Metamaterial applicator for microwave hyperthermia

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Abstract

In this work, it is studied the application in hyperthermia of a microwave focusing device based on metamaterials. The device consists of a planar array of split-ring resonators placed between two parallel metallic plates and it is fed by a small loop antenna which excites the split-rings. The device is modelled as an homogeneous uniaxial slab of negative permeability placed between two metallic plates. Both the fields and the temperature distribution in model of breast tissue and a tumor are numerically obtained. The field produced by the fabricated device inside a phantom resembling the breast tissue was measured with a probe to check the theoretical predictions.

1 Introduction

One of the most striking properties of metamaterials is the ability of negative-refractive index (NRI) slabs to focus the electromagnetic field of a source [1]. NRI slabs can provide a three-dimensional (3D) focus of energy of wavelength (λ) size in the far-field region. A 3D focus of sub-λ size is also possible in the near-field region if an antenna is placed in the focus to provide tunnel effect between this antenna and the source [2, 3]. 3D focusing of energy in the far-field region is an interesting property of metamaterials which can be exploited for non-invasive hyperthermia of tissue. Sub-λ focusing in the near-field region could find also application in invasive hyperthermia with the help of implanted antennas [4]. Microwave (MW) and radiofrequency (RF) hyperthermia is a technique used in the medical treatments of cancer and other medical therapy [5], also in combination with radiotherapy or chemotherapy, since it has been proved that hyperthermia can increase the efficiency of these therapies [5]. Tumors are heated by means of the exposure to MW or RF fields to therapeutic temperatures (43^0C–45^0C) without overheating the surrounding normal tissue. Key issues are heating, temperature measuring, and controlling the system. Biological effects depend on the field in the tissues, i.e., on the specific absorption rate (SAR), defined as power deposited in a unit mass of tissue:

$$SAR = \frac{\sigma(r)|E(r)|^2}{2\rho(r)} \tag{1}$$

Thermal effects of MW and RF fields are well known [6]. It is estimated that a SAR of 1 W/kg produces an increase of 1^0C in human body temperature, taking thermoregulation into account. Human tissues differ in their permittivity [6], varying with frequency. Cancerous tissue, except when it becomes necrotic, is highly vascular and has increased water content [7]. This finding directly accounts for an increase in the permittivity of tumors compared to that of host tissue. Typically, the permittivity is approximately 10%-30% greater for high water content tissues as skin, liver, muscle, and spleen [6]. In contrast, significantly larger dielectric constant and loss factor have been reported for tumors in breast tissue (4-10 times) [7]. Various techniques for producing localized or regional hyperthermia have been developed [5]. Each technique has its own inherent advantage and disadvantage, and its use should be chosen based on its clinical site and tissue composition. These techniques are divided basically into two categories: a single MW waveguide applicator [8] and a RF phased array of antennas [9]. One of the major problems of MW waveguides applicators is that the depth penetration is limited by the skin depth, so that MW hyperthermia is most prevalently used for external heating of superficial tumors. Moreover, the skin must be cooled, usually with circulating water. With phased arrays of antennas, a local maximum of field density can be obtained in the far field region by means of several phased sources [9]. In practice, two operating parameters controlling power and phase of the irradiated fields have to be adjusted for each of the applicator's, which can consist up to 12 channels, each comprising of two antennas [9]. It is apparent that for such a large number of resulting parameters an optimal adjustment cannot be achieved by a strategy of trial and error. Instead, precise planning is mandatory, which

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has to be based on field simulation and measurement in order to allow a reliable calibration and dosimetry similar to the prerequisites of radiotherapy.

A simpler alternative to a phased array of antennas would be a system able to focus the field of a single antenna. In this sense, metamaterials can be of a great interest. Non-invasive hyperthermia with metamaterials have been numerically investigated in a few works [10]-[12]. Hyperthermia of breast tumors have received special attention [11, 12]. Breast tumors are suitable for microwave hyperthermia since for breast tissue, the electrical conductivity of malignant breast tissue can be up to ten times higher than the conductivity of normal breast tissue [7]. In [10] it is shown that focusing with metamaterial lenses inside a lossy dielectric occurs at certain depth if conductivity is higher deeper inside. In [11] a left-handed (LH) media lens is analyzed for hyperthermia of tumors in breast tissue. All of these works dealing with LH lenses [10]-[12] show numerical analysis but no further details are given about the practical implementation of the LH slabs. In the present work, a real LH device is proposed and analyzed for microwave hyperthermia of breast tumors. Both focusing of field and temperature distribution are obtained by numerical analysis, and a experiment is carried out in order to check the ability of the proposed device to focus the field.

2 Method

Fig. 1.a shows a sketch of the proposed device. It consists of a pair of parallel metallic plates which provides negative permittivity ($\varepsilon < 0$) for the electric field ($\mathbf{E}$) which is parallel to the plates, and a two-dimensional planar array of broadside-coupled split-ring resonators (BCSRRs) [13] which provides negative permeability ($\mu < 0$) for the magnetic field ($\mathbf{H}$) perpendicular to the plates [14, 15]. The dimensions of the plates are 30×15 cm^2. The device is designed to work at frequencies close to 4 GHz. At this frequency, breast tissue permittivity is $\varepsilon \simeq 9$ [16], so that the index of refraction is $n \simeq 3$. The distance between plates is fit to 12 mm to provide $\varepsilon \simeq -9$ at that frequency, whereas the BCSRRs are designed to show $\mu \simeq -1$ [17], so that $n \simeq -3$ in the LH device. The printed circuit board with the BCSRRs is sandwiched between two foam layers of 6 mm. Fig. 1.b shows a sketch of the setup under analysis. A coaxial cable is introduced trough the center of one of the plates of the device to excite the BCSRRs by means of a resonant loop. Although it is not shown in the figure, the device is shielded at the back and at the sides to avoid leak of radiation.

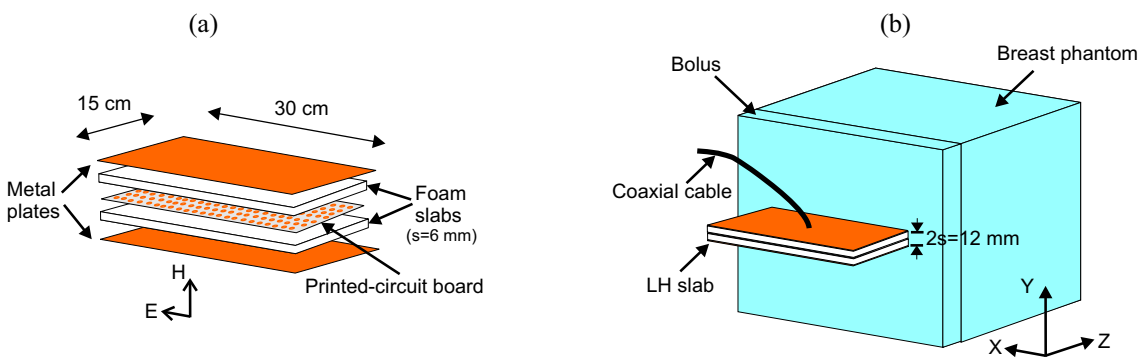

Figure 1: (a) Detail of the left-handed (LH) slab. (b) : Sketch of the experimental setup.

In the numerical simulations, breast tissue is modelled as a squared phantom and the tumor as a small sphere of 30 mm in diameter. A thin bolus of 30 mm in thickness and of the same dielectric characteristics as the phantom is placed between the LH device and the phantom for cooling of the phantom surface. The numerical analysis is carried out by means of the electromagnetic solver CST *Microwave Studio*, which also provides a solution for the bio-heat equation. The device is implemented in CST by means of an homogeneous slab placed between two metallic plates. The permeability of this slab is anisotropic to reproduce the effect of the array of BCSRRs and it is $\mu = -1$ in the direction perpendicular to the plates and $\mu = 1$ in the

other directions. The dispersive properties of the phantom and the tumor are incorporated in the simulation by means of single-pole Debye dispersion equations which take into account the different permittivity and conductivity of normal and malignant breast tissue [16]. The device is excited by a point dipole placed in the center, i.e., at a distance of 7.5 cm from the surface of the bolus. A power of 8 W is injected in the dipole. Fig. 2.a shows a side view of the setup with the distribution of the modulus of the electric field in V/m obtained in the simulation in absence of the tumor and at the frequency of 4.1821 GHz. A focus is shown in the phantom at the expected distance of 7.5 cm from the exit of the LH device, a value which corresponds to the half-width of the device.

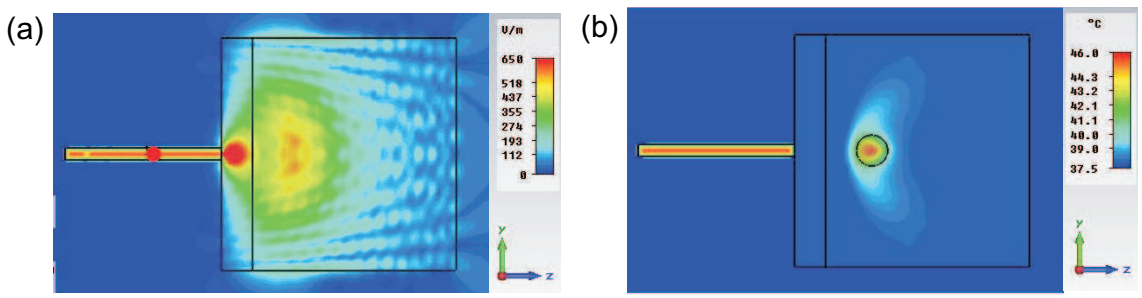

Figure 2: Side view of the setup with numerical results for (a) the modulus of the electric field in V/m inside the breast phantom and (b) the temperature distribution in the phantom with a tumor located at the main field focus.

Fig. 2.b shows the temperature distribution provided by CST when a tumor is placed at the position of the focus. The parameters for the thermal simulation are suitable chosen for both the phantom and the tumor [16]. The temperature of the bolus is fixed to 25^0. The results in Fig. 2.b show a peak of temperature in the tumor reaching 46^0 , whereas the temperature remains lower in the surrounding tissue and below 37.5^0 in the main volume. The peak of temperature is restricted to a region smaller than the field focus due to the higher conductivity of the tumor (2 S_mm) [7]. To check the predictions given by the electromagnetic simulations, the electric field produced by the fabricated device is measured with a dipole probe inside a phantom of glycerine which resembles the electrical properties of breast tissue [18]. Fig. 3 shows the field measured for several frequencies and along a line ranging between 2 and 12 cm from the surface of the phantom. The measurements confirm the presence of a focus around 4.18GHz at 7 cm.

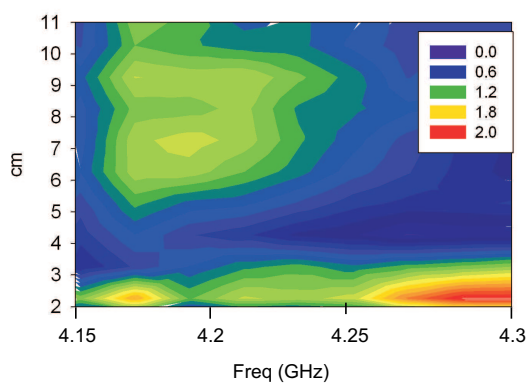

Figure 3: Measurement of the normalized field produced inside a phantom resembling the breast tissue. A focus appear at 4.18 GHz and at a distance of 7 cm.

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