation (blue: endocardial, red: epicardial); (b): left ventricular boundaries with aorta and left atrium; (c): reconstructed geometry with tetrahedral meshes

FE-IB method is a computational technique which enables the fluid structure interaction for complex multi-domain problems by treating an elastic structure immersed in a viscous incompressible fluid. It uses a Lagrangian description of the elasticity of the structure and an Eulerian description of the momentum, viscosity and incompressibility of the coupled fluid system. Let $\mathbf{x} \in \Omega(R^3)$ represent the Cartesian coordinates in the the fluid domain; let $\mathbf{s} \in U(R^3)$ denote the Lagrangian coordinates attached to the structure. For a material point $\mathbf{s}$ at time *t*, the physical position is denoted as $\boldsymbol{\chi}(\mathbf{s},t) \in \Omega$, then the deformation gradient is $\mathbf{F} = \dfrac{\partial \boldsymbol{\chi}}{\partial \mathbf{s}}$, and the formulation of the equations of motion for the coupled fluid-structure interaction system is as follows:

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$$
\begin{aligned}
&\rho[\frac{\partial \mathbf{u}}{\partial t}(\mathbf{x},t)+\mathbf{u}(\mathbf{x},t)\cdot\nabla\mathbf{u}(\mathbf{x},t)]=-\nabla p(\mathbf{x},t)+\mu\nabla^2\mathbf{u}(\mathbf{x},t)+\mathbf{f}^e(\mathbf{x},t),\\
&\nabla\mathbf{u}(\mathbf{x},t)=0,\\
&\frac{\partial \boldsymbol{\chi}}{\partial t}(\mathbf{s},t)=\int_{\Omega}\mathbf{u}(\mathbf{x},t)\delta(\mathbf{x}-\boldsymbol{\chi}(\mathbf{s},t))d\mathbf{s},\\
&\mathbf{f}^e(\mathbf{x},t)=\int_{U}\nabla_s\cdot\mathbb{P}^e(\mathbf{s},t)\delta(\mathbf{x}-\boldsymbol{\chi}(\mathbf{s},t))d\mathbf{s}-\int_{\partial U}\mathbb{P}^e\mathbf{N}(\mathbf{s})\delta(\mathbf{x}-\boldsymbol{\chi}(\mathbf{s},t))dA(\mathbf{s})
\end{aligned}
\tag{1}
$$

where $\mathbf{u}(\mathbf{x},t)$ is the fluid velocity field, $p(\mathbf{x},t)$ is the pressure field, $\mathbf{f}^e(\mathbf{x},t)$ is the Eulerian elastic force density generated by the immersed structure, $\mathbb{P}^e(\mathbf{s},t)$ is the Lagrangian elastic force density generated by the immersed structure related to the material coordinate system, ρ is the density, μ is the viscosity, $\mathbf{N(s)}$ is the outward normal along the structure boundary ∂U, and $\delta(\mathbf{x})$ is the 3D Dirac delta function. The LV myocardium is considered as a non-homogeneous, thick-walled, nonlinearly elastic and incompressible material by incorporating myocardium fibre and sheet orientations [2]. If the myocardium fiber direction is represented by $\mathbf{f_0}$, the sheet direction by $\mathbf{s_0}$, and the sheet normal by $\mathbf{n_0}$, then the four invariants of the right Cauchy-Green tensor, $\mathbf{C}=\mathbf{F}^{\mathrm{T}}\mathbf{F}$, is defined as:

$$
\mathrm{I}_1=tr(\mathbf{C}),\mathrm{I}_{4\mathrm{f}}=\mathbf{f}_0\cdot\mathbf{C}\cdot\mathbf{f}_0,\mathrm{I}_{4\mathrm{s}}=\mathbf{s}_0\cdot\mathbf{C}\cdot\mathbf{s}_0,\mathrm{I}_{8\mathrm{fs}}=\mathbf{f}_0\cdot\mathbf{C}\cdot\mathbf{s}_0 \tag{2}
$$

With this fibre structure, the strain energy is:

$$
W\left(\mathrm{I}_1,\mathrm{I}_{4\mathrm{f}},\mathrm{I}_{4\mathrm{s}},\mathrm{I}_{8\mathrm{fs}}\right)=\frac{\mathrm{a}}{2\mathrm{b}}\exp\left[\mathrm{b}\left(\mathrm{I}_1-3\right)\right]+\sum_{\mathrm{i=f,s}}\frac{\mathrm{a_i}}{2\mathrm{b_i}}\left\{\exp\left[\mathrm{b_i}\left(\mathrm{I}_{4\mathrm{i}}-1\right)^2\right]-1\right\}+\frac{\mathrm{a_{fs}}}{2\mathrm{b_{fs}}}\left\{\exp\left[\mathrm{b_{fs}}\left(\mathrm{I}_{8\mathrm{fs}}\right)^2\right]-1\right\} \tag{3}
$$

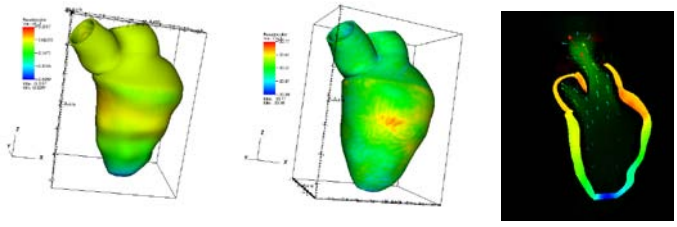
where a, b, $\mathrm{a_i}$, $\mathrm{b_i}$ ($i=f, s, fs$) are eight positive material parameters determined from experimental data[3]. Due to lack of *in vivo* information of myocardial fiber structure, a rule-based myocardial fiber generation algorithm is adopted by treating the LV as a continuum composed of laminar sheets with well-defined fibre, sheet, and sheet-normal directions. Since myocardium is contractive, the stress is the sum of passive and active stresses:

$$
\mathbf{P}^e=\frac{\partial W}{\partial \mathbf{F}}+T\mathbf{F}\mathbf{f}_0\otimes\mathbf{f}_0 \tag{4}
$$

in which T is the active tension along the fibre direction generated by the contractile apparatus of the cells, and is a dynamic function of $[Ca^{2+}]_i$ which is determined by the NHS model [4].

RESULTS

To start with, simple boundary conditions are implemented to test the proposed computational platform, where the mitral and aortic valves are not included, and the planes at the end of the left atrium and aorta are fixed in the space. Preliminary results are shown in Figure 2.



(a) ***(b)*** ***(c)***

Figure 2: (a) The the long axis displacement at the peak contracting time, where a greater shortening displacement can be found in the apex region due to the fixed boundaries at top, indicating an active contract is happening, (b) the Ca^{2+} front at the peak contracting time, , and (c) flow field inside the LV during systole.

CONCLUSIONS

In this project, a computational platform for left ventricle modelling has been proposed by integrating the clinical medial images, blood flow simulation, and structure-based nonlinear myocardium finite strain/stress analysis and myocardium active contraction together. The method has been applied to a realistic human left ventricle heart with simple boundary conditions, and the initial results are promising. It is expected the developed left ventricle simulation platform will provide more insights into heart disease in future.

References

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