Published in final edited form as:

J Vasc Interv Radiol. 2010 November ; 21(11): 1725–1732. doi:10.1016/j.jvir.2010.07.022.

High Fidelity Computer Models for Prospective Treatment Planning of RF Ablation with *in vitro* Experimental Correlation

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Abstract

Purpose—To evaluate the accuracy of computer simulation in predicting the thermal damage region produced by a radiofrequency (RF) ablation procedure in an *in vitro* perfused bovine liver model. The thermal dose end point in the liver model is used to quantitatively assess computer prediction for use in prospective treatment planning of RF ablation procedures.

Materials and Methods—Geometric details of the tri-cooled-tip electrode were modeled. The resistive heating of a pulsed voltage delivery was simulated in 4D using finite element methods (FEM) implemented on high performance parallel computing architectures. A range of physically realistic blood perfusion parameters, 3.6–53.6 kg/s/m³ were considered in the computer model. An Arrhenius damage model was used to predict the thermal dose. Dice similarity coefficients (DSC) were the metric used to compare computational predictions to T₁-weighted contrast enhanced images of the damage obtained from a RF procedure performed on an *in vitro* perfused bovine liver model.

Results—For a perfusion parameter greater than 16.3 kg/s/m³, simulations predict the temporal evolution of the damaged volume is perfusion limited and will reach a maximum value. Over a range of physically meaningful perfusion values, 16.3–33.1 kg/s/m³, the predicted thermal dose reaches the maximum damage volume within two minutes of the delivery and is in good agreement, DSC > 0.7, with experimental measurements obtained from the perfused liver model.

Conclusions—As measured by the computed volumetric DSC, computer prediction accuracy of the thermal dose shows good correlation with ablation lesions measured *in vitro* perfused bovine liver models over a range of physically realistic perfusion values.

1. Introduction

The use of radiofrequency (RF) ablation to treat hepatocellular carcinoma has become a common minimally invasive clinical alternative to surgical procedures. RF ablation techniques can create clinically relevant ablation regions, 3 – 4.5cm in diameter, and treat tumor lesions that are not otherwise surgically resectable [1]. However, good computational treatment planning tools that enable the creation of conformal ablative regions encompassing the entire clinical target volume of a large (diameter >4.5cm) irregularly shaped tumor lesion are needed [2]. Indeed, computed tomography (CT) based treatment planning systems are currently under development [3]. Treatment planning systems that combine computationally inexpensive RF ablation techniques [4] with real-time quantitative temperature MRI monitoring [5] are rapidly becoming clinically viable. These computational methodologies are intended to address the complexities of overlapping ablation regions [6], enhance the physician's ability to visualize the target lesion and surrounding critical structures, and increase delivery accuracy to treat larger volumes.

Simulation-based treatment planning that can reliably and accurately predict the extent of the thermal dose delivered during a RF ablative procedure is still in nascent stages. In addition to computationally inexpensive techniques [7,8,4], computational methodologies that capture subtleties of the RF ablation procedure through detailed modeling of the RF probe geometry [9,10] and surrounding anatomy [11] are needed. Computing requirements for this level of sophistication quickly require that the developed algorithms fully exploit the most recent parallel computing architectures to obtain a prediction in an acceptable time frame. In this paper, we present a validation study of our developed finite element algorithms for predicting the resistive heating and thermal dose resulting from a pulsed voltage emanating through a tri-cooled-tip electrode. Geometric details of the cluster tip electrode were captured in the modeling. The computational algorithms were developed on parallel computing architectures [12,13] and validated against data obtained from an *in vitro* blood perfused bovine liver model.

2. Materials and Methods

2.1. Computational Methods

The three dimensional mathematical models for RF ablation are based on the Pennes bioheat transfer model [14] coupled with the quasistatic electrical conduction of RF waves as the primary mechanism of energy transfer [15]. The quasistatic equation for the voltage, is obtained from the Maxwell equations with an irrotational electric field. The RF energy delivered over the time interval of the treatment, $[0, T]$, provides a resistive heating due to the current flow near the electrode of a 480kHz monopolar RF generator. The perfusion coefficient, thermal conductivity, and electrical conductivity, are assumed linear and spatially homogeneous model parameters. The computational domain is taken large enough so that there may be assumed no heat flux across the surface. The boundary is assumed an electrical ground for the voltage. The computational models used provides a 4D prediction of both the temperature, $u(x, t)$, and the voltage, $\phi(x, t)$. The set of governing partial differential equations that embody the computational models and as well as a summary of the model parameters used is provided in the appendix.

The temperature of the blood warmed by a heat exchanger was used as the initial temperature field, $u(x, 0) = u^0 = 37^\circ\text{C}$. Tissue damage is characterized by an Arrhenius model of the spatially varying damage field $D(x)$ resulting from the time-temperature history during the RF ablation procedure.

$$D(\mathbf{x}) = \int_0^T A e^{-\frac{E_a}{Ru}} dt$$

The Arrhenius model is a classical and empirical law used to model kinetics of chemical reaction adopted to describe tissue damage [16,17,18]. Although there are more accurate damage models available [19], Arrhenius model is commonly used due to its simplicity. In the Arrhenius model, the parameters E_a , A , and R are constants determined usually by *in vitro* experiments, which represent activation energy, frequency factor, and the universal gas constant, respectively. The values $A=3.1e98 \frac{1}{s}$, $E_a=6.28e5 \frac{J}{mol}$, and $R=8.314 \frac{J}{kg \cdot mol}$ were used in this study.

2.2. Perfused Bovine Liver Experiments

Experimental studies using an *in vitro* perfused bovine liver model [20] were used to validate computational prediction. The experimental setup of the liver model is illustrated in Figure. 1. A bovine liver was obtained from a slaughterhouse along with 7L of blood from the animal. For anticoagulation purposes, the liver was flushed with 10L of Krebs-Henseleit solution containing 30,000 USP of heparin. The viable tissue was cooled and transported using established protocols for human liver transplant. Once at the laboratory, the blood was oxygenated and heated to 37°C using carbogen (Praxair Inc., Danbury, CT), a membrane oxygenator/heat exchanger (Medtronic NT Affinity), and a heating coil. A centrifugal pump (Medtronic Bio-Medicus Bio-Console Model 540, Minneapolis, MN) was used to perfuse the blood through the liver at a rate, 300mL/min, sufficient to maintain the tissue at 37°C. A Foley catheter (24FR 5cc) was inserted into the portal vein and secured by expanding the balloon to the diameter of the vessel. Tubing connected the catheter to the pump. The return tubing was connected to the vena cava and external vasculature was clamped to ensure a closed system.

Four ablation regions were created using a 2.5cm Covidien Cooltip cluster electrode (Covidien, CTC-1525, Boulder, CO) and the Covidien 200W RF energy generator (Mo. CC-1-117). The electrode was run in impedance mode for 12 minutes per ablation and cooled with chilled water (Covidien Pump Mo. PE-PM). Figure. 2 shows the voltage time history applied by the RF generator for each ablation.

Post RF delivery, MR images of the bovine liver was acquired on a 3T MR scanner (Achieva, Philips Healthcare, Andover, MA) with a six channel cardiac coil (SENSE, Philips Healthcare, Andover, MA) to assess damage regions, Figure. 3. T₁-weighted images were obtained using a 3D RF spoiled gradient recalled echo pulse sequence. The field of view was 340mm with a 340 × 340 acquisition matrix; 120 slices were acquired each 1mm thick. A fast 3D spoiled gradient-echo sequence, with acquisition parameters similar to those used clinically for dynamic evaluation of contrast uptake in the liver ($T_r/T_e/FA=8.1 \text{ ms}/3.7 \text{ ms}/20^\circ$), was employed to assess the region of damage. The outer ring of edema in the ablation zone was segmented using semi automatic techniques that detected the image gradients resulting from the higher signal edema ring.

2.3. Study Specific Model and Data Analysis

The finite element model of the Covidien Cooltip cluster electrode is shown in Figure. 4. The insertion of cluster electrode into the bovine liver tissue was modeled using distinct regions for the electrode and the tissue. The handle of the electrode was considered to have a minimal effect on the thermal damage produced by the resistive heating and only the electrode cluster tip was modeled. The voltage history from the power generator Figure 2 was input into the finite element models with a time resolution of 0.11 seconds over the 12min delivery window for a total of 6545 time steps. At each time step, the voltage across the electrode was held

constant, i.e. the corresponding degrees of freedom within the computational models are fixed and treated as Dirichlet boundary data. Similarly, the temperature across the electrode was held at 21°C throughout the simulation of the RF energy delivery to simulate the cooled tip. A Crank-Nicolson time stepping scheme was used to propagate the governing equations (.4) in time.

Assuming a uniform mass flow of blood through the *in vitro* bovine liver perfusion model, a lower bound for the perfusion, ω^{lb} , may be estimated through dimensional arguments involving the centrifugal pump driven blood flow rate, $\dot{V}_{pump}=300\text{ml/min}$, the density of the blood, ρ_{blood} , and the volume of the liver, V_{liver} .

$$\omega^{lb} = \frac{\dot{V}_{pump}\rho_{blood}}{V_{liver}}$$

The volume of the liver was estimated from the MR images, $V_{liver}=4.0e6\text{mm}^3$. The estimated lower bound provides a reference value for the perfusion, $\omega^{lb} \approx 1.25 \text{ kg/s/m}^3$. However, a uniform mass flow of blood through the perfused liver model is highly idealized and not possible. Inherent uncertainty in this estimation is expected from the effect of the local vasculature which is expected to increase the blood flow in the immediate vicinity of the ablation regions. Consequently, several realistic values of perfusion from [21] were considered for the model, $\omega=3.6, 16.3, 23.3, 42.2$, and 53.6 kg/s/m^3 . Each surrogate value of the perfusion was evaluated independently with the parallel FEM solver to capture a range of perfusion values occurring *in vivo*. A summary of the remainder of the relevant constitutive data used is presented in Table. 1 of the appendix.

Each of the four ablation zones shown in the T_1 -weighted images, Figure. 3, were segmented, registered, and projected onto the finite element mesh. Registrations were performed using the electrode as a landmark. The measured volumes were 33.3, 41.6, 43.4, and 58.3cc. Dice similarity coefficients (DSC) [22] were computed to measure the agreement of the segmented damage region with computational prediction.

$$\text{DSC} = 2 \frac{A \cap B}{A + B} \quad 0 \leq \text{DSC} \leq 1$$

Where A is the volume of the segmented T_1 -weighted image and B is the corresponding FEM prediction. Customized Paraview [23] plugins were developed to compute the volume intersections of the finite element data. Possible values of DSC range from no overlap (DSC=0) to complete overlap (DSC=1). A DSC greater than 0.7 is assumed to be a good agreement [24]. The DSC was computed for every permutation of measured damage region against perfusion considered. At each perfusion value, the mean and standard deviation of each DSC computed from the measured data was used to characterize the statistical variation.

3. Results

The thermal and electrical fields, as well as 2D sections represented as cutplanes through the finite element prediction of the quasistatic electromagnetic bioheat transfer state, are shown in Figure. 4. The results shown in the figure correspond to the perfusion coefficient of $\omega = 23.3 \text{ kg/s/m}^3$. The cutplanes are at the axial center with respect to the cluster electrode, and the temperature profiles show no geometrical edge effects from computer simulation. The temperature and voltage predictions are shown at the left and right, respectively. The color

scales are from 21 – 100°C and 0 – 85V. The distance scale in meters is also provided as a reference. The volumetric visualization for both temperature and voltage fields are also provided in Figure. 4. The color scales are the same as mentioned previously.

Temperature and voltage profiles taken along lines through the prongs of the electrode are shown in Figure. 5. The top profile is through one of the three prongs and the bottom profile is through the remaining two prongs. The cutplanes and volumetric visualization of the system state shown in Figure. 4 is at the same time instance as the profiles shown in Figure. 5. The left axis provides the temperature scale in degrees Celsius and the right axis provide the voltages scale in volts. The profiles illustrate the effect of the cooling mechanisms and voltage application modeled by applying the Dirichlet boundary conditions (fixed temperature and voltage) on the electrode surface and decay of the temperature and voltage away from the electrode.

An illustration of the finite element thermal damage prediction and corresponding overlap with the hyperintense T_1 -weighted segmentation's are shown in Figure. 6. The distance scale is provided in mm. A particular ablation region and the electrode are shown with reference to the bovine liver at the bottom. The top left shows an early time damage model prediction. The damage region is quantified with a threshold at $D = 1$, i.e. for regions with D values greater than 1 are considered to be damaged. The early time heating by the cooled tip electrode is seen to cause no damage immediately surrounding the electrode as well as within the domain between the prongs of the cluster. The final damage prediction, illustrated as an ablation region, and the corresponding intersection is shown at the top middle of Figure. 6. Again, the threshold used was $D = 1$. A transparent view through the segmentation further illustrates the intersection used in the DSC calculations, top right of Figure. 6.

A quantitative and comprehensive summary of the damage results and the correlation with T_1 -weighted imaging data is provided in Figure. 7. Figure. 7(a) shows a 3D plot of the FEM predicted damage volume [cc] history as a function of time and perfusion; iso-damage contours are provided at the base of the plot. Plotted is the cumulative volume with $D > 1$. The damage history is further illustrated in Figure. 7(b); the damage volume as a function of time is plotted for each value of perfusion considered with the volume of each of the four T_1 -weighted segmentations provided as a reference. The FEM model predictions reveal that the time evolution of the damage region is perfusion limited. At the applied voltage, perfusion of $\omega > 16.3 \text{ kg/s/m}^3$ sufficiently carries away the thermal energy and results in a maximum lesion size within two minutes of the ablation procedure. The computed dice similarity coefficients are plotted as a function of the perfusion values considered in Figure. 7(c). The error bars provided are the standard deviations with respect to the four segmented T_1 -weighted ablation regions. A DSC value of 0.7 is provided as a reference. A linear interpolation of the DSC values presented in Figure. 7(c) provides a range of physically acceptable perfusion values, $\omega \in [16.3, 33.1] \text{ kg/s/m}^3$ that provides good damage correlation with the *in vitro* bovine liver perfusion model.

4. Discussion

Isocontours of the voltage and thermal predictions, Figure. 4, are unique to the cluster electrode and are observed to conform to the geometry of the cluster probe as expected. The contrast between the cooled tip electrode and the resistive heating is also observed. Within the circumference of the cluster, the cooling mechanism is able to keep the temperature below the boiling point threshold. The temperature rises sharply near the prongs on the outer region of the electrode. The cooling mechanisms are seen to create approximate triangular temperature contours. Despite the geometry of the electrode, the diffusive model of the quasi-static electric field creates voltage isocontours that quickly become circular with distance from the probe.

The 3D volumetric views of both the temperature and voltage fields, shown in Figure. 4, enhance the visualization of the cutplanes and show the geometrical edge effects of the electrode, which may give a physician insight into the treatment outcome prediction around the electrode.

The observed lesion size variations observed in the T_1 -weighted images due to local perfusion and conductivity indicate that there is a need for patient specific data that can be recovered and computed by inverse analysis to obtain tissue properties under MR guidance. Customized techniques for MR monitoring of the temperature [25,5] would allow recovery of nonlinear parameters for constitutive models that capture change of state effects when temperatures reach $\approx 100^\circ\text{C}$ and has potential of providing a quantitative methodology to characterize irreversible effects though nonlinear heterogeneous models [26]. These expensive bioheat transfer constrained optimization problems may even be performed on a time scale acceptable to the typical duration of an RF procedure [27] and would allow predictive computational models to simulate impedance controlled techniques [28] of delivery. In addition, the methodology has the potential to address the lack of accurate characterization of biological tissues [29] for electromagnetic driven bioheat transfer. Further work is needed to thoroughly evaluate the computational performance, efficiency, and feasibility of the developed parallel computing finite element algorithms for the purposed of routine clinical planning of RF ablation procedures.

In order to develop a treatment planning system for clinical use, more realistic factors and biological complexity need to be added to the current modeling system. The predicted time evolution of the radial versus axial extent of lesion is needed for comparison with clinical data. The electrode can be modeled in a more detailed fashion to include insulating boundary conditions for the electrode handle that may have an impact on the final prediction. Moreover, further validation of the computer model saturation time prediction of the thermal dose delivery has the potential to decrease possible risks and complications by decreasing unnecessary heated blood flow to distant vital organs. The perfusion limited finite element models developed in the work can also be extended to multiple RF ablations to create conformal 3D treatment planning of large lesions, increase the efficacy of the procedure, and reduce the likelihood that large tumor lesions may otherwise be incompletely treated.

In conclusion, a finite element model for RF ablation has been developed with predictive capabilities to correlate the damage zone with MRI measurement in terms of DSC quantification. The comparative study of the treatment outcome demonstrates that the predicted damage zone for a range of physically acceptable perfusion values, is in agreement with the experimental data obtained from the perfused bovine liver model [20]. The methodologies developed in this work provide a powerful tool and foundation for designing and validating experiments for treatment planning of RF ablation therapies.

Acknowledgments

The research presented in this paper was supported in part by NIH/NCI funding 5T32CA119930-03 and K25CA116291. The authors would also like to acknowledge the ITK [30], Paraview [23], PETSc [12], libMesh [13], and CUBIT [31] open-source communities for providing enabling software and support for scientific computing and visualization. The authors would also like to thank the technical support staff maintaining the computational and visualization facilities at UTSA and TACC, www.tacc.utexas.edu, from which the computational aspect of this work was derived.

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References

1. McGahan J, Dodd G. Radiofrequency ablation of the liver current status. *American Journal of Roentgenology* 2001;176(1):3–16. [PubMed: 11133529]
2. Wood B, Locklin J, Viswanathan A, Kruecker J, Haemmerich D, Cebal J, Sofer A, Cheng R, McCreedy E, Cleary K, et al. Technologies for guidance of radiofrequency ablation in the multimodal ity interventional suite of the future. *Journal of vascular and interventional radiology* 2007;18(1):9–24. [PubMed: 17296700]
3. Villard C, Soler L, Gangi A. Radiofrequency ablation of hepatic tumors: simulation, planning, and contribution of virtual reality and haptics. *Computer Methods in Biomechanics and Biomedical Engineering* 2005;8(4):215–227. [PubMed: 16298844]
4. Johnson P, Saidel G. Thermal model for fast simulation during magnetic resonance imaging guidance of radio frequency tumor ablation. *Annals of Biomedical Engineering* 2002;30(9):1152–1161. [PubMed: 12502226]
5. Lepetit-Coiffe M, Quesson B, Seror O, Dumont E, Le Bail B, Moonen C, Trillaud H. Real-time monitoring of radiofrequency ablation of rabbit liver by respiratory-gated quantitative temperature MRI. *Journal of Magnetic Resonance Imaging* 24(1)
6. Dodd G, Frank M, Aribandi M, Chopra S, Chintapalli K. Radiofrequency Thermal Ablation Computer Analysis of the Size of the Thermal Injury Created by Overlapping Ablations. 2001
7. Berjano E, Alio J, Saiz J. Modeling for radio-frequency conductive keratoplasty: implications for the maximum temperature reached in the cornea. *Physiological Measurement* 2005;26(3):157–172. [PubMed: 15798292]
8. Curley, M.; Hamilton, P.; Ltd, E.; Cambridge, M. Creation of large thermal lesions in liver using saline-enhanced RFablation. *Engineering in Medicine and Biology Society, 1997; Proceedings of the 19th Annual International Conference of the IEEE; 1997.*
9. Chen C, Miga M, Galloway R Jr. Optimizing Electrode Placement Using Finite-Element Models in Radiofrequency Ablation Treatment Planning. *IEEE Trans Biomed Eng* 2009;56:237–245. [PubMed: 19272862]
10. Barauskas R, Gulbinas A, Vanagas T, Barauskas G. Finite element modeling of cooled-tip probe radiofrequency ablation processes in liver tissue. *Computers in Biology and Medicine* 2008;38(6): 694–708. [PubMed: 18466889]
11. Liu Z, Ahmed M, Gervais D, Humphries S, Goldberg S. Computer Modeling of Factors that Affect the Minimum Safety Distance R equired for Radiofrequency Ablation Near Adjacent Nontarget Structures. *Journal of Vascular and Interventional Radiology* 2008;19(7):1079–1086. [PubMed: 18589323]
12. Balay, S.; Gropp, WD.; McInnes, LC.; Smith, BF. Tech Rep ANL-95/11 - Revision 215. Argonne National Laboratory; 2003. Petsc users manual.
13. Kirk, B.; Peterson, J. libMesh-a C++. Finite Element Library, CFDLab; URL <http://libmesh.sourceforge.net>
14. Pennes HH. Analysis of tissue and arterial blood temperatures in the resting forearm. *J Appl Physiol* 1948;1:93–122. [PubMed: 18887578]
15. Chang I, Nguyen U. Thermal modeling of lesion growth with radiofrequency ablation devices. *Biomed Eng Online* 2004;3(1):27. [PubMed: 15298708]
16. Henriques F. Studies of thermal injury V. The predictability and the significance of thermally induced rate processes leading to irreversible epidermal injury. *Arch Pathol* 1947;43:489–502.
17. Henriques F, Moritz A. Studies of thermal injury. I. The conduction of heat to and through skin and the temperatures attained therein. A theoretical and an experimental investigation. *Am J Pathol* 1947;23(53):1–549. [PubMed: 19970913]
18. Moritz A, Henriques F. Studies of thermal injury. II. The relative importance of time and surface temperature in the causation of cutaneous burns. *Am J Pathol* 1947;23:695–720. [PubMed: 19970955]
19. Feng Y, Oden JT, Rylander MN. A Statistical Thermodynamics Based Model for Cell Damage and its Validation in Vitro. *J Biomed Eng* 2008;130:1–10.

20. Boykin, R. PhD thesis. The University of Texas; San Antonio: 2008. Characterization of radiofrequency ablation zones for the purpose of treatment planning.
21. Diller, KR.; Valvano, JW.; Pearce, JA. Bioheat transfer. In: Kreith, F.; Goswami, Y., editors. The CRC Handbook of Mechanical Engineering. 2. CRC Press; Boca Raton: 2005. p. 4-278-4-357.
22. Dice L. Measures of the amount of ecologic association between species. *Ecology* 1945;297-302.
23. Henderson, A.; Ahrens, J. The ParaView Guide. Kitware; 2004.
24. Zijdenbos A, Dawant B, Margolin R, Palmer A. Morphometric analysis of white matter lesions in mr images: method and validation. *IEEE Transactions on Medical Imaging* 1994;13(4):716-724. [PubMed: 18218550]
25. Zhang Q, Chung Y, Lewin J, Duerk J. A method for simultaneous RF ablation and MRI. *Journal of Magnetic Resonance Imaging* 8(1)
26. Diller, KR.; Oden, JT.; Bajaj, C.; Browne, JC.; Hazle, J.; Babuska, I.; Bass, J.; Bidaut, L.; Demkowicz, L.; Elliott, A.; Feng, Y.; Fuentes, D.; Goswami, S.; Hawkins, A.; Khoshnevis, S.; Kwon, B.; Prudhomme, S.; Stafford, RJ. Computational Infrastructure for the Real-Time Patient-Specific Treatment of Cancer. Vol. Ch. 9. Taylor & Francis Group; 2008. Advances in Numerical Heat Transfer, Vol. 3: Numerical Implementation of Bioheat Models and Equations.
27. Fuentes D, Feng Y, Elliott A, Shetty A, McNichols RJ, Oden JT, Stafford RJ. Adaptive Real-Time Bioheat Transfer Models for Computer Driven MR-guided Laser Induced Thermal Therapy. *IEEE Trans BME* 57(5)
28. Kelekis A, Terraz S, Roggan A, Terrier F, Majno P, Mentha G, Roth A, Becker C. Percutaneous treatment of liver tumors with an adapted probe for cooled-tip, impedance-controlled radio-frequency ablation under open-magnet MR guidance: initial results. *European radiology* 2003;13(5):1100-1105. [PubMed: 12695834]
29. Berjano E. Theoretical modeling for radiofrequency ablation: state-of-the-art and challenges for the future. *Biomedical engineering online* 2006;5(1):24. [PubMed: 16620380]
30. Ibanez, L.; Schroeder, W.; Ng, L.; Cates, J. The ITK Software Guide. 2. Kitware, Inc; 2005. <http://www.itk.org/ItkSoftwareGuide.pdf>
31. Blacker, T., et al. Cubit Users Manual. 2008. <http://cubit.sandia.gov/documentation>
32. Demkowicz, L. Computing with Hp-adaptive Finite Elements: One and two dimensional elliptic and Maxwell problems. CRC Press; 2006.
33. Pao, R. Fluid dynamics. CE Merrill Books; 1967.
34. Jackson, J. Classical Electrodynamics. 3. New York: Jhon Willey & Sons; 1999.
35. Stauffer P, Rossetto F, Prakash M, Neuman D, Lee T. Phantom and animal tissues for modelling the electrical properties of human liver. *International Journal of Hyperthermia* 2003;19(1):89-101. [PubMed: 12519714]
36. McNichols R, Gowda A, Kangasniemi M, Bankson J, Price R, Hazle J. MR thermometry-based feedback control of laser interstitial thermal therapy at 980 nm. *Lasers in Surgery and Medicine* 34 (1)

Appendix

The equations governing the resistive heating that provides the thermal source for the bioheat equation (.4) may be obtained from Maxwell's equation [32] using Faraday's law (.1), Ampere's law (.2), and the continuity of free charge (.3).

$$\nabla \times \vec{E} = - \frac{\partial \vec{B}}{\partial t} \quad (.1)$$

$$\nabla \times \vec{B} = \mu \vec{J} + \mu \epsilon \frac{\partial \vec{E}}{\partial t} \quad (.2)$$

$$\nabla \cdot \vec{J} + \frac{\partial}{\partial t} \rho_{charge} = 0 \quad (.3)$$

Here, the medium is assumed linear and isotropic with homogeneous permittivity, ϵ , and permeability, μ . The charge density is denoted ρ_{charge} . The electric and magnetic fields are denoted $\vec{E}$ and $\vec{B}$, respectively. The current density, $\vec{J}$, is assumed related to the electric field through Ohm's law, $\vec{J} = \sigma \vec{E}$, where σ denotes the electrical conductivity. In the quasistatic case, the electric field is approximated as irrotational and Maxwell's system decouples

$$\nabla \times \vec{E} = 0 \quad \Rightarrow \quad \nabla \cdot \sigma \vec{E} = 0$$

For an arbitrary control volume, the Poynting vector defines the amount of electromagnetic power, Q_{em} , which crosses the surface

$$Q_{em} = \int_{\text{surface}} (\vec{E} \times \vec{B}) dA$$

From the conservation of energy for an arbitrary control volume, the mechanical energy and electromagnetic energy are coupled [33]

$$\rho c_p \frac{\partial u}{\partial t} + \frac{\partial u_{em}}{\partial t} + -\nabla \cdot k \nabla u + \omega c_{blood}(u - u_a) = -Q_{em}(x, t)$$

Here, the temperature is denoted u , ρ is the density of the continuum, which is liver tissue in our case, and c_{blood} is the specific heat of blood. The perfusion coefficient, ω , and thermal conductivity, k , are assumed linear and spatially homogeneous model coefficients. Combining the conservation of energy with Poynting's theorem [34] yields

$$\rho c_p \frac{\partial u}{\partial t} + \frac{\partial u_{em}}{\partial t} + -\nabla \cdot k \nabla u + \omega c_{blood}(u - u_a) = \vec{E} \cdot \vec{J} + \frac{\partial u_{em}}{\partial t}$$

hence, the energy input into the tissue is dissipated through the resistive heating.

$$Q_{RF}(x, t) = \vec{E} \cdot \vec{J} = \sigma \vec{E} \cdot \vec{E}$$

The relationship of the voltage, ϕ , as the scalar potential for the electric field, $\nabla \phi = -\vec{E}$, leads to the time dependent system of coupled equations governing the temperature and voltage.

$$\left. \begin{aligned} \rho c_p \frac{\partial u}{\partial t} - \nabla \cdot k \nabla u + \omega c_{blood}(u - u_a) &= Q_{RF}(x, t) \\ Q_{RF}(x, t) &= \sigma |\nabla \phi|^2, \quad \nabla \cdot \sigma \nabla \phi = 0 \end{aligned} \right\} \quad (.4)$$

Dimensionally, the units of Q_{RF} are consistent with power density.

$$\frac{S}{m} \frac{V^2}{m^2} = \frac{\frac{A}{V}}{m} \frac{V^2}{m^2} = \frac{AV}{m^3} = \frac{\frac{C}{s}}{m^3} \frac{J}{C} = \frac{W}{m^3}$$

Neumann and Dirichlet boundary conditions are used on the surface of the biological domain of interest, $\partial\Omega$, for the the temperature and voltage, respectively.

$$\left. \begin{array}{l} -k(u)\nabla u \cdot \mathbf{n}=0 \\ \varphi=0 \end{array} \right\} \quad \text{on } \partial\Omega$$

A summary of the constitutive data used is presented in Table. 1.

Figure 1.

Experimental Setup, reproduced with permission from [20]. A viable bovine liver is perfused with blood at 37°C. A centrifugal pump drives the blood flow at 300ml/min. The blood was oxygenated and heated using carbogen, a membrane oxygenator/heat exchanger, and a heating coil. A Foley catheter, inserted into the portal vein, allows the blood to flow from the pump into the liver without leaking. A Covidien Cooltip cluster electrode was used to deliver the RF energy to create the ablations.

Figure 2.

Voltage History. The Voltage history provided by the Covidien 200W RF energy generator is shown. The voltage delivery is pulsed with a peak delivery at a frequency of 20 seconds.

Figure 3.

Post ablation MR Contrast Images. Post ablation T_1 -weighted MR images were acquired on an MR scanner using a gradient recalled echo pulse sequence. Contrast was used to enhance the T_1 signal in the ablation regions. The MR data set shown is $352 \times 352 \times 120$ voxels with a field of view of $340\text{mm} \times 340\text{mm} \times 12.1\text{mm}$. A volume visualization of the bovine liver is shown at the top left. Axial, sagittal, and coronal planes through the MR data are shown; the locations correspond to the planes in the volume visualization. The T_1 enhanced ablation regions are shown.

Figure 4.

State Prediction. The full 3D computer model is represented by a finite element mesh consisting of 99,220 hexahedral elements and 104,580 total nodes (grid points). A 2D section through this FE mesh shows the predicted temperature and voltage resulting from the geometry of the RF electrode. The distinct regions of hexahedral mesh used to model the RF electrode tip is shown at the top. A quadrilateral mesh was extruded axially along each prong of the probe to create the base of the hexahedral finite element mesh. The mesh for the tissue conforms to the boundary of the application and extends 8cm axially and radially beyond the probe to ensure that the boundary does not influence the heating. A 3D volume visualization of the temperature and voltage distribution around the RF probe is shown at the bottom. 209,160 degrees of freedom were required to solve the coupled bioheat-RF system (.4) at each time step. Line profiles of the temperature and voltage as a function of distance along the lines indicated are provided in Figure. 5.

Figure 5.

Temperature and Voltage Profiles. The temperature and voltage along the cutlines shown in Figure. 4 are shown. The top profile is through two of the three prongs of the electrode. The bottom profile is through the remaining prong. The profiles clearly indicate the Dirichlet boundary condition enforcement of constant temperature and voltage along the electrode. Temperature and voltage are shown in degrees Celsius and volts, respectively.

Figure 6.

Damage Zone in the Liver with Magnified View Around the electrode at Different Time Instances. The FEM model of the RF electrode inserted into the liver along with the corresponding damage region is shown at the bottom. Top Left: Early time Arrhenius damage model prediction. The Cooled electrode prevents damage immediately surrounding the probe. Top Middle: The overlap (yellow) of the segmented T₁ ablation data (red) and the computation prediction (cyan) is shown. Top Right: An transparent view of the damage correlation.

Figure 7.

Damage History. The time evolution of damage volume is shown for each value of perfusion considered. The damage region growth is perfusion limited for $w_0 > 16.3 \left[\frac{kg}{m^3s} \right]$ and has saturated within ≈ 2 minutes of the delivery of RF energy. The damage volume of the segmented T_1 data is provided as a reference. The damage volume data is provided in cubic centimeters [cc]. (c) The DSC correlation between computational prediction and the T_1 -weighted contrast data show good agreement.

Table 1

Constitutive Data [21,35,36]

σ	k	ρ	u_a	c_{blood}	c_p	A	E_a	R
$\frac{\text{N}}{\text{m}}$ 0.69	$\frac{\text{W}}{\text{mK}}$ 0.5	$\frac{\text{kg}}{\text{m}^3}$ 1045	308 K	$\frac{\text{J}}{\text{kg}\cdot\text{K}}$ 3840	$\frac{\text{J}}{\text{kg}\cdot\text{K}}$ 3600	$\frac{1}{\text{s}}$ 3.1e98	$\frac{\text{J}}{\text{mol}}$ 6.28e5	$\frac{\text{J}}{\text{kg}\cdot\text{mol}}$ 8.314