Simulation Study of the Potential Hazards of Cardiac Stimulation by Induced Eddy-currents in Modern MRI

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Abstract

In modern MRI, patients are exposed to strong, rapidly switching magnetic gradient fields that may be able to elicit nerve stimulation and cardiac stimulation. Based on a new, High Definition, Finite-Difference Time-Domain (HD-FDTD) method and a realistic inhomogeneous 10-mm-resolution human body model with tissue parameters, this paper firstly provides numerical results of an investigation into induced current spatial distributions inside human tissues when exposed to pulsed z-gradient fields. Then, various cardiac arrhythmia have been simulated based on our realistic heart-torso model and supposed that the induced current reaches the cardiac stimulation threshold. Simulation results show that the induced current is not strong enough to elicit ventricular fibrillation under nowadays several KHz switch frequency of gradient fields, but if raise the gradient field switch frequency further, it is with great risk to elicit ventricular fibrillation.

1. Introduction

Because of their effectiveness in diagnostic medicine and their lower side-effect to patients compared with those of X-ray computer tomography (CT), magnetic resonance imaging (MRI) systems are routinely used in virtually every major hospital in the world. As MRI technology improves, however, patients will be exposed to stronger static magnetic fields, to more intense, higher-frequency radio-frequency (RF) electromagnetic fields, and to more rapidly switched magnetic field gradients, and thus the potential for unwanted side effects of MRI increases. Due to the fast switching of strong gradients, which is often desirable in echo-planar imaging and similar sequences, currents can be induced in excitable tissue to such a degree that may be sufficient to induce peripheral nerve stimulation (PNS) [1-4] or potentially cardiac stimulation [2]. Indeed, there is a considerable literature has reported that PNS occurred in clinical MRI scanner, and a number of patients have complained about sensations arising from MRI pulsed field gradients such as a crawling sensation on the skin and muscle twitching due to peripheral nerve stimulation, etc. For cardiac stimulation, however, no clinical findings have been reported that it was induced by nowadays used MRI machines. But one should pay special attention to cardiac stimulation, as Simunic and Renhart [4] said, since stimulation of the heart muscle could be extremely dangerous for human health. The main reasons are: 1) Stimulation process in a bundle of nerve cells is possible only if suprathresholds are reached in all of them. On the contrary, the response of only heart cell with an action potential enables spreading of excitation over all unexcited fibers. 2) Stimulation or in the worst case ventricular fibrillation of this vital human organ could be fatal and must never occur during MRI. So it is clearly important that a full understanding of the electromagnetic interaction with the patient in MRI scanner to the cardiac stimulation problem, and safety guideline for cardiac stimulation should be developed more carefully.

In this paper, firstly, we apply a new FDTD variant, namely, high-definition, finite-difference time-domain method (HD-FDTD) [5], to calculate the induced fields and currents in a human body model. Special attentions were paid to the region of heart, gradient coil configurations, and source waveforms and frequencies, in order to find out the "worst case", i.e., the case of highest induction current density. Secondly, we use a whole-heart model to simulate cardiac fibrillation under some supposed conditions, trying to estimate the "minimum" stimulation threshold for cardiac fibrillation. Finally, compare the "minimum" stimulation threshold with the calculated highest induction current density to see the potential hazards of cardiac stimulation in MRI scan.

2. Method

A realistic whole human body model is developed based on data from United States Air Force Research Laboratory [6]. The original spatial resolution of the model is 2 mm and the height of the model is 187 cm. For the computations presented here, the model is mapped into a 10-mm grid with volume-averaged dielectric properties. The body model consists of $58 \times 33 \times 187 = 357918$ cells and is utilized within a MRI environment. Tissues with higher water content such as muscle, kidney, heart and liver have larger conductivity than low water content tissues such as bone, fat and lung. Tissue conductivity increases with increasing frequency, however the permittivity decreases with increasing frequency. The magnetic permeability of tissues remains essentially constant with frequency. For the purpose of this study, only Z-coil is considered as shown in Fig. 1. The simplest Z-coil system consists of an anti-symmetric two-loop-structure (Maxwell pair), each of radius r placed parallel and coaxial. The distance between the coils is $d=\sqrt{3}r$. The body model was centred within the gradient set. Based on these model constructs, we use HD-FDTD method to calculate the electromagnetic field. The HD-FDTD technique is based on an effective time/frequency numerical conversion technique that delivers improved efficiency to the solution of the computational model. If the source field consists of frequency components W_i, $i=1,2,\cdots,n$, then a field in the solution space can be represented as

$$f(r,t)=\sum_{i=1}^{n}A_i(r)\sin(w_i t+\varphi_i(r)) \qquad [1]$$

where A is the amplitude of the field, φ is the phase, r is the position vector, and $f(\mathbf{r},t)$ represents the electric or magnetic fields. In general, w can be found by analyzing the field source. At any point in space, if A_i and φ_i were known, a full, temporal field can be determined. For high frequency problems, the time domain solution of the general field $f(\mathbf{r},t)$ can be obtained by first applying the conventional FDTD technique and then using the appropriate Fourier transform to find the amplitude and phase terms. However, for low frequency problems, it is not feasible to run the conventional FDTD technique for a full period. To obtain the solution within a fraction of the source period, a new time-frequency conversion method has been developed. Using the proposed new HD-FDTD technique, only a finite number of solutions are needed in the time domain, and then an inverse approach can be used to calculate A_i and φ_i. This method is based on suppose that the transient response of the initial EM field will die out after a time t_t. If a sequence of instantaneous solutions, $f=[f_1,f_2,\cdots,f_m]^T$, are recorded with respect to times $t_1,t_2,\cdots,t_m,t_1\geq t_t$ and $t_i<t_j$ (if $i<j,\forall i,j\in m$) at a given point in space, r, then a system of non-linear equations arises:

$$\mathrm{Ax}=\mathrm{f} \qquad [2]$$

where the matrix A has the elements

$$\begin{cases} a_{i(2j-1)}=\cos(w_j t_i) & i=1,2,\cdots,m \\ a_{i(2j)}=\sin(w_j t_i) & j=1,2,\cdots,n \end{cases} \qquad [3]$$

The problem can then be solved for each frequency using matrix algebra, resulting in significantly fewer iterations than are required for the conventional FDTD technique. The detail description of HD-FDTD could be found in reference [5].

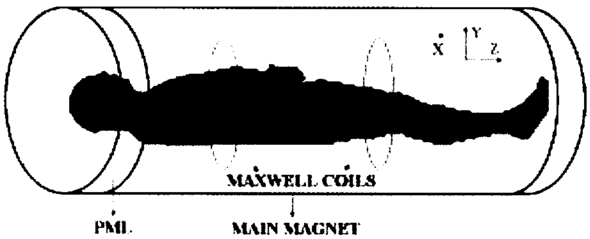


Fig. 1 The model of the human body inside the MRI device. The Maxwell coils are used to generate a gradient magnetic field, and the electromagnetic wave is absorbed by PML boundary conditions on both opened ends.

Next step, the computational domain should be defined. In "tubular" MRI devices, all gradient coils are within the long-cylinder-like main magnetic wall, which can be approximated as a perfect conductor (PEC). Usually the MRI system is open at both ends, so the electromagnetic field generated by the coil radiates into space. However, these numerical simulations can only be performed in finite space and so the region is truncated using a conducting box to enclose the solution volume. Perfected matched layers (PML) are placed at both ends as absorbing boundary conditions (ABC).

In order to estimate the eddy-current induced by the gradient field, the induced electric field in the human body must first be calculated. According to HD-FDTD scheme, the electric field can be found from a simulation run over just a fraction of a period. In the implementation of the algorithm, the modeling space is divided into different types, according to their physical properties, for example: "COIL", "AIR", "PEC", "PML", "BODY". When updating the six vector values (three magnetic field vectors Hx, Hy, Hz and three electric field vectors Ex, Ey, Ez), the coefficients are calculated according to the region type. For example, coefficients of the basis functions found in the "BODY" cells are tissue-dependent and since the human body has complex geometry and structure, approximations of the discontinuous interface boundary conditions are also considered.

After the electromagnetic field is calculated, the eddy current densities are calculated from the following equation:

$$J=\sigma E \qquad [4]$$

where J is the current density vector, E is the electric field vector, and σ is the conductivity of the medium. Here the displacement current has been neglected because it is much smaller than the conduction current.

The magnitude of the current density can be estimated as follows:

$$J=\sqrt{\sum_{j=1}^{m}c_j^{\,2}(J_{xj}^{\;2}+J_{yj}^{\;2}+J_{zj}^{\;2})} \qquad [5]$$

where the computational coefficient c_j is the Fourier coefficients of the source function that means that for a sinusoidal source, where m is the order of the Fourier harmonic expansion.

The heart model used in this study was reconstructed from computer tomography pictures of the human body and heart sections, and composed of approximately 65000 cell units which are arranged in a cubic close-packed structure spaced 1.5 mm apart. The excitation conducting system consists of sinus node, atrial-ventricular node, His bundle, left and right bundles, and Purkinje fiber is set anatomically based. Cell type, conducting velocity, and action potential waveform with variable in duration are assigned to each cell unit. The heart model is then mounted in an inhomogeneous human torso model. A excitation propagation algorithm [7] which enables multi-cycle cardiac arrhythmia simulation is used to produce excitation sequence. Electric dipoles, which are proportional to the spatial gradient of the action potential, are generated in all the cell units. These dipoles give rise to a potential distribution on the human body surface, which is calculated by means of the boundary element method. Detail description of the heart model and the excitation propagation algorithm could be found in published papers [7-8]. Based on this heart model, many kinds of heart diseases could be simulated by setting model parameters.

3. Results

Using the HD-FDTD technique, the induced electric field and local current densities within the whole body especially in some organs of interest can be determined directly. Results show that the eddy-currents can be deep within the body with larger currents occurring close to the coil windings and especially in the chest region. Although peak values are sometimes in the arm and hand, on average, high local current densities can occur in the blood and muscle organs or tissues, such as the heart. Fig. 2 shows the currents in a cross-section at the level of the heart muscles. Fig. 3 shows the peak-induced current densities for the various layers of the human body model in different conditions.

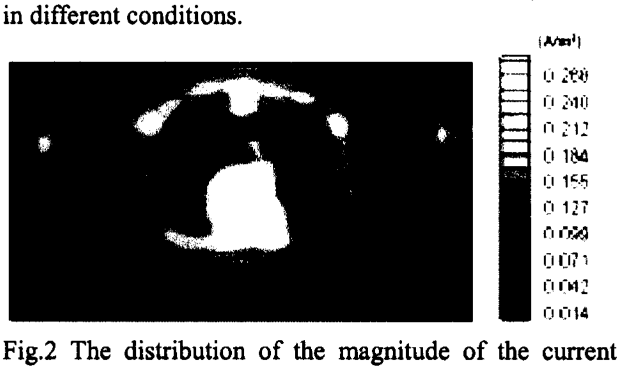
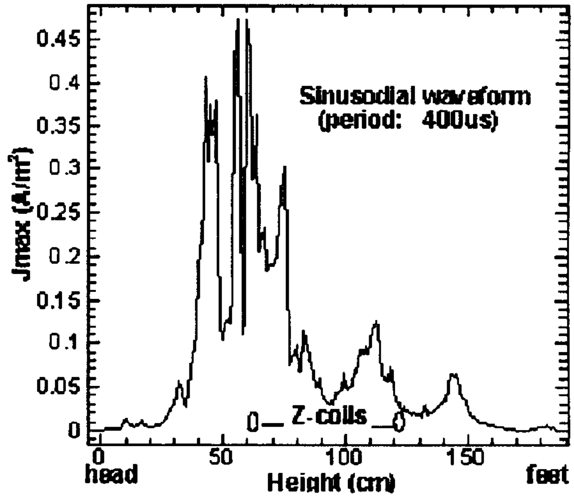
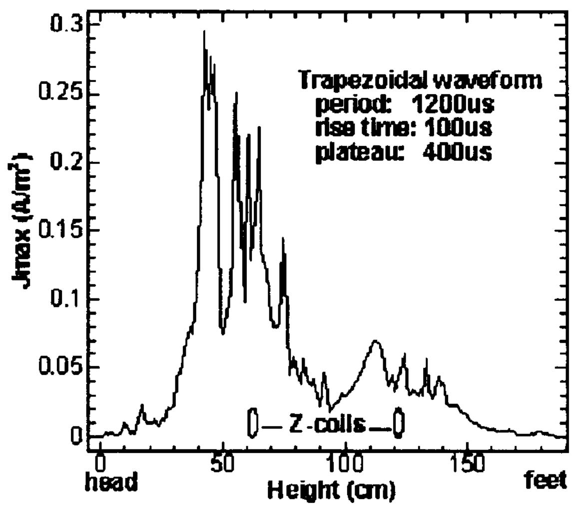


Fig.2 The distribution of the magnitude of the current density induced by trapezoidal waveform in the cross section including the heart.



(a)



(b)

Fig. 3 The peak-induced current densities for the various layers of the human body model under the conditions of sinusoidal waveform excitation (a) and trapezoidal waveform excitation (b).

Based on our realistic heart-torso model, various simulation experiments of cardiac arrhythmia have been done under some supposed conditions. In this paper, only one simulation result is presented. Fig. 4 (a) shows the simulated 12-lead ECG of normal heart based on our heart-torso model for comparison. Fig. 4 (b) shows the simulated 12-lead ECG of the supposed case that the eddy-currents are not strong enough to stimulate almost all of heart cells to be excited, but enable to stimulate some abnormal heart cells which have lower excitation thresholds to be excited continuously, and the firing site is set on the left ventricular free wall. From Fig. 4 (b), we can see that the heart could be developed into serious cardiac arrhythmia, like ventricular fibrillation.

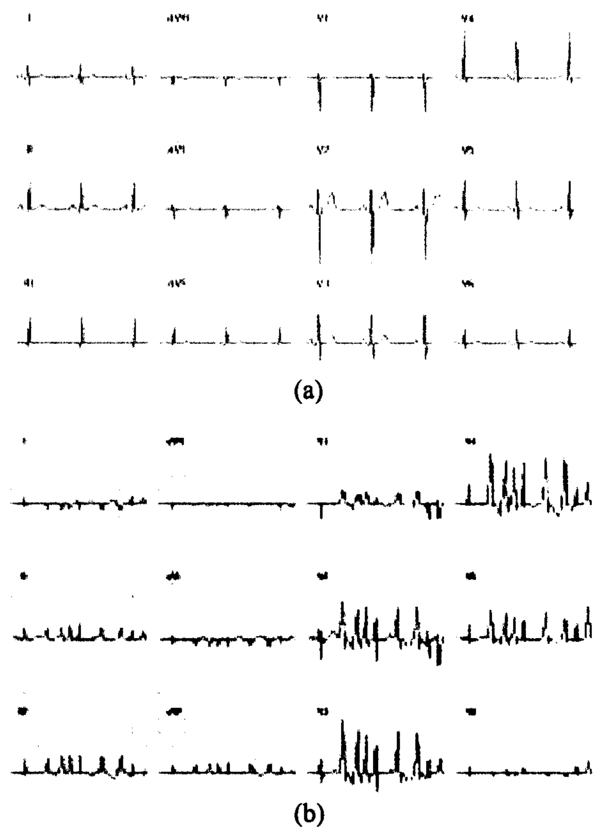


Fig. 4 Simulated 12-lead ECG of normal heart (a) and of a supposed case (b) (see the text description).

4. Conclusion

In modern MRI, the fast-switched gradient field may be cause cardiac stimulation. The simulation results of this paper show that the induced current caused by nowadays MRI is not strong enough to elicit ventricular fibrillation, but one should pay great attention to this problem because the induced current is very near to the "minimum" cardiac stimulation threshold in some "worst case". This kind of simulation is helpful for a better understanding and prediction of the potential hazards of MRI gradient fields on patients. In the future work, calculation should be repeated for whole gradient systems together with using a more refined whole body model and a refined heart model as well.

Acknowledgements

This project is supported by National Natural Science Foundation of China (30170243) and the Special Grant for the Authors of Excellent Doctoral Dissertation of Ministry of Education (199941). The authors also thank United States Air Force Research Laboratory for providing the human body data.

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