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Breast Cancer Electromagnetic Ablation using a 2.45 GHz Microcoaxial Applicator

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ABSTRACT

In this work, heating effects from the antenna during the thermal ablation process are modeled by using the Finite Element Method (FEM), which solves partial differential equations of the Bioheat equation. The FEM model assumes that the coaxial slot antenna is immersed in homogeneous breast tissue and the tumor was considered a perfect sphere. The inner and outer conductors of the antenna were modeled using perfect electric conductor boundary conditions and boundaries along the z axis are set with axial symmetry. After the Electromagnetic model is solved, we analyze the temperature distribution of ablation and the antenna efficiency.

Keywords: Ablation, Breast cancer, Microwave, Modeling

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Introduction

Globally, breast cancer is the most common cancer among women, comprising 23% of all female cancers that are newly diagnosed in more than 1.1 million women each year [1]. As a result of the implementation of mammography screening for breast cancer, the proportion of small carcinomas found at the time of screening has increased, raising a demand for new techniques that minimize changes of breast configuration. Therefore, approaches other than traditional surgery have been explored to satisfy these demands. These procedures are minimally or totally noninvasive, and include, thermal energy using microwaves, radiofrequency, interstitial laser photocoagulation, focused ultrasound, and cryotherapy.

Microwave ablation (MWA), like Radiofrequency ablation (RFA), uses localized heating to cause tissue necrosis. Within the MW energy field, water molecules in the tissue rotate with the varying electric fields, causing frictional heating. This heating is determined mainly by power deposition in tissue, often expressed as specific absorption rate (SAR), but it is also dependent on both the dielectric and thermal properties of the tissue being ablated. Lesion size is mainly limited by the available power and treatment time. Furthermore, MW energy is promising because it can preferentially heat and damage high-water-content breast carcinomas, compared with the lesser degrees of heating that occur in lower-water-content adipose and breast glandular tissues [2]. Compared to RF, MW has a theoretically broader field of power density, with a correspondingly larger zone of active heating. This may allow more uniform tumor kill, both within a targeted zone and in perivascular tissue. Nevertheless, MWA has received less attention than RFA for breast cancer treatment.

In antenna design, computational electromagnetics (CEM), a discipline that employs numerical methods to describe propagation of electromagnetic waves, is broadly used to obtain numerical results for electromagnetic problems. Though there are various ways to classify the assortment of techniques in CEM, they are classified as either differential-equation-based or integral-equation-based. The finite-element method (FEM) is a technique that is differential-equation-based and can provide users with quick, accurate solutions to multiple systems of differential equations and as such is well suited to heat transfer problems like ablation [3].

Numerous previous antenna designs specifically targeted for cardiac and hepatic applications have been presented in the literature for MW ablation [4]. These designs have focused largely on thin, coaxial-based interstitial antennas, which are minimally invasive and capable of delivering a large amount of electromagnetic power. These antennas can usually be classified as one of three types (dipole, slot, or monopole) based on their physical features and radiative properties [5]. We have investigated the coaxial-slot antenna to apply to such a technique for breast cancer treatment. Research in thermal interstitial microwave ablation process in breast tissue using a coaxial slot antenna is not yet reported.

Methodology

An electromagnetic wave propagating in a coaxial cable is characterized by transverse electromagnetic fields (TEM). The electric and magnetic fields associated with the time-varying TEM wave generated by the microwave source propagating in a coaxial cable in the z-direction is expressed in 2D axially symmetric cylindrical coordinates as:

$$\vec{E}(r) = r\frac{C}{r}e^{j(\omega t - kz)}, \tag{1}$$

$$\vec{H}(r) = \varphi\frac{C}{rZ}e^{j(\omega t - kz)}, \tag{2}$$

where z is the direction of propagation and r, φ and z are cylindrical coordinates centered on the axis of the coaxial cable. Z is the wave impedance in the dielectric of the cable and C is an arbitrary constant. The angular frequency is denoted by ω. The propagation constant k relates to the wavelength in the medium λ as:

$$k = \frac{2\pi}{\lambda} \tag{3}$$

In the tissue for interstitial coaxial-based antennas during MWA, the electric field is in the radial direction only inside the coaxial cable and in both radial and the axial direction inside the tissue whereas the magnetic field is purely in the azimuthal direction. This allows for the antenna to be modeled using an axisymmetric transverse magnetic (TM) formulation, in which the source was modeled as a low reflection boundary:

$$\hat{n} \times \sqrt{\varepsilon}\check{E} - \sqrt{\mu}H_{\varphi} = -2\sqrt{\mu}H_{\varphi 0}, \tag{4}$$

where

$$H_{\varphi 0} = \frac{\sqrt{\dfrac{P_{av}Z}{\pi r \ln\left(\dfrac{r_{outer}}{r_{inner}}\right)}}}{r}, \tag{5}$$

P_{av} is the time average power flow in the cable and r_{inner} and r_{outer} are the inner and outer radii respectively in the dielectric.

The antenna's frequency-dependent reflection coefficient and specific absorption rate (SAR) pattern in tissue are important for the performance of interstitial antennas. The frequency-dependent reflection coefficient can be expressed logarithmically as:

$$\Gamma(f) = 10 \cdot \log_{10}\left(\frac{P_r(f)}{P_{in}}\right) \; [dB], \tag{6}$$

where P_{in} is the input power and P_r indicates the reflected power (W). The frequency where the reflection coefficient is minimum is commonly referred to as the resonant frequency and should be approximately the same as the operating frequency of the generator used. Antennas operating with high reflection coefficients can cause overheating of the feedline possibly leading to damage to the coaxial line or due to the thin outer conductor damage to the tissue [6]. SAR represents the amount of time average power deposited per unit mass of tissue (W/kg) at any position. It can be expressed mathematically as:

$$SAR = \frac{\sigma}{2\rho}\left|\overset{\vee}{E}\right|^2 \quad [W / kg], \tag{7}$$

where σ is tissue conductivity (S/m), ϱ is tissue density (kg/m^3) and E is the electric field vector [V/m].

The SAR takes a value proportional to the square of the electric field generated around the antenna and is equivalent to the heating source created by the electric field in tissue. The SAR pattern of an antenna causes the tissue temperature to rise, but does not determine the final tissue temperature distribution directly. The tissue temperature increase results from both power and time, caused by direct MW heating (from SAR) and tissue thermal conduction. MW heating thermal effects can be roughly described by Pennes' Bioheat equation [7]:

$$\nabla \cdot (-k\nabla T) = \rho_b C_b \omega_b (T_b - T) + Q_{met} + Q_{ext}, \tag{8}$$

where k is the tissue thermal conductivity (W/m°K), ρ_b is the blood density (Kg/m^3), C_b is the blood specific heat (J/Kg°K), ω_b is the blood perfusion rate (1/s). T_b is the temperature of the blood and T is the final temperature. Q_{met} is the heat source from metabolism (w/m^3) and Q_{ext} an external heat source. The major physical phenomena considered in the equation are microwave heating and tissue heat conduction. Heat conduction between tissue and blood flow in tissue is approximated by the term

$\rho_b C_b \omega_b (T_b - T)$ in the equation. The temperature of the blood is approximated to the core temperature of the body. In ex vivo samples, ω_b and Q_{met} can be ignored since no perfusion or metabolism exists. The external heat source is equal to the resistive heat generated by the electromagnetic field. Dimensions and thermal properties for the materials and breast tissue are listed in Table 1, which were taken from the literature [8,9].

Table 1. Dimensions and material properties for the materials and tissue.

Parameter	Value
Center conductor diameter	0.51 mm
Dielectric diameter	1.68 mm
Outer conductor diameter	2.20 mm
Diameter of catheter	2.58 mm
Electrical conductivity of breast	0.137 S/m
Thermal conductivity of breast	0.42 W/m°K
Blood density	920 Kg/m3
Specific heat of blood	3639 J/Kg/°K
Blood perfusion rate	0.0036 s^{-1}
Electrical conductivity of tumor	3 S/m
Thermal conductivity of tumor	0.5 W/m°K
Material	**Relative permittivity**
Inner dielectric of the coaxial cable	2.03
Catheter	2.60
Breast tissue	5.14
Tumor	57

Figure 1, shows the axial schematics of each section of the antenna, and the interior diameters.

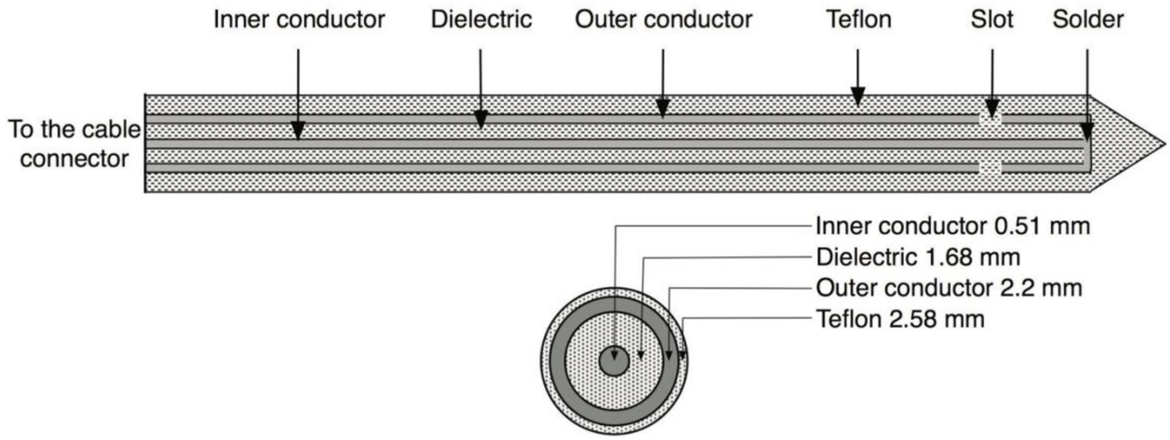

Figure 1. Cross section and schematic of the coaxial slot applicator.

Heating effects from the antenna during the thermal ablation process are modeled using commercial software [10], which solves partial differential equations using the FEM. The coaxial slot antenna exhibits rotational symmetry around the longitudinal axis; therefore, we used an axisymmetric model, which minimized the computation time.

The FEM model assumes that the coaxial slot antenna is immersed in homogeneous breast tissue. The inner and outer conductors of the antenna were modeled using perfect electric conductor boundary conditions and boundaries along the z axis are set with axial symmetry [11].

All boundaries of conductors are set to PEC (Perfect electric conductor). Boundaries along the *z* axis are set with axial symmetry and all other boundaries are set to low reflection boundaries.

Figure 2 shows the geometry of the antenna model with details near the antenna slot. Because the model is axialsymmetric, only a half of the geometry structures of antenna, breast tissue and the tumor are created in the model [12].

The axisymmetric finite element mesh presented in Figure 3, has been selected in order to achieve a compromise between accuracy of computation and reasonable dimensionality of the model. In order to obtain a finest mesh size, it was generated using the mesh parameters in Table 2. The mesh consists of 4835 triangular elements. Dense mesh zone has been generated in the vicinity of the tip of the antenna, where the temperature is more concentrated.

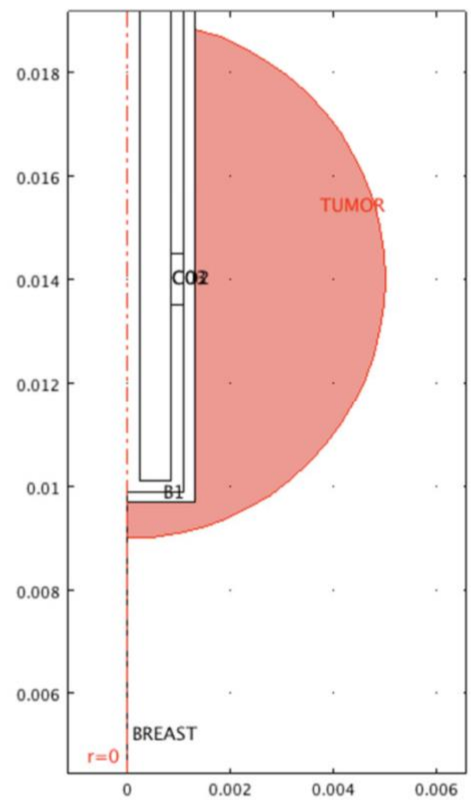

Figure 2. Axisymmetric CEM model in the vicinity of the tip of the coaxial slot antenna. The vertical axis corresponds to the longitudinal axis of the antenna while the horizontal axis corresponds to radial direction.

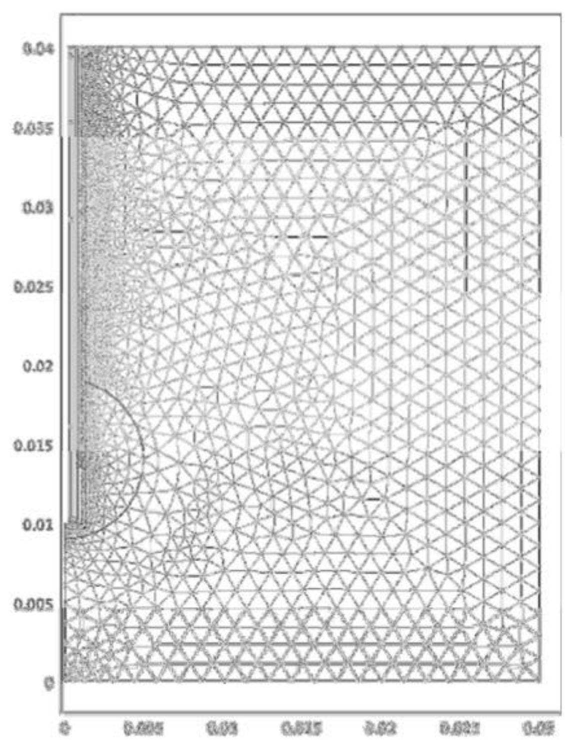

Figure 3. Plot of the whole model with mesh grids. This FEM model contains 4835 elements.

Results

Figure 4 shows the resulting steady-state temperature distribution isotherm at 60 °C, since temperatures above these values are considered ablation temperatures, for a microwave power output at 10 W. The reflection coefficient calculated for the frequency of 2.45 GHz was -4.55 dB.

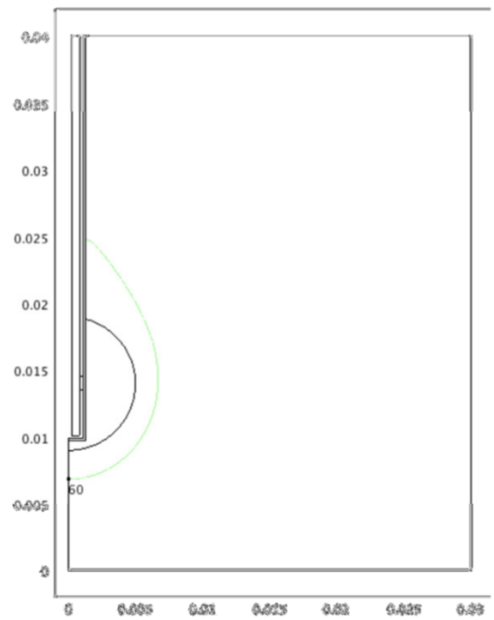

Figure 4. Temperature distribution.

An antenna for MWA was simulated using an axisymmetric electromagnetic model. Once the theoretical model has been built, and although it is based on equations which correspond to well characterized phenomena, the next step is the experimental validation conducted to guarantee the results obtained from computer simulation.

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