2019 16th International Conference on Electrical Engineering, Computing Science and Automatic Control (CCE)
Mexico City, Mexico. September 11-13, 2019

Antennas Design for Microwave Ablation in Bone Tissue: Simulation and Experimental Validation

Texar Javier Ramírez Guzmán
Department of Electrical Engineering/Bioelectronics
CINVESTAV-IPN
México City, México.
texar.ramirezg@cinvestav.mx

Arturo Vera Hernández
Department of Electrical Engineering/Bioelectronics
CINVESTAV-IPN
México City, México.
arvera@cinvestav.mx

Lorenzo Leija Salas
Department of Electrical Engineering/Bioelectronics
CINVESTAV-IPN
México City, México.
lleija@cinvestav.mx

Citlalli Trujillo Romero
National Institute of Rehabilitation, Division of Research in Medical Engineering
México City, México.
cjtrujillo@inr.mx

Abstract — **The objective of this study was to analyze the performance of two types of antennas proposed for microwave ablation as possible treatment for bone tumors by making a comparison between simulations and experiments. The design and optimization, by using computational parametric models and the Finite Element Method (FEM), allowed to predict the behavior of two antennas, a Metal-Tip Monopole antenna (MTM) and a Double Slot Choke antenna (DSC) when they are in contact with bone tissue. The parameters to evaluate the coupling of the antennas were the standing wave ratio (SWR) values, heating patterns and temperatures reached. Optimized antennas were fabricated from an UT-085 semi-rigid micro-coaxial cable in order to replicate the design used in the FEM models. *Ex vivo* porcine bones were ablated with the fabricated antennas using 10 W during 5 min. Both antennas reached ablation temperatures around 60°C-100°C.**

Keywords — **Microwave ablation, monopole antenna, double slot antenna, thermotherapy.**

I. INTRODUCTION

Bone cancer is a significant worldwide public health issue. It is estimated that about 3,500 new cases will be diagnosed and 1,660 deaths from these cancers are expected for 2019[1]. This kind of cancer affect particularly children and adolescents, which implies these tumors have a major impact on the life of patients and their family[2]. The most common treatments to treat cancer are surgery (amputation), chemotherapy and targeted therapies, however, all of them generate a lot of side effects in the patient. In the worst case it may be necessary to amputate the entire limb to eradicate the entire tumor[3]. For this reason, new minimally invasive treatments are required to help eliminate tumors while minimizing collateral damage to healthy tissue and reducing the patient's recovery time, as well as the patient's quality of life.

Thermal ablation refers to the direct application of heat in the tumor to produce its destruction. To produce real therapeutic outcome, temperatures between 60°C-100°C are required. In the microwave ablation (MWA), a thin needle that functions as an antenna is inserted to apply the electromagnetic energy at microwave frequencies, in this case 2.45 GHz. The advantages of ablative therapies include faster recovery, lower cost and a less invasive procedure[4]. The energy applied to the tissue derives in the rotation of polar molecules, and this rotational energy is transformed into heat, achieving coagulative necrosis, this treatment generates rapid heating in the tissue, allowing treatments of shorter times[5].

The MWA has an advantage over other modalities. The heating of the tissue begins slowly with MWA, but it continues to occur beyond 60 s, resulting in a greater growth of the lesion, thus allowing a better control in the size of the lesion. Different kind of antennas as monopole[6], dipole[7] and choke[8] has been proposed for thermal therapies in soft tissue; however, antennas specifically designed for bone tissue are less investigated. In this study, the goal is to validate the MTM and DSC antennas using numerical simulations and experiments carried out in *ex vivo* swine bones.

II. METHODOLOGY

A. *Antenna design*

The main advantage of MTM and DSC antennas is the easiness of construction as well as simple geometry to generate the Finite Element Method (FEM) models[9]. Fig. 1 shows the general schematic for the MTM and DSC antenna. The antennas design were based on the use of UT-085 semi-rigid coaxial cable (See Fig. 1a). In addition, the antennas are covered with a Teflon-based catheter to prevent adhesion of the antenna to the tissue destroyed by ablation [10]. Parametric studies were carried out to determinate the antennas dimensions as antenna length, inner conductor, slot spacing, choke length and slot length. These dimensions were chosen based on the effective wavelength in the medium (bone) at 2.45 GHz, calculated by using the Equation (1).

$$\lambda_{eff} = \frac{c}{f\sqrt{\varepsilon_r}} \quad (1)$$

978-1-7281-4840-3/19/$31.00 ©2019 IEEE

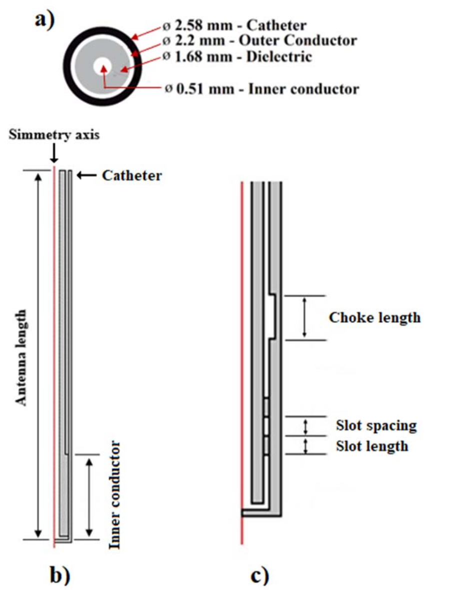


Fig. 1. Schematic design of the antennas. a) Dimensions of the coaxial cable, b) MTM antenna, c) DSC antenna.

where *c* is the speed of light equal to 300,000,000 m/s, *f* is the operating frequency of the microwave generator equal to 2.45 GHz and ε_r is the relative permittivity of bone equal to 18.5. The MTM and DSC antennas were modeled and optimized. The antennas reported here were chosen based on the lower SWR values, which means that there is a better power transmission to the tissue.

B. *Computational model*

The FEM was used to implement the electromagnetic and thermal simulations in COMSOL Multiphysics; where the Max-well´s and the bioheat equations are solved in order to evaluate the antenna performance over the surrounding tissue[11]. This model sets aside the heat source of metabolism. The external heat source is equal to the resistive heating generated by the electromagnetic field.

$$SAR = \frac{\sigma}{2\rho}|E|^2 \quad , \tag{2}$$

where σ is the tissue conductivity (S/m), ρ is the tissue density (kg/m^3) and E is the electric field generated by the antenna under study. The bioheat equation is used for heat transfer. This equation describes the problem of stationary heat transfer as:

$$\rho C\frac{\partial T}{\partial t} = \nabla(k\nabla T) + \rho Q + SAR - C_b W(T - T_b) \ , \tag{3}$$

Where k is the thermal conductivity of the tissue (W/m/K), ρ represents the density of the blood (kg/m^3), C_b is the specific heat capacity of the blood (J/(kg·K)), W is theblood perfusion rate (1/s) and T_b is the blood temperatura (K). In addition, SAR is the external heat source. The initial temperatura is equal to T_b in all domains. The initial tissue temperature in simulation and experimentation was 37°C. The parameters used in the FEM models are described in Table I.

In the 2D axisymmetric model, the inner and the outer conductors of the antenna (coaxial cable) were modeled using perfect electric conductor boundary conditions. The power point is modeled with a port boundary condition with a set power level. It is essentially a first-order boundary condition with low reflection with a field entry. The mesh used to generate each model had 0.0014 m minimum element size, with 6178 domain elements for DSC antenna and 7824 domain elements for MTM antenna.

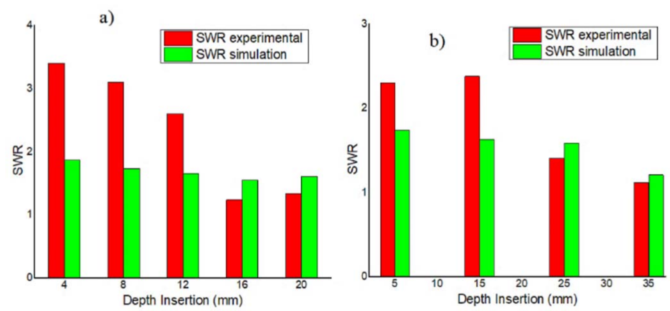
C. *Experimental validation*

The proposed antennas were built from UT-085 semi-rigid micro-coaxial cable to replicate the design used in the simulation. An Amphenol RF 132103 connector was used. Each antenna was inserted in *ex vivo* swine bones. The Technologies E5071B analyzer was used which allows us to measure the value of the SWR over a wide frequency range, the working frequency of the antennas (2.45 GHz) being of special interest. The radiation system consists of a power amplifier SSPA Aethercomm, a microwave generator Rohde & Schwarz SML03 that works at 2.45 GHz and a thermometry system Luxtron MAR05 STB that records the temperature during the experimentation. The antennas were feed with an input power of 10 W for 5 min. The MTM antenna was inserted 16 mm while the DSC antenna was inserted at 35 mm inside the bone. The sensors of the thermometry system were placed in the *ex vivo* tissue in such a way that a mapping of the temperature distribution is obtained. These sensors were placed next to the tip of the antenna at 0.5 cm, 1 cm, 1.5 cm and 2 cm. The heating pattern generated by the antenna was obtained with a thermographic camera (Fluke Ti32). Fig. 2 shows the experimental system configuration.

III. RESULTS

SWR values, reached temperatures and heating pattern were evaluated to analyze the behavior of the antennas.

TABLE I
DIELECTRIC AND THERMAL PORPERTIES USED IN THE MODEL[12]



A. *Standing Wave Ratio (SWR)*

The SWR is a measure of the impedance coupling that exists between a load and the transmission line that feeds it. The ideal SWR is 1; which indicates that all the input power had been transmitted to the tissue. A high value (> 2) indicates greater loss of power in the antenna and, therefore, greater power back to the microwave system.

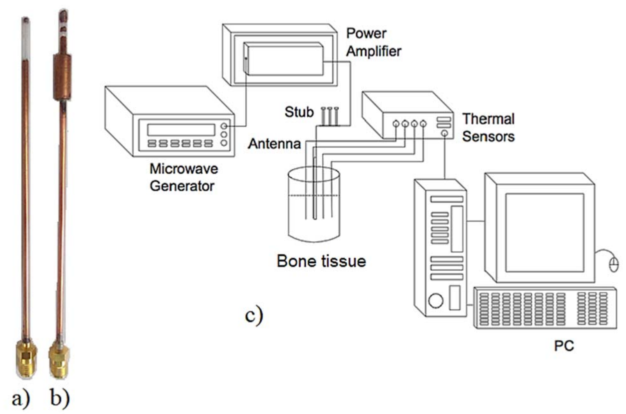


Fig. 2. MWA antennas and radiation system. a) MTM antenna. b) DSC antenna. c) radiation and thermometry system.

Table II shows the best-case dimensions of the optimized antennas obtained from the simulation and the SWR value, power loss and ablation volume for each antenna. These dimensions were used to fabricate the antennas. On the other hand, SWR values are dependent on the insertion depth of the antenna. Each optimized antenna was inserted at different levels in bone tissue. Fig. 3 shows the SWR values obtained in the simulation and in the experimentation when the optimized antennas are inserted in bone tissue at different insertion depth.

B. *Temperatures profiles*

To obtain real therapeutic outcomes, temperatures higher than 60°C are necessary. Four optical fibers (sensors) were used to acquire the temperatures reached when using the antennas. Fig. 4 shows the temperatures obtained in the simulation and experimentation during 5 min of radiation using 10 W of input power.

TABLE II
OPTIMIZED ANTENNAS DIMENSIONS

Parameters	MTM	DSC
Inner conductor	0.0142 m	--
Slot spacing	--	9.5E-4 m
Slot length	--	0.0018 m
Choke length	--	0.014 m
Choke thickness	--	0.0015 m
Input power	10 W	10 W
Power loss	1.25 W	0.52 W
Ablation volume	3.4 cm^3	5.3 cm^3
SWR	1.4	1.21

Fig. 3. SWR values at different insertion depth. a) MTM antenna and b) DSC antenna.

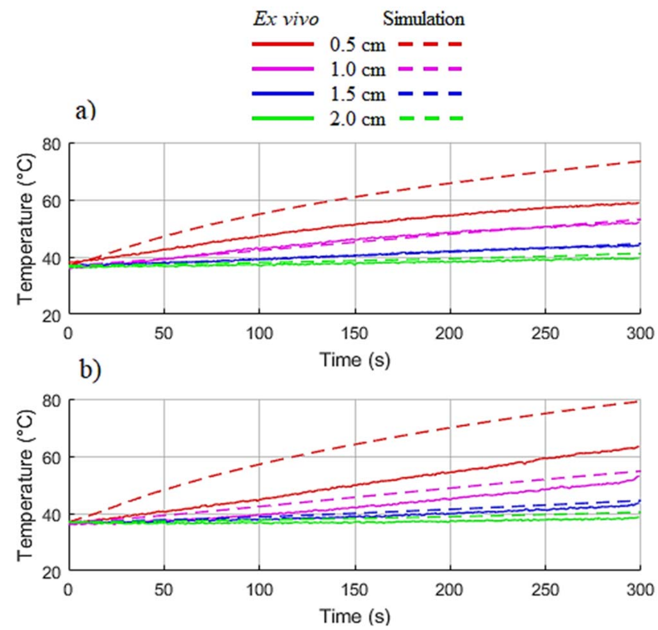


Fig. 4. Temperatures reached in the simulation and experimentation with 10 W input power and 5 min of radiation. a) MTM antenna and b) DSC antenna.

C. *Heating pattern*

Heating pattern is the measure of heat distribution in the human body or model. These patterns are a function of the amount of energy the tissues absorb; the higher the amount of energy deposited at a point, the higher the temperature will tend to be. Fig. 5 shows the heating pattern generated in the simulation and experimentation with the optimized antennas after 5 min of radiation and 10 W of input power. The maximum temperatures were obtained at the tip of the antennas. The MTM antenna generate heat distribution through the monopole, so the heating patterns are longer; on the other hand, the DS-C antenna generate heat distribution through the slots and the heating patterns have a spherical shape.

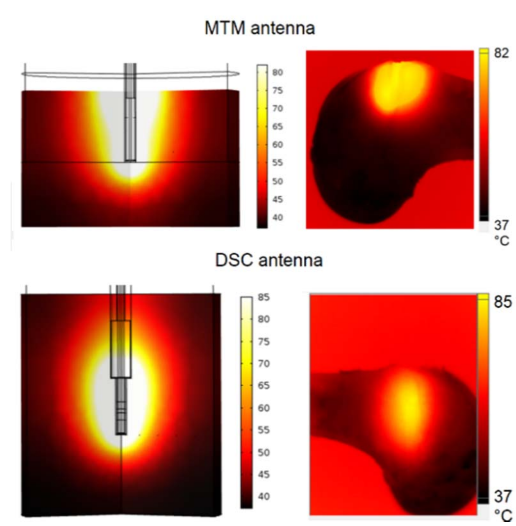


Fig. 5. Heating pattern generated in simulation (left) and experimentation (right) in bone tissue using MTM and DSC antennas.

IV. DISCUSSION

The coupling of the antennas with the tissue depends on the geometry of these as well as the depth of insertion. For the MTM antenna, the length of the inner conductor is required to be completely within the bone tissue in order to have a good coupling with the tissue and lower SWR values (<1.5); while the DS-C antenna it is necessary that both the slots and the choke are completely inside the bone to have lower SWR values (<1.5). In addition, the DS-C antenna shows better coupling at insertion depths greater than 30 mm, so this antenna can be proposed to treat tumors in greater depth in bone tissue.

Fig. 3 shows that the best SWR value for MTM and DSC antenna was obtained at 16 mm and 35 mm of insertion depth respectively, which means that DSC antenna has a better coupling with bone tissue and less power loss. Also, SWR values are lower in the greater insertion depth, if the inner conductor (MTM) and the choke (DSC) are totally inside the bone tissue, the SWR values are lower. So, the contact of the monopole and the choke of the antennas with bone tissue affects the SWR values. The percentage of maximum power loss is 12.3% for MTM antenna and 5.2% for DSC antenna. So, the DSC antenna produce the lowest percentage for 5 min of radiation and, comparing the ablation volume between MTM and DSC antenna, DSC antenna produce the highest ablation volume (5.4 cm^3) versus MTM antenna (3.4 cm^3).

The temperatures obtained (Fig. 4), as expected, are higher the closer they are from the tip of the antenna. Maximum temperatures for MTM and DSC were at 0.5 cm next to the tip of the antennas. Comparing the temperatures obtained for each sensor, its shows that temperatures between the simulation and experimentation are similar (ΔT=2°C) for the sensors farther to the tip of the antenna (1 cm, 1.5cm, 2 cm). While the sensor placed at 0.5 cm next to the tip of the antennas, shows a difference of approximately 15°C between the simulation and experimentation after 5 min of radiation; this difference of temperature may be due to the fact that in the experiment the tissue in contact with the antenna goes from being healthy tissue to necrotic tissue in a short period of time, while in the simulation all tissue is considered as healthy tissue during the entire time of radiation. In addition, the accuracy in the position of the temperature sensors could also influence the results; in the model the temperature values were taken at a precise point, while in the experimentation there may be variations in the placement of the temperature sensors.

Fig. 5 shows that the heating pattern depends of the kind of antenna, an MTM antenna shows a circular heating pattern while a DSC shows a spherical heating pattern. This indicates that the DSC antenna has the potential to reduce heating in the entire antenna body due to the presence of the choke. One of the advantages of using a choke antenna is that it allows that the heating is not generated throughout the whole antenna and it focuses the energy below the choke.

Choosing an antenna depends mainly on the size and shape of the tumor. The DSC antenna has a better coupling at a greater insertion depth compared to the MTM antenna; while the MTM antenna has a better coupling at a depth of 16 mm inside the bone. So, the DSC antenna promises to have better coupling with tumors that are located deeper in bone tissue.

V. CONCLUSION

Two antennas for MWA of bone tissue were designed and simulated by using an axisymmetric model in COMSOL Multiphysics. The results of the computational models allowed to know the optimal dimensions of each antenna in which a better coupling with the tissue to be treated is presented. Using these models, a metal-tip monopole and double slot choke antennas were fabricated and evaluated. The maximum temperatures reached were higher near the monopole (MTM) and the slots (DSC) of the antennas, and it decreases as a function of distance. With both proposed antennas, 60 ° C was exceeded, indicating that ablation temperatures were reached with an input power of 10 W applied for 5 min.

The implementation of the choke on the DSC antenna allows a better targeting of the electromagnetic energy below the choke and prevents the antenna body from heating up in its entirety. Finally, considering the analysis of the results, the DS-C antenna presents the best performance and expected coupling with the tissue with a SWR equal to 1.12 and a power loss of 0.52 W using 5W of input power in the *ex vivo* tissue experiments.

ACKNOWLEDGMENT

We appreciate the funding for the development of the work presented project: ERAnet-EMHE 200022, CYTED-DITECROD-218RT0545 and Proyecto IV-8 call Amexcid-Auci 2018-2020. Authors thanks J. H. Zepeda for his technical assistance.

REFERENCES

[1] "American Cancer Society. Facts & Figures 2019. American Cancer Society. Atlanta, Ga. 2019."

[2] A. Franchi, "Epidemiology and classification of bone tumors.," *Clin Cases Min. Bone Metab.*, vol. 9, pp. 92–95, 2012.

[3] B. Erol, T. Er, O. E. Aycan, and O. M. Topkar, "Bone Tumor: Types, causes and symptoms," *Musculoskelet. Res. Basic Sci.*, pp. 603–629, 2015.

[4] R. Ortega-Palacios. C. J. Trujillo-Romero, M. F. J. Cepeda, "Feasibility of Using a Novel 2.45 GHz Double Short Distance Slot Coaxial Antenna for Minimally Invasive Cancer Breast Microwave Ablation Therapy; Computational Model, Phantom, and In Vivo Swine Experimentation," *J. Healthc. Eng.*, 2018.

[5] P. Moroz, S. Jones, and B. N. Gray, "Status of hyperthermia in the treatment of advanced liver cancer," *J. Surg. Oncol.*, vol. 77, no. 4, pp. 259–269, 2001.

[6] T. West and T. Alster, “Percutaneous RF Interstitial Thermal Ablation in the Treatment of Hepatic Cancer,” no. September, pp. 759–768, 1996.
[7] H. Reinblod F, Lorenz WJ, “A dipole antenna for interstitial microwave hyperthermia,” *IEEE Trans. Microw Theory Tech*, 1995.
[8] J. Deshan Yang, Mark Converse, John G. Webster, David M, “Antenna design for microwave hepatic ablation using an axisymmetric electromagnetic model,” *Biomed. Eng. Online*, 2006.
[9] C. J. Trujillo-Romero. G. Rico-Martinez, L. Leija, A. Vera, J. Gutierrez-Martinez, “Microwave ablation to treat bone tumors by using a double slot antenna: a modelling study.”
[10] F. Lujan, J. Gutierrez-Martinez, A. Vera, L. Leija, C.J. Trujillo-Romero, “Antenas Micro-coaxiales de Doble Ranura y Doble Ranura con Recubrimiento Para Tratar Tumores Óseos: Modelado Teórico.”
[11] P. Prakash, “Theoretical modeling for hepatic microwave ablation,” *Open Biomed Eng*, vol. 4, pp. 27–38, 2010.
[12] N. Hasgall, E. Neufeld, M. Gosellin, A. Klingenbock, “IT’IS database for thermal and electromagnetic parameters of biological tissues,” *[Online]. Available: http://www.itis.ethz.ch/itis-for-health/tissue-properties/oerview/*, 2012.