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Hans Lamecker, Tommaso Mansi, Jatin Relan, Florence Billet, Maxime Sermesant, et al.. Adaptive Tetrahedral Meshing for Personalized Cardiac Simulations. CI2BM09 - MICCAI Workshop on Cardiovascular Interventional Imaging and Biophysical Modelling, Sep 2009, London, United Kingdom. pp.149-158. inria-00417371

HAL Id: inria-00417371

<https://inria.hal.science/inria-00417371>

Submitted on 15 Sep 2009

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Adaptive Tetrahedral Meshing for Personalized Cardiac Simulations

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Abstract. Personalized simulation for therapy planning in the clinical routine requires fast and accurate computations. Finite-element (FE) simulations belong to the most commonly used approaches. Based on medical images the geometry of the patient's anatomy must be faithfully represented and discretized in a way to find a reasonable compromise between accuracy and speed. This can be achieved by adapting the mesh resolution, and by providing well-shaped elements to improve the convergence of iterative solvers. We present a pipeline for generating high-quality, adaptive meshes, and show how the framework can be applied to specific cardiac simulations. Our aim is to analyze the meshing requirements for applications in electrophysiological modeling of ventricular tachycardia and electromechanical modeling of Tetralogy of Fallot.

Keywords: elements, modeling, mesh quality, mesh size control

1 Introduction

1.1 Motivation

With the establishment of sophisticated computer models for studying electrophysiological and biomechanical phenomena of the heart, patient-specific treatment planning in the clinical routine seems to be within reach. Personalized numerical simulations of pathological cardiac behavior and different treatment scenarios can support physicians in selecting the optimal therapeutic approach [1,2]. However, simulations are still quite time-consuming – especially when personalization is required. Indeed, for the estimation of parameters from clinical data, a rule of thumb indicates that the time needed to solve this inverse problem is proportional to the number of parameters multiplied by the time needed to run the forward simulation. To this end, it is important to find a reasonable compromise between accuracy and speed of the simulation. In finite-element-based cardiac simulations, this can be achieved on the one hand by proposing reduced models of the physiology [3] or models with better numerical schemes. On the other hand, the

discretization of the model domain in space and time [4,5] can be adapted in accordance with the scale of the underlying computational model.

For patient-specific purposes the adaptive spatial discretization (mesh) is particularly challenging due to high geometric variations of anatomical structures especially in pathological cases. The mesh should be able to resolve all relevant geometrical details with sufficient accuracy. There are two main aspects for improving the computation speed: (i) the local resolution of the mesh should be adapted to the (a-priori or a-posteriori) error made in the computations in order to reduce the total number of elements; (ii) the type and shape of the elements should be optimized to improve the convergence properties of the iterative FE solver. Details about the mathematical relations between element shape and FE matrix conditioning (and thus convergence speed) can be found in [6]. Hence, mesh generation is a multi-criteria optimization problem, where a reasonable trade-off between anatomical accuracy, mesh resolution and element quality is desired.

1.2 Contributions

Commonly used mesh elements are hexahedra and tetrahedra. However, the decomposition of an arbitrary shape into hexahedra is currently more challenging than into tetrahedra. As in personalized simulations arbitrary complex shapes may appear, we will focus on tetrahedral meshes in this work.

The *first* contribution of this paper is a pipeline for generating isotropic tetrahedral meshes that can achieve such a trade-off. As input a segmented medical image and a user-defined sizing field should be provided. Here, we do not address the issue of anisotropic mesh generation, which is subject to future work. Our pipeline consists of well-studied components which can be exchanged if needed. It is described in Sec. 2.

Our *second* contribution is the application of the pipeline to specific problems in cardiac simulations: the electrophysiological simulation of abnormal electrical activity (ventricular tachycardia) and the electromechanical simulation of congenital heart defects (Tetralogy of Fallot). Our main aim is to provide information about the mesh requirements of these specific applications. In particular, we aim at studying the effects of the mesh resolution upon the simulations in order to find the best trade-off between simulation accuracy and speed. Our preliminary results show the effectiveness of our meshing pipeline to produce spatially adapted meshes suitable for these applications. The results are described in Sec. 3.

2 Adaptive Tetrahedral Mesh Generation

There exists a plethora of approaches for generating volumetric meshes with different types of elements. It is beyond the scope of this paper to give an overview here. Instead we refer to a recent result in this field [7] and references therein. Among the most commonly employed elements in simulation applications are isotropic tetrahedral elements, because they are geometrically less rigid than hexahedral meshes, and easier to generate from segmented image data. There exist many software systems for generating such meshes, which are available for both academic

and commercial purposes. A rather comprehensive compilation can be found in the meshing research corner [8]. Such systems mostly take as input a triangle mesh of the surface of the given shape.

Hence, our proposed meshing pipeline (Fig. 1) consists of three steps, which are described in the following subsections: (i) From a given segmented image a triangle mesh is generated, which accurately reflects the input shape. (ii) The density of the triangles of this mesh is adapted to a given sizing field (in units of edge length), while optimizing the quality of the elements and at the same time preserving the geometry of the shape. (iii) The resulting triangle mesh is used as input for the tetrahedral mesh generator, which retains the remeshed surface as the boundary of the resulting tetrahedral mesh.

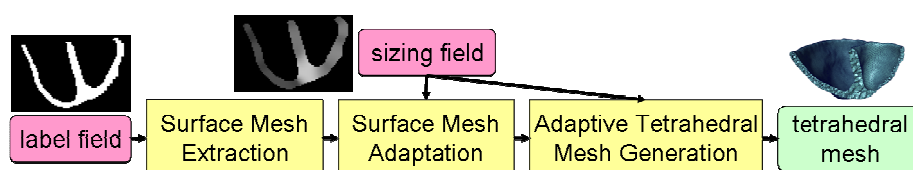


Fig. 1. Volumetric mesh generation pipeline with local mesh size control

2.1 Surface Mesh Extraction

The most commonly used method used to generate an iso-surface from segmented image data is the marching cubes (MC) method [9]. We use this method in our pipeline because it allows computing an accurate representation of the input shape. Additionally we apply shape-preserving smoothing [10] in order to remove non-anatomical staircase effects from the resulting mesh (see Fig. 2 left for a result). While the MC method has no significant parameter the smoothing parameters need to be adjusted to the smoothness of the anatomical shape.

2.2 Surface Mesh Adaptation

In this step, a user-given 3D sizing field is sampled onto the MC surface. The MC surface is remeshed using the algorithm proposed in [11] such that its edge lengths reflect the absolute values of the 3D sizing field. Simultaneously, the mesh quality is optimized to obtain triangles that are as equilateral as possible. This is done via local mesh operations (vertex movement, edge collapse/contract/flip), that can be constrained so that the geometry of the input surface is preserved up to a given tolerance. This tolerance as well as the weighting between sizing and quality can be adjusted to the requirements of the application, if needed. Since the algorithm interprets the sizing field only as a density relative to a user-given number of vertices, we pre-process the MC surface to adjust the number of vertices to approximately yield edge-lengths which equal the absolute values of the 3D sizing field. To this end we coarsen the MC surface globally to edge lengths equal to the maximal value of the 3D sizing field using the same method [11], and then refine it locally (via simple 1:4

subdivision) to match the given surface sizing field. The final remeshing process then basically accounts for mesh quality optimization since the sizing is already well-adapted. Due to the local optimization scheme of the method this also speeds up the algorithm considerably. Fig. 2 shows an exemplary input/output on a myocardium shape. The surface distance between input/output is on average 10% of the voxel size (1.5 mm), with a maximum value of one voxel size. This provides a good trade-off between sizing, quality and shape approximation.

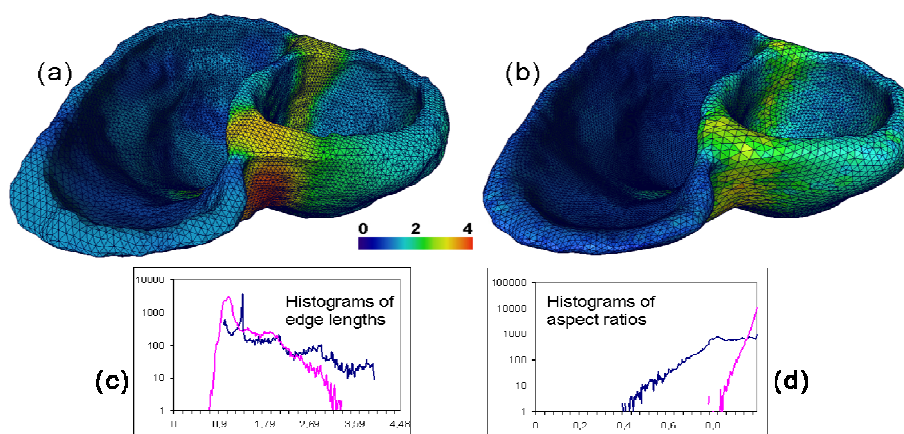


Fig. 2. From initial marching cubes surface mesh to high-quality adaptive surface mesh:
 (a) Initial surface mesh with prescribed metric (color indicates edge length in mm)
 (b) Remeshed surface with mean edge length per triangle
 (c) Histogram of prescribed metric (blue, mesh (a)) vs. remesh metric (purple, mesh (b))
 (d) Aspect ratio histogram of initial (blue, mesh (a)) vs. remesh surface (purple, mesh (b))

2.3 Adaptive Tetrahedral Mesh Generations

Based on the previously adapted triangle mesh the goal is to fill the interior of the shape with tetrahedra whose edge lengths approximately match the given 3D sizing fields, constrained by the already adapted surface mesh. We evaluated five different existing software systems by benchmarking them in terms of mesh quality and computation time on different input surfaces (3 myocardium & one aorta shape). We included three systems with (Tetgen, Netgen, Tetmesh) and two systems without (Geopack, Amira) local sizing capabilities, since built-in global grading parameters still can lead to useful results, given the previously adapted surface mesh. The results are summarized in Table 1. The quality is given by the minimum dihedral angle over all elements and all four meshes. The time is specified relative to the fastest method (in our benchmark: Tetmesh) as an average value over all times relative to the number of input surface vertices. More details (including absolute timings) can be found in [15]. We identified Tetgen to yield a reasonable compromise between speed and quality. Furthermore, Tetgen allows local size control and it is available as source code and thus easy to integrate into simulation codes, for example. There is only one relevant parameter (targeted quality), which is set to 1 in our experiments.

Table 1. Results of Tetrahedral Meshing Benchmark.

	Tetgen	Netgen	Tetmesh	Geopack	Amira
Local sizing	yes	yes	yes	no	no
Availability	free, source	free, source	commercial	free, binary	commercial
Quality [deg.]	6	6	7	12	6
Time [rel.]	4	60	1	10	39

3 Application to Cardiac Simulations

A common approach in cardiac modeling and simulations consists in segmenting the patient myocardium from clinical images and in converting the segmentation into a suitable volumetric mesh. Since the cardiac anatomy can vary tremendously between patients, tetrahedral meshes are usually preferred to hexahedral meshes, which are more complex to define. Furthermore, trabecular structures and papillary muscles of the myocardium are often neglected in these simulations. For the meshes generated in this section *all* parameters of the meshing pipeline discussed in Sec. 2 are fixed.

3.1 Electrophysiology

We first apply our meshing approach to cardiac electrophysiology, more specifically in the case of the simulation of abnormal heart activity known as ventricular tachycardia (VT). We present a case of ischemic VT, where the complex geometry of scar tissue within the myocardium can create a special pathway (isthmus). Such isthmus has been shown to be a substrate for sustained VT, as it creates a reentry circuit, where a depolarization wave can rotate around the scars. The aim here is to be able to test on a model if VT can be induced on a given patient-specific mesh, including the scars imaged with late-enhancement Magnetic Resonance Imaging. We use a simple ionic model (Mitchell-Schaeffer [12]) in the monodomain formulation. To capture the steep slope of potential during the depolarization phase, such reaction-diffusion equations usually require a small mesh size on the order of 0.1 mm. In the case presented below, we are not interested in capturing the details of the depolarization but in the propagation pattern (i.e. the conduction velocities) of the propagating wave around the scars. Therefore much larger edge lengths are considered below.

As a first experiment we generate a 3D sizing field with a metric of 0.8 mm within the scar/isthmus region and 2 mm in the rest of myocardium, and some smooth gradation between the regions. The prescribed metric is shown in Fig. 3a. The cross-section through the resulting tetrahedral mesh (Fig. 3b) reveals a high density of elements in the scar/isthmus region (total number of 622k elements). A uniform mesh with edge lengths of 0.8 mm everywhere has more than 2 million elements, thus adaptivity drastically reduces the degrees of freedom (and hence the simulation time).

We compare this adaptive mesh in a simulation scenario (fixed simulation parameters: only the mesh is different) to a uniform mesh with an average edge length of 3.8 mm (total number of 44k elements), which is previously used in personalization studies.

For the simulation each tetrahedra is assigned an electrical conductivity (scar=0 vs. myocardium=3 S/m) by a spatial lookup into the original segmented image. The adaptive mesh (Fig. 4b) naturally captures the true geometry of the scar tissue (Fig. 4c) much better than the uniform mesh (Fig. 4a). For example, the uniform mesh does not reproduce the isthmus in the upper region, while it is visible in the adaptive mesh. Fig. 4 also shows the activation times (AT) isochrones of the resulting simulations. Qualitatively, significant differences are visible especially in the main isthmus region, where in the adaptive mesh two separate isochrones meet, which is not the case in the uniform mesh. Such differences are crucial for planning RF ablations in the clinical setting.

Therefore, the next efforts will be directed in two major directions: One is to extract a surface of the scar/myocardium interface, which will lead to a multi-label mesh. Then we will perform experiments on a large range of resolutions for the different regions to find out a good trade-off between accuracy and speed for VT simulation. As ground truth for the simulations we will perform simulations on very fine uniform meshes. Furthermore, we will quantify simulation error using in-vivo measurements.

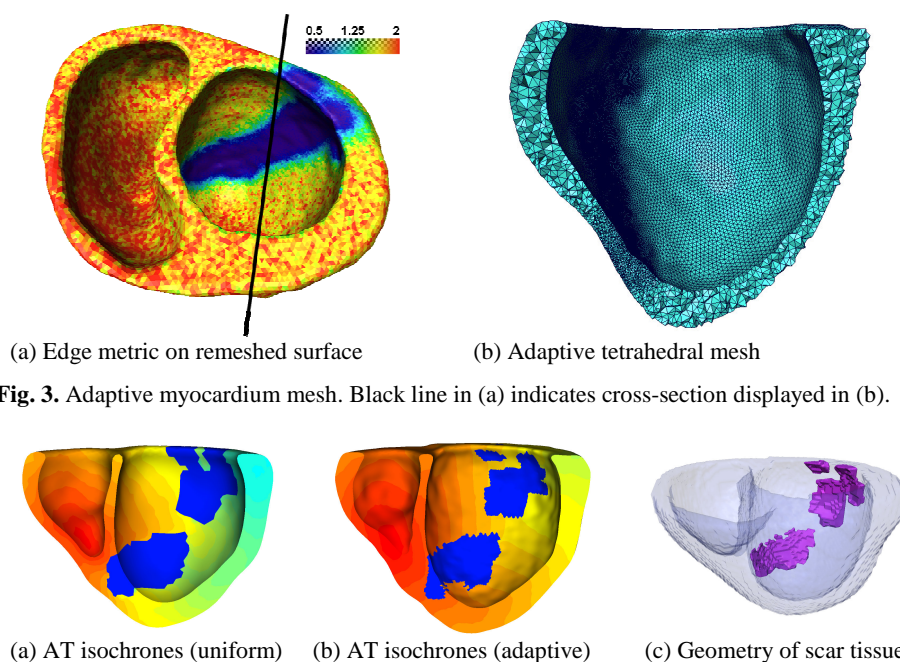


Fig. 3. Adaptive myocardium mesh. Black line in (a) indicates cross-section displayed in (b).

Fig. 4. Simulation results on (a) uniform and (b) adaptive mesh, (c) underlying scar tissue.

3.2 Electromechanics

We now apply our meshing approach on the cardiac biomechanics. As a first application, we study the effects of the discretization upon global clinical parameters such as blood pool volume and pressures, derived from simulated cardiac motions. However, as we do not have any ground truth, we only assess the added-value

provided by a finer mesh with respect to the computation time. We choose a patient with chronic pulmonary valve regurgitations due to an initial repair of Tetralogy of Fallot. Because of the regurgitations, the right ventricle (RV) is extremely dilated and its function is seriously impaired (Fig. 2). In this paper, we make use of the electromechanical model proposed by [3]. All the parameters are personalized by following the strategy described in [13] and kept constant for all the simulations: only the mesh resolution is modified. Patient cardiac function is simulated using four different meshes of the patient anatomy, with increasing number of tetrahedral layers across the wall (from 1 to 4, see Table 2 and Fig. 5) to ensure that biomechanical parameters, such as fiber directions, are equally discretized across the wall. Despite a very thin but dilated RV, our spatially adaptive meshing technique still ensures the exact number of tetrahedral layers, everywhere in the mesh. The meshes are generated based on the thickness [14] of the structure as a 3D sizing field, which is linearly scaled to different ranges for the four different layers (ranges are given in Table 2). The voxel size of the input image data is 1.3 mm.

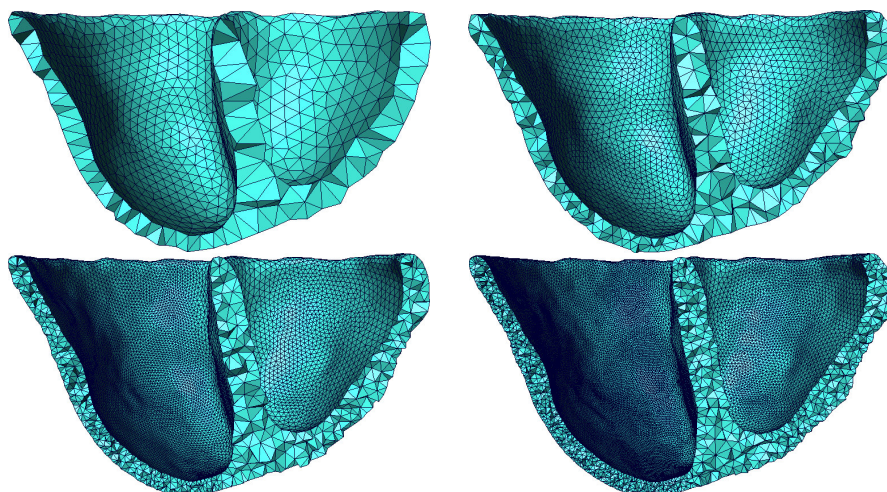


Fig. 5. Four meshes with varying number of layers in the myocardium wall.

Fig. 5 shows a longitudinal view of the meshes. Smaller edge lengths have been used within the RV to account for the very thin wall. Table 2 also provides the computation time for each simulations.

Fig. 6 shows the four pressure-volume loops for both ventricles. For the left ventricle (LV), these simulations clearly point out that one layer is not enough to correctly simulate the cardiac function. This is mainly due to the very coarse discretization of the myocardium fiber directions. Indeed, in our model, each tetrahedron is related to only one fiber direction. As a result, the variations of fiber directions across the wall cannot be correctly modeled. On the other hand, global volume and pressure variations are not significantly different using 2-layer, 3-layer or 4-layer. This suggests that a 2-layer mesh could be used to simulate the LV global function if one is mainly interested in these parameters.

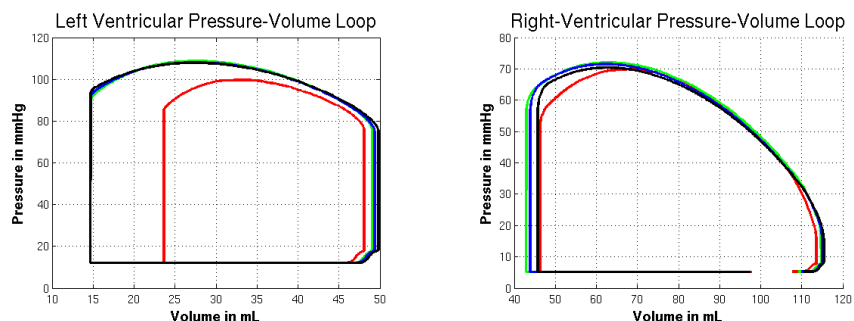


Fig. 6. Pressure-volume diagrams for 1-layer, 2-layer, 3-layer, 4-layer meshes.

Regarding the RV, the conclusions are slightly different due to the thin wall and the dilated blood pool. Surprisingly, no significant differences in the global parameters are found. If confirmed by further experiments, this result may have tremendous consequences in the meshing process. Indeed, we can use in this case another meshing strategy, where we enforce the regularity of the tetrahedra within the whole heart while ensuring at least 2 layers within the LV. This would then constitute a good trade-off between accuracy and speed.

These observations are further verified quantitatively by comparing the simulated ejection fractions (EF) and by calculating the improvements we obtain by increasing the mesh resolution with respect to the increased computational overhead, reference values being computed from the coarsest mesh (see Table 2).

In these experiments, we set the parameters to standard values as the purpose was to demonstrate the effects of mesh resolution. However, validating the simulation against clinical data is ongoing work [1]. Patients with congenital heart diseases undergoing cardiac resynchronisation therapy usually obtain pressure assessment before therapy, data that can be used to validate model predictions.

Table 2. For each mesh the following results are shown: scaled range of the input thickness sizing field; relative values with respect to 1-layer mesh for: meshing time, number of tetrahedral, simulation time, and simulated ejection fractions for LV and RV.

	1-layer	2-layer	3-layer	4-layer
Sizing range [mm]	[4,6]	[1.8,5]	[1,4]	[0.7,2.8]
Meshing time	1 (\equiv 39 sec)	1.6	3.7	9.5
Number of tetrahedra	1 (\equiv 8064)	4.1	18.7	48.2
Simulation time	1 (\equiv 453 sec)	4.6	32.5	94.9
LV ejection fraction	1 (\equiv 51%)	1.4	1.4	1.4
RV ejection fraction	1 (\equiv 59%)	1.1	1.0	1.1

4 Conclusions and Perspectives

We have presented a pipeline for generating spatially-adapted, high-quality, shape-preserving tetrahedral meshes from a given segmented image and a 3D sizing field. We showed its effectiveness on different myocardium shapes. The parameters of the

pipeline balance sizing, quality and surface errors, and are easy to adjust to a given application. Indeed, most parameters may be fixed for a wide range of applications, but a systematic investigation is subject to future work. Further comparison of the prescribed with the actual sizing over a range of different shapes shall also be carried out in the future. Regarding meshing requirements in electromechanical cardiac simulations, combining spatial with temporal adaptivity (e.g. along the wave front propagation in electrophysiology) and anisotropic elements (to better represent fiber directions) are important issues that need to be addressed in the future.

We have shown that spatially adapted meshes are useful to find a compromise between computation time and accuracy in two cardiac applications. Our preliminary experiments demonstrate the need of meshes that are tailored to the specific application requirements. This is particularly important in model personalization, where a large number of repeated simulations need to be performed. We have presented ways to investigate what mesh resolution is best suited for the simulation of ventricular tachycardia and Tetralogy of Fallot. However, these prototype experiments need to be extended in the future to a wider range of scenarios, including different models, parameters, anatomical shapes, pathologies and a larger variety of meshes.

Acknowledgements. This work has been supported the European Commission (FP7-ICT-2007-224495: euHeart). The software ZIBAmira (<http://amira.zib.de>) was used to perform surface extraction, surface mesh adaptation, and thickness computation. Simulations were performed and evaluated using the software MIPS and CardioViz3D (<http://www-sop.inria.fr/asclepios/software/CardioViz3D/> and <http://www-sop.inria.fr/asclepios/software/MIPS/>).

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