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Heating characteristics of antenna arrays used in microwave ablation: A theoretical parametric study

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ABSTRACT

A numerical study of the performance of antenna arrays used in microwave ablation (MWA) is carried out. Double-slot coaxial antennas in triangular and square configurations are studied. Clinical (healthy vs. malignant) and experimental (*in* vs. *ex vivo*) scenarios for hepatic cancer treatment are modeled, and further application in bone and lung tissue is examined. It is found that triangular arrays can create spherical ablation zones, while square configurations result in flatter ones. Thresholds in power and application times for creating continuous ablation zones are calculated, and the characteristics of the latter are quantified.

1. Introduction

Microwave ablation (MWA) is a minimally invasive technique, providing treatment in the form of coagulative necrosis of the tumor cells, which occurs when the temperature rises, due to the deposition of electromagnetic (EM) energy, and reaches a critical level of around 60 °C [1]. Along with radio-frequency ablation (RFA), it constitutes a common treatment alternative in cases where surgical resection of the tumors is not an option [2–4]. Comparing these two thermal therapy techniques, MWA has a few additional advantages, including its ability to not be limited by tissue charring, heat tissues faster and create larger ablation zones, as discussed in [5–8]. The latter is of particular interest, as incomplete necrosis of the tumorous tissue is associated with high recurrence rates [7].

MWA has found applications in different clinical cases, such as in liver (hepatocellular carcinomas and colorectal metastases), lung, kidney, pancreas and bone tumor treatment [9], in invasive cardiology (cardiac arrhythmias treatment) or in gynaecology (dysfunctional uterine bleeding). Its potential applications in the brain are also being researched [10], although additional risk factors are involved.

During MWA treatments, a single antenna or an array of antennas (applicator) is used to deliver the EM field, after being placed in proximity with the area to be treated. Frequencies of approximately 900 MHz are most commonly used in the U.S.A., and of 2.45 GHz in Europe and Asia [7]. Different antenna designs have been proposed for the delivery of MWA. The commonly used monopole and dipole antennas create an elongated ablation zone along their axes (known as the "teardrop" or "pear" shape), which is a characteristic that might be undesirable in some clinical applications [11,12]. Single- and double-slot coaxial antennas have been studied [13–16], and it has been shown that double-slot antennas create a more localized heating around their tips. Sleeve or choked designs have also been proposed in order to concentrate the power distribution in the target volume, having, however, the drawback of being thicker, thus more invasive [12,17]. For this matter, an improved design (closed tip antenna loaded with a conductor to obtain a capacitive cap) has recently been proposed by Cavagnaro et al. [18]. Its minimal dimensions allowed for percutaneous application and the antenna was able to create localized thermal lesions. Different antenna designs have been reviewed by Bertram et al. [11] and Lin et al. [12].

In cases where the region of interest is large, the use of antenna arrays has been proposed, and found preferable over single applicators. The thermal synergy that exists when antennas operate in concert leads to the creation of larger ablation zones [20,19]. Concerning liver tissues, Liang et al. [21] and Saito et al. [22] studied double-applicator configurations, Brace et al. [19] and Ryan et al. [23] studied triangular arrays, and Phasukkit et al. [24] investigated different geometrical configurations using three applicators. In specific, Ryan et al. [23] have shown that treatment efficiency is improved when antennas are operated synchronously.

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The present study aims at providing an insight into the expected characteristics of the ablation zones under different array configurations, simulating both *in* and *ex vivo* experimental conditions, and emphasising on the importance of the treatment settings (antenna spacing, heating power and duration), as well as the tissue dielectric and thermal properties. To our knowledge, such a comprehensive work on arrays does not exist in the literature, and its findings can assist with the selection of appropriate treatment parameters in case-specific applications.

Since MWA has arguably found its widest application in hepatic ablations treatments [22], the majority of the results shown in this paper concern *in vivo* liver tissue models, and, as most experiments found in the literature are conducted in normal (healthy) tissues, the corresponding parameters were used. Nevertheless, an extension to other tissues (malignant liver, bone, and lung) as well as *ex vivo* (by omitting the effect of blood perfusion) cases were also included.

2. Methods and models

The Finite-Differences Time-Domain (FDTD) method was used to solve both the EM and the thermal problems in SEMCAD X (Schmid & Partner Engineering AG, Zurich, Switzerland), a software package that allows 3D modeling and analysis.

2.1. Modeling

Coaxial double-slot antennas with the characteristics of Table 1 were modeled and used to form the arrays. The slots were formed by removing annular portions of the outer conductor of the coaxial cable. The inner and outer conductors were short-circuited at the tip of the antenna. The antennas are symmetrical around their axes, and a dielectric catheter surrounds each antenna. No layer of air between the antenna and the inner side of the catheter was assumed. The specific design characteristics (number of slots, slot distances and slot length) were chosen after taking under consideration the results of the parametric study by Karampatzakis et al. [25] which reported that such antennas were able to create ablation zones of favorable characteristics.

Triangular and square array configurations were formed by placing the antennas on the vertices of equilateral triangles and squares, respectively, with sides of lengths (S) of 5, 10, 15 and 20 mm. Fig. 1 schematically summarizes the above. All antennas were excited in synchronous phase, as Ryan et al. [23] have shown that a synchronous activation yields better efficiency. The antennas were placed at the same insertion depth (i.e., air–tissue interface to antenna tip distance) of 7 cm.

A 10 cm × 10 cm × 12 cm rectangular solid region was used to model the tissue. The dielectric (at 2450 MHz) and thermal properties used for all healthy tissues were taken from the IT'IS database [26]. The properties of the malignant liver were those found by the *ex vivo* measurements in [27], and its perfusion rate was calculated by scaling down to 13% the value for the healthy one [28]. They are presented in Table 2. All tissues were considered homogenous and isotropic.

Table 1
Design characteristics of the coaxial double-slot antennas.

Dimensions (mm)	
Antenna diameter (T_a)	1.3
Outer catheter diameter (T_c)	1.8
Slot distance (D_s)	10.0
Slot length (L_s)	1.0
Total antenna length	140
Relative permittivities	
Dielectric in the coaxial line (ϵ_{r_a})	2.03
Dielectric of catheter material (ϵ_{r_c})	2.60

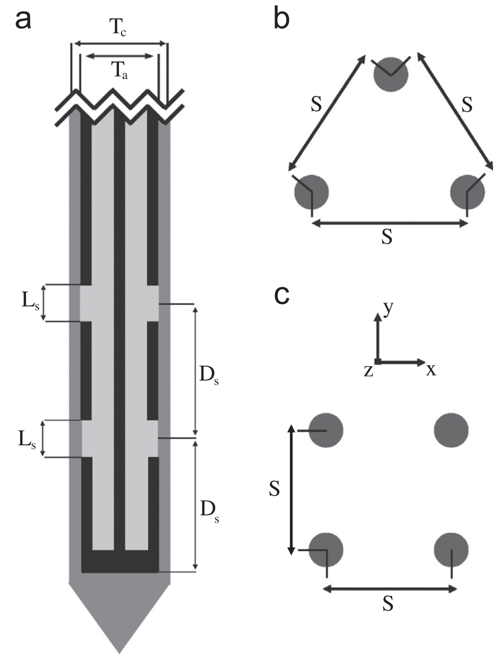

Fig. 1. Schematic outline of the (a) double-slot antenna, and the projections of the (b) triangular and (c) square array configurations.

2.2. Numerical methods

2.2.1. Electromagnetic problem

The specific absorption rate (SAR) distribution, given by (1), was calculated within the computational domain. SAR quantifies the rate at which energy is absorbed per unit mass of a biological tissue, and is the heating source in MWA [29,30]. In (1), σ is the electric conductivity, ρ the mass density of the tissue and $\mathbf{E}$ the induced electric field

$$SAR = \frac{\sigma}{2\rho}|\mathbf{E}|^2 \tag{1}$$

The electromagnetic field in the simulation was excited by a voltage applied between the inner and outer conductors of the coaxial cable at 2450 MHz. A sufficiently long coaxial cable was used, so that the transverse electromagnetic mode (TEM) was developed inside the cable, before the wave entered the tissue model. A non-uniform mesh with an overall maximum mesh size of 1 mm was used, whereas the thinner parts of the antenna were further refined by defining a maximum grid step of 0.04 mm. Perfectly matched layer (PML) boundary conditions were chosen to truncate the computational domain.

2.2.2. Thermal problem

The Pennes Bioheat Equation (PBE) was solved for the thermal problem, described by

$$\rho c\frac{dT}{dt} = \nabla\cdot(k\nabla T) + \rho Q + \rho SAR + \rho_b c_b \rho\omega(T-T_b), \tag{2}$$

where k is the thermal conductivity, SAR the specific absorption rate, ω the perfusion rate, Q the metabolic heat generation rate, ρ the density of the tissue, and c the specific heat capacity. The respective quantities for blood are denoted with the subscript b.