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Expanded modeling of temperature-dependent dielectric properties for microwave thermal ablation

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Abstract

Microwaves are a promising source for thermal tumor ablation due to their ability to rapidly heat dispersive biological tissues, often to temperatures in excess of 100 °C. At these high temperatures, tissue dielectric properties change rapidly and, thus, so do the characteristics of energy delivery. Precise knowledge of how tissue dielectric properties change during microwave heating promises to facilitate more accurate simulation of device performance and helps optimize device geometry and energy delivery parameters. In this study, we measured the dielectric properties of liver tissue during high-temperature microwave heating. The resulting data were compiled into either a sigmoidal function of temperature or an integration of the time–temperature curve for both relative permittivity and effective conductivity. Coupled electromagnetic–thermal simulations of heating produced by a single monopole antenna using the new models were then compared to simulations with existing linear and static models, and experimental temperatures in liver tissue. The new sigmoidal temperature-dependent model more accurately predicted experimental temperatures when compared to temperature–time integrated or existing models. The mean percent differences between simulated and experimental temperatures over all times were 4.2% for sigmoidal, 10.1% for temperature–time integration, 27.0% for linear and 32.8% for static models at the antenna input power of 50 W. Correcting for tissue contraction improved agreement for powers up to 75 W. The sigmoidal model also predicted substantial changes in heating pattern due to dehydration. We can conclude from these studies that a sigmoidal model of tissue dielectric properties improves prediction of experimental results. More work is needed to refine and generalize this model.

1. Introduction

Ablation techniques are applied clinically to treat benign and malignant tumors in the abdomen, chest and musculoskeletal system with other locations under investigation (Callstrom *et al* 2006, Gervais *et al* 2005, Huston and Simmons 2005, Rose *et al* 2006). Preclinical studies show that microwaves are attractive and potentially superior compared to RF ablation in several respects, such as producing more rapid volumetric heating, a greater zone of active heating, less susceptibility to the heat-sink effect of local blood flow and options for multiple-antenna arrays that exploit wave phenomena (Andreano *et al* 2010, Brace *et al* 2007, Oshima *et al* 2008, Wright *et al* 2003). However, few systems to date have capitalized on the potential benefits of microwave energy, in part, because many basic aspects of how tissue responds to high-temperature heating in the microwave regime have not been well studied. As a result, the potential of microwave ablation remains largely untapped.

To date, design and characterization of microwave ablation devices has primarily been performed by simulating electromagnetic performance using the finite-element method (FEM), and finite-difference time-domain method (Phasukkit *et al* 2009, Torres and Jecko 1997, Tungjikusolmun *et al* 2002, Yang *et al* 2007b). While useful for initial design, these

simulations have most commonly been performed using a static set of tissue properties, which do not account for changes that occur when tissue is exposed to high temperatures. Some studies have reported temperature-dependent dielectric properties of a few tissues; however, none of these investigations considered the properties of tissues at temperature above 70 °C (Chin and Sherar 2004, Gabriel *et al* 1996, Lazebnik *et al* 2006, Pop *et al* 2003, Sipahioglu *et al* 2003, Stauffer *et al* 2003). In fact, temperatures during microwave ablation easily exceed 100 °C (Brace *et al* 2005, Yang *et al* 2007a). At these high temperatures, tissue undergoes structural changes and water loss, which can substantially affect tissue properties (Foster and Schepps 1981, Sapareto and Dewey 1984, Bircan and Barringer 2002). Therefore, models based on a linear or even quadratic temperature dependence may not be adequate to predict intense microwave heating. More accurate simulation of the ablation process would facilitate more rapid design and validation of new devices or energy delivery techniques by reducing the number of experimental validation studies required.

In a recent study, tissue dielectric properties were measured simultaneously with temperature during multiple-electrode RF ablation (Brace 2008). That work demonstrated the feasibility of measuring properties during thermal ablation and that severe changes in tissue dielectric properties occur when temperatures approach 100 °C. However, that study was limited by its use of RF devices to produce the ablation zone, which cannot generate temperatures over 100 °C. The objective of this study was to measure the relative permittivity and conductivity of *ex vivo* bovine liver tissue during high-temperature microwave ablation, and then construct and validate a numerical model based on those results. We hypothesized that simulations carried out using the new model would more accurately predict microwave heating than existing models.

2. Methods

2.1. Dielectric property measurement

Fresh livers were obtained from a local slaughterhouse, sealed in an air-tight container and stored at 4 °C. Before each experiment, 6 cm × 6 cm × 9 cm sections of the liver were extracted, sealed in air-tight containers and allowed to warm to room temperature (20 °C). Each sample was then placed into a plastic container with normal saline to reduce air pockets inside the tissue (figure 1).

Microwave ablations were produced using a coaxial monopole antenna and magnetron-based microwave generator (GENLRS0.3, 2.45 GHz; CouberMuegge LLC, Norwalk, CT). The monopole antenna was fabricated from a 16 cm length of 0.085" semi-rigid coaxial cable (UT-85-C; Micro-Coax, LLC, Pottstown, PA), from which a 13 mm length of the outer conductor was stripped. PTFE heat shrink tubing (McMaster-Carr; Elmhurst, IL) covered the antenna to ease extraction from ablated liver.

Liver relative permittivity and conductivity were measured during microwave ablation of six samples by using the open-ended coaxial probe technique with open, short and water-load calibration (Gabriel *et al* 1994, Misra *et al* 1990, Stuchly and Stuchly 1980). The coaxial probe was attached to a vector network analyzer (Agilent 8753ES from Agilent Technologies, Palo Alto, CA), which was time-gated according to the technique described in Misra *et al* (1990). A fiber optic temperature sensor with 1.15 mm outer diameter (Neoptix, Quebec, Canada) was bundled flush with the open-ended coaxial probe to ensure that dielectric properties and temperature were measured at the same location. The bundled probes were placed into the tissue at about 1 cm away from the antenna flush with the expected zone of maximal ablation diameter to capture the region of most substantial temperature elevation (figure 1). Complex reflection coefficients from 500 MHz to 6 GHz in 18.3 MHz intervals were recorded along with temperatures at 1 Hz via Labview (v8.5,

National Instruments, Austin, TX). Reflection coefficient data were processed using Matlab (v7.6, Mathworks, Norfolk, MA) to reconstruct the relative permittivity and effective conductivity at each frequency using a rational function model (Anderson *et al* 1994).

Based on preliminary results that suggested a sigmoidal dependence on temperature, best-fit sigmoidal functions of relative permittivity and effective conductivity at 2.45 GHz were created with respect to temperature. The sigmoidal model parameters were obtained by nonlinear regression using the Levenberg–Marquardt algorithm implemented in the Matlab function ‘nlinfit’. The termination tolerance on the residual sum of squares was set as 10^{-8} for the converging condition. To investigate whether thermal dose, rather than instantaneous temperature, was a better predictor of dielectric property changes, relative permittivity and effective conductivity were fit to an integration of temperature over time to create the so-called temperature–time integration (TTI) model:

$$TTI(t) = \int_0^t T(t') dt', \quad (1)$$

where TTI is the temperature–time integration ($^{\circ}\text{C s}$), t is the time (s) and T is the temperature ($^{\circ}\text{C}$).

2.2. Ex vivo temperature mapping for model validation

Tissue preparation and microwave ablation system setup procedures were identical to those described in section 2.1. Fiber optic temperature sensors were placed into the liver 5, 10, 15 and 20 mm away from the antenna, flush with the zone of maximal heating (figure 2). Continuous power was then applied using 25, 50 or 75 W for 5 min ($n = 6$ samples each power) while simultaneously recording temperatures. Mean temperatures at each location and time point were used for comparison with numerical modeling, as described in section 3.2.

2.3. Computer simulation

Microwave heating inside biological tissues can be described by the following bioheat equation (Pennes 1998):

$$\rho C \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + Q_{\text{ext}} + Q_{\text{met}} - Q_p, \quad (2.1)$$

where T is the tissue temperature (K), t is time (s), ρ is the tissue density (kg m^{-3}), k is the thermal conductivity (W (m K)^{-1}), Q_{ext} is the external heat source (W m^{-3}), Q_{met} is metabolic heat generation (W m^{-3}) and Q_p is heat lost to blood perfusion (W m^{-3}). The energy generated by metabolism and taken away by blood perfusion was neglected for *ex vivo* modeling. For microwave ablation, the external heating term is given as

$$Q_{\text{ext}} = \frac{\sigma}{2} |E|^2, \quad (2.2)$$

where $|E|$ is electric field intensity (V m^{-1}) and σ is the effective conductivity (S m^{-1}).

Relative permittivity and electrical conductivity were dependent on temperature or thermal dose, making the microwave tissue heating process a coupled electro-thermal problem, which was solved using the FEM in COMSOL Multiphysics 3.5a (Stockholm, Sweden). The problem was reduced to a two-dimensional computational domain by assuming axial

symmetry about the center of the monopole immersed in liver (figure 2). Four numerical models were simulated: (1) a static model assuming no temperature variation, (2) a linear temperature variation based on published data (Lazebnik *et al* 2006), (3) the sigmoidal model developed in section 2.1 and (4) the TTI-dependent model developed in section 2.1 (see figures 3–5). After analysis of these results, the high- and low-sigmoidal fits to experimental data were also simulated to evaluate the sensitivity of parameter choice when using the sigmoidal model (figure 3). Thermal and physical properties were assumed constant over temperature to isolate and study the effects of the dielectric models independently (table 1). The parameters used in each dielectric model are listed in table 2.

To match experimental conditions, the antenna input power was set as 25, 50 or 75 W, and the ablation time was set as 5 min in all simulations. Maximum mesh element sizes were 0.4 mm in the liver domain and 0.03 mm in the antenna domains. A total of 64 056 triangular elements were generated. Electromagnetic scattering and thermal insulation were assumed on the peripheral liver boundaries. Zero electric and magnetic field and room temperature (293 K) were set as the initial conditions. The solver used in COMSOL was UMFPACK. Simulation outputs such as temperature, electric field and heating rate maps, as well as antenna reflection coefficient, were generated at 5 s intervals.

Finally, we corrected the simulated temperature curves to account for tissue contraction known to occur during high-temperature thermal ablation, especially microwave ablation (Brace *et al* 2010). Such contraction tends to pull the fiberoptic temperature probes toward the antenna, so that probes originally at 5, 10, 15 or 20 mm from the antenna end up at approximately 4, 7, 11 and 15 mm. The location of temperature probes in simulation models was adjusted using an exponential rate to achieve 20–25% contraction after 5 min, according to recent measurements of contraction rates and magnitudes (Brace *et al* 2011). Raw and corrected data were analyzed and will be presented separately, since the corrected data are based on assumed contraction.

3. Results

3.1. Dielectric property model development

Figure 3 shows the measured relative permittivity and conductivity as a function of temperature at 2.45 GHz. Overall, tissue dielectric properties changed slowly below 70 °C as observed in previous studies (Lazebnik *et al* 2006, Stauffer *et al* 2003). However, a rapid decrease in both relative permittivity and conductivity was observed from 70 to 100 °C. At above 100 °C, tissue relative permittivity and conductivity dropped to, and remained at, values about 10% of those at room temperature.

When a sigmoidal function was adopted as the regression model for both relative permittivity and conductivity, the resulting best-fit equations were

$$\varepsilon_r(T) = a_3 \left\{ 1 - \frac{1}{1 + \exp[a_1(a_2 - T)]} \right\} + 1 \quad (3.1)$$

$$\sigma(T) = a_3 \left\{ 1 - \frac{1}{1 + \exp[a_1(a_2 - T)]} \right\}, \quad (3.2)$$

where temperature T is the lone independent variable (°C) and a_i ($i = 1, 2, 3$) are the regression coefficients (table 3).

Figure 4 shows relative permittivity and conductivity versus TTI. An exponential function was adopted to fit the data in one-way nonlinear regression analysis based on the observed exponential decay trend shown in experimental data. The resulting best-fit equations for relative permittivity and conductivity were

$$\varepsilon_r(\text{TTI}) = 1 + a_1 \exp(-a_2 \cdot \text{TTI}) \quad (4.1)$$

$$\sigma(\text{TTI}) = a_1 \exp(-a_2 \cdot \text{TTI}), \quad (4.2)$$

where TTI (s °C) is the lone independent variable and a_i ($i = 1, 2$) are the regression coefficients (table 4).

Figure 5 shows the comparison of temperature-dependent sigmoidal, linear and constant dielectric models investigated in simulations. The linear model is based on linear regression of the results from Lazebnik *et al* (2006), and takes the form $aT+b$, where $a = -0.076$, $b = 50.6169$ for relative permittivity and $a = -0.0006$, $b = 1.715$ for conductivity. The TTI-dependent model is shown as the solid lines in figure 4.

3.2. Ex vivo temperature measurement and mapping

Figure 6 compares the measured and simulated temperature changes for models listed in table 2 with time during thermal ablation at the input power of 50 W. In measurements, temperatures at the sensor 5 mm from antenna (point A in figure 2) rapidly exceeded 90 °C by 60 s, with an approximately linear slope of 1.5 °C s⁻¹. Over the next 30–60 s, temperature elevations slowed to a rate of only 0.02 °C s⁻¹ as energy was absorbed to overcome the latent heat of vaporization (Yang *et al* 2007a). Temperatures then rose steadily above 100 °C at a much slower rate (~0.1 °C s⁻¹) than the initial temperature rise. Temperatures 10 mm from the antenna rose according to a similar trend but at lower slope (0.6 °C s⁻¹ before 100 s, ~0.1 °C s⁻¹ afterward). At positions more distant from the antenna—20 mm (point D figure 2)—the temperature increased at an approximately linear rate of ~0.13 °C s⁻¹.

Table 5 lists the mean percent difference of temperatures predicted by each of the four tissue models and experimental measurements during microwave heating of liver tissue, as well as for the upper and lower envelop sigmoidal models.

The sigmoidal temperature-dependent model agreed more closely with experimental data than all of the other models at 5 mm from the antenna. The TTI model tended to overestimate temperatures for over 100 °C and predicted a depression in temperatures after about 3 min of heating, which contradicts the continuously increasing trend observed in experiments. Mean percent differences between simulated and experimental temperatures over all time points were 4.2% for the sigmoidal model, 10.1% for the TTI model, 27.0% for the linear model and 32.8% for the constant model. The sigmoidal model markedly improved agreement with experimental data for temperatures above 100 °C, where the mean percent difference was 5.8%, compared to 13.0%, 39.5% and 45.4% for the TTI, linear and constant models, respectively (figure 6).

While the constant and linear models provided slightly closer agreement with experimental data at 10 and 20 mm from the antenna, this discrepancy can be compensated by taking tissue contraction into account, as shown later. Simulation results implied that the sigmoidal form was more crucial to simulation accuracy than the model parameters as shown in figure

7. The predicted temperatures agree well with experiments for the best-fit, extreme upper envelop and lower envelop sigmoidal models in figure 3. For relative permittivity, the mean percentage differences between the best-fit curve and the envelops in figure 3 are 53% and 37% for upper and lower envelops, respectively. And for conductivity the differences are 49% and 46% for upper and lower envelops. Comparing with the results in table 5 indicates that changing the sigmoidal inputs by ~50% for the upper envelop case and by ~40% for the lower envelop case resulted in ~3% changes of predicted temperatures.

Figure 8 shows the dielectric heating and electric field intensity for the sigmoidal model in the maximum heating plane at time 0, 50, and 150 s at the input power of 50 W. Compared to the change in the electric field over time, dielectric heating decreased significantly, especially near to antenna, due to the drop in effective conductivity at high temperatures. This trend was not observed in constant and linear models, which predicted a relatively constant electric field intensity spatial profile and, hence, relatively constant heat generation. Ignoring temperature-dependent effective conductivity predicted much higher temperatures than those observed experimentally.

The contour plot and reflection coefficient (S_{11}) in figure 9 also demonstrate that as a result of dielectric property changes, the heating pattern and reflecting power vary during microwave heating. The heating pattern tended to elongate along the antenna axis, which may account for the elongated ablation zones observed with many microwave ablation systems. The change in heating pattern may also reduce heating in the radial direction, further contributing to differences in temperature elevation observed between the sigmoidal and temperature-independent models.

The corrected comparison data due to tissue contraction are presented in figure 10 for the input powers of 25, 50 and 75 W. At 25 W, good agreement was observed between both the constant and sigmoidal model and experimental data. However, the constant model became increasingly inaccurate with input power, while the sigmoidal model demonstrated very good agreement at both 50 and 75 W once corrected for tissue contraction (table 6).

4. Discussion

This study evaluated how to best model dielectric property changes during microwave ablation. Experimental measurements revealed that the relative permittivity and effective conductivity at 2.45 GHz decreased substantially near 100 °C, which was best-represented by a sigmoidal model of their temperature. The sigmoidal model more accurately predicted high temperatures noted near the antenna at higher input powers. Like the other numerical models, the sigmoidal model underestimated temperatures 10–20 mm from the antenna. However, accounting for tissue contraction improved agreement between simulated and experimental data considerably. Finally, the sigmoidal model predicted spatial changes in the electric field and dielectric heat generation around the antenna, as well as the antenna reflection coefficient. Such changes coincided with spatial changes in dielectric properties at temperatures exceeding 100 °C.

Since this study was intended to identify whether an empirical, temperature-dependent model could improve simulation of high-temperature microwave ablations, the physical processes explaining the rapid change in dielectric properties were not studied in detail. Previous efforts have identified that irreversible property changes at high temperatures can be caused by protein denaturation, contraction of collagen and dehydration of the tissue (Yang *et al* 2007a, Foster and Schepps 1981, Sapareto and Deway 1984, Bircan and Barringer 2002, Brace *et al* 2009, Brace 2009, Wall *et al* 1999). In addition, near 100 °C, a large amount of gas is generated due to water vaporization. A large portion of this gas

becomes trapped inside of *ex vivo* tissue, especially near the ablation antenna. This gas is likely to decrease the local tissue density and, as a result, may reduce effective permittivity and conductivity.

An interesting finding noted experimentally which was only corroborated by the sigmoidal model was the rapid decrease in microwave heating when temperatures exceeded 100 °C. This decrease corresponded to a drop in effective conductivity and served to slow temperature elevation near the antenna. As a result, the spatial profile of microwave heating and reflection coefficient of the antenna changed markedly during the course of the procedure. Such changes in the spatial heating profile impact the growth rate and shape of microwave ablations and should be considered when developing new ablation devices or heating strategies.

In contrast to the rapid temperature elevations indicative of direct microwave heating near the antenna, temperatures increased more slowly and linearly farther from the antenna. These trends are more characteristic of thermal conduction from the central heating zone, with minimal direct microwave heating. We noted that simulated temperatures were slightly lower than the experimental temperatures at these locations. One explanation for this discrepancy may be that heat transport from expanding water vapor present in experimental measurements was not captured in our simulations.

Other transitional functions, such as a step or trapezoidal function, could represent the tissue dielectric property changes observed in experiments. However, experimental data revealed a more smooth transition, which is more effectively captured by a sigmoidal model (figure 3). In addition, the sigmoidal model contains a smooth first and second derivative, making multiphysics simulations more likely to converge. Non-smooth functions with sharp transition points require more effort to resolve computationally, which can negatively affect numerical results.

Traditional thermal dose models such as the one proposed by Sapareto and Dewey are not applicable for temperatures in excess of ~60 °C, and certainly not above 100 °C (Sapareto and Dewey 1984). However, it is plausible that a thermal dose heating history may be more generally applicable to model tissue property changes. Simulations based on the simple TTI model overestimated temperature at 5 mm from antenna and also gave an unphysical prediction of temperature drop during ablation. More investigations may be necessary to develop an applicable thermal dose model, but since the simpler sigmoidal model provided very good agreement with experimental data, such development was not included in this study.

The use of static thermal conductivity, specific heat capacity and tissue density coefficients may have also contributed to temperature discrepancies between simulations and experiments. For example, latent heat during water vaporization consumes part of the applied microwave energy (Yang *et al* 2007b). In addition, changes in thermal conductivity may increase thermal diffusion from the higher temperature inner regions to more peripheral regions (Bhattacharya and Mahajan 2003, Haemmerich *et al* 2005). In this study, we chose to keep thermal properties constant so that the dielectric models could be examined under controlled conditions. The greater discrepancy between simulated and experimental data at higher powers (50 and 75 W) may be due to significant thermal property changes at higher temperatures. A more complete tissue model would benefit from temperature-dependent thermal properties, and is the subject of ongoing investigations.

Our study was also limited by the sole use of *ex vivo* tissue and a single energy input. Considering tissue property differences and the effects of blood perfusion, studies carried out *in vivo* would complement our *ex vivo* study and provide more valuable information to

clinical applications. While only 50 W was used to develop the tissue dielectric properties, we noted considerable variation in our dielectric and temperature data. This range of data captures many of the variations in heating rate or thermal dose that would be encountered with different power inputs or tissue types, but more study is necessary to elucidate the effects of heating rate on model development.

5. Conclusion

The dielectric properties of liver change rapidly near 100 °C and can be more effectively modeled using a sigmoidal temperature variation function. As a macroscopic reflection of tissue structure and water content changes, the essential trend of the liver dielectric property variation with temperature was captured. As a result, decreased microwave absorption in the high-temperature center of the ablation leads to increased electric field penetration in the radial direction. Lingering differences between simulation and experimental results are the subject of ongoing investigations.

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Figure 1. Setup for the simultaneous measurement of bovine liver relative permittivity, conductivity and temperature during microwave thermal ablation.

Figure 2.

Temperature sensor positions (A, B, C and D) inside the liver sample. A symmetrical geometry in 2D was generated in simulation, where z is the symmetrical axis. The plane containing points A, B, C and D is defined as the maximum heating plane.

Figure 3.

Experimental results (dots) of relative permittivity and conductivity versus temperature during microwave ablation. Also shown are the best-fit sigmoidal curves (solid lines), along with the upper and lower envelopes (dashed lines) used for numerical simulation.

Figure 4. Relative permittivity and conductivity versus TTI from experimental measurements (dots), and the best-fit exponential curves used for numerical simulation (solid lines).

Figure 5. Relative permittivity and conductivity as sigmoidal, linear and constant functions of temperature. The linear model was obtained based on a fitting of the result from Lazebnik *et al* (2006), which is also shown. Constant relative permittivity and conductivity were adopted from Duck (1990) as shown in tables 1 and 2.

Figure 6. Temperature measurement and simulation results of best-fit sigmoidal, linear, constant and TTI relative permittivity and conductivity models at the positions 5, 10 and 20 mm away from the ablation antenna (positions A, B, C and D in figure 2).

Figure 7. Temperature measurement and simulation results for the best-fit sigmoidal model, and upper and lower envelopes as shown in figure 3.

Figure 8. Simulation results for the sigmoidal model at time $t = 0, 50, 150$ s for (a) resistive heating, and (b) normalized electric field as a function of radial distance in the maximum heating plane (see figure 2) from the ablation antenna.

Figure 9.

Simulation results for sigmoidal model: (a) resistive heating contour plot of $0.8 \times 10^6 \text{ W m}^{-3}$ (horizontal line in figure 8(a)) at time $t = 0, 50, 150 \text{ s}$, (b) normalized electric field contour plot of 1000 V m^{-1} (horizontal line in figure 8(b)) at time $t = 0, 50, 150 \text{ s}$ and (c) reflection coefficient (S_{11}) at the input of the antenna.

Figure 10.

Comparison of measurement and simulation results after taking the tissue contraction effect into account at input power of (a) 25 W, (b) 50 W, and (c) 75 W.

Table 1

Parameters in simulation.

	Property	Unit	Value
Liver ^a	Relative permittivity		Model dependent (table 2)
	Conductivity	(S m ⁻¹)	
	Density	(kg m ⁻³)	1050
	Specific heat	(J (kg K) ⁻¹)	3400
	Thermal conductivity	(W (m K) ⁻¹)	0.564
Copper ^b	Relative permittivity		1
	Conductivity	(S m ⁻¹)	5.87×10^7
Insulator ^b	Relative permittivity		2.05
	Conductivity	(S m ⁻¹)	0

^aValues adopted from Duck (1990).^bAntenna, insulator parameters adopted from manufactures.

Table 2

Parameters for different models.

Property	Unit	Value					TTL-dependent
		Temperature-dependent					
		Sigmoidal					
		Constant ^d	Linear	Best fit	Upper envelop	Lower envelop	
Relative permittivity		45.6	See figure 5		See figure 3		See figure 4
Conductivity	(S m ⁻¹)	1.97					

^aValues adopted from (Duck 1990).

Table 3

Regression coefficient and R^2 for figure 3.

	a_1	a_2	a_3	R^2
ε_r	0.0764	82.271	48.391	0.8570
σ	0.0697	85.375	2.173	0.7881

Table 4

Regression coefficient and R^2 for figure 4.

	a_1	a_2	R^2
ε_r	49.1026	1.5542e-4	0.7852
σ	2.2416	1.3707e-4	0.7841

Table 5

Mean percent difference in simulation models compared to experimental results at 50 W.

Position	Temp. range	Sigmoidal				Linear	TTI
		Best fit	Upper Env.	Lower Env.	Constant		
5 mm	Overall	4.2%	1.2%	11.1%	32.8%	27.0%	10.1%
	<100 °C	0.6%	3.3%	5.9%	3.4%	2.2%	3.4%
	>100 °C	5.8%	3.2%	13.4%	45.4%	39.5%	13.0%
10 mm	Overall	32.0%	29.4%	33.8%	16.8%	19.5%	24.6%
20 mm	Overall	33.9%	33.8%	33.9%	30.8%	30.4%	30.8%

Mean percent difference in simulation models compared to experimental results for input power of 25, 50 and 75 W after taking the tissue contraction effect into account.

Table 6

Input power	Simulation model	25 W		50 W		75 W	
		Sigmoidal	Constant	Sigmoidal	Constant	Sigmoidal	Constant
Probe position	A	4.4%	8.1%	10.1%	53.4%	10.3%	98.4%
	B	4.4%	2.8%	4.8%	24.6%	1.1%	64.2%
	C	4.7%	5.1%	3.4%	22.1%	2.9%	40.1%
	D	9.6%	9.4%	15.7%	6.8%	15.9%	2.9%