Towards Model-Based Estimation of the Cardiac Electro-Mechanical Activity from ECG Signals and Ultrasound Images

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Abstract. We present a 3D numerical representation of the heart which couples the electrical and biomechanical models. To achieve this, the FitzHugh-Nagumo equations are solved along with a constitutive law based on the Hill-Maxwell rheological law. Ultimately, the parameters of this generic model will be adjusted by comparing the actual patient's ECG with computational results and the deformation of the biomechanical model with the geometric information extracted from the ultrasound images of the patient's heart.

1 Introduction

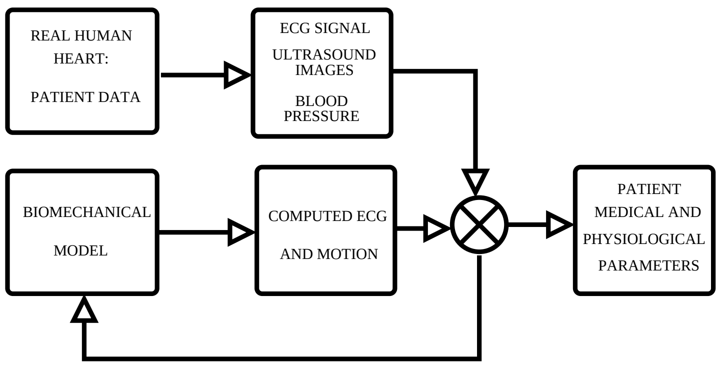
The objective of our multidisciplinary project ICEMA (standing for Images of the Cardiac Electro-Mechanical Activity, a collaborative research action between different INRIA projects) is to build a generic dynamic model of the beating heart and a procedure to automatically adjust the parameters to any specific patient. We plan to construct an identification procedure using 2 sets of relatively easy-to-access measurements on a patient: the ECG (Electrocardiogram), and a time sequence of volumetric ultrasound images. Once the generic model is adapted to a specific patient, it becomes possible to derive a set of quantitative and objective parameters useful in helping clinicians and physiologists to better understand the electro-mechanical coupling and diagnose pathological conditions. Significant results are expected in the following fields of cardiovascular pathology:

- assessment of the hæmodynamic repercussions of heart rate and electrical conduction disorders;
- assessment of the degree of heart failure (inefficiency of the cardiac pump induced by weakened cardiac contractility resulting in increased local wall thickness and decreased ejection fraction);

– assessment of the electrical and mechanical repercussions of cardiac infarction (right ventricle infarction is believed to rather lead to electrical repercussions while left ventricle infarction is believed to lead to mechanical repercussions);

Our approach combines a 3D numerical model of the electric wave propagation with a 3D biomechanical model of the cardiac muscle. The 2 models are explicitely coupled in the simulation to generate a dynamic behaviour of the heart. The model for electric wave propagation is derived from FitzHugh-Nagumo equations, while the mechanical model is based on the classical Hill-Maxwell rheological law. These models are expected to reflect on a macroscopic scale the coupling present on the cellular scale. To provide a realistic motion of a standard beating heart, the highly anisotropic nature of the muscle fibres in standard anatomy are accounted for. The electric wave is propagated from the extremities of the Purkinje network.

Two error functions will serve to adjust the parameters of this generic model to a specific patient: the first will compare the actual patient's ECGs with a set of ECGs computed from the simulation. The second will compare the deformation of the biomechanical model with the motion extracted from the ultrasound images of the patient's heart. In addition, the ventricular blood pressure, when available, is readily introduced as a constraint for the deformation of the biomechanical model. Ultimately, a retro-action procedure will be used to update the parameters of the generic model from these error functions.



In this article, we develop the current stages of this on-going research. In section 2, we describe the construction of a finite element model incorporating standard anatomical knowledge. In sections 3 and 4 we respectively describe the electrical and mechanical models. Section 5 describes a preliminary approach to account for the geometric information provided by the ultrasound image sequence. Finally, section 6 discusses the possible refinements we plan to introduce in each of these models, and the directions towards the identification of the parameters of the global system.

2 Anatomical Model

2.1 Geometry and Muscle Fibres

Our geometric model consists of a mesh of the ventricular myocardium (including both right and left ventricles) and of the myocardium fibre directions. Indeed, the fibre architecture has a great influence on the motion and propagation of the electrical excitation. The model is based on data available from the Bioengineering Research Group of the University of Auckland, New Zealand [9]. These data have been obtained by the dissection of a dog's heart and are composed of a mesh with 256 nodes, 180 hexahedra, and fibre direction at each node.

Since these data are too coarse for 3D computations, new meshes have been generated using a procedure which consists in:

1. interpolating the surface of the mesh with a mesh of triangles;
2. remeshing the 3D volume with tetrahedra;
3. interpolating the fibre directions on the new mesh.

We obtained meshes with 623, 3763, and 9623 vertices.

2.2 Conduction Network and Electrical Onset

A complete three-dimensional model of the heart should include both the cardiac muscle and special conduction network (Purkinje fibres and His bundle branches) [2, 13] and whose role is rendered through boundary conditions: the fibres of Purkinje are assumed to end up within the muscle in a region of the endocardium; a pulse-shaped boundary condition is applied on this hand-delimited region as a model of onset of the cardiac excitation.

3 Electrical Model

Each heart-beat cycle, a depolarisation wave is initiated (see above) and propagates along the myocardium during about 10% of the total cardiac cycle.

Among the various models for the electric wave propagation, the FitzHugh-Nagumo model is classical [1]. It writes:

$$\begin{aligned} \frac{\partial u}{\partial t} &= \operatorname{div}(D\nabla u) + f(u) - v, \\ \frac{\partial v}{\partial t} &= \epsilon\,(ku - v). \end{aligned} \tag{1}$$

where u represents the transmembrane potential and v is an auxiliary variable, and $f(u) = f_0\,u(1-u)(u-a)$. The system is subject to the boundary condition:

$$u(x,t) = e(x,t) \quad x \in \partial\Omega^{\mathrm{Purkinje}},$$

where e is the pulse-shaped input signal.

Anisotropic behaviour is accounted for by making the diffusion matrix D depend on the fibre orientation: $\forall x,\ D(x) = diag(d, \varepsilon, \varepsilon)$ in the orthonormal basis $(\underline{n},\ \underline{k},\ \underline{l})$ local to x ; where $\underline{n}$ is a unit vector tangent to the fibre.

The theoretical aspects of (1) have been widely studied [16]: a travelling wave of fixed shape and speed either appears or not, depending on the initial excitation being above or below a threshold. The important parameters are

- a which controls the value of the threshold for excitation;
- D and f_0 which adjust the wave speed and time scale.

A simple output, the electrocardiogram along the axis d, can be calculated as follows:

$$\mathrm{ecg}(d,\, t) \;=\; L(\nabla u(t),\, d).$$

L is a weighted integral over the volume of the heart.

Although more accurate models exist [1], we retained this one, as it correctly captures in a simple formulation, the qualitative behaviour of the wave and leads to fast three dimensional computations [14, 10, 7, 13]. This model would likely be improved by introducing an additional variable [8].

Solutions to (1) have been approximated by using a standard $P1$ Lagrange finite element procedure (with mass lumping and first order numerical integration at vertices), on the given tetrahedral anatomical mesh. Since the time-dependent phenomenon is of interest, small time-steps are used in an explicit procedure (Euler method).

The parameters a, ϵ, k, f_0 are estimated according to the results in [8].

Potential maps such as those illustrated on Figure 1 are derived at different time steps, starting after a wave has been initiated at the apex and propagated. Better initiation points are expected from data in [5].

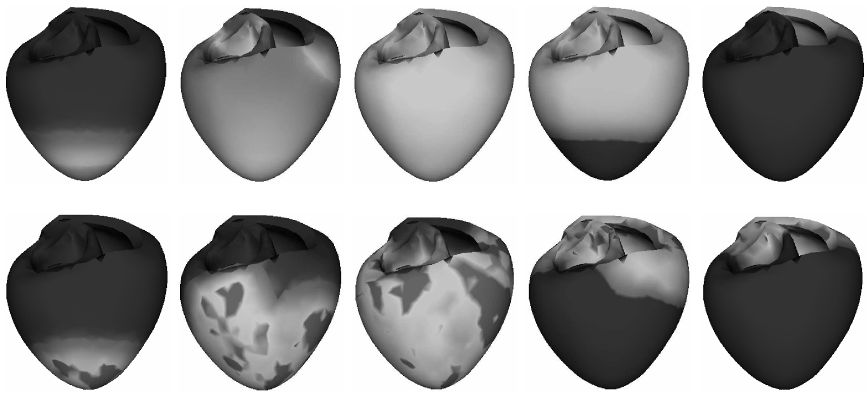


Fig. 1. (Top) Isotropic propagation ($d = 1$, $\varepsilon = 1$), (Bottom) Anisotropic propagation ($d = 1$, $\varepsilon = 0.7$)

This time-dependent computed potential can then be used as an excitation entry to the system describing the mechanical behaviour of the myocardium. No particular feedback is currently used, as proposed in [8] to strengthen the coupling between the electrical and mechanical models.

4 Mechanical Model

The constitutive law that we consider for the cardiac muscle (composed of a distribution of stacked myocardial fibres) is based on the classical Hill-Maxwell rheological model [6], schematically represented in Figure 2 where E_c, E_s and E_p symbolically denote the contractile, series and parallel element, respectively.

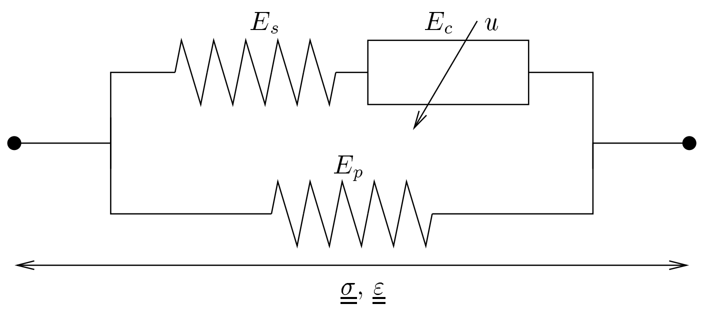


Fig. 2. Hill-Maxwell rheological model

The series and parallel elements are elastic, but non-necessarily linear. The contractile element develops an action oriented in the direction of the fibre at the point in consideration, hence the corresponding stress tensor is everywhere one-dimensional (1D) and in the form $\underline{\underline{\sigma}}_c = \sigma_c \, \underline{n} \otimes \underline{n}$, where $\underline{n}$ denotes the unit vector tangent to the fibre. This implies that the stress tensor corresponding to the series element is also 1D, since $\underline{\underline{\sigma}}_s = \underline{\underline{\sigma}}_c$. In the parallel element, the stress tensor is fully 3D (even though the corresponding constitutive law is strongly anisotropic due to the fibre architecture), so is the global stress tensor given by $\underline{\underline{\sigma}} = \underline{\underline{\sigma}}_c + \underline{\underline{\sigma}}_p$.

For the (scalar) stress quantity σ_c we use a (time-dependent) constitutive relation recently proposed and justified in [3] (see also [4]), viz:

$$\begin{aligned} \dot{\tilde{\sigma}}_c &= -\left(|\dot{\varepsilon}_c| + d|u|\right)\tilde{\sigma}_c + k_c\dot{\varepsilon}_c + \sigma_0|u|_+, \\ \dot{k}_c &= -\left(|\dot{\varepsilon}_c| + d|u|\right)k_c + k_0|u|_+, \\ \sigma_c &= k_c\varepsilon_0 + \tilde{\sigma}_c + \nu\dot{\varepsilon}_c. \end{aligned}$$

Here, ε_c denotes the (1D) strain corresponding to the contractile element, k_c and $\tilde{\sigma}_c$ represent additional internal variables, u is the input (also represented in

Fig. 2), and all other quantities are constants. The input u is a chemical quantity that depends mainly on the calcium concentration [4], and which is modelled in section 3. Hence, the resulting constitutive law for the global rheological model is of the form $\underline{\underline{\sigma}} = \Sigma\ (\underline{\underline{\varepsilon}}, u)$, where Σ denotes a functional. This constitutive law is then used in combination with the equations of solid dynamics to obtain the mechanical model.

5 Interaction with Ultrasound Images

In our current implementation, we simplify the mechanical model into anisotropic linear elasticity and the fibres activation is modelled as a stress tensor $\sigma_a = \alpha\, f \otimes f$, where α is the activation rate which is directly related to the potential u of Section 3, and f the fibre direction. It gives a force $\mathbf{F}_a$:

$$\mathbf{F}_a = \int_V \operatorname{div}(\sigma_a)\, dv = \int_V \operatorname{div}(\alpha f \otimes f)\, dv = \int_S (\alpha f \otimes f) \overline{\otimes}\, n\, ds.$$

Therefore, when a fibre is activated, its contraction is equivalent to a pressure applied to the surface of the tetrahedron in the fibre direction.

Figure 3 presents the results of a contraction proportional to the potential (which should be replaced by a differential equation controlling the activation from the potential).

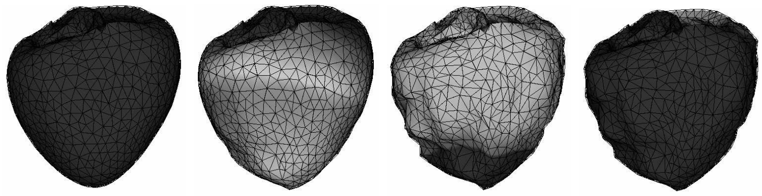


Fig. 3. Effect of the fibres contraction on the 3D model.

The model is also constrained by the ultrasound images (see Figure 4) through the external forces $\mathbf{F}_i$ which are proportional to the distance to the closest boundary point of the image from the considered point of the mesh.

We use mass-lumping in a Newtonian differential equation with an explicit integration scheme to compute the position $\mathbf{P}$ of each vertex:

$$\left(\frac{1}{\Delta t^2} - \frac{\gamma}{2\Delta t}\right) \mathbf{P}^{t+1} = \mathbf{F}_i + \mathbf{F}_a + \mathbf{F} + \frac{2}{\Delta t^2}\mathbf{P}^t - \left(\frac{1}{\Delta t^2} + \frac{\gamma}{2\Delta t}\right) \mathbf{P}^{t-1},$$

with γ the damping factor. $\mathbf{F}$ represents the forces computed from linear elasticity plus the boundary conditions (in particular the ventricular pressures). As ultrasound data is sparse and noisy, it is efficient to use a volumetric model [11, 12] together with electro-mechanical knowledge to extract information (see [15] for preliminary results).

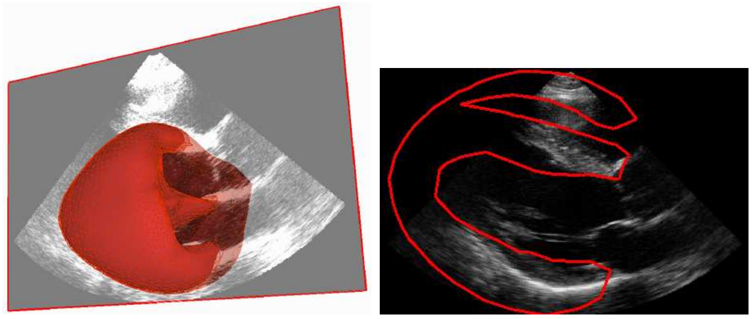


Fig. 4. Left: the 3D model in a slice of the 3D ultrasound image. Right: intersection of the model and the image.

6 Perspectives

Each part of our model is planned to be improved:

- the anatomical model (Section 2) should be based on a human heart. Diffusion MRI is probably one possible way to obtain fibre directions and maybe Purkinje network;
- the electrical model (Section 3) could include a third variable to better control the shape of the wave and a mechano-electrical feedback;
- the mechanical model (Section 4) has to be solved on a 3D mesh;
- the simplified model for ultrasound segmentation (Section 5) will become non-linear.

For each of these models we intend to identify parameters:

- for the electrical parameters, D, f_0, and the electrical entries, Purkinje fibres, ectopic foci, should be estimated by comparing computed ECGs with measured ones. To achieve this, an inverse problem has to be considered;
- for the mechanical parameters, identification techniques still have to be developed, as in vivo rheological studies for human tissues are hard to set up. Here, a criterion will be the difference between the computed motion and the one extracted from the ultrasound images.

All these points will be the topics of our future work.

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