Published in final edited form as:

Phys Med Biol. 2017 March 21; 62(6): 2070–2086. doi:10.1088/1361-6560/aa5de4.

Numerical Simulation of Microwave Ablation Incorporating Tissue Contraction Based on Thermal Dose

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Abstract

Tissue contraction plays an important role during high temperature tumor ablation, particularly during device characterization, treatment planning and imaging follow up. We measured such contraction in 18 ex vivo bovine liver samples during microwave ablation by tracking fiducial motion on CT imaging. Contraction was then described using a thermal dose dependent model and a negative thermal expansion coefficient based on the empirical data. FEM simulations with integrated electromagnetic wave propagation, heat transfer, and structural mechanics were evaluated using temperature-dependent dielectric properties and the negative thermal expansion models. Simulated temperature and displacement curves were then compared with the ex vivo experimental results on different continuous output powers. The optimized thermal dose model indicated over 50% volumetric contraction occurred at the temperature over 102.1°C. The numerical simulation results on temperature and contraction-induced displacement showed a good agreement with experimental results. At microwave powers of 55 W, the mean errors on temperature between simulation and experimental results were 8.25%, 2.19% and 5.67% at 5 mm, 10 mm and 20 mm radially from the antenna, respectively. The simulated displacements had mean errors of 16.60%, 14.08% and 23.45% at the same radial locations. Compared to the experimental results, the simulations at the other microwave powers had larger errors with 10–40% mean errors at 40 W, and 10–30% mean errors at 25 W. The proposed model is able to predict temperature elevation and simulate tissue deformation during microwave ablation, and therefore may be incorporated into treatment planning and clinical translation from numerical simulations.

Keywords

microwave ablation; numerical simulation; tissue contraction; tissue shrinkage; thermal dose

1. Introduction

Thermal ablation is an image-guided therapy for treating benign and malignant tumors in the liver, lung, kidney and bone (Goldberg *et al.*, 2000; Ahmed *et al.*, 2011). While RF ablation has been widely adopted (Rossi *et al.*, 1996; Rhim *et al.*, 2004), more recent attention has been directed toward the use of microwave energy due the more rapid heating, larger ablation volumes and less susceptibility to heat sinks microwaves can provide (Simon *et al.*, 2005; Brace *et al.*, 2011; Ahmed *et al.*, 2011). Microwave energy can heat the tissue to over 100°C, causing immense structural, chemical and physical changes to the tissue including protein denaturation, tissue shrinkage and water vaporization (Yang *et al.*, 2007a; Brace *et*

et al., 2010). As a result, tissue material properties undergo substantial changes during microwave ablation. For example, the dielectric permittivity and effective conductivity of tissues are highly temperature dependent, especially in response to water vaporization and desiccation (Chin and Sherar, 2004; Lazebnik *et al.*, 2006; Ji and Brace, 2011; Lopresto *et al.*, 2012; Cavagnaro *et al.*, 2015). Ablated tissue also undergoes a process of shrinking and mechanical stiffening that has recently been described in greater detail (Brace *et al.*, 2010; Bharat *et al.*, 2005; Masuzaki *et al.*, 2007).

The process of tissue contraction during thermal ablation has become a key issue during treatment planning and follow-up. Studies have shown that tissue volumes shrink by 40%–70% during liver ablation, 50%–60% during lung ablation and 26%–42% during kidney ablation (Brace *et al.*, 2010; Liu and Brace, 2014; Farina *et al.*, 2014; Sommer *et al.*, 2013). When looking at post-ablation imaging or explanted tissues, such contraction causes underestimation of the actual treatment effect. Tissue shrinkage has also been observed in clinical series when treating hepatic hemangiomas and renal cell carcinomas respectively, which noted 62% and 52% volume shrinkage, respectively (Ziemlewicz *et al.*, 2014; Moreland *et al.*, 2014).

Multiple groups have performed quantitative measurements of tissue contraction spatially or temporally on different organs (Liu and Brace, 2014; Sommer *et al.*, 2013), and some studies have proposed mathematical models to describe temperature dependent tissue shrinkage (Chen *et al.*, 1998; Rossmann *et al.*, 2014; Park *et al.*, 2016). However, previous studies have not incorporated the numerical models into a simulation environment that also predicts the energy-tissue interaction leading to ablation growth. Therefore, the objective of our study was to set up a temperature-dependent model of tissue contraction that could be incorporated into a simulation model to more accurately predict the tissue response to microwave heating.

2. Methods

2.1 Experimental setup

A total of 18 fresh bovine liver samples were cut into 10 cm × 8 cm × 4 cm blocks and refrigerated for up to 24 hours. Each sample was warmed to approximately 15 °C before experimentation. A microwave antenna (LK 15, NeuWave Medical Inc, Madison, WI) was inserted into the center of each sample. In fourteen samples, a total of 46 aluminum markers of 1 mm diameter were inserted around the antenna through a template of hollow needles to form a 60 mm × 45 mm grid coplanar with the antenna for quantitative measurement of tissue contraction as described in a prior study (Liu and Brace, 2014). Another four samples were prepared for both contraction and temperature measurements. A total of 23 aluminum fiducial markers were inserted into the left half of the samples, while three fiberoptic temperature sensors (Neoptics, Quebec, Canada) were inserted into the right half the sample with the tips located 8 mm, 16 mm and 24 mm radially from the antenna (Figure 1).

The prepared sample was placed onto a 64-slice CT scanning bed with the antenna, temperature probes and fiducial grid parallel to the bed (750HD, GE healthcare, Waukesha, WI). Continuous output powers of 50–100 W were applied for 10 minutes (Certus 140; NeuWave Medical). After accounting for cable losses, the estimated powers delivered to the

tissue were approximately 25 W (4 ablations), 40 W (4 ablations) and 55 W (10 ablations). A total of four temperature measurements were recorded every second with the delivered power of 55 W.

2.2 Tissue contraction and temperature measurement

Helical CT data were acquired over the whole liver sample every 15 s during the ablation (512 px × 512 px, 15 cm field of view, 120 kVp, 200 mA, 0.625 mm slices, 0.531:1 pitch). Two dimensional maximum intensity projection (MIP) CT images including fiducials, antenna and temperature sensors were generated in the coronal plane with 0.1 mm isotropic resolution, and fiducial motion was tracked by a custom thresholding and block matching algorithm in MATLAB (2014a, Mathworks, Natick, MA). The displacement of fiducials was interpolated into two orthogonal vector maps: radial (transverse to the antenna) and longitudinal (along the antenna). Similar to a prior study, three distinct metrics including total displacement (gross displacement over time), normalized displacement (gross displacement normalized by original position) and localized displacement (contraction around each marker) were created to quantify spatial tissue contraction (Liu and Brace, 2014).

Temperature sensor tips were also tracked in real time during ablation by simple thresholding. Temporal temperature maps were created by interpolating the temperature measurements from measured tip locations at the time of measurement.

2.3 Modeling of temperature-time dependent tissue contraction

Preliminary results suggested tissue contraction was not uniform spatially or temporally; it was greatest centrally and at the beginning of the ablation (Liu and Brace, 2014). While contraction was greater at higher temperatures (Rossmann *et al.*, 2014), time was also a significant factor. Therefore, a weighted temperature-time (thermal dose) dependent model was developed for simulation purposes.

Two assumptions were made to develop the model. Firstly, we assumed that contraction reached a maximum value at a radial location very near to the antenna (r_0), in the central charred zone. Secondly, only radial contraction data were correlated with temperature and time, since the temperature sensors were only allowed to move radially with our experimental setup. We assumed similar contraction in the longitudinal direction.

The radial localized contraction was set up to describe the percentage of contraction around each fiducial (Liu and Brace, 2014):

$$LC_r(r, z, t) = \overrightarrow{ND}_r(r_0, z, t) \cdot \frac{LD_r(r, z, t)}{LD_r(r_0, z, t)} \quad (2.1)$$

Where $\overrightarrow{ND}$ and LD_r represents radial normalized displacement and radial localized displacement respectively:

$$\vec{ND}_r(r, z, t) = \frac{\vec{TD}(r, z, t)}{r} \quad (2.2)$$

$$LD_r(r, z, t) = \frac{\frac{1}{r} \left(r \cdot \frac{\partial \vec{TD}(r, z, t)}{\partial r} \right)}{\left| \frac{1}{r} \left(r \cdot \frac{\partial \vec{TD}(r, z, t)}{\partial r} \right) \right|_{max}} \quad (2.3)$$

And $\vec{TD}$ is the total displacement from beginning to the each time points.

A weighted temperature-time integration term was defined as follows:

$$TTI(T, t) = \begin{cases} 0 & (T \leq T_0) \\ w1 \int_0^t (T - T_0) dt & (T_0 < T \leq T_1) \\ w2 \int_0^t (T - T_0) dt & (T_1 < T \leq T_2) \\ w3 \int_0^t (T - T_0) dt & (T > T_2) \end{cases} \quad (2.4)$$

Where TTI is the temperature-time integration ($^{\circ}\text{C s}$), t is the time in seconds, T is the temperature in degrees Celsius, $w1$, $w2$ and $w3$ are weighting coefficients for each temperature range, T_0 is the pre-ablation temperature, and T_1 and T_2 define the temperature ranges ($T_0 < T_1 < T_2$).

The mean radial localized contraction data at 5 mm, 10 mm and 20 mm were then fitted to the weighted temperature-time integration data (TTI) at each location using regression of a power function in MATLAB (2014a, Mathworks, Natick, MA). The optimal fit parameters were determined from:

$$\text{minimize} \|LC_r - a \cdot TTI^b\|_2 \quad \text{s.t. } w1 + w2 + w3 = 1, \quad (2.5)$$

where a , b , $w1$, $w2$, $w3$, $T1$ and $T2$ were the parameters calculated by least square optimization.

2.4 Computer Modeling

Since liver tissue undergoes a series of interrelated mechanical, electrical and biochemical changes during thermal ablation, some assumptions were made to simplify the numerical model:

1. Liver tissue was considered as an isotropic elastic material.
2. Only heat-induced volumetric contraction was considered in the numerical modeling; biochemical processes were not included.

3. Mass transfer of water was ignored.

The heat induced volume deformation can be described by a linear structural mechanical model with thermal strain:

$$E_{el} = E_{tot} - E_{th} \quad (2.6)$$

Where the total elastic strain tensor (E_{tot}) is the summation of the linear elastic strain (E_{el}) and thermal strain (E_{th}). Thermal strain was defined as:

$$E_{th} = \alpha (T - T_{ref}) \quad (2.7)$$

Where α is the thermal expansion coefficient (K^{-1}), and T_{ref} is the reference temperature of the tissue (generally the initial temperature). The thermal expansion coefficient characterizes the ability of the material to contract and expand due to the temperature variations; therefore, volume contraction was represented as a negative thermal expansion coefficient:

$$\alpha = - \frac{a \cdot TTI^b}{T - T_0} \quad (2.8)$$

Where a and b are optimized parameters from equation 2.5, and TTI is the temperature-time integration function in equation 2.4.

Tissue-energy interactions during microwave ablation can be described using the well-known bioheat equation (Pennes, 1948) including a thermal-mechanical coupling term W_{tm} :

$$\rho C_p \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + Q_{ext} + Q_{met} + W_{tm} - Q_p \quad (2.9)$$

Where T is the tissue temperature (K), t is the time (s), and ρ , C_p and k are tissue density ($kg\ m^{-3}$), specific heat ($J/(kg\ K)$) and thermal conductivity ($W/(m\ K)$), respectively. Q_{ext} is the external heat generation (W/m^3), Q_{met} is the metabolic heat generation (W/m^3), and Q_p is the heat loss due to blood perfusion. The thermoelastic damping term W_{tm} was defined as (Nayfeh and Younis, 2004; Coleman and Noll, 1963):

$$W_{tm} = \det(\mathbf{F})^{-1} T \frac{\partial \mathbf{E}_{tot}}{\partial T} : \frac{d\mathbf{S}}{dt} \quad (2.10)$$

Where $\mathbf{F} = \mathbf{x}/\mathbf{X}$ ($\mathbf{x}$ and $\mathbf{X}$ are current spatial coordinate and original coordinate respectively) is the deformation gradient, $\mathbf{E}_{tot}$ and $\mathbf{S}$ are total strain tensor and second Piolo-Kirchhoff stress tensor respectively (Ogden, 1997).

Heat generation during microwave ablation is given by

$$Q_{ext} = \frac{\sigma}{2} |E|^2 \quad (2.11)$$

Where σ is the effective conductivity (S/m) and E is electric field intensity (V/m).

The thermal-electrical response of tissue during microwave ablation can be described by adopting a temperature-dependent electrical conductivity (σ) and relative permittivity (ϵ_r) in the simulation model. Two models that utilize sigmoidal functions to describe temperature dependence – and the effects of water vaporization at high temperatures – have previously been evaluated and shown to improve correspondence between simulated and empirical microwave tissue heating (Ji and Brace, 2011; Lopresto *et al.*, 2012). The relative permittivity (ϵ_r) and conductivity (σ) were defined by equations 2.12–2.14:

$$\epsilon_r(T) = s_1 \cdot \left[1 - \frac{1}{1 + \exp(s_2 - s_3 \cdot T)} \right] + 1, \quad (2.12)$$

$$\epsilon_r(T) = s_1 \cdot \left[1 - \frac{1}{1 + \exp(s_2 - s_3 \cdot T)} \right], \quad (2.13)$$

$$\sigma(T) = r_1 \cdot \left[1 - \frac{1}{1 + \exp(r_2 - r_3 \cdot T)} \right]. \quad (2.14)$$

where s_1 , s_2 , s_3 , r_1 , r_2 , r_3 are the regression coefficients (Table 1). The models are also illustrated in Figure 2.

Neither model has been accepted as the single best approach (Cavagnaro *et al.*, 2015). To avoid bias caused by the choice of dielectric model, and to evaluate whether our contraction modeling approach could be broadly applied, we performed simulations using both models.

With an assumption of axial symmetry about the center the coaxial antenna, a two-dimensional liver model was set up in COMSOL Multiphysics (COMSOL 4.4, Stockholm, Sweden) covering a radial space of 40 mm and longitudinal space of 100 mm. The thermal-mechanical and electromagnetic-thermal couplings were created as described above. The other material properties were all based on published data (Table 2).

To compare with experimental results, the coaxial antenna input powers were set as 55 W, 40 W and 25 W at 2.45 GHz, and the ablation time was set to 10 minutes. Electromagnetic scattering, surface to ambient radiation (ambient = 20°C) and free deformation were assumed on the peripheral boundary. Zero electrical and magnetic field and pre-experimental temperature (~12 °C) were set up as the initial conditions. A total of 66450 degrees of

freedom were generated for the time-dependent solver. The simulation outputs were shown as a temporal deformed electrical field, temperature field and displacement field.

To validate the simulation model, the simulated temperature curves of 55 W at 5 mm, 10 mm and 20 mm from the antenna were compared with the experimental results at the same power. Simulated radial displacement curves at were compared with experimental results for all powers and available locations. The mean absolute percentage errors were calculated to discern differences between simulation and experimental data.

The whole procedure of experimental measurement, modeling, numerical simulation and model validation can be described as Figure 3.

3. Results

3.1 Tissue contraction and temperature measurements

Figure 4 shows the trajectory of fiducial markers and temperature probes during ablation. Due to ablation induced contraction, the fiducials moved towards antenna, and the temperature probes were also pushed towards the antenna. That figure also showed displacement peaked early and decayed over time. Figure 5 shows the mean localized contraction and temperature curves at 5 mm, 10 mm and 20 mm radially from the antenna. Localized contractions (LC_r) at these locations were 0.35 ± 0.05 , 0.22 ± 0.03 and 0.11 ± 0.02 , respectively. Due to tissue contraction, the temperature probes moved towards the antenna from the original radial locations of 8.7 ± 0.6 mm, 15.5 ± 0.8 mm and 21.7 ± 4.0 mm to 5.7 ± 0.2 mm, 11.2 ± 2.1 mm and 17.9 ± 2.3 mm, respectively. Figure 5 also shows both the raw temperature data assumed for each location and the data corrected for temperature probe displacement. The raw temperature data overestimated the corrected temperature by $7.0 \pm 1.4\%$, $12.6 \pm 4.9\%$ and $12.4 \pm 4.3\%$ at 5mm, 10mm and 20mm radially from the antenna, respectively.

3.2 Temperature and time dependent tissue contraction modeling

The final parameters after least squares fitting of the TTI function were:

$$TTI(T, t) = \begin{cases} 0 & (T \leq 12.0^\circ\text{C}) \\ 0.1573 \int_0^t (T - T_0) dt & (12.0^\circ\text{C} < T \leq 44.1^\circ\text{C}) \\ 0.3011 \int_0^t (T - T_0) dt & (44.1^\circ\text{C} < T \leq 102.1^\circ\text{C}) \\ 0.5416 \int_0^t (T - T_0) dt & (T > 102.1^\circ\text{C}) \end{cases} \quad (3.1)$$

Figure 6 shows linear regression results between localized contraction and the optimized TTI ($R^2=0.9953$). Localized contraction was therefore estimated by a TTI dependent term:

$$LC_r = 0.0030 \cdot TTI^{0.4684} \quad (3.2)$$

The negative thermal expansion coefficient was therefore:

$$\alpha = - \frac{0.0030 \cdot TTI^{0.4684}}{T - 285} \quad (3.3)$$

3.3 Simulation results vs experimental results

In simulations, when temperature was increasing due to microwave energy delivery, the tissue underwent a structural deformation: liver tissue contracted in both radial (transverse with antenna) and longitudinal (parallel with antenna) directions. Volume contraction was clearly visible over time (Figure 7). Figure 8 shows the radial, longitudinal and total displacement contours after 10 minutes of simulated heating. Total displacement increased with distance from the antenna. The total displacement map showed the greatest values occurring proximal to the base of the antenna, with only a small displacement noted distal to the antenna base. This finding was generally concordant with a prior study (Liu and Brace, 2014).

When comparing temperature curves with experimental data at 55W for 10 minutes (Figure 9), Model A demonstrated a very good match with experimental data with mean percent errors of 2.2% and 5.7% at 10 mm and 20 mm, respectively. However, Model A overestimated experimental data at 5 mm during the first 100 seconds. In contrast, the simulated temperature at 5mm, 10mm and 20mm in Model B overestimated experimental results with mean percent errors of 23.42%, 10.38% and 4.88% at 5 mm, 10 mm and 20 mm, respectively (Table 3).

When comparing total displacement curves with experimental data at 55W for 10 minutes, Model B demonstrated a slightly better approximation of experimental data (Table 3). Generally, both Model A and Model B underestimated the experimental results at 10 mm and 20 mm radially from the antenna with 10–20% error (Figure 10). At 40 W, both models worked well at 5 mm and 10 mm with 10%–30% error; however, larger errors (~40%) occurred at 20 mm (Figure 11). At 25 W, neither model provided good agreement with experimental data at 5 mm radially from the antenna (mean errors >100%), but they were able to predict radial displacement at 10 mm and 20 mm within 10%–30% (Figure 12).

4. Discussion

Tissue contraction during high-temperature thermal ablation has a great effect on treatment outcome, but had not been incorporated into numerical simulations of the ablation process to date. Our study proposes a temperature and time dependent tissue contraction model based on CT imaging data of contraction, with a negative thermal expansion coefficient used to produce the contraction effect. By coupling the solid mechanics model into the existing electromagnetic and heat transfer model, the thermal-mechanical response to microwave heating could be explored. The temperature and time dependent contraction model was able to reflect experimental results from room temperature (~15 °C) to extremely high temperatures (over 100 °C) within reasonable errors. The optimized TTI model implied greater tissue contraction occurred at temperatures exceeding 102 °C, which was commensurate with recent studies that found greater contraction in the central, higher-

temperature ablation zone with significant water vaporization (Liu and Brace, 2014; Farina *et al.*, 2014; Rossmann *et al.*, 2014).

The proposed simulation model coupled heat-induced volume contraction as an energy term in the bioheat equation, and improved temperature estimation during microwave ablation. Dielectric property model A proposed by Ji and Brace was previously shown to estimate experimental temperature with only 2%–8% error, which was better than most of the models that only considered a thermal-electrical effect (Cavagnaro *et al.*, 2015). The result also agreed with a recent study that stated a deformed liver model could more accurately estimate temperature elevation during microwave ablation (Keangin *et al.*, 2011). Even though model B overestimated experimental temperatures by 10%–20%, the model was still acceptable and provided an improved estimation of tissue contraction compared to model A. A more comprehensive tissue model may further improve simulation fidelity to experimental temperatures and displacements.

Our study also demonstrated that invasive temperature measurement techniques overestimate temperatures at the assumed spatial locations. As tissue contraction forces the probes to move closer to the applicator over time, the measured temperature is greater than at the assumed position. This effect may have caused errors in earlier studies of thermal ablation and should be considered if accurate temperature characterization is necessary (Yang *et al.*, 2007b; Ji and Brace, 2011; Lopresto *et al.*, 2012; Cavagnaro *et al.*, 2015). Concurrent CT imaging can allow measured temperatures to be corrected to their actual spatial location. The numerical simulation results on contraction can provide a reasonable estimation on probe location during ablation, thus improving model validation.

The simulated temperature also agreed with recent temperature measurements using the same frequency (2.45 GHz) and similar output power settings as other studies. For example, the peak simulated temperature of model A at 55 W delivered power was 114.0 °C at 5 mm, 90.6 °C at 10 mm, and 45.7 °C at 20 mm, which closely agreed with experimental results in another study: $107 \pm 2.93^{\circ}\text{C}$ at 5 mm, $87 \pm 3.21^{\circ}\text{C}$ at 10 mm and $47 \pm 4.87^{\circ}\text{C}$ at 20 mm (Sun *et al.*, 2012). The simulated temperatures followed a decreasing exponential formula along the radial distance, which also matches with another study on temperature measurement of microwave ablation on ex vivo porcine liver (Saccomandi *et al.*, 2015).

The proposed numerical model was able to simulate tissue deformation during thermal ablation. The comparison between simulated displacements and experimental results showed a reasonable agreement with 10–20% error at 55 W, 10%–40% error at 40 W and 10%–30% error at 25 W, depending on time and spatial location. The simulations also showed qualitatively similar displacement maps to those reported in a recent study using similar ablation settings (Liu and Brace, 2014). We did note a slightly increased contraction along the antenna in simulations compared to those experimental results, which may be explained by the presence of friction between the antenna and tissue in experiments that reduces tissue motion. Our model did not incorporate forces, but rather relied on a negative thermal expansion to model contraction; therefore, the model was not able to account for antenna-tissue friction. Our model was also limited in that it assumed liver tissue to be an isotropic elastic material (Chui *et al.*, 2007), with no mass transfer due to water vaporization (Yang *et*

al., 2007a; Zhu *et al.*, 2013; Chiang *et al.*, 2015). These simplifications allowed for faster simulation at the expense of reduced physical accuracy. Moreover, the present model was dependent on the accurate estimation of temperatures. Deficiencies in tissue property models might affect temperature estimation and therefore increase simulation errors compared to experimental data. For example, it was possible that at lower powers (25–40 W), the dielectric properties might not change as predicted by either model A or model B. At last, our model was set up based on ex vivo bovine tissue experiments and might be limited when applied to in vivo tissue. Some changes were suggested for estimating in vivo tissue shrinkage: initial temperature would be ~37 °C instead of room temperature and TTI model weights may be adjusted to account for differences in tissue properties.

Despite limitations in the model, our study demonstrated the ability to model thermal-mechanical interactions and to simulate tissue volume contraction during thermal ablation. The results may facilitate improved treatment planning by allowing for the prediction of contraction effects in more patient-specific models. These results may also help improve the development of novel ablation devices and treatments. For example, more accurate prediction of the entire ablation process would allow for greater utilization of computer-aided design (Chiang *et al.*, 2013). The estimation of volume shrinkage may help vendors reduce the mismatching between treatment planning and treatment outcomes. For example, the estimated ablation zone considering volume contraction produced by the treatment planning software at pretreatment would overlay well with the ablative zone under CT or MRI image at post-treatment and follow up, and therefore it helps physicians improve treatment planning and more accurately evaluate the treatment outcomes.

5. Conclusion

Tissue contraction during microwave ablation was modeled by fitting tissue contraction measurements during CT imaging with a weighted thermal dose function. A negative thermal expansion coefficient was defined based on the contraction model, and the corresponding FEM microwave ablation models were developed to predict tissue volume contraction by coupling thermal mechanical term and the bioheat equation. Numerical simulation results showed a good agreement on temperature and displacement with experimental results.

Acknowledgments

The authors would like to acknowledge Jim White for his assistance with numerical simulations and Lisa Simpson for her help with experimental setups. This work was supported by NIH grants R01 CA142737 and R01 CA149379.

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Figure 1.

Experimental setup for both real time tissue contraction and temperature measurement. Tissue contraction was measured by tracking fiducial displacement under CT imaging. Thermal contours were generated by discrete temperature sensors placed axially parallel to the antenna.

Figure 2. Dielectric properties of two simulation models: Dielectric properties of Model A were defined in Equation 2.12 and 2.14, and dielectric properties of model B were defined in equation 2.13 and 2.14.

Figure 3.
Structure of modeling and numerical simulation

Figure 4. Trajectory of fiducials and temperature probes during ablation. The displacements of finical markers (green arrows) and displacement of temperature probe tip (red arrow) within next 2.5 minutes were recorded for setting up contraction models.

Figure 5.

Localized radial contraction (left) and temperature (right) at microwave ablation 55W vs time interpolated to $r = 5\text{ mm}$, 10 mm and 20 mm . Raw temperatures were artificially elevated since contraction pulled the temperature probes toward the antenna into higher temperature tissues. Corrected temperatures account for measured contraction and more accurately portray the temperatures at fixed distances from the antenna.

Figure 6.

Localized contraction vs TTI (logarithmic scale): linear regression was used to fit the localized contraction data with the TTI function (an optimized weighted temperature- time integration function) at radial distances of 5mm, 10mm and 20mm at microwave power of 55W.

Figure 7. Numerical simulation at 55W. Deformed temperature field at 0, 3, 5 and 10 minutes of the microwave ablation.

Figure 8. Numerical simulation at 55W. Radial displacement contours (left), longitudinal displacement contours (center) and total displacement contours (right) at 10 minutes of the ablation

Figure 9.
Simulated Temperature data vs experimental data at 55W shows a good agreement with model A and an overestimation of model B.

Figure 10. Simulated total displacement (radial) vs experimental total displacement at 55W shows a slight underestimation of the experimental displacements.

Figure 11.

Simulated total displacement (radial) vs experimental total displacement at 40W shows a large underestimation of the experimental displacement at 20mm.

Figure 12.

Simulated total displacement (radial) vs experimental total displacement at 25W shows a large overestimation of the experimental displacement at 5mm.

Table 1

Dielectric Properties of the two simulation models

Model	Electrical conductivity (σ)				Relative permittivity(ϵ_r)		Reference
	$r_1(Sm^{-1})$	r_2	$r_3(T^{-1})$	S_1	s_2	$s_3(T^{-1})$	
A	1.80	5.951	0.0764	44.3	6.286	0.0765	(Ji and Brace, 2011)
B	1.80	6.583	0.0598	44.3	5.223	0.0524	(Lopresto <i>et al.</i> , 2014)

Table 2

Liver properties in simulation

Liver Properties ^a	Unit	Value
Density	$\text{Kg} \cdot \text{m}^{-3}$	1050
Conductivity	$\text{S} \cdot \text{m}^{-1}$	Temperature dependent: Table 1
Relative Permittivity		Temperature dependent: Table 1
Specific heat	$\text{J} \cdot (\text{kg} \cdot \text{K})^{-1}$	3400
Thermal conductivity	$\text{W} \cdot (\text{m} \cdot \text{K})^{-1}$	0.564
Young modulus ^b	Pa	1080
Poisson ratio		0.3
Thermal expansion coefficient	K^{-1}	See equation 2.8

^a data adopted from (Duck, 2013)^b data adopted from (Yeh *et al.*, 2002)

Table 3

Mean absolute percentage errors between simulation and experimental results

Matrices	Power (W)	Time (min)	5mm	Model A 10mm	20mm	5mm	Model B 10mm	20mm
Temperature	55	10	8.25%	2.19%	5.67%	23.42%	10.38%	4.88%
Displacement	55	10	13.95%	26.44%	23.23%	16.60%	14.08%	23.45%
Displacement	40	10	20.18%	34.27%	48.09%	31.75%	24.99%	44.15%
Displacement	25	10	>100%	31.14%	13.26%	>100%	30.92%	12.67%