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Coaxial Slot Antenna Array Design for Microwave Ablation and Monitoring

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Abstract

Tracking the ablated region in real-time can be advantageous during microwave ablation. In this study, an antenna array that ablates the tissue and monitors the ablation process is proposed. Monitoring is realized by tracking the coupling between the array elements which changes during ablation. That is because the permittivity and conductivity of organic tissues depend on temperature. The proposal is demonstrated through a hepatic tumor example with cascaded thermal and electromagnetic simulations. As the array element, a coaxial slot antenna is designed to operate at 2.45 GHz. The temperature-dependent permittivity and conductivity values are calculated through thermal simulations which contain both healthy and malignant tissues. It has been shown that the application of ablation with an output of 10 W per antenna causes a considerable change in coupling. A 100 MHz shift in frequency and 2.5 dB change in magnitude have been observed after 300 seconds.

1 Introduction

The use of thermal ablation therapy methods has become widespread in the treatment of tumors [1, 2]. The basic principle of these methods is to destroy malignant cells by changing the temperature of the target tissue. Among these techniques, radiofrequency, laser, ultrasound, and microwave ablation (MWA) aim to destroy the target tissue by increasing the temperature locally, while cryoablation does it by freezing the target area [3]. These methods are used in the treatment of liver [4], prostate [5], kidney [6], lung [7], breast [8], and thyroid [9] cancers. The treatment method is determined according to the structure of the tumor tissue and the region where it is located [10]. Higher temperature, larger ablation area, and shorter treatment time are among the advantages of MWA over other high-temperature thermal therapy methods [11].

In the MWA method, the heat released due to the vibration of molecules under the effect of changing electric field causes a rapid temperature rise above 60°C [2]. Another benefit of MWA is that a larger ablation zone can be obtained by activating more than one antenna simultaneously [12]. However, since a large ablation area is obtained in a short time, predictability and real-time visualization of the ablation area become important in order not to damage the healthy tissues around the tumor [10].

The prediction of the ablation area is not straightforward in clinical applications. For example, to characterize the MWA applicator for liver tumors, the ablation sizes for different input powers and ablation times are tabulated according to the measurements taken on the ex-vivo bovine liver. These measurements are used as guidance during a clinical procedure [13]. However, the electrical properties of the patient's liver may be quite different from that of the ex-vivo bovine liver. Hence, the measured dimensions may lead to inaccurate results for the ablation of in-vivo tissue. Therefore, the real-time monitoring of the ablation area during the application becomes important. The most commonly used real-time monitoring techniques in current MWA systems are computed tomography, magnetic resonance imaging, and ultrasound. Instead of using an additional imaging modality, it would be beneficial to conduct the monitoring by the ablation system itself.

The relative permittivity, ε_r, and conductivity, σ, of the liver are functions of the tissue temperature [14]. As the tissue temperature rises, both ε_r and σ of the liver drop significantly. Therefore, the treatment progress can be monitored by the change in the electrical properties. In [15], the ablation boundary is estimated by calculating the relative permittivity of the surrounding tissue with an open-ended coaxial slot antenna. A similar aim has been fulfilled in [13] by tracking the reflection coefficient with a water-cooled loop antenna. Following the same purpose, [16] monitors the treatment progress by utilizing resonators. All the aforementioned studies try to extract information from the reflection coefficient hence the tracking is restricted to the vicinity of the applicator. That is because the changes outside the near-field do not affect the reflection coefficient. To overcome this limitation, here, the transmission coefficient is used for tracking to enlarge the monitored area.

This paper presents the design of a double-slit dual-mode coaxial antenna system for microwave ablation and real-time tracking of the operation. The system can thermally ablate the malignant tissue and track the ablation area. To the best of the authors' knowledge, this technique has been applied for the first time in the literature. The rest of the paper is organized as follows. Section II describes the antenna design. The thermal and electromagnetic simulations are presented in Sections III and IV. Section V concludes the paper.

2 Antenna Design

In MWA, generally, needle-like antennas such as coaxial slit, dipole, or monopole antennas are preferred. The ablation zones created by these antennas are concentrated at the antenna tip. However, the electrical currents generated at the outer conductor of the coaxial cable expand the ablation zone along the antenna axis. Hence, the area becomes elliptical and damages the healthy tissue around the tumor. Increasing the number of slits reduces the backward currents in the antenna and makes the ablation region more spherical. That is the reason why a double-slit coaxial antenna system is chosen here.

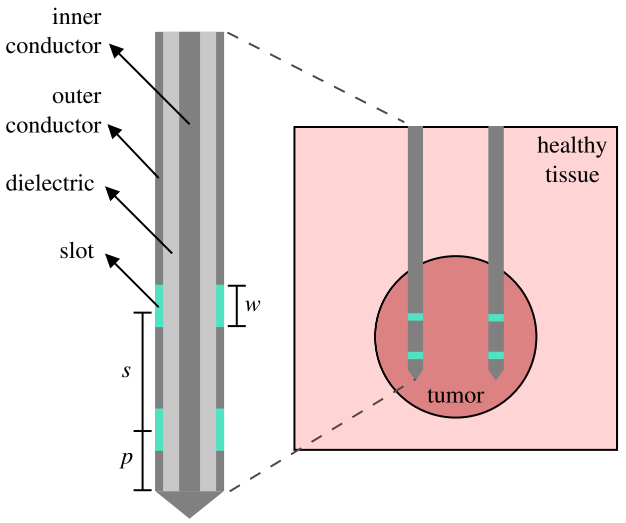


Figure 1. Antenna model and electromagnetic/thermal simulation setup.

The antenna model and the simulation setup can be seen in Figure 1. The width (w) of each slot, the distance in between the slots (s), and the distance from the tip of the antenna to the slot (p) are 0.6 mm, 9.2 mm, and 1.5 mm, respectively. The radii of the inner conductor, dielectric and outer conductor are 0.1 mm, 0.33 mm, and 0.43 mm, respectively. The conductors are chosen as stainless steel and PTFE is used as the dielectric material.

Table 1. The electrical properties of the healthy liver tissue and the hepatic tumor at 2.45 GHz at 37°C.

Properties	Healthy liver	Hepatic tumor
ε_r	43	55
σ (S/m)	1.69	1.99

The antennas are positioned inside a spherical hepatic tumor and the tumor is located in healthy liver tissue. The reflection coefficient of the antenna as it is inserted from the liver to the tumor can be seen in Figure 2. Note that the electrical properties of the healthy liver and hepatic tumor at 37°C is given in Table 1.

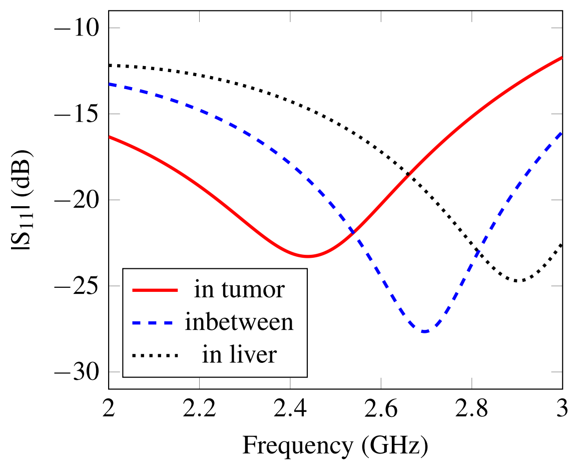


Figure 2. The reflection coefficient of the coaxial slot antenna as it is inserted from the liver to the tumor at 37°C.

3 Thermal Simulations

The temperature-dependent finite element simulations for the blood perfusion, the specific heat, and the thermal conductivity of the healthy liver and the hepatic tumor are realized as described in [14]. The healthy liver is modeled as a rectangular prism of 80 mm × 80 mm × 80 mm. The hepatic tumor, on the other hand, is modeled as a sphere of 20 mm radius. The distance between the antennas are 20 mm. The input power per antenna and the application time are set to 10 W and 300 seconds, respectively. The temperature distribution in the tissue is presented in Figure 4. The permittivity and conductivity values, which change with temperature as provided in Figure 3, are fed into electromagnetic simulations [14].

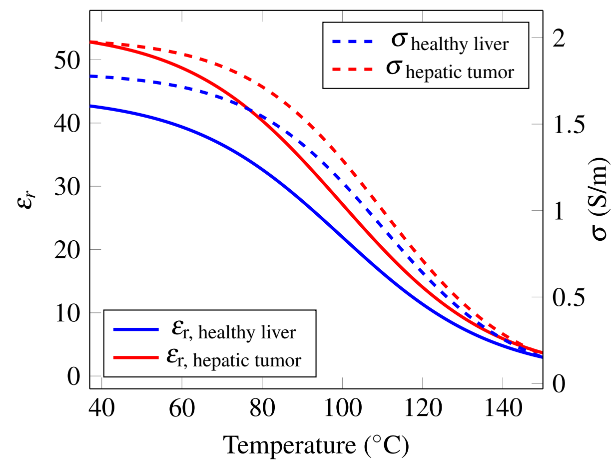


Figure 3. ε_r and σ of the healthy liver and hepatic tumor.

4 Electromagnetic Simulations

The electromagnetic simulations are conducted with temperature-dependent electrical properties. The coupling between the antennas, S_{21}, is numerically calculated for different time steps during the ablation process. Figure 5 presents how the coupling changes in time. Initially, the

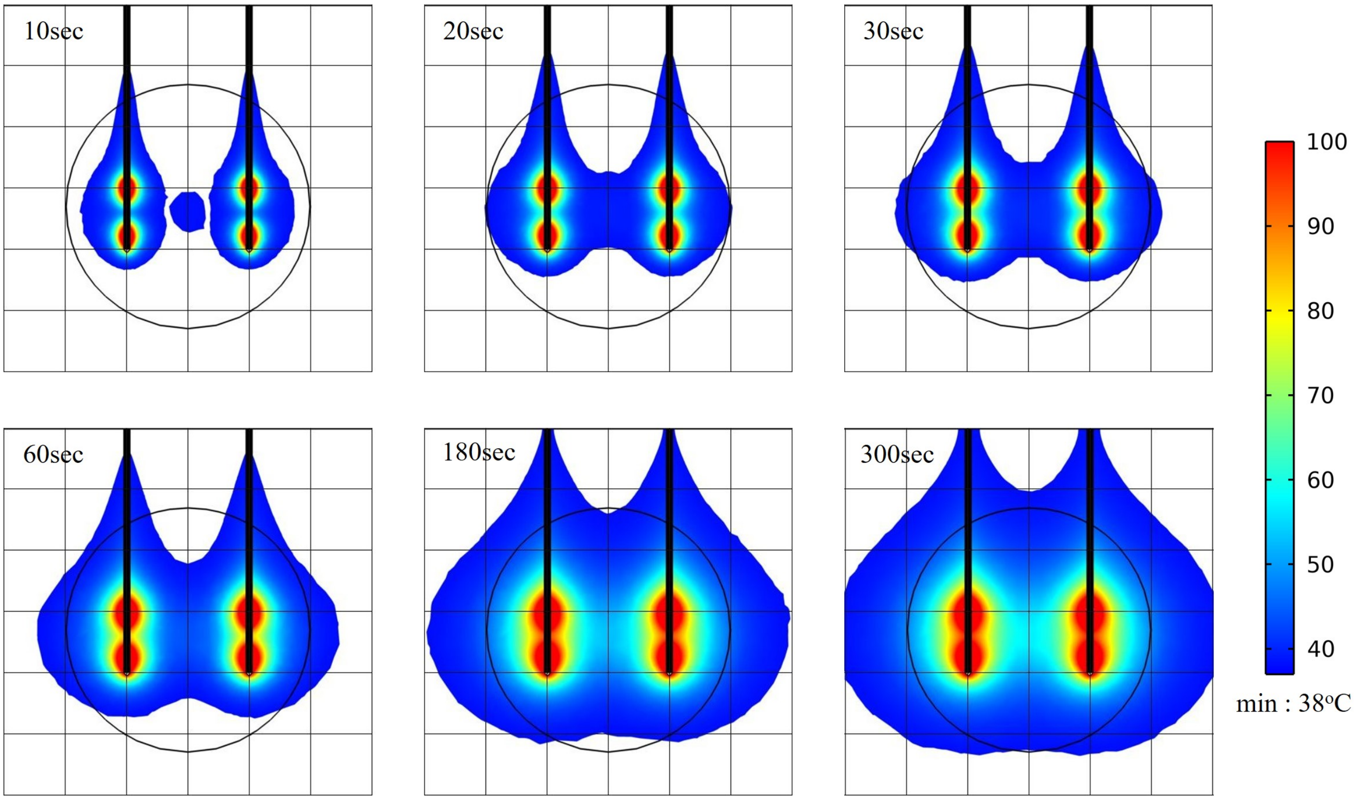
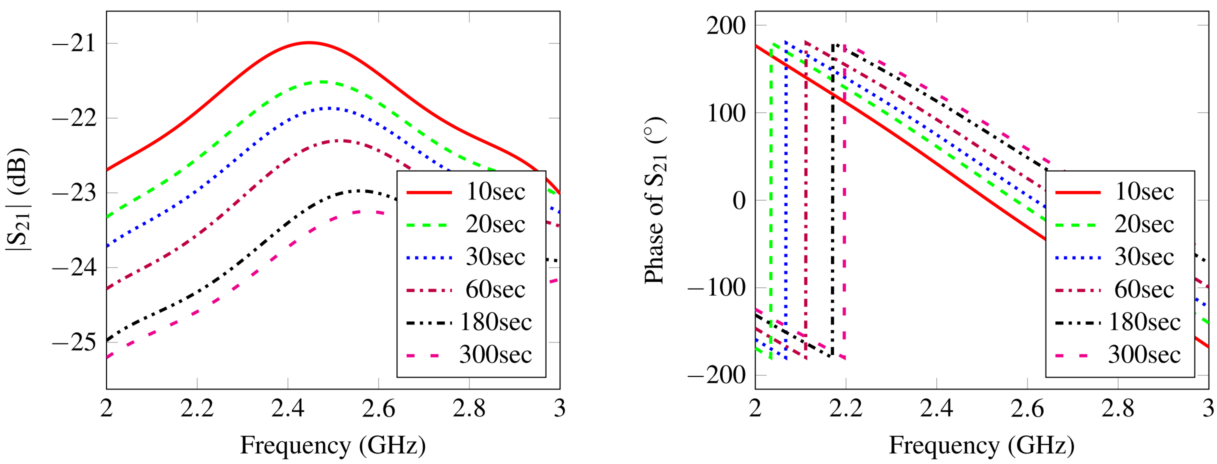


Figure 4. The thermal maps at 10, 20, 30, 60, 180, 300 seconds with 10 W input power for each antenna. Note that the circle represents the tumor and each unit cell is 1 cm in length.

ablation impacts the reflection coefficient of each antenna. The effect of this detuning decreases the transmission coefficient. As the ablated region grows beyond the near field, the reflection coefficient converges to a certain value. Then, the rate of change is expected to decrease. The monitoring is also possible by observation of the phase of the transmission coefficient as seen in Figure 5 (b).

5 Conclusion

An antenna array that can be used both for ablation and monitoring is proposed. During the operation, the relative permittivity and the conductivity of the tissues change considerably with temperature. Based on this difference, the transmission coefficient changes. Thermal and electromagnetic simulations are conducted with an output power of 10W per antenna for 300 seconds. The monitoring of the ablated area is demonstrated.



(a)

(b)

Figure 5. The magnitude (a) and the phase (b) of the transmission coefficient for different time steps.

6 Acknowledgements

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