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Human heart simulation software for parallel computing systems

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Abstract

This study demonstrates a software system for simulating the electrical and mechanical activity of the heart left ventricle (LV). The distinguishing feature of the system is the use of the heart anatomy model, which is based on an analytic description of the ventricular shape and fibre orientation. Another important system feature is that the simulation takes into account relationships between the mechanical and electrical activities that occur in the LV (cardiac mechano-electrical coupling). We describe our model integration design, the software system architecture, and the parallel implementation of the system using OpenMP technology. We evaluate the performance and scalability of the software system. In addition, we provide two examples of simulation results.

Keywords: Heart simulation, HPC, OpenMP, electro-mechanical coupling, mass-spring system

1 Introduction

Modern progress in medicine, system biology, and high performance computing have made it possible to perform computer simulations of the physiological functions of whole organs. Computer simulations can help discover new drugs and methods of treatment. This is very important for treating cardiovascular diseases, which are leading causes of mortality worldwide. Furthermore, the personalised computer heart simulations are required for predictive, preventive, personalised, and participatory (P4) medicine. Such simulations not only provide an opportunity to predict the progress of heart diseases, but they can also help prevent heart attacks and other life-threatening health conditions. The simulation performance is very important because the integration of a personalised model into health service requires suitable calculation times.

Most research groups have been working on heart simulations based on finite element method (FEM) and data from CT scans or MRI scans [1], [2], [3], [4], [5]. This approach gives flexibility

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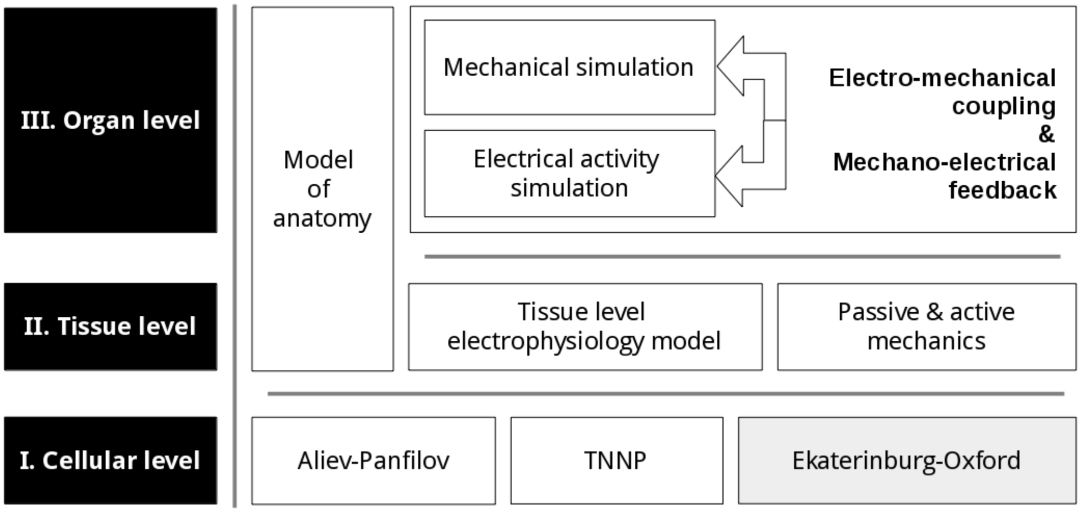

Figure 1: Levels of the simulation and the accompanying mathematical models.

and takes into account small features of patient's heart geometry. Some complex software systems are developed with this approach as prototypes for usage in health-care systems [6], [7].

In this paper, we describe the software system for simulating the left ventricle (LV) of the heart. A distinguishing feature of our system is usage of an analytic description of the LV shape and fibre orientation field [8]. In addition, the simulation takes into account relationships between LV mechanical and electrical activity (cardiac electro-mechanical coupling).

We describe our model integration design, the software system architecture, and the parallel implementation of the system using OpenMP technology. OpenMP enables easy parallelisation using compiler directives without significant code change. In addition, OpenMP program can be ported to the Intel Xeon Phi computation accelerator. We also provide examples of the simulation results and evaluate the performance of our software system.

2 Model Integration Design

A mammalian heart includes four chambers: the left ventricle, right ventricle, left atrium, and right atrium. The LV is the most powerful and important chamber of the heart, and most cardiac diseases are related to LV malfunction. For example, remodeling of the LV geometry attributes to patients with the dilated cardiomyopathy. Changes in the functional properties of the tissue may result in ventricular arrhythmias.

The LV simulation is performed on three levels: the model of myocardial cells (cardiomyocytes), the model of tissue, and the model of the whole organ (Fig 1). Below, we briefly describe the models we use in our software system and relationships between them.

The first level is a model of a single cardiomyocyte. Such the models simulate various properties of cellular activity: transmembrane potential, ionic currents through cells membrane, conductivity, length of sarcomeres, etc. The software can use the Aliev–Panfilov (AP) model [9], the ten Tusscher–Noble–Noble–Panfilov (TNNP) model [10], and the Ekaterinburg–Oxford

(EO) model [11]. The AP model is a simple qualitative model, which is important for whole system verification and preliminary analysis. This model includes the electro-mechanical coupling and can be used for simulations of the LV mechanical function. The TNNP model is a widely used ionic model simulating the electrical activity of the cardiomyocytes in the human heart. It has to be combined with a myofilament model to simulate electro-mechanical coupling in the cell. The EO model of the electrical and mechanical activity in cardiomyocytes accounts for both the electro-mechanical coupling and mechano-electrical feedback in the cell. In all these models, the cardiomyocyte behaviour is described with the help of algebraic and nonlinear ordinary differential equations (ODE).

The second level of the model hierarchy includes tissue-level models which consider the myocardium as a continuum. Partial differential equations (PDE) are used to describe the propagation of the excitation wave in the tissue. Mechanical models describe the function of potential energy in response to the deformation. Our software provides the opportunity to use two models of passive myocardium mechanics: the Hunter model [12] and the Guccione model [13].

The third level of system integration contains an anatomical LV model which simulates the LV shape and fibre orientation field. The anatomy of the LV can be described with either an individual map or an analytic model. The individual map approach requires MRI or CT data as well as a lot of computational resources. Our software uses the second approach, based on the LV analytic description that was developed earlier by our group [8]. The analytic model includes two variants of geometrical description with different complexity. The first variant is an average (population level) model. The second variant provides the ability to take into account individual features of a patient's heart. The model requires as input data some parameters of heart geometry, such as LV cavity depth, wall thickness, and so on, and generates an LV geometry description for grid generators. Also, the model provides fibre orientation field and normals to muscular layers for electrical and mechanical computation. Based on this model, the electrical and mechanical activity of the LV is calculated.

The LV electrical activity simulation is done by combining equations from the cellular, tissue, and anatomy models through the finite difference method (FDM) on a structured grid [14] using either the AP, TNNP or EO models. For the FDM, we use the explicit Euler method. The model calculates action potential and other phase variables in each node of the structured grid. If the AP or EO model is used, then active muscle tension is calculated, too.

A widely used approach for mechanical simulations is the finite element method, but we use a modified mass-spring system (MSS) [15]. This method was suggested by Lasseter and Terzopoulos [16][17] and was modified by Oussama Jarrousse [18] for cardiac-mechanical activity simulations. We consider a continuous medium in each tetrahedron of an unstructured spatial grid. The masses are placed in the grid vertices. The medium properties are described by a potential energy function from the passive and active mechanical models. This method is suitable for soft incompressible or hardly-compressible bodies such as heart tissue. Unfortunately, the medium points oscillate after the application of force. This effect requires artificial friction to be applied, and leads to several convergence steps for achieving a stable body position before applying force in the next timestep. Thus, the simulation performance depends on the force that is applied to the body. The modified mass-spring method requires less time than the finite element method if the same tetrahedral meshes are used, especially in cases where applied forces are low.

Another important property of the heart tissue is the cardiac electro-mechanical coupling. The LV mechanical contraction is induced by electrical activation and develops in parallel with excitation propagation through the tissue. Our software simulates the LV electro-mechanical

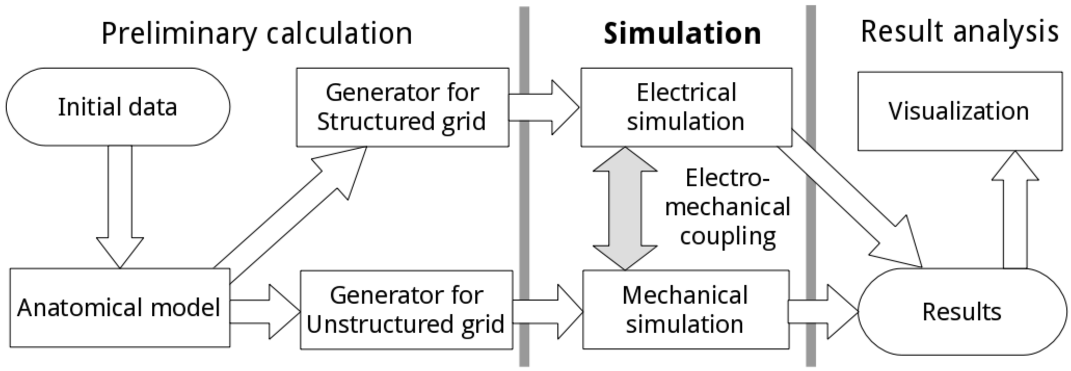

Figure 2: The software architecture

coupling using either the AP model or the EO model of cardiomyocytes. Active muscle tension is calculated in the cellular model. Together with the passive myocardial mechanics model, it enables us to find the LV deformation. In this way, the electrical excitation influences mechanical properties of myocardial tissue and induces the heart pumping function.

3 The Software System

We developed a software system based on the models described above. The architecture of our software system is shown in Fig. 2. The computation pipeline can be divided into three stages: preliminary calculation, simulation, and the analysis of the results.

During the preliminary calculation, the initial data are passed in an anatomical model, which creates equations for a structured grid and an LV shape for the unstructured grid generator. An external generator (we use Triangle [19] and Tetgen [20]) creates an unstructured tetrahedral grid. After that, the simulation stage begins. The electrical and mechanical models simulate the LV activity. Finally, the results of the simulation are analysed in the third stage.

Our architecture makes it possible to select models of electrical activity, passive and active mechanics, and mesh generators.

The structured grid is used for the simulation of the heart electrical activity by FDM. Equations for the grid are derived directly from the anatomical model.

The unstructured grid is used for the mechanical simulation. We use an unstructured tetrahedral grid based on the analytic description of the LV shape from the anatomical model. The nodes in the structured and unstructured grid do not coincide. The structured grid contains 15 times more nodes than the unstructured one.

To implement the electro-mechanical coupling, our software system calculates 10 timesteps of the electrical simulations per one step of the mechanical ones. The electro-mechanical coupling is implemented through the interpolation algorithm, which transfers the active tension value.

The software generates files with phase variable values. After analysis of the results, we can find the excitation wavefront, spiral wave filament, mechanical node coordinates, velocities, forces, blood pressure, cavity volume, etc.

The system is implemented in C language. We use OpenMP technology to provide the parallel implementation for the shared memory hardware. The ParaView system is used for visualisation and results analysis.

4 Parallelisation algorithm

The general scheme of OpenMP implementation is shown in Alg. 1.

<Preliminary calculation>
while $\Delta t * N_{step} <= T_{physical}$ **do**
  #pragma parallel for
  for *structured grid points* **do**
    <Euler alghorithm>
  end
  if N_{step} *mod* $10 = 0$ **then**
    <Data interpolation: structured to unstructured grid>
    while *MSS stability is not reached* **do**
      #pragma parallel for
      for *unstructured grid tetrahedron* **do**
        <MSS: calculate force vector in tetrahedron center>
      end
      #pragma parallel for
      for *unstructured grid points* **do**
        <MSS: sum force from tetrahedrons to points>
      end
      #pragma parallel for
      for *unstructured grid points* **do**
        <MSS: solve Newton's equation>
      end
    end
    <Write data if needed >
    <Data interpolation: unstructured to structured grid >
  end
  <Write data if needed>
end
<Write data if needed>

Algorithm 1: Implementation scheme

In preliminary calculations, the program calls external tools, allocates memory and initialises arrays. The main loop implements steps by time. It works while step number N_{step} multiplied by time step Δt is less than physical time $T_{physical}$. The loop works sequentially because of data dependency. The main loop includes two parts: electrophysiological computation and mechanical computation. The electrophysiology computation uses the Euler method to solve PDEs and is parallelised with OpenMP directives. The software system calculates 10 timesteps of the electrical simulations for every step of the mechanical ones. The MSS method implementation does the following work on each mechanical step: calculates force in each tetrahedron, sums the forces from the terahedra to the vertices, and solves Newton's second law equation at each grid point. The MSS method is implemented in three cycles. These cycles work sequentially due to data dependency, but each of them is parallelised with OpenMP.

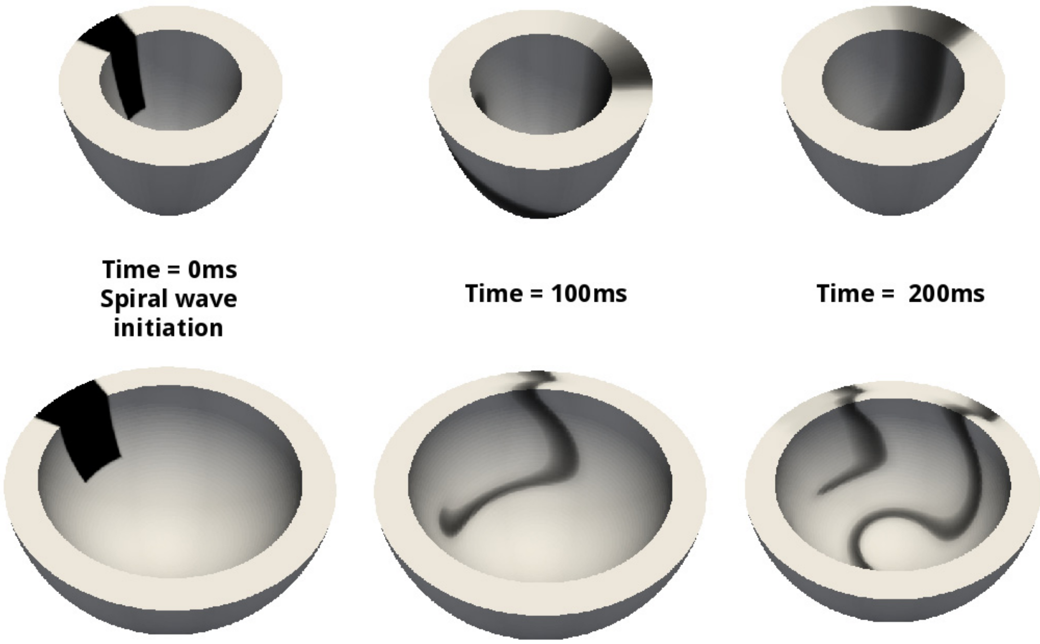

Figure 3: Spiral wave dynamics in a healthy LV and an LV with a "dilated cardiomyopathy" shape. Parameter a in the AP model was equal to 0.03 (no break-up tendency) and 0.08 (some tendency to break-up) for a healthy and non-healthy LV, respectively. The change in shape and electrical property led to the break-up of the scroll wave.

5 Examples of the Simulation Results

In this section, we demonstrate the results of two simulations produced by our software. In the first experiment, we simulated spiral waves in a healthy LV and in an LV with the shape typical for dilated cardiomyopathy (Fig. 3). The AP cellular model and the symmetrical anatomical model were used for the simulation. The mechanical simulation was turned off. A spiral wave was initiated by the electrical stimulation of a tissue block with a temporary non-conductive block near the stimulated area. This blockade disappeared after a short period of time. We saw a break-up of the wave in the heart with dilated cardiomyopathy after 200 ms of the simulation, while in the healthy heart the LV spiral is was stable (Fig. 3). Similar data were described in detail in our previous work [21].

In the second experiment, we demonstrate results of simulating the mechanical activity of the LV. Pressure and volume plots during the contractile cycle are shown in Fig. 4. In this experiment, the AP cellular model and the Hunter passive mechanics model were used. One heartbeat was simulated. The phases on the plots reflected the LV contraction phases in the real heart and reproduce the main features of the real heart mechanics. For example, the ejection fraction in the simulation is about 60% as in the normal heart. Similar data were described in detail in our previous work [22].

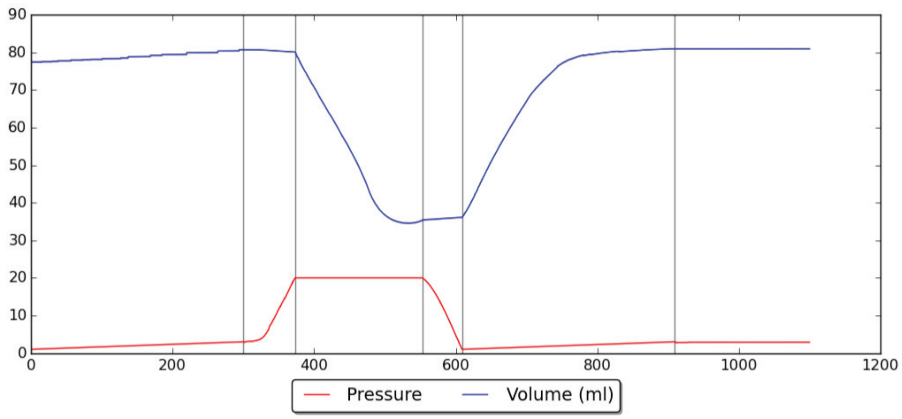

Figure 4: Simulation of the LV volume and pressure dynamics during the cardiac cycle. The electrical stimulation was applied at 290 ms to the endocardial surface, following which the electrical excitation wave propagated through the LV tissue. The following heartbeat phases are shown: the diastolic filling, isovolumetric contraction, ventricular ejection, isovolumetric relaxation, the second filling, and resting phase.

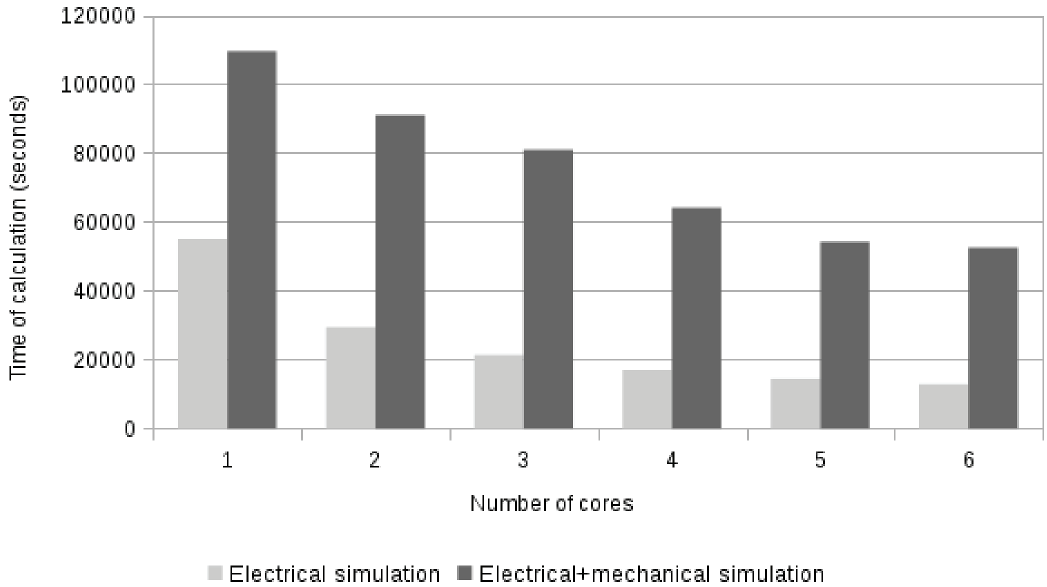

Figure 5: The relationship between the working time and the number of cores used. The Ekaterinburg–Oxford model, both with and without mechanical calculations, is used

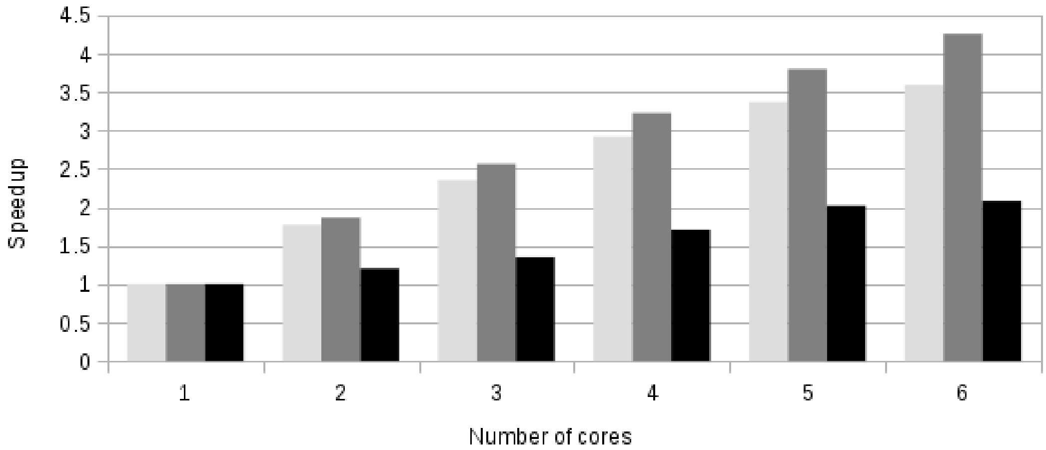

Figure 6: The relationship between the software speedup and the number of cores used. Column legend, left to right: AP model without mechanical simulation, EO model without mechanical simulation, EO model with mechanical simulation

6 Performance Evaluation

We performed a series of tests to evaluate the software scalability. The tests were carried out on the computational cluster at Ural Federal University. We used nodes with the following configurations: 2 x CPU Intel Xeon E5-2620 v2 (frequency 2.10 GHz, 6 cores, and AVX vector extensions), 32 GB RAM, and 500 GB local disk space. The server runs CentOS 7. To compile our software, Intel Composer 2015 with a maximum level of optimisation (O3) was used.

For testing, the simulation of a single heartbeat of 900 ms cycle length was performed. Either the AP cellular model or the EO electromechanical model was used with the same parameters as in the experiments shown in the previous section. The following numerical scheme parameters were used: an electrical time step of 0.005 ms; a mechanical time step of 0.02 ms; 915,000 vertices in the electrical grid with a grid step of 0.5 mm; and 58,530 elements in the mechanical model with a grid step of 2 mm.

We performed three series of tests. In the first and second series, we calculated the LV electrical activity using either the AP or EO model without the mechanical part. In the third series of the experiment, we used the EO electro-mechanical model. The results of the second and third series are presented in Fig. 5.

The relationship between the software speedup and the number of CPU cores is shown in Fig.6. We managed to achieve a 4x speedup with 6 cores for the electrical simulation (both for the AP and the EO models). For the coupled mechanical and electrical simulation, only a 2x speedup was achieved. The simulation with the electro-mechanical coupling was slower, because the MSS method requires several convergence steps for achieving a stable body position before applying force in the next timestep. Thus, the performance of the simulation depends on the force applied to the body. The mass-spring method requires less time than the finite element method in cases where applied forces are low (ventricular diastole). Unfortunately, MSS works more slowly when a larger force is applied (systole).

The implementation acceleration is nonlinear because electrophysiological simulation scalability is restricted by the small grid size. Another reason of nonlinear speedup is the parallelisation algorithm design of the modified MSS method. The method implementation uses three parallel loops that must be executed sequentially due to data dependency with the previous loop (Alg. 1).

7 Conclusion and Future Work

We presented the software system for the heart LV simulation using various state-of-the-art cellular models.

The parallel version of the system was implemented using OpenMP technology, which is the first step of parallelisation. The next steps involve porting the OpenMP version to the Intel Xeon Phi accelerator and developing a message passing interface (MPI) version for shared memory systems.

Also, this software system will be changed for personalised heart simulations.

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References

[1] F. Vadakkumpadan, V. Gurev, J. Constantino, H. Arevalo, and N. Trayanova. Modeling of whole-heart electrophysiology and mechanics: Toward patient-specific simulations, 2010.

[2] N. A. Trayanova. Whole-heart modeling : Applications to cardiac electrophysiology and electromechanics. *Circulation research*, 108(1):113–128, 2011.

[3] H. Xia, K. Wong, and X. Zhao. A fully coupled model for electromechanics of the heart. *Computational and Mathematical Methods in Medicine*, 2012, 2012.

[4] M. Sermesant, R. Chabiniok, P. Chinchapatnam, T. Mansi, F. Billet, P. Moireau, J. M. Peyrat, K. Wong, J. Relan, K. Rhode, M. Ginks, P. Lambiase, H. Delingette, M. Sorine, C. A. Rinaldi, D. Chapelle, R. Razavi, and N. Ayache. Patient-specific electromechanical models of the heart for the prediction of pacing acute effects in CRT: A preliminary clinical validation. *Medical image analysis*, 16(1):201–215, 2012.

[5] J. Aguado-Sierra, A. Krishnamurthy, C. Villongco, J. Chuang, E. Howard, M. J. Gonzales, J. Omens, D. E. Krummen, S. Narayan, R. C. P. Kerckhoffs, and A. D. McCulloch. Patient-specific modeling of dyssynchronous heart failure: A case study. *Progress in biophysics and molecular biology*, 107(1):147–155, 2011.

[6] G. R. Mirams, C. J. Arthurs, M. O. Bernabeu, R. Bordas, J. Cooper, A. Corrias, Y. Davit, S. . Dunn, A. G. Fletcher, D. G. Harvey, M. E. Marsh, J. M. Osborne, P. Pathmanathan, J. Pitt-Francis, J. Southern, N. Zemzemi, and D. J. Gavaghan. Chaste: An open source C++ library for computational physiology and biology. *PLoS Computational Biology*, 9(3), 2013.

[7] I. Larrabide, P. Omedas, Y. Martelli, X. Planes, M. Nieber, J. A. Moya, C. Butakoff, R. Sebastin, O. Camara, M. De Craene, B. H. Bijnens, and A. F. Frangi. *GIMIAS: An open source framework for efficient development of research tools and clinical prototypes*, volume 5528 of *Lecture Notes in Computer Science (including subseries Lecture Notes in Artificial Intelligence and Lecture Notes in Bioinformatics)*. Springer, 2009.

[8] S. F. Pravdin, V. I. Berdyshev, A. V. Panfilov, L. B. Katsnelson, O. Solovyova, and V. S. Markhasin. Mathematical model of the anatomy and fibre orientation field of the left ventricle of the heart. *BioMedical Engineering Online*, 12(1), 2013.

[9] R. R. Aliev and A. V. Panfilov. A simple two-variable model of cardiac excitation. *Chaos, Solitons and Fractals*, 7(3):293–301, 1996.

[10] K. H. W. J. Ten Tusscher, D. Noble, P. J. Noble, and A. V. Panfilov. A model for human ventricular tissue. *American Journal of Physiology - Heart and Circulatory Physiology*, 286(4 55-4):H1573–H1589, 2004.

[11] V. S. Markhasin, O. Solovyova, L. B. Katsnelson, Yu Protsenko, P. Kohl, and D. Noble. Mechano-electric interactions in heterogeneous myocardium: Development of fundamental experimental and theoretical models. *Progress in biophysics and molecular biology*, 82(1-3):207–220, 2003.

[12] P. J. Hunter, A. D. McCulloch, and H. E. D. J. Ter Keurs. Modelling the mechanical properties of cardiac muscle. *Progress in biophysics and molecular biology*, 69(2-3):289–331, 1998.

[13] J. M. Guccione, A. D. McCulloch, and L. K. Waldman. Passive material properties of intact ventricular myocardium determined from a cylindrical model. *Journal of Biomechanical Engineering*, 113(1):42–55, 1991.

[14] S. F. Pravdin, H. Dierckx, L. B. Katsnelson, O. Solovyova, V. S. Markhasin, and A. V. Panfilov. Electrical wave propagation in an anisotropic model of the left ventricle based on analytical description of cardiac architecture. *PLoS ONE*, 9(5), 2014.

[15] S. Pravdin, H. Dierckx, A. Panfilov, V.I. Berdyshev, L.B. Katsnelson, O. Solovyova, and V.S. Markhasin. *Simulation of the cardiac left ventricle electromechanical activity based on a computational analytical anatomical model.* Glasgow 2nd Workshop on Soft Tissue Modelling, page 41, 2015.

[16] J. Lasseter. Principles of traditional animation applied to 3D computer animation. *Computer Graphics (ACM)*, 21(4):35–44, 1987.

[17] D. Terzopoulos, J. Platt, and K. Fleischer. Heating and melting deformable models (from goop to glop). In *Proceedings - Graphics Interface*, pages 219–226, 1989.

[18] O. Jarrouse. *Modified Mass-spring System for Physically Based Deformation Modeling.* Karlsruhe: KIT Scientific Publishing, 2012, 240p., 2012.

[19] R. S. Jonathan. Triangle: Engineering a 2D quality mesh generator and Delaunay triangulator. In Ming C. Lin and Dinesh Manocha, editors, *Applied Computational Geometry: Towards Geometric Engineering*, volume 1148 of *Lecture Notes in Computer Science*, pages 203–222. Springer-Verlag, May 1996. From the First ACM Workshop on Applied Computational Geometry.

[20] H. Si. TetGen, a delaunay-based quality tetrahedral mesh generator. *ACM Transactions on Mathematical Software*, 41(2), 2015.

[21] S. Pravdin, H. Dierckx, V.S. Markhasin, and A. Panfilov. Drift of scroll wave filaments in an anisotropic model of the left ventricle of the human heart. *BIOMED RESEARCH INTERNATIONAL*, page 13, 2015.

[22] S.F. Pravdin, L.B. Katsnelson, O. Solovyova, A.V. Panfilov, V.I. Berdyshev, and V.S. Markhasin. [Matematicheskaya model' levogo zheludochka serdca]. In *[BIOMEHANIKA – 2014]*, pages 187–190, 2014.