Inhomogeneous Human Torso Model of Magnetohydrodynamic Blood Flow Potentials Generated in the MR Environment

GM Nijm, S Swiryn, AC Larson, AV Sahakian

Northwestern University, Evanston, IL, USA

Abstract

Magnetohydrodynamic (MHD) voltages resulting from blood flow in a magnetic field contribute to the ECG acquired in the MR environment. These MHD voltages may result in triggering problems for MR image acquisition, since the ECG is typically used for gating. Comsol Multiphysics software was used to model blood flow through the aorta in an inhomogeneous 3D human torso model in a 3.0 Tesla static magnetic field. These voltages were compared with experimentally acquired MHD voltages as well as MHD voltages computed using a simplified torso model. The maximum MHD voltage magnitude was 0.2 mV for the experimental data, 3.04 mV for the simplified model and 0.285 mV for the inhomogeneous torso model. Modeling MHD voltages using an inhomogeneous torso model may aid in optimizing ECG electrode placement for cardiac MRI. In addition, analysis of MHD not only as interference, but also as a physiological signal, may provide blood flow information.

1. Introduction

When a conductive fluid such as blood moves in a magnetic field, voltages are induced orthogonal to both the direction of fluid velocity and the magnetic field lines. This phenomenon is commonly referred to as the magnetohydrodynamic (MHD) effect [1]. As a result of the MHD effect, voltages are generated on the surface electrocardiogram (ECG) during cardiac magnetic resonance imaging (MRI) [2]. Consequently, MHD voltages distort the ECG, which may result in triggering problems for MR image acquisition [3,4]. In addition, since MHD voltages are related to flow, examination of the resulting voltages can theoretically provide information about the blood flow itself.

Maxwell's equations and Navier-Stokes equations fully describe the MHD phenomenon and may be used to compute MHD voltages when the equations are coupled. Maxwell's equations may be formulated as Maxwell-Ampère's law (1), Faraday's law (2), and the two forms of Gauss' law (3,4):

$$\nabla \times \mathbf{H} = \mathbf{J} + \frac{\partial \mathbf{D}}{\partial t} \tag{1}$$

$$\nabla \times \mathbf{E} = -\frac{\partial \mathbf{B}}{\partial t} \tag{2}$$

$$\nabla \cdot \mathbf{D} = \rho \tag{3}$$

$$\nabla \cdot \mathbf{B} = 0 \tag{4}$$

where **H** is the magnetic field intensity, **J** is the current density, **D** is the electric flux density, **E** is the electric field intensity, **B** is the magnetic flux density, and ρ is the electric charge density [5]. Incompressible Navier-Stokes equations may be formulated as the momentum balance equation (5) and the equation of continuity (6) :

$$\rho \frac{\partial \mathbf{u}}{\partial t} - \eta \nabla^2 \mathbf{u} + \rho(\mathbf{u} \cdot \nabla)\mathbf{u} + \nabla p = \mathbf{F} \tag{5}$$

$$\nabla \cdot \mathbf{u} = 0 \tag{6}$$

where ρ is the density, **u** is the velocity field, η is the dynamic viscosity, p is the pressure, and **F** is the volume force vector. Due to the complexity of solving these equations for 3D geometries, finite element methods are typically used to compute the solutions numerically.

We have previously presented a simplified finite element model to compute MHD voltages [6], but the results differed by an order of magnitude relative to experimental data. Consequently, we hypothesized that by making the model more anatomically complete, it may improve agreement between the theoretical model and the experimental data.

2. Methods

2.1. Inhomogeneous torso model methods

Comsol Multiphysics™ modeling software (version 3.4), which implements the finite element method to solve partial differential equations, was used to model blood flow through the aorta in an inhomogeneous human

torso model. The model geometry (Figure 1) was created using the male data from the Visible Human Project [7].

A 3.0 Tesla static magnetic field was simulated to represent the large field (B_0) generated by the MRI magnet. Magnetic field lines ran parallel to the z-axis. Physiologically realistic values were used for the tissue properties used in the model; these are shown in Table 1.

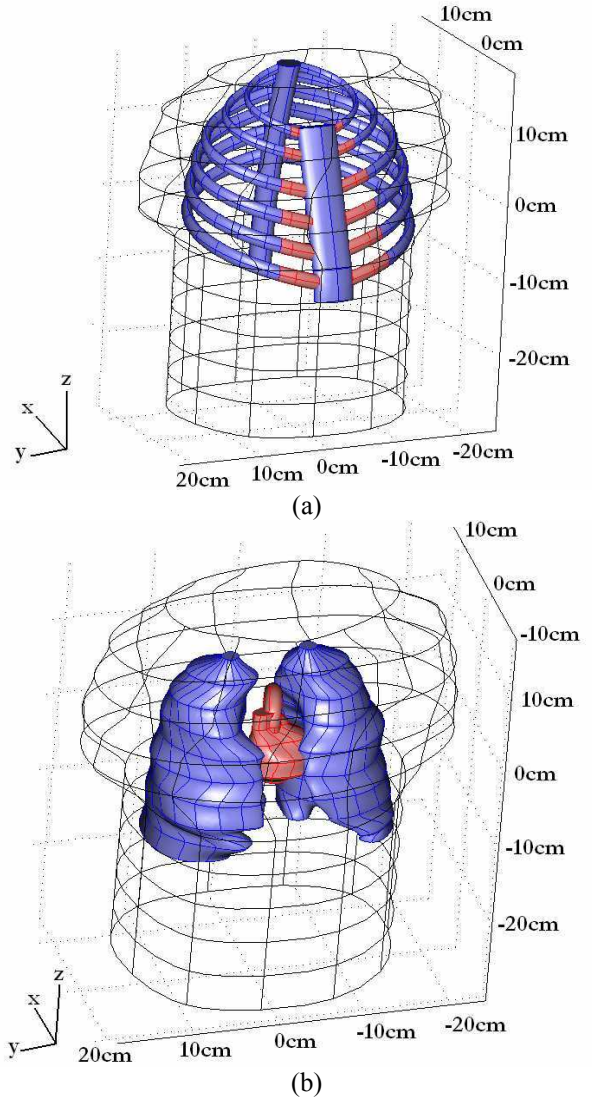

Figure 1: Model geometry. (a) skeletal system and (b) major organs.

The equation used to model the input velocity of blood, v, is expressed as:

$$v(x,y,t) = \left(\frac{R^2 - x^2 - y^2}{R^2} \right) * (3e^{-6t} \sin(2\pi t)) \quad (7)$$

where R is the blood vessel radius, t is time, and x and y are Cartesian coordinates. The first term in this equation causes the velocity to be spatially parabolic, since the center of the input face of the blood vessel is centered on (0,0,0). The second term causes the velocity to vary over time in order to simulate pulsatile flow.

Tissue Type	Conductivity (S/m)	Relative permittivity
Aorta	0.47882	56.01
Bone (cancellous)	0.17983	26.29
Bone (cortical)	0.06733	14.72
Cartilage	0.48828	52.94
Heart	0.76587	84.30
Lungs	0.31554	29.48
Skin	0.52249	65.48
Spine	0.35369	44.08

Table 1: Model parameters for tissue properties [8,9].

MHD voltages were computed using the 3D finite element model by coupling Maxwell's equations, which quantify electromagnetics, and Navier-Stokes equations, which quantify fluid dynamics. An iterative solver (GMRES) was used to compute MHD voltages for 50 ms time steps from 0.0 to 1.0 seconds. The mesh used to compute the solution consisted of 30,627 elements; the model was solved for 282,068 degrees of freedom.

MHD voltages were computed for the entire surface of the torso. The magnitudes of the voltage difference computed from two virtual electrodes separated by 10cm and centered on the fourth intercostal space on the left sternal border were recorded in order to facilitate comparison with the experimentally acquired data (see Section 2.2).

2.2. Experimental methods

In a previous study [11], ECGs were recorded from both inside and outside of an MRI magnet for nine healthy subjects. ECGs were acquired with two electrodes which were spaced 10cm apart and centered on the fourth intercostal space on the left sternal border. The electrode pair was oriented in the tranverse plane. ECGs were obtained with subjects oriented in both the supine and prone positions. The MHD voltages were computed by subtracting the ECGs taken from outside the MRI magnet (no MHD contribution) from the ECGs taken from inside the magnet (including MHD contribution). The result of the subtraction was the MHD signal.

Due to the orientation of magnetic field lines in the MRI magnet and the electrode orientation, MHD voltages resulted primarily from blood flow in the segment of the aortic arch oriented in the sagittal direction. The magnitude and duration of the resulting MHD signals were computed.

2.3. Simplified model methods

In a previous study [6], a highly simplified model of the torso was constructed using Comsol Multiphysics. The torso was modeled as a large cylinder with diameter of 20 cm. The aortic arch was modeled as another cylinder with a radius of 0.1 cm, placed directly in the center of the large cylinder. The torso was assigned a conductivity of 0.2 S/m, which was derived as a weighted approximation of the conductivity of surrounding tissues in the torso [10]. Velocity of the blood was simulated using the same equation (7) as the inhomogeneous model, and MHD voltages were computed for the same time steps as well. MHD voltages were recorded from the outer radius of the uniform volume conductor.

3. Results

3.1. Inhomogeneous model MHD voltages

The resulting model showed that the largest magnitude voltages resulting from the MHD effect occurred in the transverse direction (along the y-axis). In Figure 2, these voltages are shown for several time steps. As expected, the magnitude and direction of MHD voltages varied as the velocity changed over time. Of the calculated time steps, the maximum MHD voltage was 0.285 mV, which occurred in the y-direction at t = 0.15 seconds. The minimum voltage was –0.014 mV, which occurred at t = 0.65 seconds. The zero-crossing occurred in between time steps t = 0.5 seconds and t = 0.55 seconds. Time steps after 0.5 seconds produced voltages in the opposite orientation of those before 0.5 seconds. The MHD voltages observed from the model are plotted vs. time in Figure 3 for all computed time steps.

3.2. Comparison with experimental data

The mean of the maximum MHD voltages observed from the experimental data was 0.2 mV [11], and the maximum from the inhomogeneous human torso model was 0.285 mV. Consequently, the amplitudes of the resulting signals differed by 0.085mV.

3.3. Comparison with simplified model

The maximum MHD voltage magnitude computed by the simplified model was 3.04 mV [6], while the voltage magnitude computed by the inhomogeneous model was 0.285 mV. Therefore, the values of the MHD voltages produced by the simplified model and the inhomogeneous model differed by 2.755 mV.

4. Discussion and conclusions

There was strong agreement between the MHD voltages computed by the inhomogeneous torso model

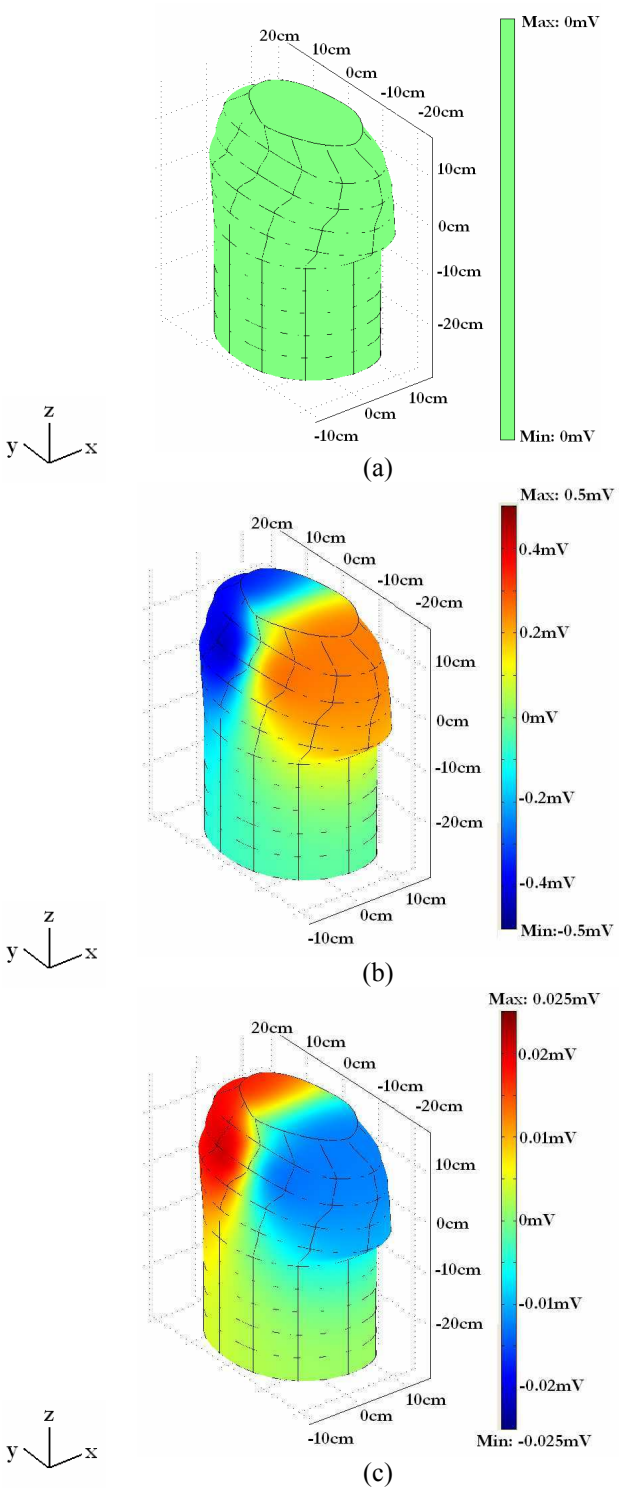

Figure 2: MHD voltages observed from surface of the torso for: (a) t = 0 s, (b) t = 0.15 s, and (c) t = 0.65 s. The bar to the right of each figure shows the color scale corresponding to the different values of voltages.

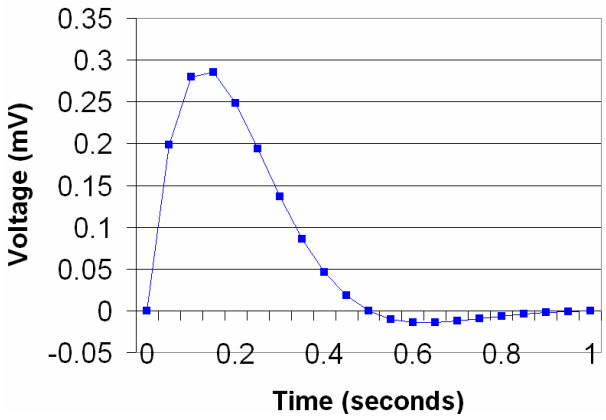

Figure 3: Plot of maximum MHD voltages vs. time observed from the torso surface. Of the calculated time steps, the maximum MHD voltage occurred at t = 0.15 seconds with a value of 0.285 mV, and the minimum voltage occurred at t = 0.65 seconds with a value of –0.014 mV.

and the experimental observations. This result is in contrast to the previous model, which differed from experimental data by an order of magnitude.

The differences between the MHD voltages calculated by the inhomogeneous model and the experimental observations most likely occurred as a result of simplifications of the model and inter-subject variability. Though the model geometry was based upon data from the Visible Human Project, some simplification of the structures was done in order to smooth the surfaces and ease mesh generation. In addition, only blood flow through the aorta was considered in this model, but the experimental observations obviously also included blood flow from other vessels and the heart itself, which may also have contributed to the MHD voltages observed on the surface ECG.

There were several significant differences between the inhomogeneous model and the simplified model. In the simplified model, only two structures were represented, the sagittal segment of the aortic arch and the surrounding tissue which was treated as a uniform volume conductor. Furthermore, the geometries were highly simplified; the uniform volume conductor was a large cylinder, and the aortic arch was a small cylinder in the center of the uniform volume conductor. While the parameters of the blood and velocity of the blood were realistic, the oversimplification of the geometry as well as the representation of so few structures clearly affected the computation of MHD voltages.

In conclusion, by modeling MHD voltages in 3D, we can quantify the distortion of the ECG during cardiac MRI. This information may be useful in selecting optimized lead locations to minimize or maximize (depending on the objective) MHD voltages which appear on the ECG. In addition, modeling of MHD voltages may lead to additional information about blood flow through examination of these voltages.

Acknowledgements

This material is based upon work supported under a National Science Foundation Graduate Research Fellowship, in part by National Institutes of Health grant number NHLBI HL079148, and in part by a grant from the Dr. Scholl Foundation.

References

[1] Togawa T, Okai O, Oshima M. Observation of blood flow e.m.f. in externally applied strong magnetic field by surface electrodes. Med Biol Eng 1967 Mar;5(2):169-70.

[2] Dimick RN, Hedlund LW, Herfkens RJ, Fram EK, Utz J. Optimizing electrocardiograph electrode placement for cardiac-gated magnetic resonance imaging. Invest Radiol 1987 Jan;22(1):17-22.

[3] Tenforde TS. Magnetically induced electric fields and currents in the circulatory system. Prog Biophys Mol Biol 2005 Feb-Apr;87(2-3):279-88.

[4] Tenforde TS, Gaffey CT, Moyer BR, Budinger TF. Cardiovascular alterations in Macaca monkeys exposed to stationary magnetic fields: experimental observations and theoretical analysis. Bioelectromag 1983;4(1):1-9.

[5] COMSOL Multiphysics™ Software and Manuals, www.comsol.com.

[6] Nijm GM, Swiryn S, Larson AC, Sahakian AV. A 3D model of magnetohydrodynamic voltages: comparison with voltages observed on the surface ECG during cardiac MRI. Computers in Cardiology 2007;34:45-48.

[7] Visible Human Project, National Library of Medicine, Bethesda, MD, USA.

[8] Gabriel S, Lau RW, Gabriel C. The dielectric properties of biological tissues: II. Measurements in the frequency range 10 Hz to 20 GHz. Phys Med Biol 41:2251-2269;1996.

[9] Gabriel S, Lau RW, Gabriel C. "The dielectric properties of biological tissues: III. Parametric models for the dielectric spectrum of tissues." Phys Med Biol 41:2271-2293;1996.

[10] Kinouchi Y, Yamaguchi H, Tenforde TS. Theoretical analysis of magnetic field interactions with aortic blood flow. Bioelectromagnetics 1996:17(1);21-32.

[11] Nijm GM, Swiryn S, Larson AC, Sahakian AV. Extraction of the magnetohydrodynamic blood flow potential from the surface electrocardiogram in magnetic resonance imaging. Med Biol Eng Comput 46(7):729-733;2008.

Address for correspondence

Prof. Alan V. Sahakian
EECS and BME Departments
Northwestern University
2145 Sheridan Road
Evanston, IL 60208, USA