

Electric Field Computation in Defibrillation Studies on Realistic Cardiac Geometries

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Abstract

In many computational studies of cardiac defibrillation homogenous electric field distributions are utilized. However, inside the heart, the electric field depends strongly on the underlying cardiac geometry and the configuration of the defibrillation electrodes. Computing this field on an anatomically realistic geometry, however, is non-trivial. The aim of this work is to demonstrate a pragmatic workflow in which COMSOL Multiphysics® is used to import a realistic four-chamber heart mesh, place realistic electrode patterns, compute the resulting electric field density, and export the field for further processing. We use a heart geometry from a human CT-scan, which is publicly available. A central challenge is that COMSOL Multiphysics® does not support the mesh format in which these geometries are distributed. We therefore convert the mesh into the STL surface format, which COMSOL Multiphysics® can import directly. After further processing within COMSOL Multiphysics®, the imported cardiac domain can be used along-side the geometric modelling of the defibrillation electrodes in the Geometry node as well as for the Physics def-initions. Our approach of using COMSOL Multiphysics® lowers the barrier to incorporating anatomically realistic geometries and electrode patterns into defibrillation research.

Keywords: cardiac defibrillation, S-ICD, patient-specific geometry, mesh conversion

1 Introduction

The central tension in working with published simulation geometries is this: on the one hand, the expensive part of the modeling chain – image segmentation, smoothing, and mesh generation – has already been carried out by someone else, validated, and made freely available, so that a geometry of a quality one could not reasonably produce in-house is a download away [1]. On the other hand, what one receives is not a geometry at all but a finished discretization in a solver-specific format: the shape arrives frozen into its own mesh. Whether that mesh can be thawed back into an object which COMSOL Multiphysics® treats like any other geometric entity - one that participates in Boolean operations, parametric sweeps, and selections – is not guaranteed, though.

The application that motivates the present work is the computation of the electric field during cardiac defibrillation. It is not uncommon in computational studies of defibrillation to represent the applied shock as a spatially homogeneous electric field of prescribed strength [2], [3], [4]. For the questions those studies address this is an entirely reasonable simplification, and indeed a productive one: it isolates the interaction between the field and tissue-scale heterogeneities, and it renders the termination

probability a function of a single scalar, so that dose-response curves are well defined [4].

The simplification breaks down as soon as one asks where in the heart a shock is strong enough to act. Inside the heart the field is the solution of a volume conduction problem whose boundary conditions are set by the geometry and placement of the electrodes, and whose coefficients vary with the surrounding medium. Curved surfaces, the finite size and separation of the electrodes, and the conductivity contrast to the medium all shape it. A homogeneous field can say nothing about any of this, which is precisely why anatomically realistic geometries are wanted in the first place.

Two obstacles stand in the way, the first being the missing import: the mesh cohort we use is distributed in the EnSight Gold Case format [1], for which COMSOL Multiphysics® offers no native reader. The second, and the more troublesome one, is that the obvious de-tour - exporting a surface and importing that instead - does not by itself yield a solid geometry object: its defects prevent a successful conversion. In the pre-sent work, we present our approach to deal with these obstacles and show first results of electric field simulations with COMSOL Multiphysics® on such a imported, realistic cardiac geometry.

The paper is structured as follows. Section 2 introduces the setup, namely, the electrode

configuration, the material parameters, and the physics interface, together with the argument that justifies a stationary study for an intrinsically transient shock. Section 3 covers the conversion of the source mesh, the repair of the imported surface, the remeshing inside COMSOL Multiphysics®, and a brief mesh convergence study. Section 4 presents the computed electric field distribution. At last, conclusions and an outlook are provided together with a look at the limitations of our model.

2 Setup

This section specifies the electrode configuration and the excitation, lists the material parameters, and states the physics interface together with the assumptions it rests on.

2.1 Electrode configuration and excitation

We mimic the geometric asymmetry of a subcutaneous implantable cardioverter-defibrillator (S-ICD), in which the shock is delivered between a coil electrode on a subcutaneous lead and planar electrode. The key features can be reproduced with simple geometric primitives, namely, a cylinder of length 80 mm and diameter 3 mm as the coil electrode, and a square plate of 70 mm × 70 mm as the counter-electrode. The corresponding areas are 7.5 cm² and 49 cm². The resulting area ratio of roughly 1:6.5 is what crowds the current into a narrow neck at the coil and thereby sets a field asymmetry across the whole model. The electrodes are positioned according to Figs. 1 and 2 relative to the heart, with a center-to-center separation of about 225 mm.

We apply a potential difference of 1200 V between the two electrodes, which lies within the range of peak voltages delivered by commercial S-ICD devices at maximum output. The cylinder is assigned a fixed potential of 1200 V and the plate is grounded, implemented as a *Terminal* (voltage) and a *Ground* boundary condition on the respective electrode surfaces. The electrodes themselves are not meshed as conducting domains but enter the model exclusively through these boundary conditions, which should be an excellent approximation since the metal conductivity exceeds that of the surrounding medium by many orders of magnitude.

It is worth noting that the assignment of the two terminals is arbitrary as far as the field problem is concerned. The reference level of the potential is pure gauge, and exchanging the terminals maps $V \rightarrow (1200\text{ V} - V)$ and hence $\mathbf{E} \rightarrow -\mathbf{E}$ at every point. Every scalar quantity reported in Section 4 is therefore independent of this choice. We nonetheless

state our convention explicitly - the coil electrode is the anode - and carry it through to the exported data, because the tissue response that the field feeds into is not sign-symmetric [4].

No thoracic structures are included in the present study. Instead, the heart is immersed in a bath of Tyrode's solution, that is, a physiological saline routinely used in *ex vivo* cardiac experiments. Two considerations motivate this choice. First, it avoids introducing a second layer of uncertain material parameters before the geometric workflow itself is established. Second, it makes the setup directly comparable to bath experiments on explanted hearts, in which the field is likewise established between electrodes in a saline bath. The bath is modeled as a 35cm × 35cm × 40cm box surrounding the heart, see Fig. 2. All outer boundaries other than the electrode surfaces are set to *Electric Insulation*.

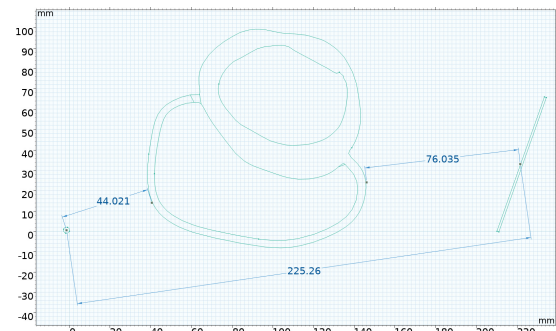


Figure 1. Geometric dimensions and electrode positioning; shown is the cut-plane defined in Fig. 2.

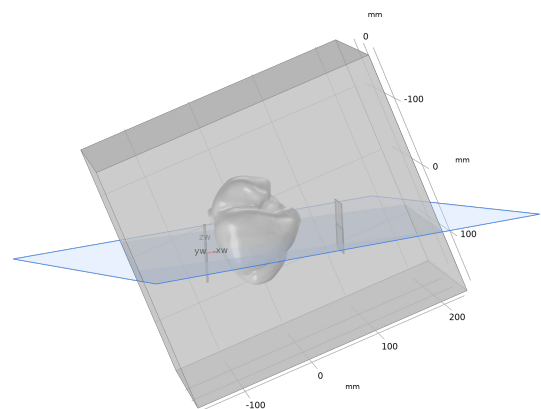


Figure 2. 3D geometry setup with the cut-plane highlighted which is used in Figs. 1 and 7.

Table 1: Material parameters used in our model. The relative permittivities are listed for completeness and do not enter the stationary solution; see Section 2.3.

Domain	ϵ_r (1)	σ (S/m)
Cardiac tissue	40	0.2
Tyrode's solution	71	2.0

2.2 Material parameters

We treat the heart as a single homogeneous, isotropic conducting domain, characterized together with the surrounding Tyrode's solution by a relative permittivity ϵ_r and an electrical conductivity σ ; see Table 1. The myocardial conductivity of 0.2 S/m represents an effective isotropic bulk value derived from bidomain current-flow considerations [5], [6]. The value for Tyrode's solution corresponds to a physiological saline of the composition used in [6], [7].

We remark that the relative permittivities of Table 1 do not enter the stationary solution at all: the computed field depends only on the conductivities and on the geometry. They are listed because (a) COMSOL Multiphysics® enforces the input and (b) they become active parameters in the time-dependent extension discussed in the outlook.

2.3 Implemented physics

We solve the problem with the *Electric Currents* in-interface of the *AC/DC Module* in a *Stationary Study*. This interface enforces current conservation,

$$\nabla \cdot \mathbf{J} = 0, \mathbf{J} = \sigma \mathbf{E}, \mathbf{E} = -\nabla V.$$

Here, V denotes the electric potential, $\mathbf{E}$ the electric field, $\mathbf{J}$ the current density, and σ the electrical conductivity of Table 1. The equations from above are solved subject to $V = 1200$ V on the cylinder surface, $V = 0$ on the plate surface, and $\mathbf{n} \cdot \mathbf{J} = 0$ (electric insulation) on the remaining outer boundaries. Continuity of the normal current density at the tissue-bath interface is enforced automatically.

One might wonder why a stationary study should suffice for an intrinsically transient shock. This comes down to a separation of time scales. The charge relaxation time $\tau = \frac{\epsilon_0 \epsilon_r}{\sigma}$

evaluates to 1.8 ns for the cardiac tissue and 0.31 ns for Tyrode's solution. Here, ϵ_0 denotes the vacuum permittivity, while ϵ_r denotes the relative permittivity of Table 1. Both values are eight to nine orders of magnitude shorter than the duration of a defibrillation shock, which is of the order of 10 ms. That is,

the field settles almost instantaneously compared to the shock and thus can be treated stationary.

With the formulation considered here, the field distribution scales linearly with the applied voltage. A single stationary solve therefore covers any shock amplitude by rescaling, which is what makes it a useful input for studies that sweep the shock strength [4].

3 Mesh

In this section we trace the path from the published source mesh to the discretization that is actually solved on, concentrating on the two steps that are not automatic, namely, the repair of the imported surface and the remeshing inside COMSOL.

3.1 Source geometry

Our geometry is taken from the publicly available cohort of four-chamber human heart meshes of Strocchi *et al.* [1]. These meshes are generated from computed tomography images and provided as tetrahedral meshes with a target edge length of 1 mm. Of the twenty-four available anatomies, mesh 01 is used throughout (see Fig. 1); it comprises 481 066 nodes and 2 349 414 tetrahedra at an average edge length of 1.06 mm. For present purposes the mesh is simply an externally generated tetrahedral discretization of a geometrically demanding domain - thin walls, high curvature, and a nontrivial topology - and none of the anatomical annotations shipped with it are used.

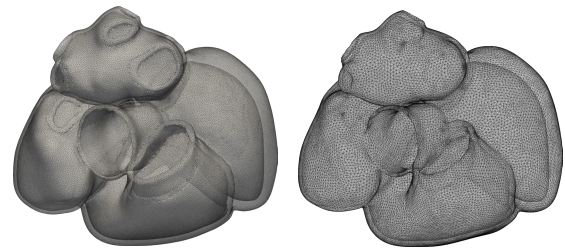


Figure 3. Original mesh (left) and COMSOL Multiphysics® created mesh (right) after importing and repairing an intermediate STL surface mesh.

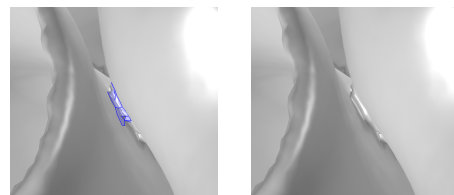


Figure 4. Detail of a surface artefact (left) directly after the STL mesh import. The repair procedure described in the text yields a smooth surface (right).

3.2 Format conversion and repair

Since COMSOL cannot import the EnSight Gold Case format in which the cohort is distributed, we take a two-stage detour. The .case file is opened in ParaView, the boundary surface of the volume is extracted using a built-in filter, and the result is written as a STL surface mesh, which COMSOL Multiphysics® imports directly; see Figure 3 for the original mesh and the resulting COMSOL Multiphysics® geometry side by side.

The import itself is unproblematic. What follows is not. The extracted surface is not a watertight manifold: it contains surface artefacts; see Figure 4 for a detail view of one representative site. Left as they are, they prevent the conversion of the imported mesh into a solid geometry object and hence prevent the domain from being used for material assignment and physics definition at all.

We therefore employ the following strategy to yield a suitable heart geometry. Firstly, *Detect Faces* with a *Maximum Neighbor Angle* of 160° is used to identify the critical faces. Subsequently, they are deleted via *Delete Entities*. The resulting holes in the cardiac surface are healed by the *Fill Holes* operation. Finally, *Create Domains* yields the cardiac muscle domain which in turn is made available in the *Geometry* node via the *Mesh Part* functionality. This is the step that makes the whole detour worthwhile. The imported geometry is no longer a rigid mesh to be worked around but an ordinary geometric object alongside the parametrically modeled electrodes and bath.

Table 2: Statistics of the imported mesh and of the COMSOL Multiphysics® remesh of the cardiac surface used for the simulations. The quality measure is the one reported by the Mesh Statistics node.

Mesh size parameter	Triangles	Quality (Min)	Quality (Avg)
Imported	365 648	0.284	0.802
1.2 mm	254 082	0.421	0.862
1.3 mm	217 482	0.487	0.861
1.4 mm	188 506	0.274	0.859
1.5 mm	164 840	0.535	0.858
1.6 mm	145 434	0.488	0.857
1.7 mm	129 808	0.344	0.855
1.8 mm	116 350	0.512	0.854
1.9 mm	104 844	0.510	0.853
2.0 mm	95 280	0.498	0.852

3.3 Remeshing in COMSOL

The remeshing proceeds in two steps. Firstly, the cardiac surface is discretized with a Free Triangular mesh. Secondly, a Free Tetrahedral mesh is created with the *Element Quality Optimization* level set to *high*. For later usage, we parameterize *Minimum* and *Maximum element size* of both operations with the same mesh size parameter. Table 2 lists the resulting surface mesh statistics of the imported STL mesh and the COMSOL Multiphysics® remeshed one. The remeshing pays off: One obtains a smaller and a better mesh in a single operation.

3.4 Mesh convergence

To check our remeshing, we performed a small mesh convergence study. That is, we repeat our simulation from Section 4 for a mesh size parameter between 1.0 mm and 4.0 mm in steps of 0.1 mm. As functionals of the solution, we monitor the average of the electric field norm over the epicardial surface together with its standard deviation. That is, one measure of the overall field level and one of its spatial inhomogeneity; see Fig. 5.

Both vary remarkably little across the sweep. The average spans about 10 V/m, or roughly 0.3% of its value, and the standard deviation about 8 V/m, or 0.5%. Both increase under refinement, as one expects when a coarse discretization smooths gradients across the strongly curved epicardium. Toward the fine end the curves flatten out: refining from 1.5 mm to 1.2 mm yields a plateau-like structure. The increase for even smaller mesh sizes we attribute to the fact that resolution of the imported surface mesh is reached. Overall, we regard the solution as converged at our working mesh size.

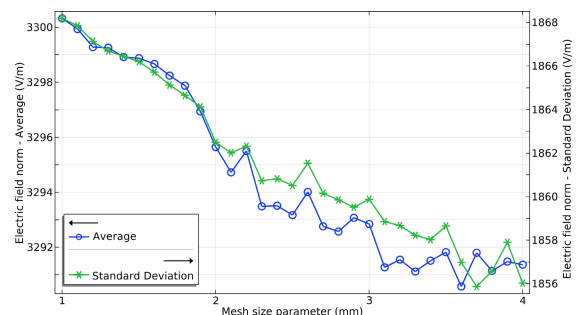


Figure 5. Convergence of the electric field norm on the epicardial surface under mesh refinement. Shown are the surface average (blue circles, left axis) and the standard deviation (green crosses, right axis) as functions of the mesh size parameter. Note the suppressed zero of both ordinates: the total variation amounts to 0.3 % of the average and 0.5 % of the standard deviation.

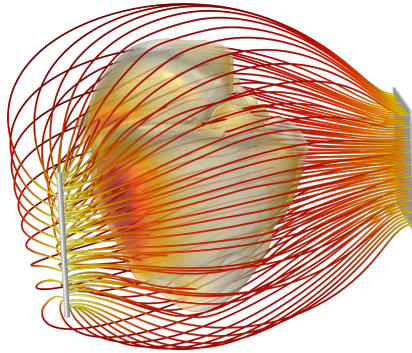


Figure 6. Visualization of the electric field magnitude on the cardiac surface; the semi-transparent rendering shows the field on the outer epicardial surface, with the inner endocardial surface visible through it. The electrodes positioning and resulting field lines are also shown.

The residual scatter - in particular for mesh sizes above 2.0 mm - we attribute to the nature of the refinement. Each mesh size parameter produces an independent tetrahedralization rather than a nested subdivision of its predecessor, so that successive points differ not only in resolution but in element topology, and the discretization error need not decrease smoothly. Accordingly, we take the envelope of the sweep, some 0.3 %, as a realistic bound on the discretization error, rather than the difference between neighboring points. This bound is far smaller than the uncertainty in the material parameters of Table 1 and than the effect of the bath idealization of Section 2.1. The discretization is thus not what limits the accuracy of our model.

4 Simulation Results

Figure 6 shows the computed magnitude of the electric field on the surface of the cardiac tissue for the electrode configuration of Section 2.1. The distribution is smooth throughout and, in particular, carries no imprint of the STL triangulation or of the individual repair operations of Section 3.2. This is the qualitative result that justifies using the converted geometry for quantitative analysis at all: whatever the repair did to the surface, it left no obvious signature in the solution.

Figure 7 shows the electric field magnitude in a cut-plane through the ventricles as defined in Fig. 2. We find an accumulation of high electric field values on the aspect facing the coil electrode. This follows directly from the previously established electrode area ratio of 1:6.5: The small coil electrode yields a field an order of magnitude stronger in its vicinity than the plate does in its own.



Figure 7. Norm of the electric field in the cut-plane defined in Fig. 2.

Turning to the global quantities, the simulation yields a total terminal current of 229 A, corresponding to a lumped resistance of $5.2 \, \Omega$ between the electrodes. This is only about 10% of shock impedance of 50-80 Ω typically measured for implanted S-ICD systems. We attribute this discrepancy to the replacement of the thorax by Tyrode's solution: at 2.0 S/m the bath is roughly a factor of five to ten more conductive than the average of the intervening thoracic tissues, so that both the total current and the fraction shunted around the heart are overestimated relative to the implanted situation: The current flows around the heart in preference to through it. Accordingly, the fraction of the total delivered current that actually traverses the myocardium is expected to be correspondingly small. Similarly, the dissipated electric power in the myocardial domain is only 1.5% of the total dissipated electric power of 274 kW. The absolute field values reported here are thus not transferable to the clinical setting.

Finally, the field is exported in the *Export* node in spreadsheet data format. In the current proof-of-principle approach we are using the plain text format. We export the three signed Cartesian components of $\mathbf{E}$ rather than the norm. This is essential rather than cosmetic, since the sign of $\mathbf{E}$ determines which face of a conductivity heterogeneity acts as a virtual cathode and which as a virtual anode [4]; exporting $|\mathbf{E}|$ would discard exactly the information that the tissue model requires.

5 Conclusions and Outlook

In the present work we have established a workflow that turns an externally generated tetrahedral mesh into a COMSOL geometry suitable for a field computation. We have converted a source mesh

published in the EnSight Gold Case format to an STL surface in ParaView, imported it into COMSOL Multiphysics®, and repaired it into a watertight manifold using the CAD Import Module. Via the Mesh Part functionality, we have combined the re-sulting domain with parametrically modeled elec-trodes and used it for the physics definition in the Electric Currents interface of the AC/DC Module. We have found that the computed field distributions carry no imprint of the repair operations and that mesh convergence behavior well-behaved. For the application at hand this should lower the barrier to incorporating anatomically realistic geometries and realistic electrode patterns into defibrillation re-search considerably.

So far, the tissue is modeled as isotropic, whereas cardiac tissue is anisotropic with a conductivity ratio along and across the fiber direction of roughly three to ten. We expect the latter to matter more for the downstream tissue simulation than for the field computation itself, though this is to be investigated in further studies. A transient study becomes of interest once the finite rise time and the biphasic shape of a realistic shock waveform are to be resolved, and it is at that point that the permittivities of Table 1 become active parameters. Since monophasic and biphasic shocks differ measurably in efficacy [4], resolving the shock waveform on a realistic geometry will be included in future investigations.

Another natural next step is statistical evaluation of the field over the tissue volume. Rather than report extremal values, we intend to compute the cumulative distribution of $|\mathbf{E}|$ over the myocardium and thereby quantify what fraction of the tissue experiences a field above the strength required for termination, which we can take from the dose-response curves of Ref. [4]. Locating the low-field regions in space then identifies the candidate sites of defibrillation failure, which is precisely the information a homogeneous field cannot provide.

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