A computational investigation of atrial fibrillation treatment using HIFU energy source

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Abstract. Atrial fibrillation (AF) is the most prevalent arrhythmia of the heart, originating usually from ectopic atrial activity of the pulmonary veins. In recent technology, RF ablation has been part of clinical practice for more than two decades and has become an important treatment option for most clinically relevant cardiac arrhythmias [1]. The electrical isolation by ablation of the pulmonary veins (PVs) in the left atrium (LA) of the heart has been proven as an effective cure of atrial fibrillation (AF). The advantage of HIFU is that it could cause deep tissue lesions without damaging intervening tissues and prevent from thrombosis. Thus, it may lead to reducing the risk of stroke. The computer model uses the Pressure Acoustics, Frequency Domain interface to model the stationary acoustic field. The wave equation solved is the homogeneous Helmholtz equation in 2D axisymmetric cylindrical coordinates. With calculated acoustic pressure field, the generated heat combines with the Pennes' Bioheat Transfer equation to solve the temperature field. The use of extracorporeal HIFU ablation has a problem of vibration motion of heart and it causes focal energy unfocused. The results showed the impact of the thickness of traversing medium to the focal area and heating power with pulse time on the focal area. The varying thickness is to simulate vibration of heart motion. Frequency ranging from 1 MHz to 4 MHz and different geometrical configurations are studied. The preliminary computer modeling and test are feasible. There are in consistent.

Keywords: atrial fibrillation, HIFU (high intensity focused ultrasound), heart rhythm disorder, computer simulation, Pennes' Bioheat Transfer equation.

1 Introduction

The main cause of atrial fibrillation (AF) is due to pulmonary vein. Seminal work from Haissaguerre et al demonstrated that the predominant trigger for AF is ectopic activity originating in the pulmonary veins (PVs) [2]. Thus, pulmonary vein isolation (PVI) using radiofrequency (RF) ablation is used nowadays for curative treatments. PVI techniques during early 1980s have two approaches [3]. The first is focal ablation to try to identify the longitudinal fibres carrying excitation in and out of the vein, and to ablate these. The second is to achieve complete encirclement of the PV orifice aiming at isolation. The latter one is the popular standard AF radiofrequency ablation treatments which the lesion is called as continuous circular lesion (CCL) [4].A circumferential lesion was placed around the left and right PVs > 5 mm from orifices [5]. The formation of CCL is an important technique to disrupt abnormal electrical signals from PVs. In most recent AF treatments using radiofrequency ablation, they additionally used electrogram to check cardiac arrhythmias [6]. Pulmonary vein isolation guided by electrogram with the AF patients is more reliable.

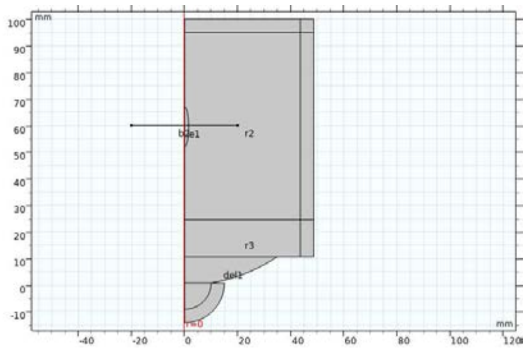
Extracorporeal HIFU ablation has been proposed and applied as early as 1999 for ablating cardiac tissue within the beating heart by use of this type of therapeutic ultrasound [7].The advantage of this type of ablation is being applied from outside the body to create a lesion of essentially any size and geometry [8]. High-intensity ultrasound can also be focused up to 15-cm within the body without heating or damaging the intervening tissue [8]. Furthermore, focused ultrasound is noninvasive, so it does not carry the risks associated with invasive procedures such as placement of catheters within the vascular system and complications associated with vascular access. Focused ultrasound can reach the desired target without damaging surrounding tissue. There are a lot of extracorporeal HIFU machines and computer simulations exist in the world but not specific to the treatments of atrial fibrillation.

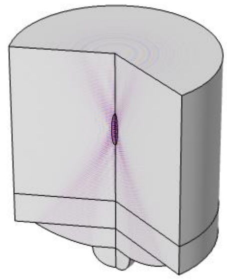
2 Methods

2.1 Formulation of the Problem

The ultrasonic system is particularly designed with the MR imaging-guided, so the system configuration is set up differently as compared with one with ultrasound-guided system. The ideal geometry model setup for computer simulation is described in Figure 1. The transducer frequency ranges from 1 to 1.5 MHz and the focused length is approximated at 10 cm. The transducer is enclosed in the water for cooling purpose. MRI probe is typically located at the top shooting downward. To save computational time, 2D axisymmetric cylindrical coordinates is used.

(a)





(b)

Figure 1 (a) A r-z cross section indicating the computational domain model with the symmetric axis. The tissue domain of the rectangular area which is above z=25 mm approximately and water domain is the rectangular area which is above z=10 mm but below z=25 mm. The bowl-shaped acoustic transducer is between z=0 and z=10 mm has a hole of radius 10 mm in the center. (b) the generated 3D computational geometry model.

Finer meshes are applied in the oval-shaped focal region to resolve the sharp gradients in the pressure field. There are two types of meshes: one mesh for pressure field and one for temperature field as shown in Figure 2. The mesh construction for pressure needs to be finer according to the high acoustic frequency.

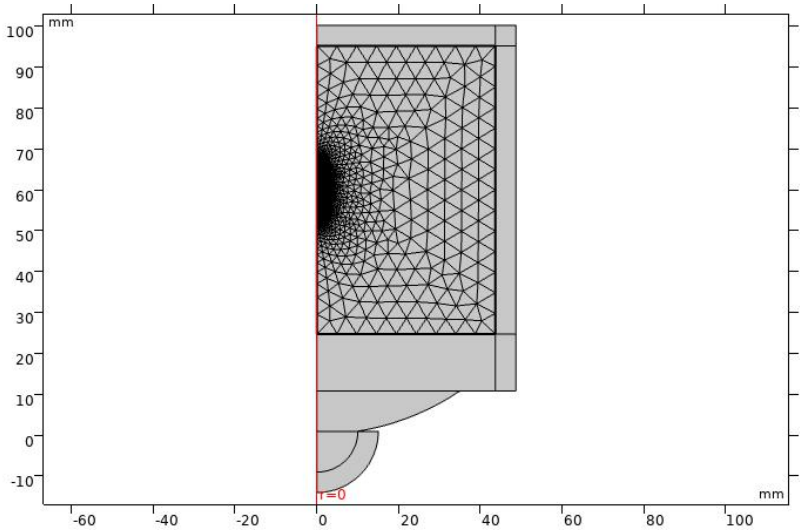


Figure 2. meshes for the temperature calculation

2.2 Mathematical Equations for the Wave and Thermal Models

The model uses the Pressure Acoustics, Frequency Domain interface to model the stationary acoustic field in the water and the tissue domain to obtain the acoustic intensity distribution in the tissue phantom. The absorbed acoustic energy is calculated and used as the heat source for the Bioheat Transfer interface model. The acoustic focal region (i.e. heated area) is much smaller than the size of tissue region, so the thermal simulation is performed only in the tissue domain.

The Wave Equation

The homogeneous Helmholtz equation is used as the following:

$$\frac{\partial}{\partial r}\left[-\frac{r}{\rho_c}\left(\frac{\partial p}{\partial r}\right)\right]+r\frac{\partial}{\partial z}\left[-\frac{1}{\rho_c}\left(\frac{\partial p}{\partial z}\right)\right]-\left[\left(\frac{\omega}{c_c}\right)^2\right]\frac{rp}{\rho_c}=0 \qquad (1)$$

Here r and z are the radial and axial coordinates, p is the acoustic pressure, and ω is the angular frequency. The density, ρc, and the speed of sound, cc, are complex-valued to account for the material's damping properties.

The Equation 1 used with the assumption that the acoustic wave propagation is linear and also that the amplitude of shear waves in the tissue domain are much smaller than that of the pressure waves. Nonlinear effects and shear waves are therefore neglected.

The acoustic intensity field is readily derived given the acoustic pressure field. The heat source Q for thermal simulation, given in the plane-wave limit, is then calculated as:

$$Q = 2\alpha_a I = 2\alpha_a \left|\mathrm{Re}\left(\frac{1}{2}p\vec{v}\right)\right| \qquad (2)$$

where α_a is the acoustic absorption coefficient, I is the acoustic intensity magnitude, p is the acoustic pressure, and $\vec{v}$ is the acoustic particle velocity vector. The volumetric acoustic heat source Q will be inserting into the Bioheat transfer equation.

The Pennes' Bioheat Transfer equation

$$\rho c_p \frac{\partial T}{\partial t} = \frac{1}{r}\frac{\partial}{\partial r}\left(rk\frac{\partial T}{\partial r}\right)+\frac{\partial}{\partial z}\left(k\frac{\partial T}{\partial z}\right)+Q-wc_b(T-T_a) \qquad (3)$$

where T is the temperature, ρ is the density, C_p is the specific heat, k is the thermal conductivity, ρ_b is the density of blood (it is assumed to be ρ), C_b is the specific heat of blood, w_b is the blood perfusion rate, T_b is the temperature of the blood, Q is the heat source (the absorbed ultrasound energy calculated from Equation 2). In this model, assume that the tissue properties do not change when the temperature rises.

To accurately resolve the sharp pressure gradient in the focal region, the model uses a fine mesh with size $\lambda/6$ (where λ is the wavelength) within that region. A coarser mesh with size $\lambda/4$ is used for the other domains. Quartic elements are used to discretize the acoustic pressure, and quadratic elements are used to discretize temperature. The material properties used in the model is shown in Table 1.

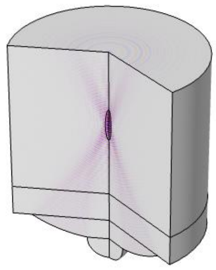
AF is the most common serious heart rhythm disorder. Normally it is treated by catheter ablation or radiofrequency ablation (RFA). The objective of this study is to numerically evaluate and design a high intensity focused ultrasound (HIFU) energy heating source instead of radiofrequency energy source which is commonly used in Taiwan. The advantage of HIFU is that it could cause deep tissue lesions without damaging intervening tissues and prevent from thrombosis. Thus, it may lead to reducing the risk of stroke. The computer model uses the Pressure Acoustics, Frequency Domain interface to model the stationary acoustic field. The wave equation solved is the homogeneous Helmholtz equation in 2D axisymmetric cylindrical coordinates. With calculated acoustic pressure field, the generated heat combines with the Pennes' Bioheat Transfer equation to solve the temperature field. The commercial finite element method was used to numerically solve the equations. Frequency ranging from 1 MHz to 4 MHz and different geometrical configurations will be studied. The preliminary computer modeling and test are feasible.

Table 1: Material properties used in the model [9]

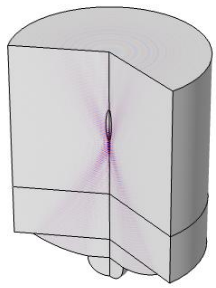
Property	Density (kg/m^3)	Speed of sound (m/s)	Attenuation (Np/m/MHz)	Specific heat (J/(kg·K))	Thermal conductivity (W/(m·K))
Water (at 293.7 K)	1000	1483	0.025	N/A	N/A
Tissue phantom	1044	1568	8.55	3710	0.59
Human tissue	1000–1100	1450–1640	4.03–17.27	3600–3890	0.45–0.56

3 Results

In the treatment of heating circular region in left atrium (LA) of the heart for curing atrial fibrillation (AF), the water depth thickness is making acoustic energy path different and thus the tissue thickness from water-contact interface to the focal point that has 1-cm (10-mm) difference which is to simulate the heating region of left heart atrium vibration. The thickness of tissue from the water-tissue interface to the focal point varied. Figure 3 showed 3D acoustic pressure with (a) shorter water thickness (b) longer water thickness. The difference is 1-cm (10-mm) in thickness. With the power transducer focusing the same angle, through different thickness of mediums, the intended focal point is shown in Figure 3(a). However, when water medium thickness increased, making the tissue traveling shorter that makes the focal region below the previous one as shown in Figure 3(b).



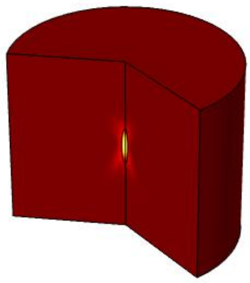
(a)



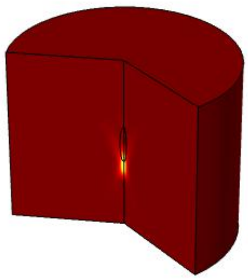
(b)

Figure 3. 3D acoustic pressure with (a) shorter water thickness (b) longer water thickness. The difference is 1-cm (10-mm) in thickness.

Figure 4 showed 3D temperature with 1 sec power pulse in the case of (a) shorter water thickness (b) longer water thickness. The difference is 1-cm (10-mm) in thickness. The bright spot in the Figure 4(a) indicated hot spot (temperature rise) in the focal region. For the case of longer water thickness, that caused the focal bright spot lower than the expected case (a) as shown in Figure 4(b). The temperature spots for both cases are different and the case (b) has higher temperatures than the case (a) due to tissue attenuation effect was not larger than the case (a). They can be seen in Figure 5. Figure 5 showed 2D temperature with 1 sec power pulse in the case of (a) shorter water thickness (b) longer water thickness. The difference is 1-cm (10-mm) in thickness.

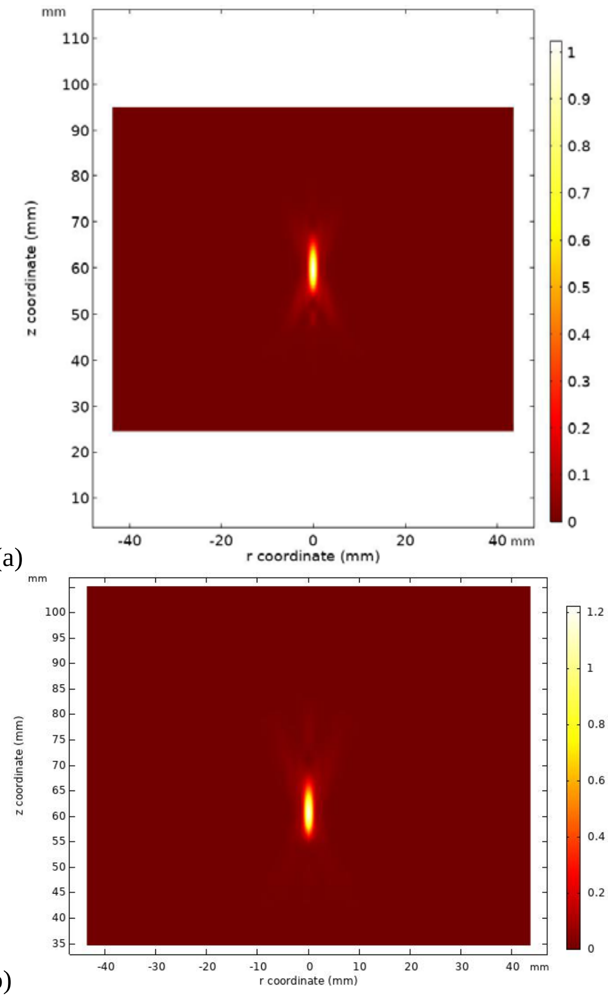


(a)



(b)

Figure 4. 3D temperature with 1 sec power pulse in the case of (a) shorter water thickness (b) longer water thickness. The difference is 1-cm (10-mm) in thickness.



(a)

(b)

Figure 5. 2D temperature with 1 sec power pulse in the case of (a) shorter water thickness (b) longer water thickness. The difference is 1-cm (10-mm) in thickness.

5 Conclusions

During the HIFU treatment of atrial fibrillation, the focal spot lesion may vary with different thickness due to heart rhythm of vibration. The case of 1-cm thickness variation is used in the computer model. The results showed the focal region temperatures with acoustic energy traveling shorter tissue medium that indicated hotter focal region temperatures.

Acknowledgement

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