

Microwave Ablation With a Triaxial Antenna: Results in *ex vivo* Bovine Liver

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Abstract—We apply a new triaxial antenna for microwave ablation procedures to an *ex vivo* bovine liver. The antenna consists of a coaxial monopole inserted through a biopsy needle positioned one quarter-wavelength from the antenna base. The insertion needle creates a triaxial structure, which enhances return loss more than 10 dB, maximizing energy transfer to the tissue while minimizing feed cable heating and invasiveness. Numerical electromagnetic and thermal simulations are used to optimize the antenna design and predict heating patterns. Numerical and *ex vivo* experimental results show that the lesion size depends strongly on ablation time and average input power, but not on peak power. Pulsing algorithms are also explored. We were able to measure a 3.8-cm lesion using 50 W for 7 min, which we believe to be the largest lesion reported thus far using a 17-gauge insertion needle.

Index Terms—Ablation, electromagnetic heating, microwave (MW) ablation, MW heating, RF.

I. INTRODUCTION

SURGICAL resection remains the gold standard treatment for primary or metastatic liver cancer. However, a majority of patients are not eligible for surgery due to a high number of tumors, difficult location of tumors, or associated comorbidities [1]–[4]. This has led to the development of minimally invasive alternatives for the treatment of liver tumors, notably focal tumor ablation. The most popular modality, RF ablation, has been plagued by high post-treatment local recurrence rates [5]–[7]. RF ablation has several drawbacks related to its mode of heating. It uses alternating electrical current to create ionic agitation, which produces frictional heat and subsequent tissue necrosis [8]. Ground pads must be placed on the patient to create a conduction path, leading to a risk of ground pad burns due to improper pad placement or high-input powers. With RF, long treatment times may be necessary due to tissue desiccation and charring. Charring increases the RF impedance and decreases the amount of energy that can be deposited into the tissue, resulting in long treatment times. Finally, current RF systems are

limited to the use of only one probe at a time due to electrical interference between probes. This means that all but the smallest tumors must be treated with sequential overlapping ablations.

Microwave (MW) ablation, like RF ablation, uses localized heating to cause tissue necrosis. However, MW systems can produce greater and more rapid heating and can easily support the use of multiple probes. Compared to RF, MWs have a theoretically broader field of power density, with a correspondingly larger zone of active heating [9]. This may allow for more uniform tumor kill both within a targeted zone and in perivascular tissue. Furthermore, MW heating does not rely on a conduction current path, which eliminates ground pad and charring concerns. Lesion size is mainly limited by the available power and treatment time. Despite these theoretical advantages, current MW systems use probes that are larger than desirable for percutaneous use (13–15 gauge), and create relatively small zones of necrosis (1.6 cm) relative to insertion size [10]. This has led to the need for multiple overlapping ablations or antenna arrays, both of which require additional procedure time and invasiveness.

Most ablation antennas are fed by coaxial lines, which have an unbalanced design that allows return current flow on the outer conductor. These currents restrict impedance matching. If the antenna's input impedance is not matched to the feed line, too much of the applied power is reflected from the antenna and, hence, not deposited in the tissue. A mismatch also causes standing waves that can overheat the coaxial feed, causing superficial burns or cable failure.

Numerous previous antenna designs have been presented in the literature for MW ablation [11]–[18]. Several of these designs are specifically targeted for cardiac applications [11]–[13], while others are intended for hyperthermia applications, where lower powers and longer treatment times are acceptable [11], [14]. The tipped monopole designs of [11] and [12] have good SAR patterns, but relatively large reflection coefficients that may restrict power handling of the feed and deposition by the antenna. It is well known that power-handling ability of a coaxial line is strongly related to cable diameter. The dipoles presented in [14] and [15] and the choked designs presented in [13] and [16]–[18] are characterized by low reflections and good SAR patterns, but require an insertion up to 1.5 mm larger in diameter than the feed cable. The smallest profile choke requires a 14-gauge (2.1 mm) introduction [17].

The triaxial antenna presented in this paper is optimized for power distribution with the smallest possible insertion. The impedance match between the feed line and antenna allows the maximum power rating of a given feed to be used. Feed-line

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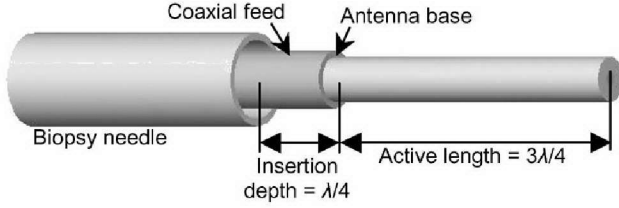


Fig. 1. Schematic representation of a coaxial monopole antenna inserted through a biopsy needle. The active length of the antenna and the insertion depth through the needle may be tuned to account for changes in tissue properties.

TABLE I
MATERIAL PROPERTIES USED FOR SIMULATION AT 37 °C, 2.45 GHz [20]

		Liver	Saline (0.9%)	PTFE	Copper
Relative Permittivity	ϵ_r	45.6	78.0	1.8	1.0
Electrical conductivity (S/m)	σ	1.97	1.8	$1.6 \cdot 10^{-5}$	$5.8 \cdot 10^7$
Thermal conductivity (W / m K)	k_T	0.564	0.7	0.24	385
Specific heat (J / kg K)	c_p	3400	4183	1050	386
Density (kg / m ³)	ρ	1050	998	1200	9000

heating is minimized. By using the full power rating, more energy may be deposited into the tissue and larger lesions realized. In addition, since the introducing needle is integral to the design, the total insertion diameter remains small enough for safe percutaneous use.

II. ANTENNA DESIGN AND NUMERICAL MODELING

A. Antenna Design

The triaxial antenna design is created from a coaxial monopole and needle together (Fig. 1) [19]. The active length of the antenna loaded in tissue (see Table I for material properties) is nominally $(2n - 1)\lambda/4$, where n is an integer. The needle sheath creates a triaxial structure and is positioned $(2n - 1)\lambda/4$ away from the antenna base. This is the insertion depth. Positioning the needle in this way improves return loss and reduces fields flowing back on the coaxial outer conductor. In turn, more energy is deposited in the tissue and less heating of the feed line occurs.

In order to introduce the antenna percutaneously, a biopsy needle with a removable introducer (Bard Medical, Covington, GA) is inserted into the patient. Once the proper needle position has been achieved, the introducer is removed. The coaxial antenna is then inserted through the needle and positioned correctly to provide the required tissue heating.

The coaxial feed lines and antennas used with 18- and 17-gauge insertion needles are created from UT-34-M (0.95 mm) and UT-47-M (1.26 mm) semirigid cable (Micro-Coax, Pottstown, PA), respectively. The entire outer conductor is copper, while the inner conductor is made from silver-plated copper wire. The coaxial dielectric used is a

TABLE II
TEMPERATURE COEFFICIENTS USED IN SIMULATIONS. UNITS ARE %°C⁻¹ UNLESS NOTED OTHERWISE [20], [21]

	ϵ_r	σ	k_T	c_p	ρ
Liver	-0.2	+1.2	0.2	-0.01	$0.4 \cdot 10^{-4\dagger}$
PTFE	-0.003	+0.01	0	0	0

† Estimate based on bovine and canine values for muscle.

low-loss polytetrafluoroethylene (PTFE), which results in a still-air power rating of 34 W for the 0.95-mm cable and 62 W for the 1.26-mm cable.

The biopsy needle material is stainless-steel coated with $\sim 10 \mu\text{m}$ of Parylene “C” to reduce the coefficient of friction on the needle surface. Parylene “C” was used because of its biocompatibility, conformal coating, and ease of application.

B. Numerical Modeling

Heating effects from the antenna are modeled using the finite-element method (FEM) with multiphysics software (COMSOL Inc., Burlington, MA). The coupled system to be solved is given by the time-harmonic wave equation

$$\nabla^2 \mathbf{E} = \omega^2 \left(\epsilon_r(T) - j \frac{\sigma(T)}{\omega} \right) \mathbf{E} \quad (1)$$

and heat-transfer equation

$$\rho c_p \frac{\partial T}{\partial t} = k_T \nabla^2 T + \frac{\sigma}{2} |\mathbf{E}|^2 - Q_h \quad (2)$$

where ω is the angular frequency (in radians per second), $\mathbf{E}$ is the electric field vector (volts per meter), T is local temperature (kelvin), and Q_h represents the heat lost to perfusion or gained by metabolic processes. In *ex vivo* samples, Q_h can be ignored since no perfusion or metabolism exists. Material parameters are assumed to be isotropic and homogeneous. Temperature variances of PTFE and liver are included in the model, as outlined in Table II. Temperature dependencies of metallic components and mechanical changes in PTFE are neglected since the resulting effects are relatively small. Since the antenna is axisymmetric, the model is reduced to a two-dimensional space to limit computational burdens.

The simulation is run for application times up to 15 min with average input powers from 10 to 80 W. Pulsed algorithms using peak powers up to 300 W and pulse times from 0.1 ms to 120 s are also simulated for comparison to *ex vivo* measurements. Care must be taken when comparing numerical and experimental data since the computer simulation calculates lesion size based on the 60 °C contour, where cell death occurs within seconds, while post-ablation measurements rely on visual inspection of tissue necrosis. Visual inspection has been shown to be in good agreement with histology [18].

III. EXPERIMENTAL METHODS

After validation of the antenna design, several ablation protocols are tested in order to map out the lesion size dependence on independent variables (IVs) such as application time, average power, peak power, and pulse on/off times (Table III).

TABLE III
ABLATION PROTOCOLS TO BE TESTED *ex vivo*.
“@” INDICATES THE TESTING INTERVAL

Protocol IV	Application time	Average power	Peak power	Pulse times
Application time	1-12 min @ 1 min	25 W (18 ga) 50 W (17 ga)	N/A	N/A
Average power	10 min	5-30 W @ 5 W (18 ga) 5-65 W @ 10 W (17 ga)	N/A	N/A
Peak power	8 min	40 W	40-300 W @ 20 W	0.1 ms – 120 s
Pulse times	8 min	25-40 W	40-300 W @ 20 W	0.1 ms – 120 s

During the testing of each protocol, only the IV is altered, while all other variables are kept as fixed as possible. The forward and reflected powers are continuously measured from the source with ± 0.3 W accuracy.

Finally, a repeatability study is conducted using seven samples and the following parameters.

- $P_{in} = 50$ W.
- $t = 7$ min.
- $T_0 = 37$ °C.

This study is used to evaluate the variability in lesion size due to any spatial and temporal variations in tissue and cable properties. By keeping the initial temperature constant, we remove the large variance in lesion size caused by changing initial temperature. The results are used in conjunction with thermal modeling to predict *in vivo* lesion sizes.

At the end of each ablation, tissue temperature is measured from a thermocouple probe inserted into the middle of the lesion to monitor maximum lesion temperature. The lesion is then divided longitudinally along the insertion track. Measurements are taken of the lesion diameter using a straight edge ruler or caliper and are based on visual inspection of the area of necrosis. Measurement accuracy is estimated at ± 0.5 mm.

IV. RESULTS

Both numerical and experimental results show that, for this triaxial design, Γ is minimized for a probe length of 12.3 mm and insertion depth of 3.5 mm in liver tissue (Fig. 2) and that the distance between the outer conductor and insertion needle is not a critical factor for antenna performance. In fact, having the needle in contact or near contact with the outer conductor provided lower reflection coefficients in both simulated and empirical data [19].

For the coaxial monopole in saline, Γ was measured to be -35.0 dB at 2.45 GHz. When inserted 3.5 mm through a biopsy needle, Γ was reduced to -46.2 dB. Thus, the triaxial design improved the reflection coefficient by 11.2 dB over the simple coaxial monopole, as seen in Fig. 3.

Γ was then measured to be -25.0 dB (0.32% reflected power) for an active length of 12.3 mm and insertion depth of 3.5 mm in an *ex vivo* bovine liver. Reflection coefficient is most likely

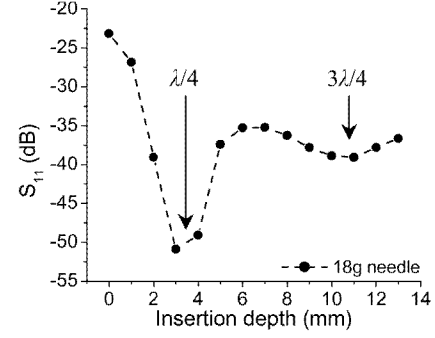


Fig. 2. Insertion depth versus Γ measurements. Γ is minimized for a 3.5-mm insertion depth in liver tissue [19].

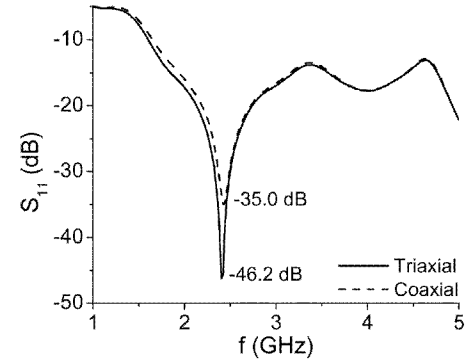


Fig. 3. Measured S_{11} of a coaxial monopole and a triaxial antenna. The triaxial design reduces reflections by 11.2 dB [19].

limited by inhomogeneity in the dielectric constant of the liver tissue.

A. Lesion-Size Dependence on Application Time

Experimental data for this study were taken on two successive days. On the first day, initial tissue temperatures were measured to be 18.0 °C ± 0.5 °C, while on the second day initial temperatures were 11.0 °C ± 0.5 °C. Shortly after ablation, temperatures in the zone of necrosis were measured to be in excess of 80 °C, far surpassing the commonly used metric of ~ 60 °C for instantaneous cell death. Numerical simulations were carried out with the same initial parameters as in the experiments.

Fig. 4 shows the results of the application time protocol. Lesion diameters increased with time with diminishing increases as application time increased according to the model

$$Y = Y_0 + Y_1 t + Y_2 \sqrt{t} \quad (3)$$

where Y is the lesion diameter and t is the application time, a trend that is also noted in simulations. The largest lesion measured was 3.9 cm in diameter and was limited in size by the initial tissue temperature.

Discrepancies between numerical and experimental values are most likely due to the metrics used and an insufficient tissue model. Recall that simulated lesion size is defined from the 60 °C contour line, which corresponds indirectly to actual lesion size. More importantly, data from Table II are only characterized in the literature for tissue temperatures up to ~ 70 °C. Better characterization of tissue properties above 100 °C, where dehydration and charring are likely to reduce heat conduction

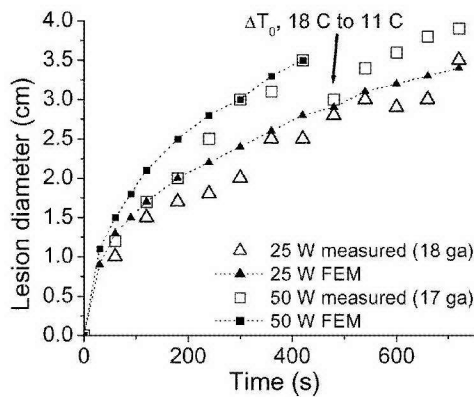


Fig. 4. *Ex vivo* measurements of lesion diameter versus application time. Note the change of T_0 in the 18-gauge data set. Lesion size in numerical simulations was defined as the maximum diameter of the 60 °C contour.

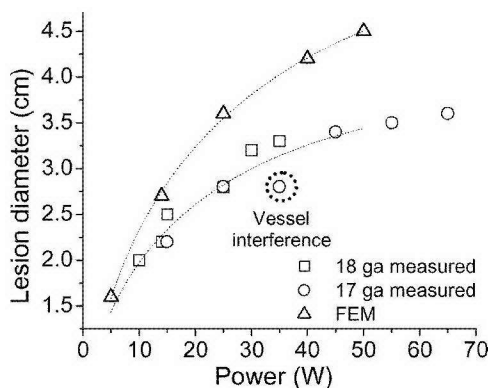


Fig. 5. *Ex vivo* measurements of lesion diameter at 10 min versus average input power. Simulations were identical for both antenna sizes. The circled point was limited in size due to a large vessel in the active region of the antenna. Dotted lines represent fits of the data to (3).

and bulk electrical conductivity in the tissue, may produce more congruent data between simulation and experiment.

Numerical data also predicts that the same trend will occur for *in vivo* studies, but that the slope change will decrease faster with time due to the larger effective thermal conductivity, i.e., graphs like Fig. 4 will appear flatter at longer time points, where thermal conduction most directly controls lesion size.

B. Lesion-Size Dependence on Average Input Power

For this protocol, initial ablation temperatures ranged from 15.0 °C ± 0.5 °C to 18.0 °C ± 0.5 °C. Again, final temperatures in the core of the lesion were measured to be in excess of 80 °C.

Fig. 5 demonstrates the dependence of lesion size on average input power. We find the same functional dependence as (3) for lesion size and input power in both simulation and experimental results. Again, simulations tended to overestimate the size of the lesions as power increased. The 17-gauge outlier at 35 W was likely due to a vessel running through the active region of the antenna, as in Fig. 6.

Vessels in *ex vivo* samples are filled with fluid and air, which create material inhomogeneities, increase reflected power, and reduce the energy deposited in the tissue. This likely caused the smaller lesion size. Many of the data points in these protocols had very inhomogeneous active regions—mostly due to empty vessels and bile ducts—and were not the ideal cases studied in

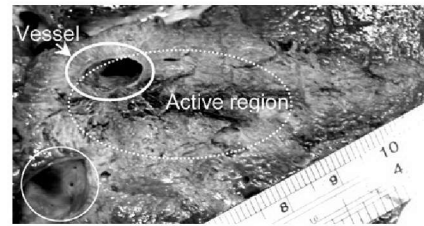


Fig. 6. Example of vessels running through the active region of the antenna. Vessels cause inhomogeneity and anisotropy of the antenna medium and, thus, alter the radiation characteristics, reduce the impedance match, and reduce power deposition into the tissue.

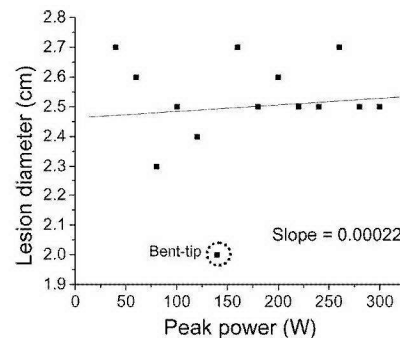


Fig. 7. *Ex vivo* measurements of lesion diameter versus peak power with average power held constant at 25 W. The slope of the best linear fit is nearly zero. Thermal simulations give similar results—that the lesion diameter depends primarily on the energy applied, not the power.

computer simulations. The effects of inhomogeneities on the resulting lesions can be seen, but since no liver will present a perfectly homogeneous medium, all of the data points measured are included here.

C. Lesion-Size Changes With Pulsed Power and Pulse Time

The final protocols provided very little change from previous results. There seems to be no dependence of lesion diameter on peak applied power (Fig. 7). Similar amounts of energy were deposited into the active region of the antenna because of the fixed average power and application time. This makes qualitative sense since total energy, not power, is the appropriate metric for measuring heat generation. Thus, in this case, we have a large sample of 25-W lesions created in tissues with similar initial temperatures of 16 °C–18 °C. The mean diameter in this case is 2.5 cm and the standard deviation is 0.184. One data point is considered an outlier because the antenna was unknowingly bent during insertion. If we ignore the bent-tip data point at 140 W, the average increases slightly to 2.54 cm, while the standard deviation improves to 0.119 cm.

Changing the pulse times did very little to improve lesion diameter, which depended mostly on the average power applied. When pulses from 15 to 120 s were used with duty cycles above 50%, lesions decreased somewhat in size as duty cycle decreased. The most notable feature of this experiment was that lesion diameter decreased more rapidly when duty cycle used was less than 50%. These trends were also observed in the thermal simulations. The largest lesions were created with large duty cycles, indicating that simply using constant power is a better option.

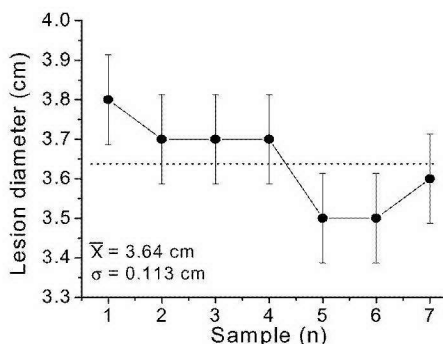


Fig. 8. Repeatability of lesion size. Lesions were created with 50 W, continuous, for 7 min using the triaxial antenna. The dashed line at 3.64 cm represents the mean value.



Fig. 9. 3.8-cm lesion created with 50 W for 7 min.

The effect of this type of pulsing *in vivo* has yet to be studied, but because the cooling times are assumed to be faster *in vivo*, some sort of pulsing algorithm may be warranted. For example, a relatively short high-power pulse might be applied to coagulate a blood vessel and reduce the effects of perfusion. Lower CW power might then be used to create the appropriate zone of necrosis. Using pulsed powers may also reduce fatigue in the feeding cables and antennas, which may result in a more robust system.

D. Repeatability Study

Results of the repeatability study may be found in Fig. 8. In this study, the mean lesion diameter ($\bar{x}$) was found to be 3.64 cm with a standard deviation (σ) of 0.113 cm, similar to the results obtained in the peak-power study. We also note that the lesion size tends to decrease slightly with each successive ablation and attribute this fact, in part, to temporal changes in the feed cable induced by temperature variations. An image of the largest lesion from this study is shown in Fig. 9 for reference.

V. CONCLUSION

A triaxial antenna design for clinical MW ablation was tested in an *ex vivo* bovine liver. Previous numerical and experimental

data provided show that the triaxial antenna achieves return losses as low as -46.2 dB in saline and -25.0 dB in an *ex vivo* bovine liver tissue, with the triaxial design improving return losses by up to 11.2 dB over a coaxial design. This matching allows high powers to be used in relatively small diameter antennas for procedures at least 12 min in length. The antenna is capable of being presented inside 18- or 17-gauge needles, making percutaneous introduction feasible.

Ex vivo data compiled show that lesion size is a function of both average input power and application time. Since average input power is limited by the power-handling capability of the coaxial feed line, improvement is possible with larger cables, cables with lower loss, or better heat transfer characteristics. However, larger cables are not desirable due to their invasive nature.

No significant dependence was observed on peak power. Using pulse on/off times ~ 10 s was not beneficial in *ex vivo* simulations or measurements and, in fact, decreased lesion size when the duty cycle fell below 50%.

We found that an average lesion diameter of 3.64 cm is possible using a conservative 50 W of input power for 7 min, which makes MW ablation a viable surgical alternative. Thermal modeling predicts that the lesion size will decrease $\sim 25\%$ for this time-power protocol *in vivo*. If 65 W is able to be used *in vivo*, then lesion diameters near ~ 3.0 cm are predicted. Further testing *in vivo* is warranted and ongoing.

Comparisons between datasets must be taken with care. Both simulated and measured data suggest that changing the initial tissue temperature from $\sim 10^\circ\text{C}$ to $\sim 25^\circ\text{C}$ results in a $\sim 30\%$ change in lesion size, clearly demonstrating the influence of initial conditions.

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