Three-dimensional Finite Element Modelling of Current Density in Maternal Transthoracic Defibrillation

Aleksandar Jeremic[1], Elham Khosrowshahli[2]
[1]Department of Electrical & Computer Engineering
McMaster University, Hamilton, ON, Canada
[2]Department of Biomedical Engineering
McMaster University, Hamilton, ON, Canada

***Abstract*— Although resuscitation during pregnancy is relatively uncommon and rarely cause deaths they have a particularly large impact in terms of the mortality of the unborn child and long-term effects on families and society as whole. Although modelling of current density distribution and conductive anatomy of the human transthoracic defibrillation has been subject of certain research interest the attempts to model this phenomenon in pregnant women using 3D finite element have been nonexistent. In this paper, we present simplified 3D finite element model of a pregnant female torso and calculate current density distribution in the uterus. We evaluate the current density a function of voltage difference in order to obtain maximum allowed voltage. The purpose of our study therefore is to evaluate maximum possible voltage for a given position of electrodes which can still be considered safe for a fetus.**

I. INTRODUCTION

Although it has been acknowledged that the cardiac arrest in pregnancy is a rare event [1] it can have significant impact in terms of age of mother, mortality of unborn children (especially with potential loss of two lives) and consequently long-term effect on a family. One of the commonly used procedures in resuscitation is defibrillation which is routinely used for treating ventricular arrhythmias.

With recent advances in understanding pathophysiologies behind electrical shock in pregnant women [2] it became more obvious that previous studies of current conduction in human body should be extended to account adequately for changes in maternal body that affect conduction pathways. Namely it has been shown [3] that physiological changes in pregnancy affect transthoracic impedance and thus affect transmyocardial current which depolarizes heart (myocardium) as a part of resuscitation. However due to the physiological changes (e.g. size of uterus, increased intra- and extra-cellular fluid, increased blood volume, increased thoracic volume, and presence of amniotic liquid) the transthoracic impedance changes may affect current pathways in an unpredictable way.

In this paper we present a three-dimensional simplified model for finite-element analysis of maternal transthoracic defibrillation. In this procedure an electrical pulse is applied to the torso through electrodes commonly called paddles. One of the most important aspects is the energy or current density generated on the surface (aforementioned transthoracic current) and corresponding current density in the heart (transmyocardial current) which needs to be above certain threshold, sufficient for stimulation of myocytes that are inexcitable [4]. Three-dimensional models of human defibrillation have been previously studied and quite extensive list of previous reports is listed in [6]. However to the best of our knowledge this is a first attempt to model defibrillation in pregnancy. In this preliminary work we propose the simplified model in which the uterus and stomach are modelled as a single area with larger conductivity. In order to account for frequency dependent properties of biological tissues we decompose the biphasic pulse into frequency component and perform frequency-domain analysis resulting in corresponding current harmonics. We then calculate the amplitude of the current density harmonics in the lower abdomen and analyze these values with respect to position of electrodes and/or energy delivered by defibrillator.

The organization of the paper is as follows. In Section 2 we present frequency dependent mathematical equations and electric properties of physiological tissues. In Section 3 we discuss finite-element implementation using COMSOL. In Section 4 we present numerical results. Finally in Section 5 we discuss our results and future research directions.

II. MATHEMATICAL MODEL

In order to describe the mathematical model of the defibrillation we first describe the electric pulse that is applied to the patient's torso. Commonly used defibrillators commonly apply rectilinear biphasic pulse which consists of constant current pulse followed by truncated exponential decay pulse. In Figure 1 we show typical waveform of a rectilinear biphasic form for energy of 120 KJ into 50 Ω load (common value for standardly used defibrillator pads).

Due to the time-dependant nature of the problem the current propagation is governed by Maxwell's equations

$$\nabla \times \vec{E}(\vec{r},t) = -\frac{\partial \vec{B}(\vec{r},t)}{\partial t} \tag{1}$$

$$\nabla \times \frac{\vec{B}(\vec{r},t)}{\mu} = \vec{J} + \varepsilon \frac{\partial \vec{E}(\vec{r},t)}{\partial t} \tag{2}$$

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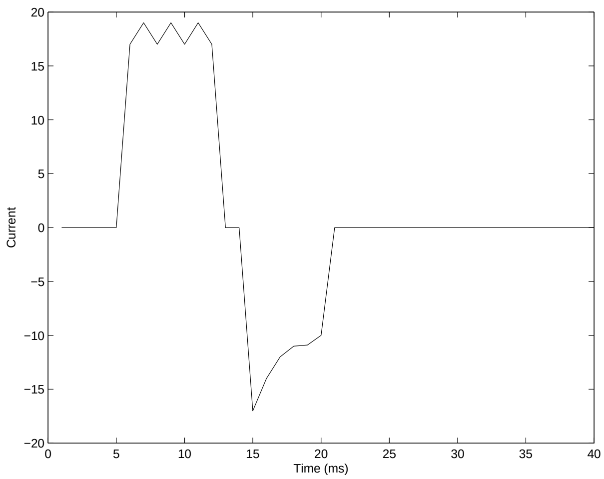

Fig. 1. Time waveform of rectilinear pulse

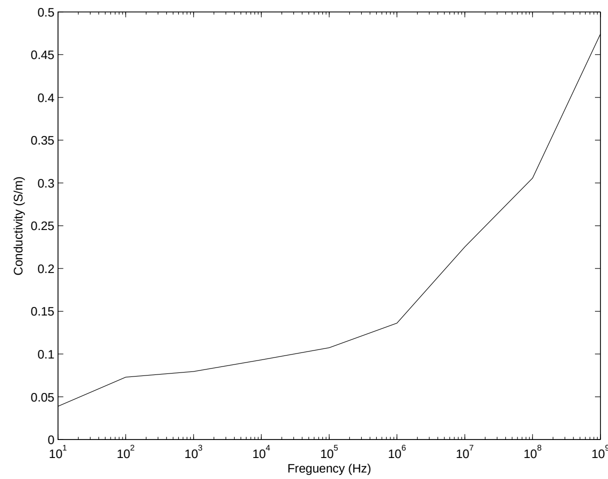

Fig. 3. Average conductivity of lungs

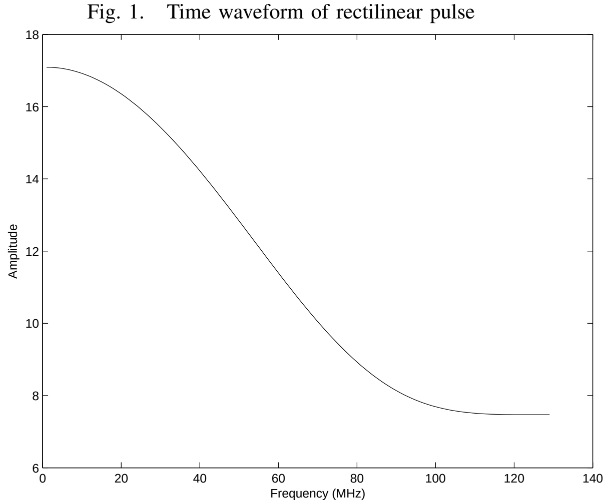

Fig. 2. Short-time Fourier transform of the pulse

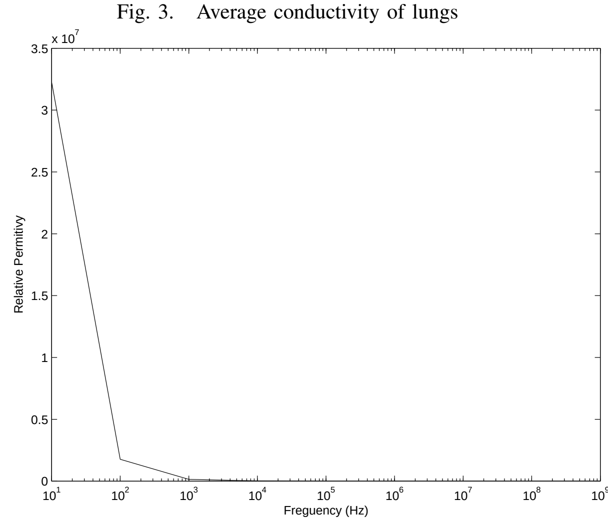

Fig. 4. Average permittivity of lungs

Note that in the above equations the electric properties (conductivity and permittivity) depend on frequency. In this paper we use several different types of tissue: left and right lung, interstice, diaphragm, equivalent lower torso and skin/fat tissue. We approximate tissues of the abdominal compartment (liver, gut, spleen, uterus) using a homogeneous equivalent tissue with slightly larger conductivity. For illustrational purposes in Figures 3 and 4 we illustrate frequency dependence of lung conductivity and permittivity. Similarly Figures 5 and 6 show same values for muscles. These values were selected from reported estimates of tissue properties. [5].

It has been shown [7] that using a piecewise homogeneous torso model consisting of the closed surfaces $S_i, i = 1, \ldots, M$ can be obtained in the analytical form. Let σ_i^- and σ_i^+ be the conductivities of the layers inside and outside S_i respectively. We will denote by G_i the regions of different conductivities, and by G_{M+1} the region outside the torso, which behaves as an insulator. Therefore, in our model $\sigma_{M+1}^- = \sigma_M^+ = 0$. The magnetic field and potentials are then given by

$$\begin{aligned}
\vec{B}(\vec{r},t) &= \vec{B}_0(\vec{r},t) + \frac{\mu_0}{4\pi}\sum_{i=1}^{M}(\sigma_i^- - \sigma_i^+)\int_{S_i}\phi(\vec{r}',t)\frac{(\vec{r}-\vec{r}')}{\|\vec{r}-\vec{r}'\|^3}\times \mathrm{d}S(\vec{r}'),\\
\vec{B}_0(\vec{r},t) &= \frac{\mu_0}{4\pi}\int_G \frac{\vec{J}(\vec{r}',t)\times(\vec{r}-\vec{r}')}{\|\vec{r}-\vec{r}'\|^3}\mathrm{d}^3r', \qquad (3)
\end{aligned}$$

where μ_0 is the magnetic permeability of the vacuum.

Similarly, the potential $\phi(\vec{r},t)$ is given by [7]

$$\begin{aligned}
\frac{\sigma_k^- + \sigma_k^+}{2}\phi(r,t) &= \phi_0(\vec{r})(\sigma_i^- - \sigma_i^+) + \frac{1}{4\pi}\sum_{i=1}^{M}(\sigma_i^- - \sigma_i^+)\cdot\\
&\cdot \int_{S_i}\phi(\vec{r}',t)\frac{(\vec{r}-\vec{r}')}{\|\vec{r}-\vec{r}'\|^3}\cdot \mathrm{d}S(\vec{r}'),\\
\phi_0(\vec{r},t) &= \frac{1}{4\pi}\int_G \frac{\vec{J}(\vec{r}',t)\cdot(\vec{r}-\vec{r}')}{\|\vec{r}-\vec{r}'\|^3}\mathrm{d}^3r', \qquad (4)
\end{aligned}$$

where we use subscript k is such that $\vec{r} \in G_k$.

The above equations are amenable to inverse modelling and signal processing techniques when estimation of certain unknown parameters is needed. In this paper we focus our attention on so called forward modelling in which all

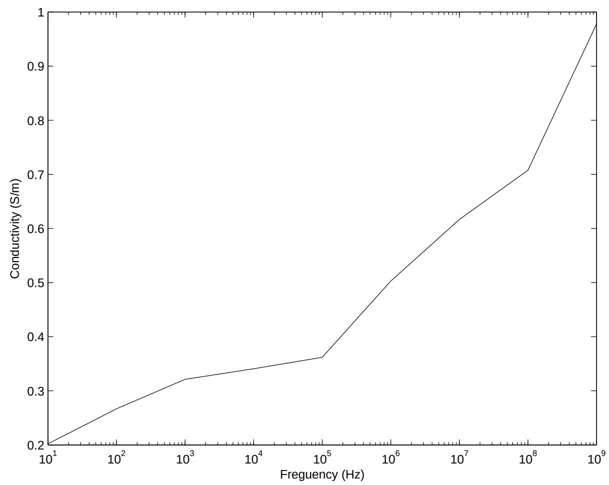

Fig. 5. Average conductivity of muscle

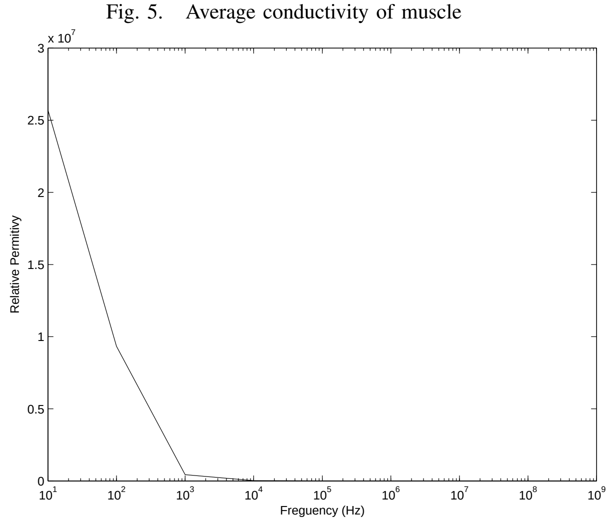

Fig. 6. Average permittivity of muscle

the parameters are to be known. In this case due to the complexity of the geometry it is more beneficial to calculate solution directly using commercially available software such as COMSOL.

III. FINITE-ELEMENT MODEL

In this Section we discuss model construction and solution. As mentioned before in this paper we use a cylindrical model consisting of the following compartments: left and right lung, interstice, diaphragm, equivalent lower torso and skin/fat tissue . We approximate tissues of the abdominal compartment (liver, gut, spleen, uterus) using a homogeneous equivalent tissue with 30% larger conductivity to account for the presence of amniotic fluid.

Defibrillation pads were simulated as groups of nodes on torso surface that were constrained with the same potential. No elements were used to simulate the physical structure of the pads. Solutions were obtained for the potential and current density distribution in the lower abdomen (uterus). In order to model different boundaries within the torso we enforced continuity conditions for all the interior boundaries.

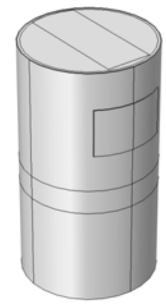

Fig. 7. Model geometry

Fig. 8. Finite-element mesh

To properly model exterior boundary we place the torso model in a large (significantly larger than torso) volume and enforce zero potential on the external boundary. We then apply adaptive meshing in which we enforce very coarse mesh for the sphere which ensures reasonable computational time. The resulting mesh consists of approximately 16000 nodes with 21000 degrees of freedom and was implemented using AC/DC module in COMSOL Multiphysics software.

IV. NUMERICAL RESULTS

n order to illustrate the applicability of the proposed model in Figures 9 and 10 we illustrate the current density in the lower abdomen at frequencies of 1KHz and 1MHz respectively. Observe that the current density at higher frequency is higher than the current density at lower frequencies. This is a consequence of two opposite effects: decreasing power amplitude of biphasic waveform and increasing conductivity at higher frequencies.

To evaluate the effect of the pad position we perform the analysis for two different positions: anterior-posterior (AP) and anterior-apex (AA). For a maximum voltage difference of 1000V (peak-to-peak of the biphasic waveform) we obtain maximum values of current density at 18 mA and 32 mA respectively.

In Figure 11 we illustrate the maximum current density as a function of frequency. These result can be potentially

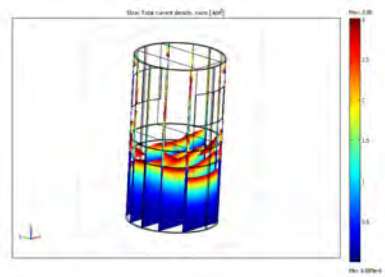

Fig. 9. Current density at 1KHz

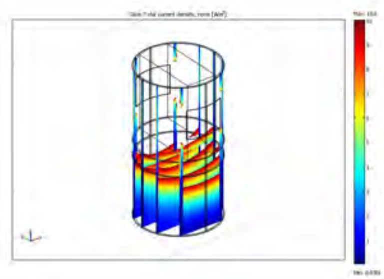

Fig. 10. Current density at 1MHz

used to determine maximum allowed voltage which can still produce myocardial current density above inexcitability threshold.

V. CONCLUSIONS AND FUTURE WORKS

In this paper we presented a three-dimensional finite-element model for calculating current density distribution during pregnancy. We approximated presence of uterus by increasing conductivity of lower abdomen for 30 % to account for the presence of amniotic fluid. These result can be potentially useful for analyzing if energy requirements for defibrillation of pregnant women should be potentially changed. Future research will include more precise geometry with respect to detail resolution (e.g. including uterus as a separate compartment) as well as anatomically correct torso. Our preliminary results indicate that the current density distribution in lower abdomen may become significant since the levels that are considered potentially dangerous for the fetus are low and therefore warrant further investigation of this topic.

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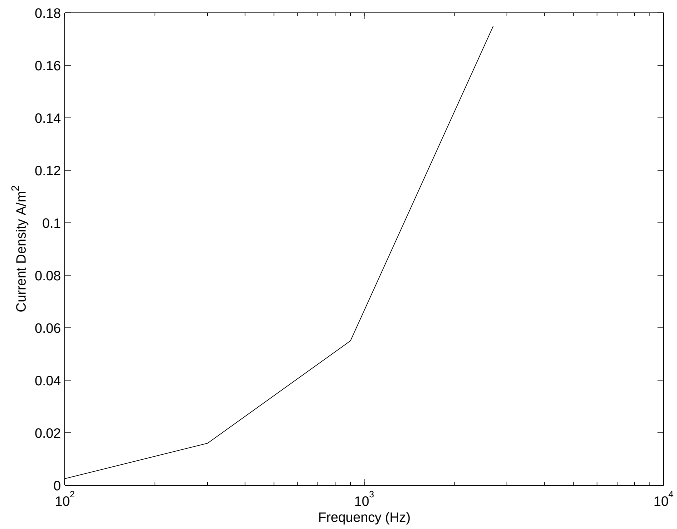

Fig. 11. Current density as a function of voltage

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