Influence of Tissue Anisotropy on the Distribution of Defibrillation Fields

SA Seitz, G Seemann, O Dössel

Institute of Biomedical Engineering, Universität Karlsruhe (TH), Karlsruhe, Germany

Abstract

The development of new devices used for defibrillation and cardioversion is often supported by numerical simulations of the induced electric potentials and current distributions. The commonly used tools incorporate isotropic models of the tissue properties present in the human torso. A comparative study was conducted to characterize the influence of anisotropic compared to isotropic tissue modeling. Defibrillation shocks with amplitudes of 1000 V and 2000 V were simulated and a set of varying conductivity values and anisotropy ratios was examined.

The inclusion of tissue anisotropy produced significantly smaller values for current density compared to isotropic calculations especially in the myocardial tissue.

1.   Introduction

In clinical practice defibrillators are a widely used technique to bring fibrillating hearts back to normal sinus rhythm. By applying high energetic electrical shocks to the myocardium, the reentrant electrical activation can be terminated. The necessary currents are injected through electrodes placed on the patient's torso in various locations.

Recently, new defibrillators or electrode positions have been evaluated by means of computer simulations. The efficacy of the defibrillation concerning the electrode placement and design can be examined by numerically calculating the generated electric fields and induced currents in torso and myocardial tissue. Yoon et al. showed that an isotropic representation does not completely cover the realistically occuring current flows and potential distributions [1]. The findings of Eason et al. [2] support this theory. The larger conduction along the skeletal muscle fibers that are oriented mainly parallel to the body surface leads the applied currents away from the intended trans-thoracic direction. This may cause wrong assumptions about the effectively induced current distribution in the myocard by the examined defibrillation system. The goal of this work is to determine the influence of tissue anisotropy on the field distribution and current densities.

2.   Methods

A wide range of software packages has been developed to numerically calculate the field distribution in the human body caused by electrodes on the body surface with voltage pulses. The frequencies used for defibrillation and cardioversion are in the low kilohertz range. The signals examined here can be regarded as electro-quasistatic. This allows the use of Poisson's equation (1) with conductivity tensor σ, potential distribution Φ and the current density $\mathbf{J}$.

$$\mathbf{J} = \sigma\mathbf{E} = \sigma(-\mathrm{grad}\Phi) \tag{1}$$

The finite element method is used to implement the system matrix of this equation describing the anisotropic electrical properties of the tissue. The numerical computing package PETSc is than used to iteratively calculate the inverse of the system matrix [3]. The sparse system matrices were preconditioned using Jacobi's method, which produced input for the GMRES (Generalized minimal residual method) solver. The electrical potentials at the electrodes were integrated into the system as boundary conditions. All computations to determine the defibrillation effects were done on a subset of the Visible Human dataset [4] with a resolution of 2 millimeters. It is completely segmented in the region of interest and identifies 23 tissue types and additionally provides fiber orientation information for skeletal and myocardial muscles [5].

A literature survey on electrical properties of tissue resulted in varying information regarding the conductivity parallel and perpendicular to the fibers in skeletal and myocardial muscles (see Table 1). Rush et al. [6] determined an anisotropy ratio of 1:15.33 in skeletal muscles while Burger et al. [7] measured only 1:2.76. To examine the effects of anisotropy the results were compared with simulations based on an isotropic setup.

The isotropic conductivities were derived from the anistropic ones under the following assumption: The parallel and perpendicular conductivity σ_l and σ_t form an ellipsoid of volume A. The tensor of isotropic conductivities can be interpreted as a sphere with radius r_b. In order to achieve the same global conductivity, the sphere needs to

Figure 1. Thorax of the Visible Man dataset with applied electrode patch.

Figure 2. Conductivity measurements of electrode gel in the frequency range of 50 Hz to 1 MHz.

have the same volume as the ellipsoid (for values see table 1).

Table 1. Conductivity values and anisotropy ratios of muscle tissue at low frequencies (perpendicular and parallel to fiber orientation), σ in S/m, T=37 $°$C , h=human, c=canine, b=bovine.

σ	Burger (h) [7]	Rush (c) [6]	Epstein (c) [8]	Gabriel (h/b) [9]
Skeletal				
perpend.	0.148	0.043	0.076	–
parallel	0.408	0.66	0.524	–
isotropic	–	0.108*	–	0.202
Ratio	*1:2.76*	*1:15.33*	*1:6.89*	–
Myocard.				
perpend.	–	0.178	–	–
parallel	–	0.397	–	–
isotropic	–	0.232*	–	0.054
Ratio	–	*1:2.23*	–	–

* computed using the constant volume approach

The fields were applied as potentials on quadratic silver electrodes with a size of approx. 77 cm^2 each and placed on the front and back of the Visible Man model (1000 V/2000 V anterior, 0 V posterior). The reason for this placement was to achieve straight field gradients between electrodes and to maximize the coverage of the heart region (see figure 1). The transition between electrode and tissue was approximated by one voxel layer (2 mm) of electrode gel. For all simulations, the conductivity of the gel was set to σ=1.697 Sm^{-1}, corresponding to physiological saline solution. Additionally, the conductivity has been investigated based on measurements of an electrode gel (*Elektroden Gel* by Sonogel, Germany) with an impedance analyzer (HP4192A, Hewlett Packard, Wilm-
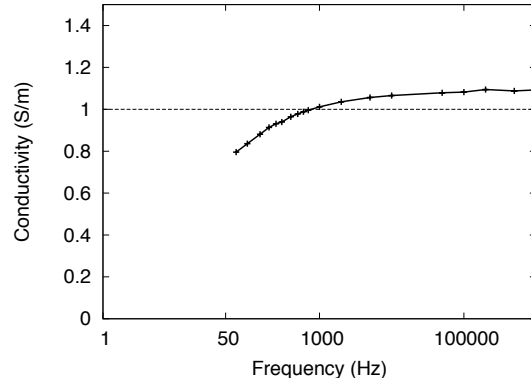
ington, DE) (see figure 2).

The main purpose of the numerical simulations is the evaluation of defibrillation pulses with a duration of few milliseconds. The signals are discretized in time and the potential distribution is computed for every time step. A test of consistency was carried out by comparing two calculations: in the first one two voltages are applied to the electrodes and in the second only one voltage is applied and second solution is calculated by scaling the results according to the ratio of the voltages.

3.   Results

Seeding the defibrillation pulses as an electrical potential in a few voxels within the electrode, the potential discretization was homogenous due to the electrode's high conductivity. Immediately when reaching the gel layer and especially when reaching the tissue layers the potential is clearly attenuated (see figures 7 and 8). The heart is located between x_1=150 mm and x_2=230 mm. In this region, we could observe a relatively constant potential distribution. The high conductivity led to an increased current density compared to the surrounding myocardium. In the latter, significant differences between isotropic and anisotropic calculations were found for both simulated conductivity models (see figure 7).

The presented approach of calculating the isotropic conductive does not match the results achieved with the anisotropic modeling. This effect occurred again in the spinal area due to the high contingent of muscles there. In the anterior (x > 278 mm) and posterior region (x < 4 mm) the calculation domain was modeled as vacuum and therefore the field becomes completely attenuated resulting in an absence of currents.

The *a posteriori* scaling of the potential distribution from an initial simulation showed exactly the same results throughout the whole calculation domain compared to cal-
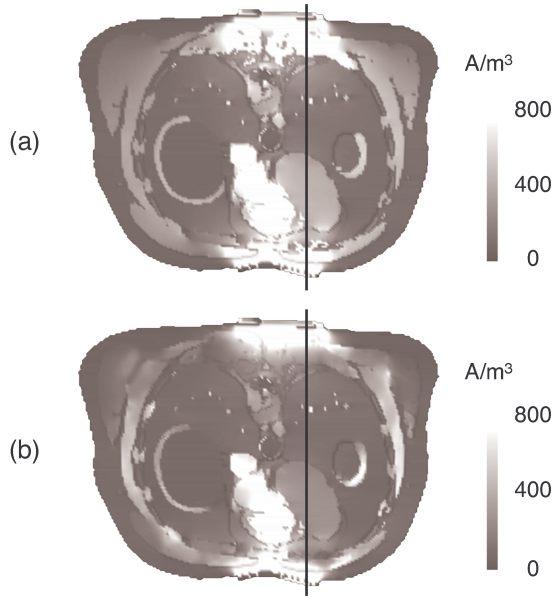
Figure 3. Current density distribution inside the Visible Man model in a heart plane, conductivity values were set according to Rush [6], (a) isotropic, (b) anisotropic model.
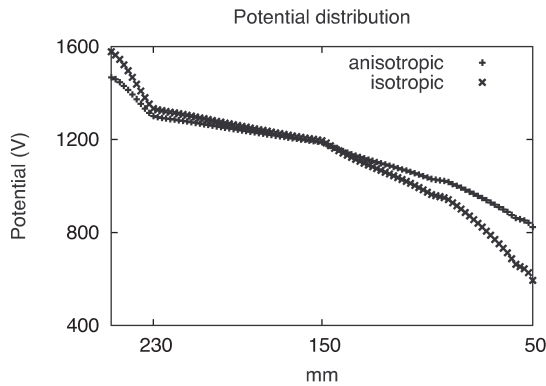

Figure 4. Potential distribution from anistropic and isotropic calculation, extracted at the line indicated in figure 3, conductivity values were set according to Rush [6].
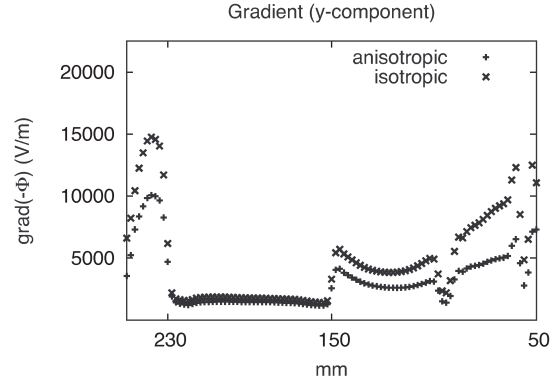

Figure 5. Gradient of potential from anistropic and isotropic calculation, extracted at the line indicated in figure 3, conductivity values were set according to Rush [6].
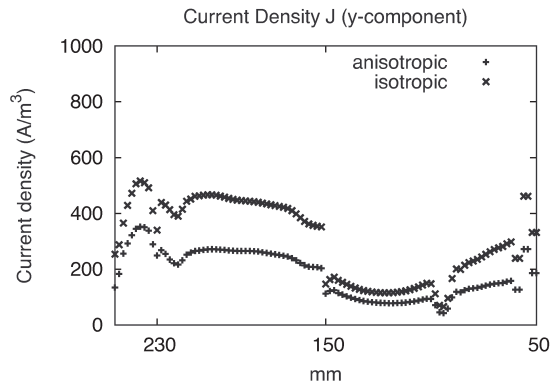

Figure 6. Current density resulting from anistropic and isotropic calculation, extracted at the line indicated in figure 3, conductivity values were set according to Rush [6].

culations with *a priori* adjusted potentials.

Including a more realistic formulation for the electrode gel with the measured conductivity values shown in figure 2 did not produce remarkable effects (see figure 8). This may be attributed to the small thickness of the layer (2 mm).

4. Discussion and conclusions

Incorporating information describing anisotropy in skeletal and myocardial tissue can lead to significantly dif-ferent results compared to isotropic calculations with conductivity values from Gabriel et al. [9]. Especially in the thicker regions of the myocardium, a prominent decrease of current densities could be identified. In contrast to the findings of Karlton et al. [10] we were able to show that even by transforming the anisotropies identified by Burger and others [7, 6, 8] into isotropic conductivity values in a more advanced way, the isotropic simulations only roughly approximate the anisotropic results. As a next step we would like to evaluate our findings with more detailed anisotropy data. The influence of the electrode gel on the other hand can be neglected.

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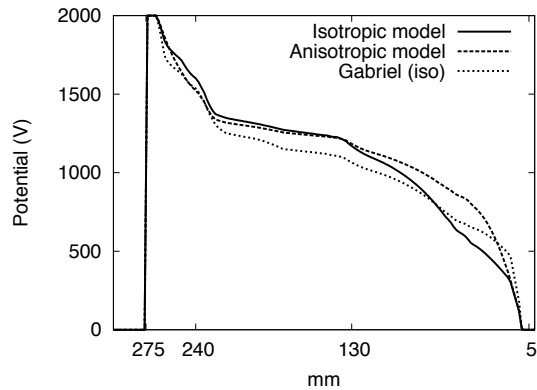
Figure 7. Potential distribution resulting from an anisotropic and isotropic model based on measurements of Rush [6] compared to a model with isotropic conductivity values presented by Gabriel [9].
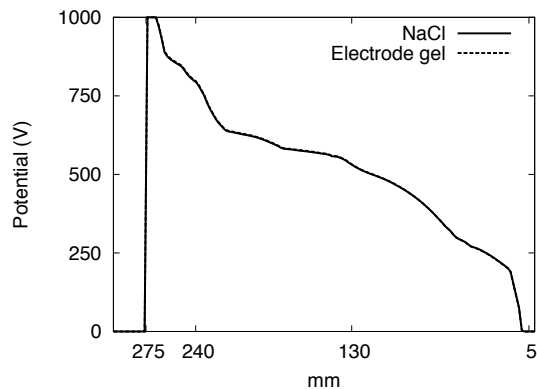

Figure 8. Potential distribution resulting from NaCl and an exemplary commercial product as electrode gel.

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Address for correspondence:

Dipl.-Ing. Sebastian Seitz

Institute of Biomedical Engineering, Universität Karlsruhe (TH)

Kaiserstr. 12, 76128 Karlsruhe, Germany

tel./fax. ++49 721 608 7182/2789

sebastian.seitz@ibt.uni-karlsruhe.de