MEASUREMENTS, ANTENNA DESIGN AND ADVANCED COMPUTER MODELING FOR MICROWAVE TISSUE ABLATION

by

Deshan Yang

A dissertation submitted in partial fulfillment of the requirements for the degree of

Doctor of Philosophy

(Electrical Engineering)

at the

UNIVERSITY OF WISCONSIN-MADISON

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# Table of Contents

Acknowledgements .............................................................................................................. i  
Table of Contents ................................................................................................................ ii  
Abstract ............................................................................................................................. vii  

Chapter 1  Introduction to microwave tissue ablation ...................................................... 1  
A  Liver cancer and treatments ................................................................................ 2  
B  Microwave tissue ablation fundamentals .............................................................. 3  
C  Current research status for MWA .......................................................................... 7  
C.1  Current clinical trials ...................................................................................... 7  
C.2  Problems and challenges of current MWA technologies .................................... 8  
D  References ............................................................................................................. 12  

Chapter 2  Theories and fundamental physics .................................................................. 14  
A  Electromagnetism and microwave-tissue interaction ............................................ 15  
A.1  Maxwell’s equations ....................................................................................... 15  
A.2  EM waves in material medium ....................................................................... 18  
A.3  Power flow of EM waves ................................................................................ 21  
A.4  Interactions of microwave and biological materials ....................................... 22  
B  Dielectric properties of biological tissues ............................................................. 23  
B.1  Frequency dependence ................................................................................... 23  
B.2  Temperature dependence ............................................................................... 28  
B.3  Tissue water content dependence .................................................................. 29  
B.4  Dielectric properties of liver tumor tissues ..................................................... 35  
C  Thermal responses of biological tissues during MWA ............................................ 36  
C.1  Tissue responses versus temperature ............................................................... 36  
C.2  Tissue damage versus thermal dose ................................................................ 37  
C.3  Pathological analysis ...................................................................................... 38  
D  Heat transfer and the bioheat equation .................................................................. 39  
D.1  The bioheat equation ...................................................................................... 39  
D.2  Blood perfusion and effects ............................................................................ 41  
D.3  Thermal properties and dependent factors ...................................................... 41  
E  References ............................................................................................................. 42  

Chapter 3  Preliminary Studies and Results ..................................................................... 46  
A  Review of current antenna designs .......................................................................... 47  
A.1  Monopole coaxial antenna ............................................................................... 47  
A.2  Dipole antenna .................................................................................................. 49  
A.3  Slot antenna ...................................................................................................... 52  
A.4  Tri-axial antenna ............................................................................................... 53  
A.5  Cap-choke antenna .......................................................................................... 54  
A.6  Other antennas ................................................................................................ 55
Appendix 1 A Floating Sleeve Antenna Yields Localized Hepatic Microwave Ablation
I. Abstract................................................................................................................... 199
II. INTRODUCTION .................................................................................................. 200
III. Design of the floating sleeve antenna ................................................................. 202
IV. Computer Simulation and Experimental Results................................................ 203
  A. The computational electromagnetics (CEM) model........................................... 204
  B. Frequency sweep for antenna power reflection .................................................. 205
  C. Ex-vivo experiments and results ........................................................................ 206
V. Discussion and conclusion...................................................................................... 207
Acknowledgment ............................................................................................................ 208
References....................................................................................................................... 208

Appendix 2 Expanding the bioheat equation to include tissue internal water evaporation
during heating ............................................................................................................... 216
1 Abstract ................................................................................................................... 216
2 Introduction ............................................................................................................. 216
3 Methods ................................................................................................................... 219
  3.1 Theoretical solution of tissue water evaporation with the bioheat equation... 219
  3.2 Numeric Simulation ......................................................................................... 221
  3.3 Experiment setup and procedures ................................................................. 226
4 Results and Discussion ........................................................................................... 227
5 Conclusion .............................................................................................................. 231
6 References............................................................................................................... 232

Appendix 3 Measurement and Analysis of Tissue Water Content during Microwave
Tissue Ablation ............................................................................................................... 234
1. Abstract ................................................................................................................... 234
2. Introduction ............................................................................................................. 234
3. Experimental setup............................................................................................... 236
Appendix 4 Measurement and Analysis of Tissue Temperature during Microwave Liver Ablation
1 Abstract ........................................................................................................ 244
2 Introduction .................................................................................................. 244
3 Experiment setup ......................................................................................... 245
4 Results .......................................................................................................... 248
5 Discussion .................................................................................................... 252
  5.1 Differences between regions ................................................................. 252
  5.2 Tissue temperature versus tissue water content .................................. 253
6 Conclusion .................................................................................................... 256
7 References .................................................................................................... 257

Appendix 5 Computer Simulation of Microwave Liver Ablation .................... 258

Appendix 6 MATLAB and FEMLAB programs for the comprehensive computer models ........................................................................................................ 259
Abstract

MEASUREMENTS, ANTENNA DESIGN AND ADVANCED COMPUTER MODELING FOR MICROWAVE TISSUE ABLATION

Deshan Yang

Under the supervision of Prof. John G. Webster and Prof. David M. Mahvi at University of Wisconsin-Madison

Microwave hepatic ablation is a new promising technology to treat both primary and metastatic tumors of the liver by delivering microwave power with an antenna to heat and destroy the tumors. My Ph.D. research is to study the fundamental physical mechanisms of microwave ablation (MWA) of liver tissue and to utilize the results to improve microwave tumor ablation systems.

I designed the floating sleeve antenna—a new coaxial-based antenna, which is able to generate a localized heating pattern to overcome the backward heating problem of MWA. I performed experiments to measure tissue temperature and tissue water content changes for MWA. The measurement results suggested tissue water content and thermal energy movement during the course of MWA. I developed an expanded bioheat equation to cover both thermal conduction and tissue water evaporation at high tissue temperature. I think the consideration of tissue water evaporation, condensation and diffusion could be the key to integrate antenna performance and tissue thermal responses at higher temperature situations during MWA. I implemented a new comprehensive computer model to predict complete tissue physical responses during ablation. The computer model
simulates most important physical phenomena in MWA, including antenna performance, conductive heat transfer, blood perfusion, tissue water evaporation and diffusion, water vapor condensation and diffusion. It calculates and utilizes tissue physical properties depending on the tissue water content. Compared to the previous simple antenna EM model plus simple thermal conduction model, the comprehensive computer model generates results that match better to experimental results. Its results are not only lesion size and shape, but also tissue temperature and tissue water content time history and spatial distribution.

Chapter 1 gives introduction of microwave tissue ablation and the current research status.

Chapter 2 introduces EM theory and fundamental heat transfer theory for MWA. It also introduces tissue physical properties and their dependency factors.

Chapter 3 presents the preliminary studies and results. Current antenna designs for MWA, basic antenna fabrication procedures, basic MWA experiment procedures and initial results and basic computer models are also explained.

Chapter 4 introduces the expanded bioheat equation and the very important hypothesis regarding tissue water and thermal energy movement during MWA.

Chapter 5 covers tissue water content measurement procedures, results and analysis.

Chapter 6 covers tissue temperature measurement procedures, results and analysis.

Chapter 7 presents the new floating sleeve antenna, the design and theoretical analysis.
Chapter 8 covers the comprehensive computer model.

I have prepared 5 papers for publication during my Ph.D. program years. The sleeve antenna paper was accepted by IEEE-TBME. All other papers have either been submitted or will be submitted. I attached all the 5 papers at the end of the thesis as appendices. Important MATLAB and FEMLAB programs for the comprehensive computer model(s) are stored on a floppy disk as a part of the thesis.

Appendices

1: Sleeve antenna paper
2: Bioheat equation paper
3: Tissue water content measurement paper
4: Tissue temperature measurement paper
5: Comprehensive computer model paper
6: MATLAB and FEMLAB programs for the comprehensive computer models on the floppy disk
Chapter 1

Introduction to microwave tissue ablation
A Liver cancer and treatments

Hepatocellular carcinoma (HCC) is one of the most common malignant tumors with an estimated 1,000,000 worldwide deaths per year. Persistent or recurrent liver disease is the major cause of both morbidity and mortality in patients with HCC. The liver is the commonest site of distant metastasis of colorectal cancer and nearly half of the patients with colorectal cancer ultimately develop liver involvement during the course of their diseases. Nearly 150,000 new cases of colorectal cancer will be diagnosed in the US each year with 57,000 deaths. Among men 40 to 79 years old, colorectal cancer is the second leading cause of cancer mortality. Primary and secondary malignant hepatic tumors are among the most common tumors worldwide [1-4].

Though the ultimate control of this disease rests with the treatment of at-risk populations with vaccines for both hepatitis B and C, extirpation of tumor is the only potentially curative therapy for established cancers. Chemotherapy and radiation therapy are ineffective to treat liver tumors. Surgical resection is the gold standard for the treatment of patients with respectable isolated hepatic metastases with 40% 5 year and 26% 10 years survival. However, only about 25% patients are surgical candidates. In addition, the morbidity and mortality associated with surgical resection are relatively high. For most patients, tumors may be too close to the major hepatic blood vessels to be resected, or too many tumor spots to be resected and the remained portion of the liver would not be enough to sustain normal liver functionality for the patients. Such patients can not be surgical candidates. Patients without treatment will usually die in 1 to 5 years.
Ablative treatments have started to become viable alternative methods to treat patients who cannot be treated by surgery. Such ablative treatments include cryoablation, radiofrequency ablation (RFA), microwave ablation (MWA) or also called microwave coagulation therapy (MCT), and ethanol ablation, etc. [2, 3] These ablation methods can be performed in either open-hepatic operations or minimally invasive percutaneous operations [4].

Ablative treatments can treat most nonsurgical candidate patients. These ablative treatments do not work equally for all patients. Different ablation methods have different mechanisms, different rates of complication and reoccurrence. They are suitable for different patient groups.

Among these ablative technologies, RFA and MWA are similar in many ways. Both of them use heat to treat and kill tumor tissues. RFA is much more mature than MWA. RFA has been used in clinical operations in the USA for years while MWA is still undergoing major improvements and is actively researched. Clinical trials for microwave liver ablation have been carried out in Asian countries.

B Microwave tissue ablation fundamentals

The basic principle of microwave hepatic ablation is to apply microwave power to the liver tissue through the microwave applicator—the antenna. The power of the EM wave is absorbed by the liver tissue and heats the tissue. Liver tissue is destroyed after the tissue is heated to a high enough for a long enough time.

Figure 1 shows basic devices to perform a MWA are a microwave generator, a microwave applicator—the antenna, and a section of flexible coaxial cable to connect the
antenna to the microwave generator. Ultrasound scanners are often used in the MWA procedures to guide the placement of the applicator. Fiber-optic thermometers can be used to measure tissue temperature. MRI scanners can be used to examine lesion size after the procedures.

![Figure 1: Schematic of experimental setup of microwave liver tissue ablation. For a clinical procedure, the Luxtron fiber-optic thermometer and the temperature probes are not used.](image)

In a clinical MWA procedure, position of the tumor is determined in advance with medical imaging devices, including MRI, CT or ultrasound devices. A MWA probe is placed into the tumor with an open surgery or a percutaneous procedure, guided by ultrasound or other medical imaging device. The probe is connected to the microwave power generator. Microwave power level and heating duration are selected in advance according to the shape and size of the tumor. Microwave power is then applied for the selected duration. A thermal lesion of predicted volume is created by the applied microwave heat to cover the entire tumor with 1 cm margin. The MWA probe is then safely retrieved. Before the clinical procedure is finished entirely, imaging devices can be used to verify the lesion size and shape.
The ultimate goal of ablation technology, including MWA, is to kill the liver tumor while preserving healthy liver tissue effectively. In order to achieve the goal, an ablation method needs to:

**Kill the liver tumor completely and effectively**
- MWA needs to create a thermal lesion large enough to enclose tumors of large sizes with about 1 cm margin.
- MWA needs to be able to overcome the heat sink effect of large blood vessels and kill tumors right next to such blood vessels

**Minimize damage to healthy liver tissue and liver function**
- The thermal lesion created by MWA should be spherical in shape because liver tumors are generally spherical in shape
- The thermal lesion should cover the tumor only and completely in order to reduce the thermal damage to adjacent normal liver tissues
- MWA should selectively heat tumor tissue only instead of heating both tumor and normal tissues
- MWA procedure should be easy to control to generate thermal lesions of desired sizes

**Easy, fast and less costly procedures**
- MWA procedures should be easy to perform
- MWA procedures, devices and probes should be cost effective
- MWA procedures should be performed quickly in order to reduce operational time
MWA is one of the available thermal ablation technologies. There are other similar thermal ablation technologies, including radiofrequency ablation (RFA), laser ablation, hot-saline injection, focused ultrasound ablation, etc. RFA is the most used to treat liver tumors.

All thermal ablation technologies deliver heat energy to the targeted tissue by some sort of applicators and destroy tissue by heating, but the heating mechanisms and the abilities to create thermal lesions are different from one technology to another. Besides thermal conduction and blood perfusion in the liver tissue, which affect lesion creation for all thermal ablation technologies, the intrinsic heating mechanisms and the dependence on thermal conduction determine the lesion creation ability.

Among all the thermal ablative technologies, MWA is the most similar to RFA.

**The difference between MWA and RFA**

- MWA can heat tissue to a higher temperature
- MWA can deliver energy further into the tissue
- MWA is less dependent on thermal conduction
- MWA can still deliver energy into the tissue even when tissue is desiccated
- MWA works faster
- MWA does not need a grounding pad

**Advantages over RFA**

- Faster procedure, shorter duration
- Possible to generate larger lesions with single probe
• Able to create lesions better next to blood vessels
• No grounding pad needed, and less risk of skin burning due to the grounding pad
• Multiple MWA probes can work simultaneously without interfering with each other

Disadvantages
• MWA is less controllable on lesion size
• Detrimental backward heating
• Inability of generating lesion large enough for large tumors

C Current research status for MWA

C.1 Current clinical trials

Seki used 60 W for 120 s with a dipole antenna probe to raise temperature over 56 °C in a 3.5 cm × 2.5 cm area. The antenna probe was 30 cm long and 1.6 mm in diameter. The power level and duration were selected in order to keep the temperature on the shaft of the probe within 50 °C to reduce risk of skin burn [5]. The operations were performed percutaneously, guided by ultrasonography. A surgical specimen was obtained 30 days after the operation from one patient who had a small HCC measuring 1.4 cm in greatest dimension. Histopathologic findings revealed that thermal coagulation caused necrosis of the tumor and the surrounding parenchyma, leaving no viable cancer cells. A fibrous capsule had formed around the necrotic area.
Lu reported a MWA trial on 60 patients [6], guided by ultrasound imaging devices. He used a monopole antenna probe, 60 W for 300 s to create a coagulation volume of 3.7 cm $\times$ 2.6 cm $\times$ 2.6 cm. Multiple probes were used to treat large tumors. The probes were 24.7 cm long and 1.6 mm in diameter.

**C.2 Problems and challenges of current MWA technologies**

MWA is one of the new ablative technologies. It is not mature yet. Clinical trials of MWA were mainly carried out in Asia. Researches on MWA are now going on. Despite many promising advantages over other thermal ablative technologies, MWA still has many problems to be solved and technical challenges.

**Lesion size limitation**

In clinical trials or in-vivo experiments, a single MWA probe can only create thermal lesions of limited sizes in one pass [5, 6]. A common lesion size is 3.5 cm $\times$ 2.5 cm $\times$ 2.5 cm with 60 W and 120 to 300 s power application. The goal of an ablation procedure to treat liver tumor is to create a lesion covering the entire tumor with a 1 cm margin. For even a middle size tumor of 2 cm $\times$ 2 cm $\times$ 2 cm, a thermal lesion size of 4 cm $\times$ 4 cm $\times$ 4 cm is required to safely cover the entire tumor. Current MWA probes are apparently not powerful enough for such a requirement.

Tumors in human liver could be in sizes up to 10 cm in diameter. Tumors are usually spherical shaped, except the ones close to the liver surface. RFA is able to create lesions of up to 7 cm in diameter with multiple RFA probe configurations. It is desirable for MWA to generate a thermal lesion as large as possible with one antenna probe, in one pass of treatment. It is also possible to use multiple MWA probes to treat large tumors,
but multiple probe MWA procedures are more complicated and slower. In general, a 5 cm × 5 cm × 5 cm lesion to cover a 3 cm × 3 cm × 3 cm tumor is a desirable goal for MWA to achieve.

Strickland created lesions of large sizes in in-vivo pig experiments with larger antennas and higher power levels [7]. The antenna probes were customer designed with 6.8 mm diameter. He could push 180 W power through the antenna for 300 s to generate lesions up to 6.8 cm in diameter.

Large antennas are able to deliver high power and create large thermal lesions, but the large antenna size could conflict with the desirable size required by percutaneous operations.

**Detrimental backwards heating**

Detrimental backward heating is one of the major problems for MWA, especially for percutaneous treatments. The backward heating problem refers to the undesired heating that occurs along the coaxial feedline of the antenna. This detrimental heating causes damage to the liver outside the desired treatment region and can lead to burning of the skin during percutaneous treatment. This was the reason why Seki had to limit clinical trials to 60 W and 120 s in order to reduce the risk of skin burn.

There are three potential causes of detrimental heating along the coaxial feedline. First, any impedance mismatch between the antenna and the surrounding medium will create reflections that set up standing waves within the coaxial feedline. Under such conditions, the local currents on the inside of the outer conductor can become large enough to cause local heating. If the wall of the outer conductor is thin, the heat may transfer to the surrounding tissue. Second, an impedance mismatch between the antenna
and surrounding medium may also result in unbalanced currents on the inner and outer conductors of the coaxial feed. In this case, a remainder current flows along the outside of the outer conductor of the coaxial feedline. The ‘tail’ seen in many of the specific absorption rate (SAR) patterns computed from simulations of MWA antennas is attributed to this current flow. Finally, most antenna designs are based upon copper coaxial cables. Since copper is a good thermal conductor, heat generated near the distal tip may be conducted along the feedline.

The backward heating problem posts a huge challenge for MWA antenna designs. Most antennas used in MWA are unbalanced coaxial-based antennas, which transmit the microwave power out of their tips. The active radiation region of the antennas is usually from the antenna tip all the way back to the end of the long tail. Thermal lesions usually have tear-drop shapes. The backward heating problem will become more serious when power levels and application durations are increased in order to achieve larger lesions.

Fig. 1: Demonstration of the backward heating problem. The photo was presented by Brace in his triaxial antenna design paper in 2004 [8]. Brace’s experiments were performed ex-vivo with cow liver tissues. This photo clearly shows a tear-drop shape lesion with a long tail.
This detrimental backward heating problem has been studied in recent antenna designs. The cap-choke antenna, proposed by J. C. Lin, was one of the new antenna designs in an effort to address this problem [9]. I will discuss different antenna designs in a later section.

**Control of lesion generation**

Clinical treatment with MWA needs to control the lesion generation accurately in order to ensure destroying tumor tissue and minimizing damage to the normal liver tissue. Current MWA technology cannot provide such managed control over the MWA procedure because of the inhomogeneous liver tissue mechanical structure and lack of knowledge about the tissue thermal responses at higher temperature.

**Unknown tissue physical changes**

The tissue physical responses to MWA at high temperature are not well understood. MWA can heat tissue to much high temperatures than RFA. At high temperatures, tissue undergoes many physical changes, including loss of tissue water, changes of the tissue dielectric properties, thermal properties and other physical properties because of changes in temperature and in tissue water content, protein denaturalization, tissue charring, etc. All of such physical responses of tissue affect the MWA procedure.

**Computer simulation**

RFA is a much more mature technology. Computer simulations for RFA have been achieved with satisfactory accuracy. Due to the lack of knowledge about tissue physical responses for MWA, complete computer simulation of MWA is not achievable.
Computer simulation is very necessary to design and optimize the MWA antennas. Computer simulation also helps to optimize the MWA procedure by predicting the lesion size, shape versus power level and duration. Without a good computer simulation at reasonable accuracy, such optimizations have been done through unreliable experimental trials.

D References


Chapter 2

Theories and fundamental physics
A Electromagnetism and microwave-tissue interaction

A.1 Maxwell’s equations

The microwave electromagnetic frequency spectrum is from 300 MHz to 300 GHz, or wavelength from 1 m to 1 mm. In 1873, James Clerk Maxwell published the famous Maxwell’s equations, which mathematically describe the interdependence of the electric field and the magnetic field. Maxwell not only summarized previous scientists’ separate works on electric and magnetic fields in his equations, but also introduced the new concept of displacement current. Maxwell suggested the existence of electromagnetic waves, which were discovered later by other scientists.

Differential form of time-dependent Maxwell’s equations:

\[ \nabla \times \vec{E} = -\frac{\partial \vec{B}}{\partial t} \]  
(2.1)

\[ \nabla \times \vec{H} = \frac{\partial \vec{D}}{\partial t} + \vec{J} \]  
(2.2)

\[ \nabla \cdot \vec{B} = 0 \]  
(2.3)

\[ \nabla \cdot \vec{D} = \rho \]  
(2.4)

where \( \vec{E} \) is electric field intensity (V/m), \( \vec{H} \) is magnetic field intensity (A/m), \( \vec{D} \) is electric flux density (C/m\(^2\)), \( \vec{B} \) is magnetic flux density (Wb/m\(^2\)), \( \vec{J} \) is electric current density, \( \partial \vec{D} / \partial t \) is displacement current, \( \rho \) is the electric charge density.

\[ \vec{D} = \varepsilon \vec{E} \]  
(2.5)
\[ \mathbf{B} = \mu \mathbf{H} \]  

(2.6)

where \( \varepsilon \) is the electric permittivity (F/m) and \( \mu \) is the magnetic permeability (H/m) of the surrounding medium.

In free space:

\[ \varepsilon = \varepsilon_0 = 8.854 \times 10^{-12} \text{ (F/m)} \]  

(2.7)

\[ \mu = \mu_0 = 4\pi \times 10^{-7} \text{ (H/m)} \]  

(2.8)

In a material medium:

\[ \varepsilon = \varepsilon_r \varepsilon_0 \]  

(2.9)

\[ \mu = \mu_r \mu_0 \]  

(2.10)

where \( \varepsilon_r \) is the relative permittivity, or dielectric constant and \( \mu_r \) is the relative permeability.

Electromagnetic fields are completely described by Maxwell’s equations. All EM waves are time-dependent electromagnetic fields and obey the above Maxwell’s equations. If the EM waves are time-harmonic fields with sinusoidal time-variation, they have the form as:

\[ \mathbf{E}(t) = \mathbf{E}_0 \cos(\omega t + \varphi_E) \]  

(2.11)

\[ \mathbf{H}(t) = \mathbf{H}_0 \cos(\omega t + \varphi_H) \]  

(2.12)

where \( \omega \) is the angular frequency (rad), \( \varphi_E \) is the initial phase of \( \mathbf{E} \) and \( \varphi_H \) is the initial phase of \( \mathbf{H} \).

They can also be written in complex phasor representation:
\begin{align*}
\bar{E}(t) &= \text{Re}\left\{ E_0 e^{j(\omega t + \phi_0)} \right\} \\
\bar{H}(t) &= \text{Re}\left\{ H_0 e^{j(\omega t + \phi_0)} \right\}
\end{align*}

(2.13)

(2.14)

Phasors of \(\bar{E}\) and \(\bar{H}\) are defined as:

\begin{align*}
\bar{E}(t) &= E_0 e^{j(\omega t + \phi_0)} \\
\bar{H}(t) &= H_0 e^{j(\omega t + \phi_0)}
\end{align*}

(2.15)

(2.16)

The time derivatives can be greatly simplified with the phasor forms as:

\begin{align*}
\frac{\partial \bar{E}}{\partial t} &= j\omega \bar{E} \\
\frac{\partial^2 \bar{E}}{\partial t^2} &= -\omega^2 \bar{E} \\
\frac{\partial \bar{H}}{\partial t} &= j\omega \bar{H} \\
\frac{\partial^2 \bar{H}}{\partial t^2} &= -\omega^2 \bar{H}
\end{align*}

(2.17)

(2.18)

(2.19)

(2.20)

Maxwell’s equations can be written in phasor form as:

\begin{align*}
\nabla \times \bar{E} &= -j\omega \mu \bar{H} \\
\nabla \times \bar{H} &= j\omega \varepsilon \bar{E} + \sigma \bar{E} \\
\nabla \cdot \bar{B} &= 0 \\
\nabla \cdot \bar{D} &= \rho
\end{align*}

(2.21)

(2.22)

(2.23)

(2.24)

where \(\sigma\) is the electric conductivity of the surrounding medium (S/m) and
If vacuum is considered as the medium, the equations can be transformed to the Helmholtz equations, which are also the general solution for EM waves propagating in 3D space.

\[
\nabla^2 \mathbf{E} + \omega^2 \varepsilon \mu \mathbf{E} = 0 \tag{2.26}
\]

\[
\nabla^2 \mathbf{H} + \omega^2 \varepsilon \mu \mathbf{H} = 0 \tag{2.27}
\]

\(\varepsilon\) and \(\mu\) are complex tensors in the Helmholtz equations. These two equations are uncoupled. Only one of the two equations needs to be solved in order to completely describe the whole problem.

Please refer to [1, 2] for further discussion about Electromagnetism theory and Maxwell’s equations.

### A.2 EM waves in material medium

If the medium is a conductive medium with \(\sigma \neq 0\), the equations become:

\[
\nabla^2 \mathbf{E} + \omega^2 \mu \varepsilon \left( 1 - j \frac{\sigma}{\omega \varepsilon} \right) \mathbf{E} = 0 \tag{2.28}
\]

A complex relative permittivity is defined as:

\[
\varepsilon_i = \varepsilon'_i - \varepsilon''_i = \varepsilon'_i - j \frac{\sigma}{\omega \varepsilon_0} \tag{2.29}
\]

where \(\varepsilon'_i\) is the real relative permittivity of the medium. The new \(\varepsilon_i\) is the complex relative permittivity, a combination of the real relative permittivity and the conductivity of the medium.
Using equation 2.29, equation 2.28 could be written into the same form as equation 2.26, which has simple solutions. For a simple case that the wave is traveling in the \( z \) direction only, the solution is:

\[
\bar{E} = \bar{E}_0 e^{-j\gamma z}
\]  

(2.30)

where \( \gamma \) is the wave propagation constant, which is given as:

\[
\gamma = \alpha + j\beta = \sqrt{j\omega\mu(\sigma + j\omega\varepsilon)} = j\omega\sqrt{\mu\varepsilon} \sqrt{1 - j\frac{\sigma}{\omega\varepsilon}}
\]  

(2.31)

where \( \alpha \) is the attenuation constant (Np/m) and \( \beta \) is the phase constant (rad/m) and

\[
\alpha = \frac{\omega\sqrt{\mu\varepsilon}}{\sqrt{2}} \left[ \sqrt{1 + \left(\frac{\sigma}{\omega\varepsilon}\right)^2} - 1 \right]^{1/2}
\]  

(2.32)

\[
\beta = \frac{\omega\sqrt{\mu\varepsilon}}{\sqrt{2}} \left[ \sqrt{1 + \left(\frac{\sigma}{\omega\varepsilon}\right)^2} + 1 \right]^{1/2}
\]  

(2.33)

Phase velocity and wavelength are functions of frequency as:

\[
\nu_p = \frac{\omega}{\beta} = \frac{\sqrt{2}}{\sqrt{\mu\varepsilon}} \left[ \sqrt{1 + \left(\frac{\sigma}{\omega\varepsilon}\right)^2} - 1 \right]^{-1/2}
\]  

(2.34)

\[
\lambda = \frac{2\pi}{\nu_p} = \frac{\sqrt{2}}{f\sqrt{\mu\varepsilon}} \left[ \sqrt{1 + \left(\frac{\sigma}{\omega\varepsilon}\right)^2} - 1 \right]^{-1/2}
\]  

(2.35)

where \( \nu_p \) is the wave phase velocity, \( \lambda \) is the wavelength and \( f \) is the frequency.

The solution of \( \bar{E} \) in phasor form can be written as:

\[
\bar{E}(z) = \bar{E}_0 e^{-j\gamma z} = \bar{E}_0 e^{-\gamma z} e^{-j\beta z}
\]  

(2.36)
The time-dependent form of \( \vec{E} \) is:

\[
E(z,t) = E_0 e^{-\alpha t} \cos(\omega t - \beta z + \varphi_0)
\]  
(2.37)

For perfect dielectric medium in which \( \sigma = 0 \), it can be shown that:

\[
\alpha = 0
\]  
(2.38)

\[
\beta = \omega \sqrt{\mu \varepsilon} = \omega \sqrt{\mu_0 \varepsilon_r \varepsilon_0}
\]  
(2.39)

\[
\nu_p = \frac{\omega}{\beta} = \frac{1}{\sqrt{\mu_0 \varepsilon_r \varepsilon_0}}
\]  
(2.40)

\[
\lambda = \frac{2\pi}{\beta} = \frac{1}{f \sqrt{\mu \varepsilon}}
\]  
(2.41)

For an imperfect dielectric medium in which \( \sigma \neq 0 \) but \( \sigma / \omega \varepsilon << 1 \), it can be shown that:

\[
\alpha \approx \frac{\sigma}{2} \sqrt{\frac{\mu}{\varepsilon}}
\]  
(2.42)

\[
\beta \approx \omega \sqrt{\mu \varepsilon}
\]  
(2.43)

\[
\nu_p = \frac{\omega}{\beta} = \frac{1}{\sqrt{\mu \varepsilon}}
\]  
(2.44)

\[
\lambda \approx \frac{1}{f \sqrt{\mu \varepsilon}}
\]  
(2.45)

For a medium in which \( \sigma / \omega \varepsilon >> 1 \), it can be shown that:

\[
\alpha \approx \sqrt{\pi \mu \sigma}
\]  
(2.46)

\[
\beta \approx \sqrt{\pi \mu \sigma}
\]  
(2.47)
The quantity of \( \sigma / \omega \varepsilon \) is called the loss tangent. It describes how lossy the medium is. If the loss tangent \( \sigma / \omega \varepsilon \leq 0.1 \), the medium is called a good dielectric material. If the loss tangent \( \sigma / \omega \varepsilon \geq 10 \), the medium is called a good conductor.

In good conductors, the fields attenuate very rapidly. Skin depth is defined as the distance over which the fields are attenuated by a factor of \( e^{-1.0} \):

\[
\text{skin depth} = \delta = \frac{1}{\alpha} = \frac{1}{\sqrt{\pi f \mu \sigma}}
\]

\[ (2.50) \]

A general medium is referred to as a material that is neither a good dielectric nor a good conductor. For a general medium, it follows:

\[ 10 > \frac{\sigma}{\omega \varepsilon} > 0.1 \]

\[ (2.51) \]

### A.3 Power flow of EM waves

The time-dependent power flow density of the EM wave is given by the instantaneous Poynting vector:

\[
\mathbf{P}(t) = \mathbf{E}(t) \times \mathbf{H}(t)
\]

\[ (2.52) \]

The time-average power flow density for time-varying fields is given by:

\[
\langle \mathbf{P}(t) \rangle = \frac{1}{T} \int_0^T \mathbf{P}(t) \, dt = \frac{1}{T} \int_0^T \mathbf{E}(t) \times \mathbf{H}(t) \, dt
\]

\[ (2.53) \]
where $T$ is the period of the EM wave.

If the EM waves are time-harmonic fields and $\vec{E}$ and $\vec{H}$ are represented in phasor forms, the complex Poynting vector is defined as:

$$\vec{P} = \vec{E} \times \vec{H}^\ast$$  \hspace{1cm} (2.54)

where $\vec{H}^\ast$ is the complex conjugate vector of $\vec{H}$.

It can be shown that:

$$\langle \vec{P}(t) \rangle = \text{Re} \{\vec{P}\}$$  \hspace{1cm} (2.55)

In a lossy medium, the field $\vec{E}$ and $\vec{H}$ are attenuated as the factor of $e^{-j\alpha z}$, the power density is attenuated as the factor of $e^{-2\alpha z}$.

### A.4 Interactions of microwave and biological materials

Biological materials are generally lossy mediums for EM waves with finite electric conductivity. They are usually neither good dielectric materials nor good conductors. When EM waves propagate though the biological materials, the energy of EM waves is absorbed by the materials. The specific absorption rate (SAR) is defined as the power dissipation rate normalized by material density \[3\]. It can be shown that:

$$\text{SAR} = \frac{1}{\rho} \vec{J} \cdot \vec{E} = \frac{\sigma}{\rho} |\vec{E}|^2$$  \hspace{1cm} (2.56)

EM energy absorbed by biological materials becomes heat and causes the temperature of materials to increase. Heating and temperature increase cause other mechanical and chemical changes to the biological materials. This phenomenon was first noticed by Percy Spencer in 1946. He noticed that a candy bar in his pocket melted while
he stood near a microwave radar source. His observation of microwave heating led to the invention of the microwave oven. Liver tissue is one type of biological material. The effect of microwave heating in liver tissue is the basis of this research on microwave liver tissue ablation.

B Dielectric properties of biological tissues

Different tissues have differences in their dielectric properties. Tissue dielectric properties have very important roles in microwave tissue ablation. Dielectric properties of tissue directly affect the performance of the ablation probes—the microwave antennas. Equation 2.56 showed that tissue conductivity is directly associated to the EM wave energy dissipation in tissue.

B.1 Frequency dependence

Tissues’ dielectric properties change with microwave frequency. Many studies have been conducted to determine the relationship between dielectric properties and frequency for different types of biological tissues. Gabriel summarized most of the previous researches in his 1996 publications [4-6].

Figure 1 shows the graph of measured dielectric properties of liver tissue versus frequency. The graph is from Gabriel’s first of their three papers. The data in the graph summarized experimental results on different liver tissues from many previous researchers.

Figure 2 shows the graph of measured dielectric properties of human and ovine liver tissues versus frequency. The measurements were performed for a continuous
frequency spectrum. The graph is from Gabriel’s second of their three papers. The measured data are also compared to the individual experimental results from Figure 1.

In their third paper of the series [6], Gabriel suggested the empirical parameterized equation to approximate the measured dielectric properties of different tissues. By using the suggested parameters to different tissue types, the complex relative permittivity of tissues could be calculated as:

\[
\hat{\varepsilon}(\omega) = \varepsilon_\infty + \sum_n \frac{\Delta \varepsilon_n}{1 + (j \omega \tau_n)^{\alpha_n}} + \frac{\sigma_i}{j \omega \varepsilon_0} \quad \text{(2.57)}
\]

Gabriel gave parameters for liver tissues as Table 1. Figure 3 shows the plots and comparison between the measured values and calculated values for liver tissue relative permittivity and conductivity.
Figure 1: Summary of experimental measured dielectric properties of liver tissue. The graph is from [4].
Figure 2: Measurement of liver tissue dielectric properties versus continuous frequency spectrum.

The graph is from [5].
Table 1: Parameters for liver tissue for Gabriel tissue dielectric equation

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\varepsilon_\infty$</td>
<td>4.0</td>
<td>$\sigma_1$</td>
<td>0.0200</td>
</tr>
<tr>
<td>$\Delta\varepsilon_1$</td>
<td>39.0</td>
<td>$\Delta\varepsilon_2$</td>
<td>6000</td>
</tr>
<tr>
<td>$\tau_1$ (ps)</td>
<td>8.84</td>
<td>$\tau_2$ (ns)</td>
<td>530.52</td>
</tr>
<tr>
<td>$\alpha_1$</td>
<td>0.10</td>
<td>$\alpha_2$</td>
<td>0.20</td>
</tr>
<tr>
<td>$\Delta\varepsilon_3$</td>
<td>$5.0 \times 10^4$</td>
<td>$\Delta\varepsilon_4$</td>
<td>$3.0 \times 10^7$</td>
</tr>
<tr>
<td>$\tau_3$ (µs)</td>
<td>22.74</td>
<td>$\tau_4$ (ms)</td>
<td>15.915</td>
</tr>
<tr>
<td>$\alpha_3$</td>
<td>0.20</td>
<td>$\alpha_4$</td>
<td>0.05</td>
</tr>
</tbody>
</table>

Figure 3: Dielectric properties of liver tissue. Black lines are calculated values by the empirical equation. Gray lines are measured values.
B.2 Temperature dependence

It is well known that dielectric properties of biological tissues are dependent on temperature changes. Temperature dependence is difficult to measure and related publications are rare.

Chin and Sherar measured dielectric properties of bovine liver tissues at 915 MHz in their ex-vivo experiments by heating the liver tissues to different temperatures [7]. They suggested that changes in liver tissue dielectric properties due to heating are caused by the relaxations of two tissue components: tissue water and proteins. Changes due to tissue water content were found to be reversible and changes due to protein denaturization were found to be irreversible. The temperature dependence was only measured to the lower 50s °C. For reversible changes, the temperature coefficients were found to be 1.82 ± 0.28% °C⁻¹ for conductivity and –0.130 ± 0.0059% °C⁻¹ for relative permittivity respectively. The results reasonably agreed with the experimental results published by Duck 1990 [8]. However Chin and Sherar pointed out that the irreversible changes in tissue dielectric properties due to tissue protein denaturization were unpredictable because such changes were not only temperature dependent but also heating duration dependent.

Jaspard published his research on temperature dependence of dielectric properties of human and animal blood in 2002. He measured from 25 °C to 45 °C. At 1 GHz, his results showed that the temperature coefficients are –0.3% °C⁻¹ for relative permittivity and 1% °C⁻¹ for conductivity.

Such temperature dependence publications are rare and the published measurements are often performed in the temperature region much lower than the tissue
temperature region in which microwave tissue ablation is effective. In later chapters of this document, measurement results showed that the tissue temperature could reach much higher than 125 °C, which is the maximum temperature I could measure with my current fiber-optical thermometer.

**B.3 Tissue water content dependence**

In the microwave frequency range, tissue dielectric properties can be studied as suspensions of proteins in water solution and could be analyzed by using the Fricke dielectric mixture equation [9]. Because of the complexity of the biological tissue, the equation cannot be applied directly to tissues, especially to tissues with high water content. The values of dielectric properties can be calculated from the equation for comparison with measured tissue values.

\[
\varepsilon = \varepsilon_w \left( \frac{1 - P}{1 + (K - 1)P} \right) \left( 1 + \frac{KP\varepsilon_p}{\varepsilon_w (1 - P)} \right) \tag{2.58}
\]

where \(\varepsilon\) is the complex permittivity of the suspension of proteins in water, \(P\) is the volume fraction of suspended proteins, \(\varepsilon_w\) is the complex permittivity of water, and \(K\) is a factor dependent on the geometry and permittivity of the protein molecules.

\[
K = \frac{1 + \chi}{\chi} + \frac{\varepsilon_p}{\varepsilon_w} \tag{2.59}
\]

where \(\varepsilon_p\) is the permittivity of the protein molecules and \(\chi\) is the geometry factor for the shape of the protein moles. \(\chi\) equals 2 for spheres, 1.5 for prolate spheroids and 1.0 for oblate spheroids.
The ratio of \( \varepsilon_p / \varepsilon_w \) is quite small for frequency over 1.0 GHz, can be ignored from the equation and results in:

\[
K = \frac{1 + \chi}{\chi} \quad (2.60)
\]

Relative permittivity and conductivity of electrolyte can be calculated as:

\[
\varepsilon_w = \varepsilon_\infty + \frac{(\varepsilon_\infty - \varepsilon_s)}{1 + (f / f_c)^2} \quad (2.61)
\]

\[
\sigma_w = \sigma_s + \frac{2\pi f^2 (\varepsilon_\infty - \varepsilon_s)\varepsilon_0 / f_c}{1 + (f / f_c)^2} \quad (2.62)
\]

where \( f_c \) is the relaxation frequency of pure water, \( \varepsilon_s \) is permittivity at low frequency, \( \varepsilon_\infty \) is the permittivity at high frequency. \( f_c = 25 \) GHz, \( \varepsilon_\infty = 4 \) and \( \varepsilon_s = 74 \) for temperature at 37 °C. \( \sigma_s \) is the frequency independent conductivity due to the ions in the electrolyte solution. \( \sigma_s \) is dependent on the concentration of the ions.

For frequency higher than 1 GHz, the ratio \( \varepsilon_p / \varepsilon_w \) is very small. The equation 2.58 can be rewritten as:

\[
\varepsilon = \varepsilon_w \left( \frac{1 - P}{1 + (K - 1)P} \right) \quad (2.63)
\]

For frequency higher than 1 GHz, relative permittivity and conductivity of tissue can be computed as:

\[
\varepsilon_t = \varepsilon_\infty + \frac{(\varepsilon_\infty - \varepsilon')}{1 + (f / f_c)^2} \quad (2.64)
\]
\[ \sigma_t = \sigma_s^m + \frac{2\pi f^2 (\varepsilon_s^m - \varepsilon_{\infty}^s) \varepsilon_s}{1 + (f / f_c)} \]  

(2.65)

where \( \varepsilon_t \) is the permittivity of tissue, \( \sigma_t \) is the conductivity of tissue, \( \sigma_s^m \) is the contribution of the conductivity of tissue electrolytes and other dielectric relaxation processes occurring below 1 GHz to the tissue conductivity at the microwave frequencies. \( \varepsilon_s^m \) is proportional to the parameter \( \varepsilon_s \) of pure water, with the proportional factor equal to \((1 - P)/(1 + (K - 1)P)\). For no-fatty tissue, \( \varepsilon_{\infty}^s \) should be comparable to that of pure water, which equals 4. \( \sigma_s^m \) and \( \varepsilon_s^m \) are the two major variables depending on the tissue types.

Jonathan Schepps and Kenneth R Foster tabulated important parameters for different tissue types [9]. Parameters for liver tissues are:

<table>
<thead>
<tr>
<th>Volume fraction of water ((1 - P))</th>
<th>Extrapolated microwave permittivity (\varepsilon_s^m)</th>
<th>Conductivity at 0.1 GHz, (\sigma_{0.1}) (\text{mS/m})</th>
<th>Extrapolated microwave conductivity (\sigma_s^m) (\text{mS/m})</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.795</td>
<td>43</td>
<td>6.7</td>
<td>23</td>
</tr>
</tbody>
</table>

Equations 2.61 and 2.62 can be simplified for frequencies between 0.1 GHz to 27 GHz as [9]:

\[ \varepsilon' = 1.71 f^{-1.13} + \frac{\varepsilon_s^m - 4}{1 + (f / 25)} + 4 \]  

(2.66)

\[ \sigma = 1.35 f^{0.13} \sigma_{0.1} + \frac{0.0222 (\varepsilon_s^m - 4) f^2}{1 + (f / 25)^2} \]  

(2.67)
where \( f \) is the frequency in GHz, \( \sigma \) is the conductivity in mS/cm, \( \varepsilon' \) is the calculated tissue permittivity, but \( \varepsilon^m_S \) and \( \sigma_{0.1} \) are tabulated in table 1 in [9]. \( \varepsilon^m_S \) and \( \sigma_{0.1} \) can also be determined from the function of the volume fraction of water in tissue, \((1 - P)\).
Figure 4: Extrapolated (a) microwave permittivity $\varepsilon_m$ (b) conductivity $\sigma_{0.1}$ at 0.1 GHz, against the volume fraction of water in tissue $(1 - P)$ [9].

Susan Smith and Kenneth Foster in 1985 measured dielectric properties of low-water-content tissues [10]. They predicted the permittivity and conductivity of tissues by using the Maxwell mixture formula.

$$
\varepsilon^* = \varepsilon_d^* + \frac{2\varepsilon_d^* + \varepsilon_i^* - 2p(\varepsilon_d^* - \varepsilon_i^*)}{2\varepsilon_d^* + \varepsilon_i^* + p(\varepsilon_d^* - \varepsilon_i^*)}
$$

(2.68)

where $\varepsilon^*$ is the complex permittivity of the suspension spheres (proteins and lipids) in continuous medium (tissue water electrolyte), $\varepsilon_d^*$ is the permittivity of the continuous medium, $\varepsilon_i^*$ is the permittivity of the spheres, $p$ is the volume fraction.

The above equation was shown to be excellent approximation to the measured results at 100 MHz.
Figure 5: (a) Tissue permittivity normalized by the permittivity of water. (b) Tissue conductivity. Both permittivity and conductivity are plotted against the volume fraction of water. The data were measured at 25 °C at 100 MHz. The curves are the predicted values according to equation [10].
The results of the equation shown by Susan Smith seemed to be a very good approximation to the experimental measured data, but her research was for frequency below 1 GHz.

**Tissue water volume fraction versus weight fraction**

Both [9] and [10] present the relationship between tissue dielectric properties versus tissue water volume fraction. It is easier to measure tissue water weight fraction instead of volume fraction. The two fractions can convert to each other by assuming the average densities of lipid and protein fractions to be 0.9 and 1.3 g/cm$^3$ [10].

If liver tissues consist of protein instead of lipid, then:

\[
P_v = \frac{1.3P_w}{1 + 0.3P_w} \quad (2.69)
\]

\[
P_w = \frac{P_v}{1.3 - 0.3P_v} \quad (2.70)
\]

where $P_v$ is the volume fraction and $P_w$ is the weight fraction.

If $P_v = 0.795$, then $P_w = 0.749$.

**B.4 Dielectric properties of liver tumor tissues**

Liver tumor tissue has higher tissue water content. It has higher permittivity and higher conductivity than normal liver tissue. Stauffer reported that relative permittivity is 12% higher and electric conductivity was 24% higher for human liver tumor tissue than the surrounding normal liver tissue [11]. The differences of dielectric properties between normal tissues and tumor tissues agree with the relationship between tissue dielectric properties and tissue water content in the previous sections.
C Thermal responses of biological tissues during MWA

C.1 Tissue responses versus temperature

Tissue temperature elevates when microwave power is applied. Tissue near the active radiation region of the microwave antenna absorbs more microwave wave energy and has higher temperature than tissue further away from the antenna. Heat is also transferred from tissue at higher temperature to tissue at lower temperature by thermal conduction and blood perfusion in the liver tissue. The overall effect of the microwave power is to raise tissue temperature in a limited region near the antenna active radiation region.

Liver tissue undergoes a few steps of different physical responses to the temperature elevation.

Table 3: Physical responses of tissue and cells to heating [12]

<table>
<thead>
<tr>
<th>Temperature</th>
<th>Tissue responses</th>
</tr>
</thead>
<tbody>
<tr>
<td>&lt; 48 °C</td>
<td>Increase of tissue blood perfusion because of intrinsic response of tissue by enlarging blood vessel in order to reduce the tissue temperature.</td>
</tr>
<tr>
<td>48 °C</td>
<td>Cell depolarization Heat-caused pore formation on cell membrane and an increase of membrane fluidity leading an overwhelming number of extracellular ions rushing into the cells and causing cell depolarization</td>
</tr>
<tr>
<td>&lt; 50 °C</td>
<td>Cell physical changes are reversible</td>
</tr>
<tr>
<td>&gt; 50 °C</td>
<td>Heat causes cell transmembrane ion pump activities to stop and eventually cell dies.</td>
</tr>
<tr>
<td>&gt; 50 °C</td>
<td>Protein denaturization, tissue coagulation Tissue shrinking because of breaking down of collagen and other structure protein molecules Tissue color changes from red to white due to denaturization of myoglobin</td>
</tr>
<tr>
<td>&gt; 90 °C</td>
<td>Tissue water evaporates</td>
</tr>
<tr>
<td>&gt; 300 °C</td>
<td>Tissue charring</td>
</tr>
</tbody>
</table>
C.2 Tissue damage versus thermal dose

When considering the thermal damage of tissue during a thermal ablation procedure, the temperature history has to be taken into account [13]. The most commonly used model to describe the tissue thermal damage mathematically is the Arrhenius model [14]. It has been shown that there is an exponential relationship between the necessary treatment time and temperature to cause tissue damage for many tissue types. This rule applies for temperatures over 43 °C for most cases, that necessary treatment duration to cause tissue damage cuts in half with every 1 °C increase of treatment temperature. This rule can be mathematically described by the isoeffect equation:

\[ t_1 = t_2 \times R^{T_2 - T_1} \]  \hspace{1cm} (2.71)

where \( t_1 \) and \( t_2 \) are the necessary treatment durations at temperatures \( T_1 \) and \( T_2 \) respectively, \( R \) is the constant, equal to 0.5 for temperatures above 43 °C and 0.25 for temperatures below 43 °C [14]. Sapareto and Dewey suggested to quantify the thermal damage by a thermal dose – cumulative equivalent minutes at 43 °C, as CEM_{43}.

\[ CEM_{43} = \int R^{43-T(t)} dt \]  \hspace{1cm} (2.72)

Once the thermal dose exceeds a certain limit, the tissue is considered to be damaged. The critical value of thermal dose is about 340 min for liver tissue [15].
Figure 6: Necessary treatment duration versus treatment temperature [13]

Figure 6 shows the critical treatment time required to cause tissue damage depends on the treatment temperature assuming the treatment temperature is constant through the whole treatment duration. To damage liver tissue, the duration needs to be 340 min for temperature at 43 °C, 5.3 min for temperature at 49 °C, 1.3 min for temperature at 51 °C.

C.3 Pathological analysis

According to Yamashiki 2003, the area of ablation was histologically rimmed by a palisading, histiocytic, giant cell, inflammatory reaction associated with fibrotic bands. Coagulative necrosis with faded nuclei and eosinophilic cytoplasm were the predominant findings in the ablated areas. There were also areas in which the tumor cells had cytoplasmic eosinophilia, but nuclei were present and the cells seemed to be viable. Most
of the treated areas after microwave ablation develop coagulative necrosis accompanied by a foreign body–like inflammatory reaction and fibrosis.[16]

D Heat transfer and the bioheat equation

The transport of thermal energy in biological tissue is a complex process. It involves multiple physical mechanisms including heat conduction, convection, radiation, metabolism heat generation, tissue water evaporation, condensation, etc. Such a complex process is difficult to be studied; especially when tissue temperature is high enough for tissue water evaporation and water vapor condensation to take place. Unfortunately, MWA is one of the thermal ablative technologies that heats tissue to a temperature high enough for all the phenomena to happen.

D.1 The bioheat equation

Pennes’ Bioheat equation effectively describes how heat transfer occurs in biological tissue.

\[ \rho c \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + \text{SAR} - \rho_{bl}c_{bl}w_{bl}(T - T_{bl}) \]  

(2.73)

where \( \rho \) is the tissue density (kg/m\(^3\)), \( c \) is specific heat capacity (J/kg·K), \( k \) is thermal conductivity (W/m·K), \( \rho_{bl} \) is blood density (kg/m\(^3\)), \( c_{bl} \) is the specific heat capacity of blood (J/kg·K), \( w_{bl} \) is blood perfusion (kg/m\(^3\)·s), \( T_{bl} \) is blood temperature (K), SAR is the microwave power per unit volume applied by MWA (W/m\(^3\)).

Some important thermal and physical properties of normal liver tissue are listed in Table 4.
The Bioheat equation is a simple model for heat transfer in biological tissues. The major physical phenomena considered in the equation are microwave heating and tissue heat conduction. Heat conduction between tissue and blood flow in tissue is approximated by the item $\rho_{bl}c_{bl}w_{bl}(T - T_{bl})$ in the equation. Heat radiation and metabolism heat generation are assumed to be minimal during MWA and are ignored.

Since the Bioheat equation does not cover convective heat transfer, tissue water evaporation and water vapor condensation, it is only valid when thermal effects of these phenomena are minimal, particularly when temperature is relatively low and there is no major convective thermal transport. When applied under valid conditions, the Bioheat equation has proved to be a viable approximation for heat transfer in biological tissues [22-26].

The Bioheat equation cannot be applied to the situations when tissue temperature so high, over 90 °C, when evaporation of tissue water becomes the major factor for the overall thermal responses of tissues. It cannot be applied when convective heat transfer is one of the major considerations when fluid convection cannot be ignored during heating. The Bioheat equation can be safely applied for tissue temperatures below 70 °C.

### Table 4: Thermal properties of liver tissue and blood [8, 17-21]

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thermal conductivity [W/m-K]</td>
<td>0.5</td>
</tr>
<tr>
<td>Specific heat [J/kg-K]</td>
<td>3600</td>
</tr>
<tr>
<td>Density [kg/m³]</td>
<td>1060</td>
</tr>
<tr>
<td>Blood flow rate [m³/kg-s]</td>
<td>5×10⁻⁶</td>
</tr>
<tr>
<td>Density of blood [kg/m³]</td>
<td>1060</td>
</tr>
<tr>
<td>Specific heat of blood [J/kg-K]</td>
<td>3960</td>
</tr>
</tbody>
</table>
D.2 Blood perfusion and effects

The effects of blood flow on heat transfer in biological tissue must be considered when thermal ablations are performed in-vivo to living tissues. A common approach for small blood vessels is to consider them as uniformly distributed heat sinks in the whole tissue, based on the basic assumption that blood enters the local tissue volume at the arterial temperature $T_{bl}$ and leaves this volume at local tissue temperature [27]. Under such an assumption, the term $\rho_{bl}c_{bl}w_{bl}(T - T_{bl})$ is introduced into the Bioheat equation.

An alternative method to treat small vessel blood flow is to use the concept of effective thermal conductivity $k_{\text{eff}}$, instead of the concept of heat sink [27].

\[
\rho c \frac{\partial T}{\partial t} = \nabla \cdot (k_{\text{eff}} \nabla T) + \text{SAR} \tag{2.74}
\]

Large blood vessels cannot be treated as uniform heat sinks or enhanced thermal conductivity. They have to be considered individually because convective heat transfer inside the blood vessel cannot be ignored [24, 27, 28].

D.3 Thermal properties and dependent factors

Tissue thermal and physical properties are not constant. There are a few factors that could affect tissue properties. Temperature and tissue composition are two of the most important factors. Researchers have reported on the relationship between these affecting factors and tissue properties.

Bhattacharya studied the temperature dependence of thermal conductivity ex-vivo with cow liver in 2003 [29]. He showed that thermal conductivity of cow liver had
reversible temperature dependence for temperatures up to 90 °C. The change with thermal conductivity was irreversible if temperature was over 90 °C. His results showed:

\[
k(T) = 0.4475 + 0.0033T
\]

(2.75)

where \( k \) is the tissue thermal conductivity and \( T \) is the temperature. Bhattachary’s equation was good for 25 °C < \( T \) < 80 °C. The temperature coefficient was higher than the previous results on human liver by Valvano 1985 [30]:

\[
k(T) = 0.4692 + 0.0012T
\]

(2.76)

Valvano’s results were good for 3 °C < \( T \) < 45 °C.

According to Bhattacharya, the thermal conductivities of cow liver are 0.57 and 0.74 for temperatures at 37.5 °C and 90 °C respectively. The accuracy of the empirical equation is still questionable, especially for the situation that the temperature coefficients are different for different tissue types and from in-vivo to ex-vivo, but the temperature dependence of tissue thermal conductivity is confirmed.

Similar to the tissue water content dependence of tissue dielectric properties, tissue thermal properties also depend on the tissue water content. Empirical equations are available to calculate tissue thermal properties from tissue water content. Duck has summarized results from different researchers in [8]. I will discuss more about the dependence of tissue properties on tissue water content in the later chapter on comprehensive computer models.

E References


Chapter 3

Preliminary Studies and Results
This chapter presents the preliminary studies and results. Currently available antenna designs for MWA are reviewed. Basic antenna fabrication procedures, basic MWA experiment procedures and initial results are presented. Basic EM model and thermal model are also explained.

A Review of current antenna designs

The microwave antenna, as the applicator for MWA, is the most important part of the MWA system. Many different antennas designs have been proposed, optimized and verified. Coaxial-based antennas are extremely important for MWA application because of their low cost and small dimensions. Coaxial antennas differ in many ways: the allowed power level, thermal lesion size, lesion shape, antenna dimension, etc. Different antennas have different tolerances to minimize backward heating.

I worked together with John Bertram on the paper “A review of coaxial-based interstitial antennas for hepatic microwave ablation” which was submitted in Dec. 2004. This section is basically a concise reiteration of related contents of this paper.

A.1 Monopole coaxial antenna

The monopole coaxial antenna is one of the very basic and commonly used antennas in MWA. The basic configuration of a monopole antenna is to have the center conductor of the coaxial cable extending further than the outer conductor of the coaxial cable, with or without the cable dielectric extended together with the center conductor. The EM field is between the extended center conductor and the whole outer conductor. The most active radiation region of the antenna is near the center conductor extension.
The length of the extension of the center conductor is \( \frac{\lambda_{\text{eff}}}{4} \) where \( \lambda_{\text{eff}} \) is the effective wavelength in the medium [1].

Figure 1: Three basic configurations of monopole coaxial antennas [2-4]. (a) open-tip monopole (OTM), (b) dielectric-tip monopole (DTM) and (c) metal-tip monopole (MTM).

Figure 1 shows three basic configurations of monopole coaxial antennas. Figure 2 shows the plots of SAR patterns for OTM and MTM. Figure 3 shows the corresponding measurements of antenna frequency responses. The metal-tip helps to move the antenna resonant frequency lower.

The monopole antennas are very easy to fabricate, but one can see from the figures that the performance of these monopole antennas are not very desirable for MWA. Their SAR patterns have long tails, which could contribute to backward heating. The antenna resonant frequency of the OTM antenna is at the desired working frequency of MWA at 2.45 GHz or 915 MHz.
A.2 Dipole antenna

The dipole antenna is another commonly used antenna for MWA. The remarkable structures of the dipole antenna are the antenna slot and the antenna termination tip, which is an enlarged metal structure of the coaxial center conductor. The whole coaxial antenna structure is usually sealed in a catheter to be used in MWA procedures.
Figure 4: Schematic of the dipole antenna I have fabricated for our MWA experiments.

A coaxial dipole antenna is an unbalanced dipole structure. The antenna tip is one pole and the long outer conductor of the coaxial cable is the second pole. The length of the tip is usually short and the outer conductor tail of the antenna is much longer. The unequal lengths of both poles make the EM fields of the antenna unbalanced between the two poles.

Figure 5: Plot of the SAR pattern for the dipole antenna. SAR contours are in dB scale, normalized to maximal SAR value of the EM field.

The dipole antenna has very low power reflection when it is immersed in liver tissue at a frequency of 2.45 GHz. The SAR pattern shows that it does not do a good job to minimize the backward heating. The SAR pattern apparently has the tail towards to the insertion point of the antenna.
Another problem with both the dipole antenna and the monopole antenna is that their SAR patterns depended on the antenna insertion depth. Hurter described this problem in 1991 [5].
Figure 6: Demonstration of the problem that SAR patterns depend on the antenna insertion depth [5]. Three SAR patterns from computer simulations are shown for different antenna insertion depths.

### A.3 Slot antenna

Slot coaxial antennas are the most popular antennas in MWA. For a slot antenna, the outer conductor and the center conductor are soldered at the end of the antenna tip and a ring of metal is cut off the antenna outer conductor to be the antenna slot. Slot antennas are easy to fabricate. Looking from outside of the antenna, a slot antenna is actually very similar to a dipole antenna. The tip of the dipole antenna is a whole piece of metal, while the inside of the tip of a slot antenna is still the dielectric of the coaxial cable. Like all other antennas, a slot antenna is usually sealed in its catheter to be used in a MWA procedure.
For frequency at 2.45 GHz, a slot antenna works very similarly to a dipole antenna. It has very low power reflection and it has a tail on its SAR pattern. With such a tail, a slot antenna would have the problem of backward heating. Its SAR patterns also depend on the antenna insertion depth.

### A.4 Tri-axial antenna

Brace proposed the triaxial antenna design in 2004 [6]. Figure 9 shows an open-tip monopole coaxial antenna inserted through an 18-gauge biopsy needle. The needle is placed 1/4 wavelength from the antenna base. The monopole antenna and the biopsy needle together form a triaxial structure.
The triaxial antenna does not work better than other antennas for most aspects of antenna performance. It does not solve the backward heating problem, and it does not create larger lesions. Nevertheless, this antenna is very easy to use in real clinical operations, especially for percutaneous treatments. The biopsy needle with introducer can be easily inserted and placed into the liver tissue. The introducer is then redrawn and the monopole antenna is inserted into the needle and advanced to the desired position.

### A.5 Cap-choke antenna

The cap-choke antenna was first reported by J. C. Lin in 1996 [7]. The purpose of the cap and choke is to constrain the EM field to the tip of the antenna and to reduce the
backward heating problem. A cap-choke antenna can provide a better localized and less
insertion depth dependent SAR pattern.

Figure 11: Schematic of the cap-choke antenna

Figure 12: SAR pattern of the cap-choke antenna. SAR contours are in dB scale, normalized to
maximal SAR value of the EM field.

A.6 Other antennas

John Bertram designed a modified cap-choke double slot antenna. This antenna is
optimized to have its SAR pattern better localized to the active radiation region in order
to further ease the backward heating problem.
Figure 13: Schematic of the double slot cap-choke antenna, designed by John Bertram in 2004.

Compared to the cap-choke antenna, the double-slot cap-choke antenna generates a better SAR pattern. However it has higher power reflection.

Figure 14: SAR pattern of the double slot cap-choke antenna. SAR contours are in dB scale, normalized to maximal SAR value of the EM field.

There are other antennas designed for MWA. However all other antennas are either not suitable for MWA, or perform less well. They haven’t received much attention for MWA applications.

A.7 Antenna array

An array of antennas could be used simultaneously in order to create larger thermal lesions for large tumors [8-14]. A liver tumor could be as large as 10 cm in diameter. A single antenna is not likely able to treat such a large tumor in one pass.
Multiple treatments with one antenna are possible, but such treatments will take long time to finish. Simultaneous treatment with multiple antennas is a better and faster idea.

Figure 15 shows that simultaneous treatment by an antenna array could generate larger thermal lesions than sequential multiple treatments.

Figure 15: Demonstration of MWA with antenna array. (a) MWA lesion with one antenna, 40 W for 10 min (b) MWA lesion of sequential treatments with 3 parallel antennas, antennas are separated by 1.1 cm, (c) simultaneous treatment by 3 antennas, separated by 1.6 cm.

**B Antenna fabrication procedure**

It is necessary to fabricate a coaxial antenna after the antenna is designed. This section shows the basic steps that I used to fabricate a simple dipole coaxial antenna in the lab. Other antennas can be made using similar steps.

Materials and hand tools required include UT-085 semirigid microwave coaxial cable, SMB 21 coaxial connectors, Teflon tape, fine needle files, coaxial cable striper and cutter, caliper and ruler, and soldering iron and solder.
1. Use a coaxial cable cutter to cut a section of UT-085 coaxial cable. The length of the section is about the desired length of the whole antenna, and it should be longer than 8 cm. The final finished antenna will be about 6 mm longer than the coaxial cable section.

2. Use the coaxial cable striper to cut off a 13 mm long section of the outer conductor and the dielectric at one end of the coaxial cable (the end of the antenna distal tip) to expose the center conductor. If the cutting surface is rough, use a fine flat needle file or other available tools to smooth it. The exposed center conductor should be 13 mm long.

3. The section of the outer conductor and the dielectric, which is stripped from the coaxial cable section, should be about 13 mm long. Use a small tool to push the dielectric off the outer conductor.

4. The outer conductor section now becomes a copper tube 13 mm long. Cut it down to 9.5 mm long and use fine needle files to finish both of its surfaces and shorten it to 9 mm long. Or the 13 mm copper tube can be directly filed and shortened to 9 mm long by using a flat needle file.

5. If either end of the outer conductor tube is a little flat, use a round needle file to expand the end carefully and gently, and make the end round.

6. The dielectric should be about 13 mm long. Use a sharp knife to cut both ends of it down to 6 mm long. Reinsert the dielectric back to the finished 9 mm outer conductor copper tube. Advance the dielectric to the middle of the outer conductor. Since the dielectric is 6 mm long, the outer conductor is about 9 mm
long. The dielectric should appear 1.5 mm deep into the outer conductor at both ends as in Figure 16 (b).

7. Push the whole part back onto the center conductor. This will be the tip portion of the coaxial dipole antenna. It should appear like the left half of Figure 16 (a).

8. Use a 15 to 30 W soldering iron to solder both ends of the outer conductor to the center conductor. Let the whole assembly cool down and then use a fine needle file to remove any excess solder, especially any excess solder inside the antenna slot and outside the outer conductor tube. Be careful not to break the inner conductor.

9. After soldering, the whole assembly should look like Figure 16 (a). The antenna slot side should be clean and flat, the distal tip side should have a round end.

10. Install the SMB connector according to the SMB connector installation guide from the manufacturer of the SMB connector.

11. The whole procedure can actually start from a preassembled kit that already has the SMB connector installed onto the coaxial cable section by the coaxial cable-connector assembly supplier.

![Figure 16: Fabrication of a simple dipole coaxial antenna](image-url)
C Basic MWA Experiments and Results

C.1 Ex-vivo Experiments

All our ex-vivo experiments were performed on cow liver tissues. Liver tissues were obtained from a local slaughter house and transported to our lab in coolers. Liver tissues obtained were excised from live cows. Experiments were performed on either fresh liver tissue, or liver tissue stored in refrigerator over night. If liver tissue was stored in the refrigerator, it was sealed in plastic bags to prevent extensive loss of tissue water.

Initial temperatures for fresh liver and for refrigerated liver are different. Tissue temperature also drops for fresh liver in the cooler. Tissue temperatures were always measured before each experiment.

Ex-vivo experiments were performed on a selected piece of liver tissue, which was cut off from the whole cow liver. The tissue piece was usually selected to be large enough to accommodate the predicted lesion size with a reasonable margin. The tissue piece also needed to be free of large blood vessels, and as homogeneous as possible. Large blood vessels could affect the lesion formation because hot water vapor could travel via the blood vessel out from the center of the lesion, heat tissue along the vessel, and reduce the lesion size at the center of microwave heating.

Figure 17 shows the general lab configuration for ex-vivo MWA experiments. A microwave generator supplied power to a custom built coaxial antenna inserted into the liver. Four fiberoptic temperature probes were placed alongside the antenna to measure real-time temperature during ablation. These probes have a minimal effect on the EM radiation patterns in liver. The four probes were connected to a Luxtron 3100 Fluoroptic
thermometer, which monitored the temperature and saved the data to a computer. Power and reflection levels were carefully monitored during ablation. When tissue temperature was not measured in real time, the Luxtron thermometer, the temperature probes and the personal computer were not needed.

![Diagram of ablation setup]

**Figure 17: Configuration of ex-vivo experiment**

After ablation, liver tissues were usually cut by using a meat slicer into thin slices, in either the transverse direction or the longitudinal direction, with respect to the direction of antenna insertion. Tissue slices were scanned into digital images at selected resolution by using a HP scanner. Images files were stored in our lab PC in organized folder structures. Lesion size was measured on the image files by using NIH ImageJ software. The thicknesses of the slices were measured with a ruler. The height of a stack of slices was measured and thickness of each slice was calculated by dividing the height by the number of slices, assuming slices cut by the meat slicer were equally thick. After slice thickness was known, the lesion volume was easily calculated by multiplying the lesion size of each slice by the slice thickness.

A simpler step was taken sometimes instead of the whole slicing and scanning procedure. Ablation liver tissue was cut open by using a knife to expose the generated
thermal lesion, still in either transverse direction or longitudinal direction with respect to the antenna insertion direction. A digital camera was used then to take a photo of the exposed lesion, together with a ruler. The lesion photo image was then transferred and stored in the lab computer. ImageJ software was used to measure the lesion size with the help of the ruler in the same image of the lesion.

C.2 In-vivo Experiments

Domestic pigs of various weights were used for the in-vivo experiments. Pre-approval for all animal experiments was obtained from the Institutional Animal Care and Use Committee, University of Wisconsin, Madison. All procedures were performed with the animals under general anesthesia. Induction of anesthesia was achieved by using an intramuscular injection of tiletamine hydrochloride, zolazepam hydrochloride, and xylazine hydrochloride. The animals were then intubated and maintained on inhaled halothane. Once adequate anesthesia was achieved, the abdomen was opened and liver was exposed [15].
Figure 18: Photos of thermal lesions. (a) ex-vivo cow liver. (b) in-vivo pig liver. In the in-vivo lesion photo, the lesion has a clear boundary in dark red color, caused by increased blood perfusion during heating.

Compared to the cow liver tissue used in ex-vivo experiments, pig livers are usually much smaller in size. We usually use less power and shorter duration for in-vivo experiments. After ablations were finished, the pig was euthanized and the liver excised and transported back to our lab. The lesion was then measured in the same way as in ex-vivo experiments.

### C.3 Basic observations of ablated liver tissue

#### Lesion color

The photos of thermal lesions obtained from ex-vivo and in-vivo experiments, show that tissue color changes from the center of the lesion to the outside.

Normal tissue outside of the thermal lesion appears red. The color is believed to be dominated by the hemoglobin protein. Lesion color changes to light red, light gray, gray, even to dark from the outside of normal tissue to center of the lesion. The lesion color indicates the degree of tissue ablation. If the power level is high enough and the ablation duration is long enough, the center of the lesion could be very well cooked and charred. In such a case, the lesion is completely black.

#### Tissue hardness

The softness of the lesion can be felt with the hand. Tissue softness certainly changes along with the change of the tissue color or the degree of ablation. Normal tissue outside the lesion is soft. Light red partially ablated tissue also feels very soft. The
hardness of the tissue increases from the normal tissue outside of the lesion to the center of the lesion. Light gray tissue is harder than light red tissue. Gray tissue is harder but still with some softness. Completely charred black tissue is very hard.

Lesion water content

Like the softness of the tissue, the dryness of ablated tissue can also be felt. The black charred tissue feels very dry. Normal tissue or light red tissue feels wet.

Lesion water content, color and softness are certainly correlated. All of them correspond to the degree of ablation.

C.4 Measurements of lesion size and volume

Determination of the lesion size

Determination of lesion size for in-vivo experiments is fairly easy. When tissue temperature rises during thermal ablation, the peripheral blood vessels are enlarged and blood perfusion increases. At the boundary of the thermal lesion, partially ablated tissue becomes redder in color than normal tissue because of increased blood perfusion. Such partially ablated tissue forms a distinct dark red ring and clearly marks the boundary of the lesion as shown in Figure 18 (b). Inside the ring, completely ablated tissue is light red. Outside the ring, normal tissue has normal color.

Determination of lesion size for ex-vivo experiments is much more difficult. A lesion can still be roughly determined by the tissue color. Because there is no blood perfusion, there is no such distinct ring to clearly mark the lesion boundary. In fact, the change in color from light red to red is so smooth that it is very difficult to tell the ablated tissue from normal red tissue.
Image processing techniques could help. Digital images of scanned lesion slices can be improved by applying image processing filters. I have prepared a MATLAB program to automate the image processing procedure.

**Lesion volume versus power level and ablation duration**

I have performed basic ex-vivo experiments to measure lesion size and volume versus power level and ablation durations. Figure 19 and Figure 20 show the results. Results from experiments are shown in solid lines. Experiments results are compared to the simulated results generated by the simple thermal model, which will be discussed in later sections. The simulated lesion volume was determined from the 50 °C contour.

**Lesion size versus initial tissue temperature**

![Figure 19: Lesion volume versus power level for MWA duration = 180 s](image1)

![Figure 20: Lesion volume versus MWA duration for power level at 120 W](image2)

In ex-vivo experiments, the generated thermal lesion sizes are apparently related to the tissue initial temperature. This is also the reason why we always measured tissue temperature for each ablation experiments. Tissue cells start to die at about 50 °C and proteins start to denature at about 60 °C. Since we use color to determine the lesion
boundary for ex-vivo experiments and the tissue color changes are dominated by the
denaturization of hemoglobin protein, I believe the tissue color changes require tissue
temperature as high as 60 °C for ex-vivo experiments.

Tissues with higher initial temperature require less time and less thermal energy
to increase to 60 °C. This is the reason why we can obtain a larger lesion with tissues
with higher initial temperature with the same antenna, same power level and same
ablation duration.

**C.5 Basic tissue temperature measurement**

We used Luxtron Fluoroptic thermometer model 3100 to measure tissue
temperature during ex-vivo bovine liver tissue MWA. A fluoroptic thermometer was
selected because its fiber-optic temperature sensors are unaffected by microwave
radiation and have minimal disturbance on the antenna SAR.

Figure 17 shows the ex-vivo experiment setup. The fiber-optic sensors are
inserted into the liver alongside the antenna probe. Figure 21 shows a photo of in-vivo
temperature measurement. A glass template was used to guide the antenna and the fiber-
optic sensor and ensure their relative positions.
C.6 Tissue dielectric property measurement

I reviewed the literature on tissue dielectric properties. It is also necessary to measure tissue dielectric properties on our own because we are interested in the dielectric properties of not only normal tissue, but also partially or completely ablated tissues. Accurate measurement of tissue dielectric properties will ensure the accuracy of the EM computer model.
Our dielectric spectroscopy measurement technique is based upon a method currently being used by Prof. Hagness’s group to measure the dielectric properties of normal and malignant breast tissue [16]. A hermetic, open-ended coaxial probe attached to a vector network analyzer (VNA) is placed directly in contact with the tissue. Measurements of the reflection coefficient $\Gamma$ at the aperture of the probe are recorded and converted to complex permittivity using a rational function model [17]. Data are acquired from 50 MHz to 20 GHz. Figure 22 shows one of the results measured with fresh bovine liver ex-vivo.

**C.7 Antenna S11 measurement versus frequency**

Antenna power reflection coefficient can be measured with a vector network analyzer (VNA). Compared to the tissue dielectric properties measurement, the antenna
to be measured is used instead of the standard open-ended coaxial probe. The antenna under measurement is inserted into the center of the tissue, measurement of the reflection coefficient is then recorded for frequencies from 50 MHz to 20 GHz. Figure 26 shows one of the measurement results.

D Basic computer simulations for MWA

I use FEMLAB version 2.3 and version 3.1, together with MATLAB, as my primary computer simulation tools. FEMLAB is a commercial software package, which solves partial differential equations using the finite element method (FEM). FEMLAB™ can be used standalone. It can also be fully integrated with MATLAB™. This feature of integration with MATLAB makes the pre- and postprocessing of data straightforward, and also permits using an external (MATLAB™) program to take control of FEMLAB™ simulations. This allows us to create integrated models involving EM and thermal simulations resulting in highly accurate treatment of the complex coupled physical processes that occur during MWA.

D.1 Axial-symmetry computer models

All my basic EM models and thermal models are built in axial-symmetrical mode. Support of axial-symmetrical models is another major factor for my choice of FEMLAB as my simulation software. All the coaxial-based antennas have rotational symmetric. I assumed that the antenna is immersed in a medium that also exhibits rotational symmetry. Under these conditions, the azimuthal dependence of the vector EM field components and scalar thermal distribution can be accounted for analytically and factored out of the
governing equations, eliminating the need for gridding in the azimuthal direction. Therefore, our preliminary EM and thermal models have been created using 2D computational grids while preserving the full 3D nature of the vector EM fields. The axial-symmetrical models are sometimes called 2.5D models because they are created in 2D but different from either plain 2D models or 3D models. An axial-symmetric model is computed in 2D, but it represents the fields fully in 3D. This is much more computationally efficient than conducting the simulations in 3D. Using axial-symmetric models allows me to reduce the computation time from hours to seconds, as well as improve the grid resolution of my models 10 times.

Axial-symmetric models have their limitations. They are very good to represent a coaxial antenna in a simple homogeneous medium, and they are not good to model media with complex structures, such as large blood vessels. If I need to study the thermal field behavior in a medium with a few large blood vessels, I have to use a full 3D model instead of an axial-symmetric model.

FEMLAB does support full 3D models. I have found that FEMLAB has limited computational power for large 3D models with many mesh elements. It often runs out of memory for large models, even when there is a significant amount of system memory still available. This problem has been improved from FEMLAB version 2.3 to version 3.1, but it is still not completely solved. Abaqus and other FEM simulation software have been used to solve larger 3D models better than FEMLAB, but the much better GUI, data post-processing features as well as the ability to integrate with MATLAB make FEMLAB still the best choice for my basic models.
D.2 The basic EM model

My simple EM model is basic in the way that it does not consider the possible changes of tissue dielectric properties because of changes in tissue temperature and tissue water content during heating. Therefore the basic model is good to study the antenna performance in a static tissue dielectric environment.

My basic EM model was adapted from a basic slot antenna EM model [18] from Comsol, the company who created the FEMLAB software. I learned how to define the power source in my axial-symmetric models from the Comsol model. The EM models are powerful tools for me to study the antennas’ static performance. I can create new antenna designs, modify the antenna geometrical parameters, change the dielectric properties of the medium or the dielectric material of the coaxial cable, etc. I can then observe the antenna performance changes instantly after I modify the model parameters, and to select the optimal antenna parameters according to my predetermined rules. I would like to use the basic EM model of a simple dipole antenna to show the most important elements of the axial-symmetric FEMLAB EM computer models. The schematic of the dipole antenna is shown in Figure 4 and its SAR pattern is shown in Figure 5.

Axial-symmetrical EM waves

In axial-symmetric mode, there is no variation in the azimuthal direction—the $\phi$ direction. We use transverse magnetic (TM) waves in our model. TM waves have a magnetic field with only a $\phi$ component and electric field in the $r$–$z$ plane. The fields can be written:

$$\mathbf{H}(r, z, t) = H_\phi(r, z)e^{j\omega t}$$

(3.1)
\[ E(r, z, t) = (E_r(r, z) + E_z(r, z))e^{j\omega t} \]  \hspace{1cm} (3.2)

**Geometry of the EM model**

Figure 23 shows the geometry of the EM model of the dipole antenna. Because the model is axial-symmetric, only a half of the geometry structures of antenna and liver tissue are created in the model. The half of the geometry rotates along the \( z \) axis at \( r = 0 \) to form the whole 3D geometry of the antenna immersed in the liver tissue.

![Figure 23: The geometry of the dipole antenna EM model.](image)

(a) Figure 23: The geometry of the dipole antenna EM model. (a) the whole EM model, (b) the EM model details near the antenna slot. \( z \) is the axis in the longitudinal direction and \( r \) is the axis in the radial direction. Liver tissue is the big light gray area. The catheter and dielectric material of the coaxial cable, both are Teflon, are shown in dark gray. The antenna slot is 2 mm wide, left aligned to \( z = 0 \).

**Boundary definitions**

Important boundaries in the model are the external boundaries, including the outer boundaries of the liver tissue, and all boundaries between copper and Teflon. Boundaries along the \( z \) axis are set to be axial-symmetric boundaries. External boundaries of liver tissue and Teflon, except the boundary at the \( z \) axis, are set to be low-reflection
boundaries. All copper boundaries are set to be PEC (Perfect Electric Conductor) boundaries.

Perfect electric conductor (PEC):

$$\mathbf{n} \times \mathbf{E} = 0$$  \hspace{1cm} (3.3)

Low-reflection boundary:

$$\sqrt{\varepsilon - \frac{j\sigma}{\omega}} (\mathbf{n} \times \mathbf{E}_\phi) - \sqrt{\mu} H_\phi = -2\sqrt{\mu} H_{0\phi}$$  \hspace{1cm} (3.4)

Basic EM model constants and domain parameters

<table>
<thead>
<tr>
<th>Teflon</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Relative permittivity</td>
<td>2.1</td>
</tr>
<tr>
<td>Conductivity</td>
<td>0</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Liver tissue</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Relative permittivity</td>
<td>43.09</td>
</tr>
<tr>
<td>Conductivity</td>
<td>1.69 [S/m]</td>
</tr>
</tbody>
</table>

Frequency: $f$  \hspace{1cm} $2.45 \times 10^9$ Hz

Angular frequency: $\omega$  \hspace{1cm} $2\pi \times 2.45 \times 10^9$ (rad/s)

Definition of the power source

Impedance of empty space:

$$Z_0 = \sqrt{\mu_0 / \varepsilon_0}$$  \hspace{1cm} (3.5)

Impedance of Teflon:

$$Z_t = Z_0 / \sqrt{\varepsilon_t}$$  \hspace{1cm} (3.6)

The source is defined at the low-reflection external boundary of the coaxial cable dielectrics. In Figure 23, the power source boundary is at the far right end of the coaxial cable. With TM waves, there is no variation in the azimuthal direction—the $\phi$ direction.
The magnetic field $H$ is only in the $\varphi$ direction. The electric field $E$ is only in the $r$ and $z$ directions.

$$H_\varphi(r) = \frac{1}{Z_T r} \sqrt{\frac{Z_T P_{in}}{\pi \ln(r_{outer}/r_{inner})}}$$  \hspace{1cm} (3.7)

$$E_\varphi(r) = 0$$  \hspace{1cm} (3.8)

where $P_{in}$ is the total input power, $r_{outer}$ is the outer radius of the coaxial cable dielectrics and $r_{inner}$ is the inner radius of the coaxial cable radius. $\ln$ is the function of the natural logarithm.

**Mesh elements**

It is always better to use fine mesh elements to obtain results with better accuracy. Using too fine mesh elements will increase the computational time without further increasing the computation accuracy. The size of mesh elements should be selected so the solutions converge and provide enough accuracy. It is common to use mesh element size at 1/10 to 1/8 of the wavelength. For frequency at 2.45 GHz, the wavelength in liver tissue is about 18.6 mm. I usually set maximal mesh size in FEMLAB to 2.3 mm, which is slightly smaller than 1/8 of the wavelength. Then FEMLAB will generate mesh sizes no larger than 2.3 mm. Such a mesh element size gives me a solution with enough accuracy within reasonable computation time.
Figure 24: Mesh elements of the dipole antenna EM model. The elements are generated by FEMLAB with maximal element size at 2.3 mm. Elements near the antenna tip are manually refined. There are 7749 nodes and 13645 mesh elements in the model.

Results

After the EM model is solved, I can use the postprocessing features in FEMLAB to analyze the solution by plotting graphs for all field variables.
Figure 25: Demonstration of graphs that are generated from the solution of the dipole antenna EM model. (a) Plot of SAR in dB scale, normalized to the maximal value, (b) Plot of magnitude of \( H_\phi \) in dB scale, (c) Plot of equal level contours of the magnitude of power flow, in dB scale, (d) Plot of arrows to show the direction of power flow, near the slot of the antenna. The arrows show the power flow direction only, not the magnitude.

Besides plotting the field variables, I can also perform boundary and volume integrations for the field variables in order to calculate the values of power being delivered, reflected, deposited and leaked.

Total power deposition is calculated by volume integration of SAR in the whole liver tissue, plus the power leaked from the external boundaries of liver tissue. The power leaked at a boundary is calculated by the boundary integration of the normal vector of the power flow vector on the boundary.

Total power delivery by the power source is calculated by boundary integration of the normal vector of the power flow vector on the power source boundary. The total power delivery is the input power subtracted by the reflected power. Since we have defined the total input power, we can calculate the total reflected power by subtracting the total delivered power from the input power. After knowing the reflected power, we
can calculate the power reflection coefficient $\Gamma$ for the antenna at the simulated frequency.

\[
P_{\text{deposited}} = P_{\text{absorbed}} + P_{\text{leaked}} \tag{3.9}
\]

\[
P_{\text{in}} = P_{\text{delivered}} + P_{\text{reflected}} \tag{3.10}
\]

\[
\Gamma = \frac{P_{\text{in}} - P_{\text{delivered}}}{P_{\text{in}}} \tag{3.11}
\]

Power delivered by the power source should equal the power deposited. These two values usually slightly mismatched due to the limited accuracy of the computation and limited size of mesh elements. Reducing the mesh element sizes will increase the accuracy.

### D.3 Compute the antenna frequency response with the basic EM model

The frequency response of an antenna can be measured by using a vector network analyzer, as mentioned in section C.7. We can calculate the frequency response of a design antenna using the basic EM model. Antenna power reflection coefficient $\Gamma$ can be calculated as in the previous section. The spectrum of $\Gamma$ can be computed at discrete frequencies from 0.5 GHz to 10 GHz. At each discrete frequency, the dielectric properties of bovine liver tissue were adjusted in the model to account for the frequency dependence of the dielectric properties. Liver tissue dielectric properties were computed over the frequency range of interest using equation 2.57 and the parameters given in Table 1 in chapter 2. If the simulated antenna dimensions were revised to match the exact fabricated
dimensions of the antenna, the simulated antenna frequency spectrum will match the measured results very well.

![Antenna reflection in liver tissue](image)

**Figure 26:** Plot of power reflection coefficient versus frequency for a floating sleeve slot antenna, for both measured results and simulated results.

I have taken the advantage of FEMLAB and MATLAB integration in the antenna frequency response computation. I saved the basic EM model of the antenna from FEMLAB into a MATLAB script file. The MATLAB script is basically the sequential MATLAB functions calls into FEMLAB for the whole EM model, including geometry creation, boundary/domain parameter setting, meshing, problem solving and results postprocessing. I modified the MATLAB script into a MATLAB function, which is able to accept frequency-dependent input parameters, compute the EM model at a single frequency and return the calculated the antenna power reflection coefficient at the frequency. I created another MATLAB main program to control the main loop in which I
controlled the frequency at each loop step, calculated the frequency-dependent parameters, and called the previous created function of the EM model by passing the parameters, and accepting the return values. After the whole loop is finished, the main program will generate a finished plot for the antenna power reflection spectrum. The MATLAB main program also allows me to select the frequency lower/upper boundaries, set the frequency steps, and even a special frequency region to be calculated in finer frequency steps. With FEMLAB version 2.3, it takes about 40 min to finish such an antenna frequency response computation with 120 individual frequency steps.

D.4 The basic thermal model

I implemented the basic thermal model by using the heat transfer modeling feature in FEMLAB. The basic thermal model accommodates the basic EM model in many ways. Both models share the same axial-symmetric geometry structures and meshing elements. They are different on the boundary/domain parameter settings, and the principal physical PDE (Partial differential equation) solver. Instead of solving Maxwell’s equations as EM models for static resolutions, the thermal model solves the Bio-heat equation for transient solutions at time steps.

Geometry

The whole geometry of the basic thermal model accommodates the geometry of the basic EM model in Figure 23 (a). I show only the detail plot near the tip of the antenna as Figure 27. A floating sleeve dipole antenna is used in the thermal model instead of the basic dipole antenna in the EM model.
Besides the differences caused by the different antenna, the most important geometric difference with the thermal model is to include the metal parts into the simulation. In the EM model, all metal parts of the antenna are modeled as empty space and ignored by the simulation, and all boundaries on the metal parts are set as PEC (perfect electric conductor). All PECs can be safely ignored in the EM model because EM fields in PECs are simply 0. Such metal conductors must not be ignored by the thermal model because they are also good thermal conductors.

Figure 27: The detail plot of the thermal model, near the slot of the antenna, with a floating sleeve dipole antenna. The antenna slot is at $z = 0$. The dark gray areas are all Teflon, including the dielectrics of the coaxial cable, the antenna slot and the Teflon catheter. The large light gray area above the antenna is the liver tissue. Other light gray areas are all copper, including the solid metal tip of the antenna, the inner and outer conductor and the coaxial cable and the floating sleeve.

**Boundary definitions**

All external boundaries are defined as symmetry/insulation boundaries, including the axial-symmetric boundaries at $r = 0$.

**Domain parameters**

<p>| Liver tissue |   |</p>
<table>
<thead>
<tr>
<th></th>
<th>Density [kg/m³]</th>
<th>Heat capacity [J/kg·K]</th>
<th>Thermal conductivity [W/m·K]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Blood</td>
<td>1060</td>
<td>3600</td>
<td>0.5</td>
</tr>
<tr>
<td>Teflon</td>
<td>1200</td>
<td>1050</td>
<td>0.24</td>
</tr>
<tr>
<td>Copper</td>
<td>8700</td>
<td>385</td>
<td>400</td>
</tr>
</tbody>
</table>

**Definition of the heat source**

I loaded SAR from basic EM model into the thermal model as the heat source by calling MATLAB functions from FEMLAB.

![Diagram of EM Simulation and Thermal Simulation](image)

**Figure 28:** The basic thermal model loads the results from the basic EM model to compute tissue temperature

Referring to the Bio-heat equation, I defined the heat source for liver tissue in the thermal model as:
\[ Q = \text{sar}(r,z) - \rho_{bl} c_{bl} w_{bl} (T - 37.5) \times \text{floor}(50/T) \]  

(3.12)

where \( \rho_{bl} \) is blood density (kg/m\(^3\)), \( c_{bl} \) is the specific heat capacity of blood (J/kg·K), \( w_{bl} \) is blood perfusion (kg/m\(^3\)·s). Values for the three constants are in Table 1. \( T \) is the temperature variable in the thermal model.

The term \( \text{sar}(r,z) \) is a MATLAB function call. The function \( \text{sar}() \) is defined in MATLAB. It takes the matrix parameter \( r, z \) passed from FEMLAB, calculates the corresponding SAR value at each \( r \) and \( z \) pair according to previous calculated results by the basic EM model, and returns the SAR values in matrix form back to FEMLAB.

The term \( \text{floor}(50/T) \) is a MATLAB math function which returns 1 if \( 50/T > 1 \) and returns 0 if \( 1 > 50/T > 0 \). The initial value of \( T \) is 37.5 and value of \( T \) is already not less than 37.5. Such conditions can be expressed as:

\[
50/T > 1 \Rightarrow T < 50 \Rightarrow \text{floor}(50/T) = 1 \tag{3.13}
\]

\[
1 > 50/T > 0 \Rightarrow T > 50 \Rightarrow \text{floor}(50/T) = 0 \tag{3.14}
\]

The term floor \( (50/T) \) will ensure that subtraction from the heat source because of the blood flow only occurs when the tissue temperature is less than 50 °C. This implementation corresponds to the physical phenomenon that tissue is coagulated and blood perfusion is stopped when tissue temperature is higher than 50 °C.

**Time-stepped solutions**

Unlike the EM model, which generates static solutions of the EM fields, the thermal model simulates heat conduction in the liver tissue and antenna system and the
solution has time-dependent temperature distributions because the heating source continuously applies heat energy into the system.

Figure 29: Demonstration of the solution of the thermal model for applying 50 W for 600 s with a slot antenna. Four temperature contours are for 50, 60, 80 and 100 °C.

The limitations of the basic thermal model

The results of the basic thermal model are questionable. The basic thermal model is based on static tissue thermal properties and a static heating source—the static SAR from the basic EM model. Neither temperature dependences nor tissue water dependences of tissue physical properties are considered. The more important issue is the validity of the Bio-heat equation. The basic thermal mode is the FEM solver for the Bio-heat equation. From the result we can see the tissue temperature could reach much higher than 100 °C. The Bio-heat equation is only valid for temperatures below 80 °C. It is invalid for temperatures above 80 °C.
E References


Chapter 4

Tissue water content movement during MWA and the expanded bioheat equation
This chapter introduces the very important hypothesis about tissue water content and thermal energy movement during MWA, and also introduces the expanded bioheat equation, which works into the high temperature range by including tissue water evaporation together with basic thermal conduction.

A Tissue water content and thermal energy movement during MWA

Current MWA research status has been discussed in chapter 1. Current MWA technologies have problems and challenges because of unknown tissue physical changes, lesion size limitation, the backwards detrimental heating problem, unavailable computer simulation and uncontrollable lesion generation. The lack of understanding of the tissue physical changes during a MWA procedure, especially when tissue temperature is high (>80 °C), is one of the fundamental reasons for all other problems and challenges. The problems with lesion size limitation and backward heating are relatively isolated to MWA antenna designs, but it is difficult to design an optimal antenna for MWA to work the best in tissue environments undergoing unknown physical changes. In order to design the best antennas for MWA, we need to have a better understanding of the tissue physical property changes, not just for the liver tissue at normal temperature with static physical properties, but for the entire MWA procedure.

The basic MWA experiments show that the tissue temperature rises over 120 °C during MWA [1]. The tissue lesion slice after MWA shows changes in tissue color and tissue water content. Such tissue physical changes could be quantified and associated. I have performed quantitative measurements of tissue water content and tissue temperature
changes with ex-vivo experiments [1, 2]. The results, in the following chapters, suggest that tissue water content moves during MWA. With the quantitative knowledge obtained through the experiments, I would like to develop a new and more accurate computer model. In addition to including EM for antenna and thermal conduction, the new computer model covers tissue water content movement and associated thermal energy movement. Such a comprehensive computer model, in chapter 8, will eventually help us to design the best optimal antenna for microwave liver tumor ablation and to guide MWA procedures.

A.1 The hypothesis about tissue water and thermal energy movement during MWA

Tissue water movement occurs during MWA when tissue is heated by microwaves to a temperature high enough for tissue water to evaporate. Gas pressure increases along with the evaporation. Water vapor diffuses from high-pressure region to low-pressure region where the tissue temperature is also lower. Water vapor condenses back to water liquid and releases latent heat when reaching the tissue at lower temperature. Released latent heat energy heats the surrounding tissue and increases tissue temperature. Tissue gains water content during the condensation process. Tissue also gains heat energy from the releasing of water vapor latent heat. The entire process of water evaporation, water vapor diffusion and condensation is a process of water movement and energy movement and is as significant as direct thermal conduction. The movement of tissue water is critical for MWA. It directly affects all the tissue physical
properties. It also causes heat energy redistribution because of the uptake and release of water latent heat during evaporation and condensation.

**Figure 1: Demonstration of the entire MWA procedure**

**A.2 Effects and significances**

MWA uses microwaves to heat and destroy the tissue. When the MWA energy is absorbed by the liver tissue, the tissue is heated and its temperature rises. According to the thermal dose theory in chapter 2, tissue is dead with temperature at 50 °C for over 1 min [3]. Complete protein denaturization occurs at a slightly higher temperature. Tissue temperature in MWA could reach temperatures much higher than 120 °C. When the tissue temperature is higher than 80 °C, tissue water starts to vaporize [4, 5]. Conversion from water liquid for to water gas form requires energy—water latent heat—of 2260
kJ/kg. After tissue water starts to vaporize, the rate of tissue temperature increase slows down as more and more absorbed microwave energy is converted to water latent heat instead of being used to raise the tissue temperature. Massive tissue water evaporation occurs at 100 °C under normal air pressure. As more water vapor is generated, the partial pressure of water vapor builds up and water vapor starts to diffuse to the lower pressure portion of the tissue where the tissue temperature is also lower. After the water vapor arrives at the lower pressure and lower temperature part of the tissue, water vapor could condense back to water liquid and give up the latent heat, which in turn is absorbed by the surrounding tissue and heats the tissue. At high temperature, the whole physical process of MWA is a combination of microwave energy absorption, heating conduction, water evaporation, water vapor and water liquid diffusion, and water vapor condensation.

The overall result of the tissue water evaporation, water vapor diffusion and condensation is the movement of tissue water content and the movement of thermal energy. Tissue water and water vapor movement during the MWA procedure increases effective heat conduction, reduces the highest temperature near the microwave antenna, and increases the resultant lesion size.

Tissue water evaporation directly changes the heating mechanism. At higher temperature, microwave energy from the antenna is utilized for both water evaporation and tissue temperature elevation. Our previous studies suggested that the fraction of energy used for water evaporation increases as tissue temperature increases from 90 °C to 100 °C [1]. For temperatures from 100 to 104 °C, almost all microwave energy is converted to latent heat. The movement of tissue water and latent heat energy may become the dominant phenomena instead of direct heating and heat transfer.
The antenna heating pattern itself is affected by tissue water movement. Tissue dielectric properties are directly dependent on tissue water content [6-8]. As tissue loses its water content, both relative permittivity and electric conductivity decrease. Such changes directly result in a change of antenna heating pattern. After tissue loses its water content, it becomes more transparent to microwaves as tissue with lower water content absorbs less microwave energy. Microwaves are attenuated less by tissue close to the antenna slot and are able to propagate further into the tissue. Losing water in the tissue close to the antenna slot causes the intensity of antenna heating to decrease close to the antenna slot and increase away from the antenna slot.

Heat conduction is affected by changes of tissue water content. Tissue density, tissue thermal conductivity and tissue thermal capacity are all directly dependent on tissue water content. All these tissue thermal properties directly affect tissue temperature elevation and heat conduction.

The hypothesis is based on my experiment results, my understanding of the whole thermal process, and previous researchers’ studies on water movement in porous media during heating. The idea about water evaporation, diffusion and condensation has not been pursued much for MWA, but it has been studied in other research areas. Water movement in porous media during heating is a general phenomenon [5, 9-15].

B Expanding the bioheat equation to cover tissue water evaporation

Pennes’ bioheat equation, based on the heat diffusion equation, is a much used approximation for heat transfer in biological tissue [16-18]. Many publications have
shown it is a valuable approximation [19, 20]. However, it does not include a method to account for the evaporation of tissue water, which is expected to occur at tissue temperatures greater than 80 °C.

I have developed a new method to study tissue water related processes together with heat transfer processes for tissue at high temperature. I first map tissue temperature to changes of tissue water content caused by heating and evaporation. With such a mapping from temperature to water content, I define a new term—tissue effective specific heat—and use it instead of the normal tissue specific heat in the bioheat equation. The modified bioheat equation, which can be solved in the same way as the normal bioheat equation, expands coverage of bioheat equation to include tissue water evaporation, and expands the working temperature range to above 80 °C.

**B.1 Theoretical solution of tissue water evaporation with the bioheat equation**

Below is the Pennes bioheat diffusion equation

\[ \rho C \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + Q + Q_B + A \]  

(4.1)

where \( \rho \) is density [kg/m\(^3\)], \( C \) is specific heat [J/kg°C], \( T \) is temperature [°C], \( k \) is thermal conductivity [W/m·°C], \( Q \) is the microwave power density [W/m\(^3\)], \( Q_B \) is a term which accounts for the effects of perfusion (see equation 4.2) [W/m\(^3\)], and \( A \) is the metabolic heat generation term [W/m\(^3\)] which is considered insignificant with respect to the heating term and will be ignored for the purposes of this study.

\[ Q_B = \rho_B C_B \omega_B (T - T_{wb}) \]  

(4.2)
where $\rho_B$ is the blood mass density (kg/m$^3$), $C_B$ is the blood specific heat [J/kg·C], $\omega_B$ is the blood perfusion rate [1/s] and $T_{in}$ is the ambient blood temperature [°C] before entering the heating tissue region.

Note, all variables but $t$ are spatially dependent. For purposes of clarity the spatial dependence is left out of the equations and is implied.

Evaporation requires energy, specifically termed the latent heat. To account for the energy needed to vaporize water we add a term to the bioheat equation, $Q_E$ [W/m$^3$], yielding a modified bioheat equation

$$\rho C \frac{\partial T}{\partial t} = \nabla \cdot k \nabla T + Q + Q_{in} - Q_E$$  \hspace{1cm} (4.3)

Note, here we have dropped the metabolic heat generation term.

The power density used for evaporation is related to the change in water content of tissue as a function of time.

$$Q_E = -\alpha \frac{dW}{dt}$$  \hspace{1cm} (4.4)

where $\alpha$ is the water latent heat constant, which is 2260 [kJ/kg] and $W$ is the tissue water density [kg/m$^3$] which is assumed to be only a function of temperature.

From the chain rule the derivative of $W$ with respect to time is

$$\frac{dW}{dt} = \frac{\partial W}{\partial T} \frac{\partial T}{\partial t}$$  \hspace{1cm} (4.5)

Substituting this into Equation 4.3, yields

$$Q_E = -\alpha \cdot \frac{\partial W}{\partial T} \frac{\partial T}{\partial t}$$  \hspace{1cm} (4.6)
The modified bioheat equation becomes

$$\rho C \frac{\partial T}{\partial t} = \nabla \cdot k \nabla T + Q + Q_w + \alpha \left( \frac{\partial W}{\partial T} \right)$$

(4.7)

Pulling the last term in the above equation to the left hand side,

$$\left( \rho C - \alpha \frac{\partial W}{\partial T} \right) \frac{\partial T}{\partial t} = \nabla \cdot k \nabla T + Q + Q_w$$

(4.8)

Examining the above equation we can define an effective specific heat,

$$C' = C - \frac{\alpha}{\rho} \frac{\partial W}{\partial T} = C - \frac{\alpha}{\rho} W_{r}$$

(4.9)

which yields a new modified Pennes bioheat equation

$$\rho C' \frac{\partial T}{\partial t} = \nabla \cdot k \nabla T + Q + Q_w$$

(4.10)

Equation 4.10 is in the same format as the original bioheat Equation 4.1, with the effective specific heat used instead of normal specific heat. Since $W_{r}'(T)$ is 0 when evaporation doesn’t occur and is negative when evaporation occurs, effective specific heat $C'$ is never less than normal specific heat value $C$.

Tissue effective specific heat $[\text{J/kg-K}]$ is the only new term in equation 4.10. It has a similar physical meaning as the normal specific heat. It is defined for a unit mass of tissue as the amount of energy required to increase temperature by 1 °C, including the water latent heat energy required if tissue water evaporation occurs, which regular specific heat does not account for.
B.2 Discussion about the expanded bioheat equation

The most important change with the new bioheat equation is the new term $C'$. The new bioheat equation needs a tissue temperature to water content mapping function in order to calculate the tissue effective specific heat term $C'$. Unfortunately $C'$ is not constant, but changes with tissue temperature.

I have developed a mapping equation based on my experimental results [1]. The mapping function, as equation 4.11, is plotted in Figure 2. The differential of the mapping function is plotted in Figure 3. With equation 4.11, I have calculated tissue effective specific heat and regular specific heat, plotted in Figure 4. Note that tissue regular specific heat is not constant, either. It is calculated based on tissue water content at the given temperature. The method to calculate it will be introduced in chapter 8.
Figure 2: Plot of mapping from tissue temperature to tissue water content. Tissue water evaporation is assumed to start at 70 °C and completes at 160 °C. Tissue originally contains 73% water content. There is about 10% drop by 100 °C and 35% drop by 104 °C.

\[
W(T) = \begin{cases} 
0.778 - 0.778 \times \exp \left( \frac{T - 106}{3.42} \right) & T \leq 103 \\
0.0289T^3 - 8.924T^2 + 919.6T - 31573 & 103 < T \leq 104 \\
0.778 \times \exp \left( \frac{T - 80}{34.37} \right) & T > 104 
\end{cases} \quad (4.11)
\]
Figure 3: Plot of $W'(T)$ which is continuous on $T$. The peak is at $T = 104$.

Figure 4: Calculated tissue effective specific heat and tissue regular specific heat versus tissue temperature.
I have presented the new expanded bioheat equation, numerical solution and experimental verification in [21]. The new bioheat equation will be used in the comprehensive computer models, which will be covered by chapter 8 and in the paper [22].

C References


Chapter 5

Measurement and Analysis of Tissue Water Content
during MWA
I have observed that tissue water content changes for the slices of thermal lesions. The centers of the lesions seem to have less water than the outer portions of the lesions and the normal tissue surrounding the lesions. By measuring the tissue water content of the thermal lesions after MWA, I quantitatively analyze the movement of the tissue water during the MWA procedures in order to understand the movement of thermal energy and the changes of tissue physical properties caused by the tissue water movement. I think such analysis of tissue water and thermal energy movement, as well as the related tissue physical property changes would help to understand tissue thermal responses quantitatively and eventually help to implement the comprehensive computer model of MWA for lesion prediction and antenna optimization.

Results of this chapter have been presented in [1] which was submitted to IEEE TBME in Nov. 2005.

A Method

A.1 The wet-dry procedure

Liver tissue can be dried completely by using a tissue dryer. The weight difference between the weight before drying and weight after drying is the total weight of lost tissue water. The weight fraction $w$, or the ratio of weight of tissue water before drying to weight of normal tissue, can be calculated as:

$$w = \frac{M_w - M_d}{M_B}$$

(5.1)
where, \( M_w \) is the measured tissue weight before drying, \( M_d \) is the measured dry tissue weight, and \( M_B \) is the weight of normal tissue before ablation. \( M_B \) is calculated as:

\[
M_B = \frac{M_d}{0.27}
\]  

(5.2)

Calculation of \( M_B \) is based on the fact that water mass fraction for normal tissue is about 0.73 and the mass fraction of solid material in normal tissue is 0.27. These numbers were obtained through our previous bulk measurements on larger tissue pieces.

Note that tissue water mass fraction ratio \( w \) is not equal to volume fraction ratio \( w_v \), because mass densities of water and tissue solid material are different. Many research publications were using volume fractions in various equation and other purposes. In fact, these two ratios can be converted as:

\[
w = \frac{w_v}{1.3 - 0.3 \times w_v}
\]  

(5.3)

\[
w_v = \frac{1.3 \times w}{1 + 0.3 \times w}
\]  

(5.4)

These conversion equations are based on the facts that water density is 1000 kg/m\(^3\) and density of tissue solid materials (protein) is 1300 kg/m\(^3\) [2].

**A.2 Basic experiment setup and procedures**

All experiments were performed ex-vivo with bovine liver, which was obtained from a local slaughter house. Liver tissue was refrigerated overnight and the initial temperature at the beginning of the experiments was approximately 6 °C. Initial water content was about 0.73 by weight.
A homogeneous block of liver tissue, free of major blood vessels, was selected. The coaxial antenna was inserted into the tissue and connected to a MW generator through a 1 m long flexible coaxial cable. A lesion was created after 75 W of MW power was applied to the liver tissue for a controlled duration. After the antenna was smoothly drawn out, the liver tissue was cut into 2 pieces along the trace of the antenna insertion. A 5 mm thick slice was cut from one of the two pieces, at a distance 5 mm from the slot, perpendicular to the antenna insertion direction, as shown in Figure 1 (a). A strip of tissue was then cut from the slice and cut into 1 mm thick pieces, as shown in Figure 1 (b). The plane 5 mm above the slot was chosen as this is the position along the axis of the antenna with the largest lesion radius, based upon simulation and experimental observation.

Weights of all tissue pieces were measured with an Ohaus Explorer E11140 digital analytical balance with 0.1 mg resolution. Tissue pieces were vacuum-dried.
overnight with an Edwards ETD4 Tissue Dryer, and weighed the next morning. All measurements were performed as quickly as possible to minimize the effects of evaporation from the surfaces of the tissue.

A.3 Detail procedures and explanations

A.3.1 Tissue preparation

Using refrigerated tissue helped to enforce that all experiments were performed under similar conditions including the initial tissue temperature and initial tissue water content. When a bovine liver was stored in the refrigerator, it was stored in a plastic bag to prevent extensive water content loss due to the cold air circulation in the refrigerator. We also made sure the bovine liver was placed away from the back plate of the refrigerator, which was colder and could freeze the liver tissue if the liver tissue touched it.

Ex-vivo experiments enabled us to ignore the blood perfusion effect in live tissue, which otherwise would introduce more uncontrollable variants and system errors because of the difference in blood vessel size and blood flow rate.

Blocks of tissue were from the most homogeneous part of the thick leaf of the beef liver. The tissue blocks must be free of major blood vessels. There was no blood in the refrigerated beef livers and all major blood vessels were empty. If major vessels were located near the site of MWA lesion to be created, such vessels could be good ducts for water vapor to escape from the ablated region and result in inconsistency of lesion size and shape between experiments.
The tissue block must be large and thick enough. A lesion would be created symmetrically around the inserted antenna and the lesion should be centered in the tissue block with enough distance from the lesion boundary to the tissue block edge. Too small a tissue block could allow the lesion boundary to reach to the block surface and deform the shape of the MWA lesion. Despite the minimal tissue block requirement, it was always safe to use larger tissue blocks. Table 1 gives a list of experimental setup parameters and minimal tissue block sizes.

Table 1: Lesion size estimation

<table>
<thead>
<tr>
<th>Power</th>
<th>Duration (min)</th>
<th>Estimated lesion diameter (cm)</th>
<th>Minimal tissue block size (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>75 W</td>
<td>3</td>
<td>3</td>
<td>6×6×10</td>
</tr>
<tr>
<td>75 W</td>
<td>6</td>
<td>3.5-4</td>
<td>7×7×12</td>
</tr>
<tr>
<td>75 W</td>
<td>12</td>
<td>5</td>
<td>8×8×14</td>
</tr>
</tbody>
</table>

A.3.2 Ablation procedure

The same antenna was used so that the lesions created by different MWA experiments followed the same SAR pattern, similar tissue temperature and tissue water distribution during the MWA procedures.

A.3.3 Handling ablated tissue

Ablated tissue was processed quickly and accurately. Cutting and weighing of the tissue after ablation took approximately 1 to 3 min. During this delay, tissue water may have diffused from a high water content area to low water content area—the highly ablated region. The overall effect would be to artificially increase the tissue water content near the antenna body, and artificially decrease the water content near the lesion.
boundary. The estimated water diffusion constant in normal tissue is approximately $10^{-9}$ to $10^{-10}$ m$^2$/s [3]. Diffusion rates may be even higher in the highly ablated area because the tissue is porous. Nevertheless our data do not suggest that the amount of diffusion was quantitatively significant and initial computer simulations utilizing the base water diffusion constant support this hypothesis.

I used a meat slicer to cut the ablated liver block into about 5 mm thick slices perpendicular to the antenna insertion direction. All slices are laid down on a plastic cutting board, which was free of water and did not absorb water. I examined both sides of each slice to make sure there was no free water on it. If there was some free tissue water left on either side of a slice, I used a soft paper towel to quickly wipe it clean and I also avoided having the paper towel absorb water from the slice extensively. Among all the tissue slices, we select the master slice, which had the largest lesion size. Lesion shape and size appearing on both sides of the master slice was about the same.

![Image](image.png)

**Figure 2:** The lesion was created by using 75 W for 4 min. The black hole in the center was where the antenna was inserted. The lesion was generated symmetrically around the antenna hole.
The thickness of the slices is important. Slices could be cut thinner than 3 mm. But it was difficult to cut slices thinner than 3 mm with a meat slicer. The slices were thin enough so that the master slice had about the same lesion shape and size on both sides. Figure 2 shows a photo of a tissue lesion slice.

In the master slice, the MWA lesion was usually round. The antenna insertion hole was in its center, surrounded by dark charred tissue. The width of the charred tissue depended on the applied power level and the duration. For short durations or very low power levels, there was no charred tissue at all. The tissue color changed from dark, to light dark, gray, light red and red from the slice center outwards. Different colors represent different degrees of tissue ablation. Tissue color changes are caused by protein coagulation and loss of water content. The width of each region depended on the applied power level and duration. Normal tissue outside the lesion appeared red because of remaining blood. It was very easy to distinguish normal tissue from the well ablated tissue, but it was very difficult to determine the exact boundary of the ablated region.

The slice thickness was determined by the setting on the meat slicer and was calibrated previously by measuring the thickness of a stack of slices. When picking the master slice, other slices were also checked. The first slice with the antenna hole was marked as the reference of the antenna starting point. The distance from the master slice to the antenna tip was determined by the number of slices between the master slice and the first slice with the antenna hole. Since the master slice corresponded to the slice with maximum SAR in the antenna longitudinal direction and the position of the maximum SAR didn’t change from procedure to procedure, the position of the master slice was about the same relative to the antenna tip.
We use a scalpel to cut a strip of tissue from the master slice from the antenna hole in the center of the slice to the slice edge. The strip width was about 3 mm. If the surface of the master slice appeared very homogeneous as it should be, it did not matter which direction to cut the strip, otherwise, we selected a direction in which we obtained a strip with the most homogeneous appearance. Both tissue slice and the strip were scanned with HP scanner at resolution of 600 DPI so that we could measure lesion size later using the scanned images.

The tissue strip was then laid down on the transparent bottom of a plastic examining dish. The dish was 100 mm in diameter and 15 mm deep. The dish was put on the top of a ruler (or a sheet of paper with ruler lines). The strip was cut into 1 mm width pieces with a scalpel according to the ruler.

The master slice was selected such that lesion sizes on both sides of the slice were the same. The degree of ablation was uniform from one side to the other. The tissue strip was only 3 mm wide, and tissue pieces were cut 1 mm wide, so it was reasonable to assume the degree of tissue ablation and tissue desiccation was uniform for each tissue piece.

A.3.4 Special attention

Some special attention was taken to handle the tissue slice, strip and pieces. First, water loss and water gain was avoided. We used a dry plastic surface to place the tissue slices, the strip and pieces. Dry plastic surfaces ensure that no external water is introduced and no tissue water is absorbed. It is unavoidable that tissue pieces touch the plastic disk surface and some tissue water is left on the plastic surface. Nevertheless such
problems only happen with high water content tissue pieces in which tissue water is saturated. The low water content pieces are quite dry. They won’t lose water onto the dish surface. The tissue loss on the wet tissue pieces will be analyzed in later sections.

Second, water loss by exposing to the dry air was avoided. After the tissue strip was cut into pieces, the pieces were kept in the covered plastic dish to avoid exposure to dry air.

Third, tissue pieces were cut precisely. It was very necessary to use a very sharp scalpel (with a new blade) in order to precisely cut the tissue pieces from the strip. The dark to light gray part of the strip was more desiccated than the other part of the strip, which was light red to normal red. The desiccated part of the strip was hard. It was easy to cut with good precision. The soft part of the strip deformed easily when pressed during cutting with a dull scalpel. Pressing the tissue strip could also cause tissue water draining out onto the dish surface.

A.3.5 Weighing

We used a 0.1 mg accuracy digital analytical balance to measure the weight of each tissue piece. It was necessary to use a high accuracy balance. Tissue pieces usually weigh from a few milligrams to 10s of milligrams. The 0.1 mg accuracy measurement error is acceptable.

Weights of each tissue piece were recorded for later data processing. Weight measurement was taken quickly in order to avoid water exchange between tissue pieces and the air.
A.3.6 Drying

I used an Edwards Tissue Dryer ETD4 to dry the tissue pieces. The tissue dryer was a vacuum dryer with cooling capability. It was able to cool the tissue to -40 °C. It usually took overnight for the tissue pieces to be completely dried.

Since the drying is very slow, we dried more than one group of tissue pieces in a single drying cycle. After the tissue pieces had been weighed, it was fine to let them sit aside and wait for another group of tissue pieces to be generated, which needed to go through the ablation, slicing, cutting and weighing steps.

A.4 Results

We performed lesion water content measurements for MWA durations at 1, 2, 3, 4, 5, 6, 9 and 12 minu. Figure 3 shows the results. Water weight fraction was calculated for each tissue piece. We performed at least 4 measurements for each duration. Mean results and standard deviations were calculated from the results from multiple measurements for the same MWA duration.

A.5 Discussion and conclusions

We used the ANCOVA (Analysis of Covariance) statistical model to analyze the measured data and show statistical significance. The results show that tissue water content changed significantly ($p < 0.001$) with ablation duration by controlling the tissue piece position. Tissue water content also changed significantly ($p < 0.001$) with tissue piece position by controlling the ablation duration. The statistical results are consistent with the plots.
Figure 3: Plots of postablation water content mass fraction $w$, calculated according to equation 5.1. Mean results from multiple measurements for the same duration are plotted as solid or dashed lines. The vertical bars represent the standard errors at the corresponding data points. The data points are plotted at the position where the center of each tissue piece is located on the lesion strip. Position value is the distance from the edge of the antenna insertion hole to the center of the tissue piece. 0 is at the edge or the surface of the antenna when the antenna is in the tissue.

The results suggest that microwave ablation has several effects on tissue water properties. First, tissue is desiccated to a greater extent near the antenna. This is to be expected both from an observation of the lesion and from a basic understanding of the SAR pattern (i.e. where energy is deposited) and heat transfer. Second, even after only 1 min there is significant water loss in the tissue up to 3 mm away from the edge of the antenna. This illustrates the volume heating effect we expect from MWA. Microwaves emitted from the antenna propagate through the tissue, resistively heating the tissue as the
wave attenuates. This leads to a direct deposition of energy in a larger volume of tissue than seen during RF ablation. Third, the change in water content seen early and over a significant volume can be expected to affect both the thermal and electromagnetic properties. This may affect them enough to significantly alter the SAR pattern and/or the dynamics of heat flow that occurs in conjunction with the direct deposition of energy due to the microwaves. Finally, water content drops to near 0 for tissue very close to the antenna for durations of greater than 6 min. This indicates charring of the tissue near the antenna.

**A.6 Discussion about the experimental design, restrictions and system errors**

Under the restrictions of available lab experiments, there are some limitations of our experimental design and such limitations result in unavoidable measurement and system errors.

We can obtain only limited spatial resolution with this experimental design. The water content measurement is only performed on a strip of tissue from the slice of ablated tissue in which the ablated region is the largest. This measurement gives only very limited measurements on less than 20 individual spots in the whole ablated tissue. In fact, the ablated tissue is not just a slice, but a 3D volume. Nevertheless, we use the measurement from the master slice to judge our hypothesis regarding the movement during MWA and the limited measurement results work fine for the purpose.

Tissue slices, tissue strips and tissue pieces are placed on a dry surface of plastic dishes. Different regions of the tissue slices and tissue strips have different water
contents. Normal liver tissue has a water content near 0.73. When the tissue pieces are on the dry surface, some tissue water drains onto the surface. This isn’t a big problem for desiccated tissue, but it is a bigger problem for wet and normal tissue, especially when such tissue is cut into very small pieces. This is the reason why the measured water content for wet and normal tissue differs from the normal value of 0.73. Using large tissue pieces can reduce possible measurement errors.

Each tissue piece is still too large. The water fraction is not completely uniform in each tissue piece, and the water fraction measurement result is actually the average water fraction of the whole piece even if water in the piece is not uniformly distributed.

Regarding measurement errors: Results of the water content measurements above 0.5 mass fraction are not presented here due to water dripping from the cut tissue. During the cutting stage of the experiments we noticed a significant loss of water from the higher water content tissues pieces. This led to a discrepancy between the measured values of the high water content (normal) tissue outside the lesion and our bulk water content measurements taken for calibration purposes. Water loss was only observed in tissue pieces with higher than 0.5 water content, therefore measurements on such tissue pieces were unreliable and are not presented in Figure 3. For tissue pieces of lower water content, below 0.5, no visible water loss was detected, nor was expected, so the measurement results are considered acceptable. For the purposes of understanding water-related phenomena near the antenna these are clearly the most significant.
B References


Chapter 6

Measurement and analysis of tissue temperature during MWA
Tissue temperature measurement during a MWA procedure is very desirable. The measurement results can directly validate the results from computer simulations. Evaluation of the results from a few accurately known positions in the tissue can help to quantitatively reveal tissue physical responses to the MWA heating. Nevertheless, tissue temperature measurement during MWA seems to be trivial, but it is not so easy to carry out, especially when the exact measurement positions need to be controlled.

I measured tissue temperature changes during ex-vivo microwave ablation (MWA) procedures for bovine liver tissue. Tissue temperature started to increase rapidly at the beginning of the MW power application. It came to a plateau after 100 to 104 ºC before it increased again. I divided the changes of tissue temperature versus time into 4 phases. They suggested that tissue temperature changes may be directly related to tissue water related phenomena during MWA, including evaporation, diffusion, condensation and tissue water composition. Changes of tissue temperature versus position in the lesion showed a similar plateau at 100 to 104 ºC. I also studied the relationship of tissue water evaporation and tissue temperature by mapping temperature to remaining tissue water after ablation. Other analysis indicated that lesion color was closely related to tissue temperature and remaining tissue water content, and lesion boundary was at about 60 ºC temperature and at where tissue water content just started to evaporate.

Results of this chapter are also presented in [1] which is going to be submitted to IEEE TBME for publication in 2005.
A Method

I used a Luxtron Fluoroptic thermometer model 3100 to measure tissue temperature during MWA of ex-vivo bovine liver tissue. A fluoroptic thermometer was selected because its fiber-optic temperature sensors are unaffected by microwave radiation and have minimal affects on the SAR pattern of the antenna.

All experiments were performed ex-vivo with bovine liver, which was obtained from a local slaughter house. Liver tissue was refrigerated overnight and the initial temperature at the beginning of the experiments was approximately 6 to 9 °C. Homogeneous blocks of bovine liver were used for the experiment. They were large and thick enough, at least 8 cm long and wide, at least 5 cm thick, to accommodate the lesion for a 6 min microwave heating.

Figure 1 shows placement of the microwave coaxial antenna and fiber-optic temperature sensors. In addition to the antenna, fiber-optic temperature sensors were inserted into the liver tissue guided by 14 gauge (2.1 mm in diameter) biopsy needles through holes in a 1 cm thick plastic template. This ensured that the antenna and the needles were placed parallel to each other in the tissue. The antenna was inserted 7 cm deep into the tissue, referenced to the antenna tip. The needles were 15 cm long, with a needle guide and a 15 gauge (1.83 mm in diameter) stylet. The needle guide and the stylets were pushed together through the liver block and the stylets were pulled out. Temperature sensors, each paired with a section of thin plastic fiber (0.5 mm), 12 mm long, were introduced into the needle guides. The temperature sensor fibers were inserted to a position 5 mm proximal to the antenna slot. This is the axial location of maximum
power deposition and lesion radius, see Figure 2 and Figure 6 (a). The plastic fibers were inserted 2 cm beyond the temperature sensor fibers. The needle guides were then withdrawn from the tissue on the template side while the temperature sensors and the plastic fibers were left in the liver tissue.

**Figure 1:** The fiber-optic temperature sensors are placed 5 mm away from the antenna slot in the longitudinal direction, and various distances away from the antenna body in the radial direction.

The temperature sensors were positioned at various distances away from the surface of the antenna body. Each was connected to a Luxtron 3100 thermometer, which was connected to a PC via its serial port for data collection. Temperature values from all sensors were updated by the thermometer 4 times a second and were recorded by the PC for the entire ablation procedure. After ablation, the temperature sensors and the antenna were removed, leaving the plastic fiber sections behind to mark the temperature sensor locations. Tissue was then cut open with scissors, perpendicular to the antenna insertion.
direction, along the marked line in Figure 1. The tissue was cut so the plastic fibers remained together with the tissue. The distances from each fiber position to the near edge of antenna hole were measured on the cross section. Temperature sensors were assumed to be at the same positions as the plastic fibers.

![Figure 2: Antenna SAR plotted in dB scale from EM computer simulation. I use a coaxial-based sleeve antenna, to be introduced in [2] or chapter 7, which creates a localized SAR pattern. The longitudinal position of the antenna slot is 10. The longitudinal position of the maximum SAR is near 17 mm.](image)

**B Results**

Figure 3 shows some measurement results. The shape of the curves in Figure 3 shows that the temperature curve can be separated into four regions, labeled 1, 2, 3, 4 on the graph. What separates the regions are the temperature ranges and shape of the curve. Region 1 of the temperature curve is slightly concave, starting at the initial temperature and ending between 70 to 90 ºC. Region 2 is convex and appears to be a transition region between regions 1 and 3. In region 3 the temperature curve becomes almost flat and slowly increases from 100 ºC to 104 ºC. After the temperature reaches 104 ºC, the temperature increases quickly again in region 4. Most curves showed these different
regions although the slopes changed as distance from the antenna increased and the transition region 2 was less sharp.

Figure 3: Plots of measured temperature versus time. The MWA procedure was performed with 75 W for 360 s. Sensors were 5 mm above the antenna slot longitudinally. Temperature curves are labeled by the distance of the temperature sensors to the antenna surface, measured after heating.

Figure 4: Plot of measured tissue temperature versus the measurement positions for different ablation durations. Data points are derived from the same measurement results used in Figure 1.
Figure 4 shows measured temperature data as a function of position for various times during ablation. The interesting feature here is the flat spatial region, corresponding to the 100 to 104 °C plateau seen in the time domain plots. Tissue temperature is virtually flat at 100 to 104 °C in the region of 4 mm to 6 mm on the 3 to 6 min curves.

In previous experiments [3] I have measured tissue water content versus ablation duration. Figure 5 shows a plot of water content and temperature as a function of time for a number of positions to examine correlations. Values of tissue water content are the ratios of the remaining tissue water mass after ablation to the normal tissue mass before ablation. According to bulk measurements, normal tissue had about 0.73 water content. Figure 5 shows that for all locations as tissue temperature increases to 100 °C, water content drops to 0.45 to 0.50. For locations near the antenna (2, 3, and 3.5 mm), as the temperature reaches 104 °C, water content drops to about 0.30 to 0.35. For locations 4 and 5 mm, although the temperature does not reach 104 °C, water content at those locations eventually drops to 0.35.
Figure 5: Plot of tissue water fraction and measured temperature versus ablation duration. Curves are labeled by the distance of the temperature sensors to the antenna surface. Curves without markers correspond to plots of tissue temperature. Curves with markers are plots of tissue water content with the markers being data points.

Figure 6: Photos of ablated bovine liver tissue. (a) Longitudinal section, the maximum lesion radius is located at about 5 mm from the slot along the longitudinal direction. (b) Cross section. Both lesions were created separately by applying 75 W for 360 s. The lesion sizes were smaller than reported for
this antenna due to a low initial tissue temperature of 8 ºC. Lesion size is approximately 26 to 28 mm in diameter.

Figure 6 shows a scanned photo of ablated tissue. The lesion is identified as the region in which tissue color changes from dark red to light red. The antenna hole is approximately 3.5 mm in diameter. Based upon 9 experiments, the length from lesion boundary to the edge of antenna hole had a mean value of 14.5 mm and standard deviation of 0.9 mm. The mean lesion diameter was 32.5 mm. Figure 4 shows that tissue temperature was 60 ºC at 13.5 mm after 6 min ablation. This suggests that ex-vivo liver lesion boundary has temperature of 55 to 60 ºC which is close to 50 ºC, the previously published lesion boundary temperature for radiofrequency ablation (RFA) [4].

C Discussion

C.1 Differences between regions

Tissue temperature and its changes are closely related to the associated physical phenomena occurring during the heating. Tissue is heated by different heating mechanisms, including microwave power heating, thermal conduction from adjacent tissue and energy conversion related to water evaporation and water vapor condensation. Microwave power heating is of course the original energy source of all heating mechanisms.

In phase 1, the tissue temperature is relatively low. I think that tissue water evaporation is not significant. The concave curve shape indicates acceleration of tissue temperature and increasing heating strength. There could be a few reasons for such
acceleration. One reason is the increase of microwave power density in the tissue because tissue closer to the antenna was desiccated. Desiccated tissue is less attenuating to microwave propagation because tissue electric conductivity decreases as tissue water content decreases [5]. A second reason is that water vapor diffusing from the high temperature area condensed in the lower temperature area and the released latent heat energy heats the surrounding tissue. A third reason is the increase of heat conduction flux from the high temperature area because of the increase of temperature gradient. These possible reasons could combine to cause the acceleration of tissue temperature. The importance of different reasons could be different from tissue closer to the antenna to tissue away from the antenna. I think this was the reason why the ending temperatures of phase 1 were different from curve to curve. It appeared that the ending temperature was higher if tissue was closer to the antenna where the increase of microwave power density could be more significant. For curves measured over 4 mm away, the ending temperatures were at about 80 ºC. This was the temperature where tissue water evaporation was expected to start [6].

Phase 2 is from the end of phase 1 to 100 ºC. The convex curve shape indicates deceleration of temperature increase. I think the most important reason is tissue water evaporation. Evaporation starts at temperature about 80 ºC and becomes more significant after 90 ºC. When tissue water evaporates, absorbed microwave energy is partially used for water vapor latent heat. Tissue temperature still rises because the remaining part of the energy is still heating the tissue. The closer tissue temperature is to 100 ºC, the larger part of the energy is used for latent heat. Tissue heating mechanisms are more
complicated in this phase. There are not only microwave power heating and heat conduction, but also energy loss because of tissue water evaporation.

Phase 2 on the curves measured close to the antenna body had sharper deviations than curves measured farther away from the antenna. If the measuring spot was further than 4 mm, phase 2 became very smooth. It was very flat if measured over 10 mm away, or may not be distinctive. I think the deceleration of tissue temperature was dominated by tissue water evaporation, which was closely dependent on tissue temperature. Thus the smooth phase 2 curves may very well reflect the relationship between tissue water evaporation and tissue temperature. Tissue temperature changes slowly at positions further from the antenna body, so phase 2 of the curves were very flat.

In phase 3, tissue temperature almost stopped increasing after the temperature reached 100 ºC. Under normal air pressure, water boils at 100 ºC where the water vapor partial pressure is equal to the atmospheric air pressure. When such massive water evaporation occurs, tissue temperature won’t rise until all free tissue water has vaporized. Absorbed microwave energy is converted almost completely to water vapor latent heat. A small amount of the energy is used to maintain the tissue temperature because there could be conductive heat loss. The plateau is not a pure flat line. The tissue temperature actually increases very slowly, from 100 ºC to around 104 ºC. I think the slight increase of the temperature is caused by the build up of water vapor pressure. Water vapor, limited by the fluid-filled tissue environment, could not escape freely, thus gas pressure would build up. If the gas pressure is higher than the atmospheric pressure, the water boiling temperature becomes higher than 100 ºC. The gas pressure of 876 mmHg is needed for the water boiling temperature increase from 100 ºC to 104 ºC [7].
Phase 4 only exists if the temperature can get higher than 104 °C when measurements are taken close enough to the antenna and the ablation duration is long enough. The tissue temperature starts to increase again constantly from 104 °C. I could not measure the maximum temperature that the tissue could reach because of the measurement limit of the Luxtron 3100 thermometer, but the temperature increases over 120 °C and continues to rise. In fact, liver tissue near the antenna is completely charred when the ablation duration is long enough. The charred region gets larger and larger if the microwave power is continuously applied. Temperature was estimated to be as high as 300 to 500 °C for tissue to char, so we expect that maximal tissue temperature could reach as high as 300 °C.

One important fact was that tissue was not completely desiccated after 104 °C, the end of phase 3. One apparent reason is that tissue would not be able to effectively absorb microwave energy if tissue completely loses its water content, because water molecule vibration under microwave radiation is the main reason of microwave heating. Because tissue temperature continuously increased after 104 °C in phase 4, one can see that tissue was still efficiently absorbing microwave energy and still retaining enough of its water content.

C.2 Tissue temperature versus tissue water content

I expect tissue water content to play a significant role in MWA. To include the effects of water evaporation and movement into a thermal model of MWA I have come up with a function that correlates tissue temperature to water content based upon experiments measuring temperature and water content during ablation [3]. Figure 7
shows the proposed functional relationship between tissue temperature and tissue water evaporation. Ramachandran measured that half of the tissue water content was lost at temperature of 104 °C [6]. The approximation, as equation 6.1, was based on the previous analysis of tissue water content decrease at 100 °C and 104 °C, as well as the expectations that evaporation starts near 80 °C [8].

$$W(T) = \begin{cases} 
0.778 - 0.778 \exp \left( \frac{T - 106}{3.42} \right) & T \leq 103 \\
0.0289 T^3 - 8.924 T^2 + 919.6 \times T - 31573 & 103 < T \leq 104 \\
0.778 \exp \left( \frac{T - 80}{34.37} \right) & T > 104
\end{cases} \quad (6.1)$$

Figure 7: Mapping tissue temperature to remaining mass of tissue water after ablation. Tissue water and tissue temperature from matched ablation durations and positions in lesion are plotted as thin solid lines with asterisks. Only data points lower than 0.5 were used to create the approximation line.
because data points higher than 0.5 were not reliable [3]. The dotted line is the approximation.

Normal tissue has about 0.73 water content in mass.

C.3 Free water versus bounded water

Tissue water exists in two forms—free water and bounded water. Bounded water refers to water molecules combined with other chemical molecules. Liver tissue contains 0.70 to 0.75 water in mass fraction. Our bulk experiments showed it was about 0.73. Some publications have suggested that free water and bounded water each account for half of the total water content [6]. Unlike free water, bounded water needs extra kinetic energy to break the chemical bonds to get free, so bounded water vaporizes at a higher temperature than free water.

Figure 7 show that tissue water content dropped to 0.32 at the end of the plateau at 104 °C. We speculate that free tissue water has all been vaporized and bounded water starts to evaporate after the plateau. With such a speculation, our results suggest that free water and bounded water account for 0.56 and 0.44 respectively of the tissue water. Such a ratio seems to match very well with expectations.

D Conclusion

In this chapter, I have presented results of tissue temperature measurement during MWA. Analyzed together with tissue water content measurement data, our results have indicated important relationships among tissue temperature, tissue water content and thermal lesion. Even though the measurements were performed under particular experimental conditions and the measurement data are not directly applicable for other applications, the results improved the fundamental knowledge about tissue thermal
responses during MWA. A few very important general conclusions have been obtained and will be useful to guide and help future studies of MWA, including antenna design and computer thermal simulations.

E References


Chapter 7

Design, verification and theoretical analysis of the floating sleeve coaxial antennas
I have discussed the problems and challenges of current MWA technologies in chapter 1. The problems with the lesion size limitation and detrimental backward heating are both directly related to inefficient design of microwave antennas. It is very necessary to design a new antenna in order to have improved performance over the current antenna designs.

I have designed a novel coaxial antenna—the floating sleeve antenna. This new antenna uses a floating sleeve, that is, a metal conductor electrically isolated from the outer connector of the antenna coaxial body, to achieve a highly localized SAR pattern that is independent of insertion depth. This floating sleeve coaxial dipole antenna has low power reflection in the 2.4 GHz IMS band. Ex-vivo experiments confirm our numeric simulation results.

I have presented the design of the floating sleeve antenna in [1]. This paper was accepted by IEEE TBME. The paper is currently in press and will be published in 2006.

A The design of a floating sleeve antenna

Our goal was to design a coaxial antenna with a highly localized SAR pattern and low reflectivity for higher power transmission. Figure 1 shows the design of the floating sleeve antenna. The antenna is based on a 50 Ω UT-085 semirigid copper-Teflon coaxial cable. A standard coaxial dipole antenna was constructed from the coaxial cable and wrapped with a thin layer of Teflon tape. The metal sleeve, a section of copper tube (3.2 mm outer diameter, 2.5 mm inner diameter), was slid onto the Teflon-coated coaxial dipole antenna and positioned behind the antenna slot. The whole antenna assembly was then tightly wrapped with Teflon tape. The Teflon tape was heated during wrapping in
order to prevent air from being trapped in the tape layers. The longitudinal dimensions of each section of the antenna along with the overall diameter are in Figure 1 (a), while the interior diameters in the region of the sleeve are in Figure 1 (b).

![Diagram of the floating sleeve antenna](image)

Figure 1: (a) schematic of the floating sleeve antenna and (b) cross section of the antenna at the sleeve.

The floating sleeve antenna differs from existing laboratory and clinical devices (such as the cap-choke antenna) in that the sleeve is electrically isolated from the outer conductor of the coaxial feedline. This floating sleeve is similar to the open sleeve antenna [2] which also uses a floating sleeve. However the floating sleeve of the open sleeve antenna in [2] is quite long. We have determined the length of the sleeve is critical. If the sleeve is too long or too short the response of the antenna is suboptimal, and the antenna is less effective than previously reported antenna designs.

Design of the floating sleeve was accomplished using computer simulations once a qualitative understanding of the importance and effect of the following parameters was determined.
We have determined that the SAR pattern is affected by both the length of the sleeve and the thickness of the Teflon layer. If the sleeve is not covered by a Teflon layer, it needs to be approximately half of the effective wavelength in liver tissue. If the sleeve is covered by a Teflon layer, the sleeve needs to be longer, with the length depending on the thickness of the Teflon layer. This length is critical, and if significantly longer or shorter than the ideal length, the sleeve can be less effective than other reported antenna designs. As long as the sleeve is half a wavelength (adjusted for the presence of Teflon) in length and is not covering the antenna slot, it best constrains the tail of the SAR pattern. In fact the edge of the SAR pattern seems tied to the termination of the sleeve, to the extent that by sliding the floating sleeve, we can control and change the SAR pattern from a spherical shape to an elliptical shape. This may make it possible to control the shape of the lesion to fit different tumor shapes. The thickness of the Teflon isolation layer (the Teflon layer between the floating sleeve and the outer conductor of the coaxial cable) does not seem to affect the SAR pattern as long as it is at least 0.1 mm.

In this design, we used a 2 mm wide slot which was easily fabricated, and also gave good power reflection. The length of the antenna tip also slightly affected the power reflection and shape of the SAR pattern. The length in the design was adjusted to give a good trade off between the power reflection and a spherical SAR pattern.

The antenna design is based on commercially available coaxial cable and utilizes readily available and inexpensive construction materials. This easily fabricated antenna is suitable for open or laproscopic operative therapies. However, the diameter of 3.5 mm precludes this antenna from being useful for percutaneous therapies.
The floating sleeve antenna addresses many of the critical problems with current MWA antennas. It has the following important advantages over most current antennas. It has a localized SAR pattern and minimizes backward heating. It also allows higher power output and creates a large lesion.

B Computer simulation

I used computational electromagnetics simulations to compute the SAR distribution and input reflection coefficient, or $S_{11}$, as a function of frequency for the proposed antenna design. Simulated $S_{11}$ was compared to experimentally measured $S_{11}$ to validate the model, and the sizes and shapes of lesions were evaluated after ex-vivo ablation.

B.1 The computational electromagnetics (CEM) model

Simulations of the floating sleeve antenna were performed using the electromagnetic modeling capabilities of FEMLAB™ version 2.3. Simulations were carried out using an axially symmetric model which minimized the computation time while maintaining good resolution and the full 3D nature of the fields. The 2D axi-symmetric model requires 180 MB memory and 50 s of CPU time for each FEMLAB simulation on an Intel P4 2.8 GHz desktop computer. Figure 2 shows the structure of the floating sleeve antenna near the tip in our CEM model. The model assumes that the floating sleeve antenna is immersed in homogeneous bovine liver tissue. The horizontal $z$ axis is oriented along the longitudinal axis of the antenna and the vertical $r$ axis is oriented along the radial direction. The liver is assumed to be infinite in extent, which is
accomplished using low-reflection boundary conditions provided in FEMLAB™. The computational domain corresponds to a physical domain size of 60 mm in radius and 110 mm ($z = -30$ mm to $z = 80$ mm) in length.

![Diagram](image.png)

**Figure 2**: The axisymmetric CEM model in the vicinity of the tip of the floating sleeve antenna. The vertical axis ($r$ axis) corresponds to the radial direction while the horizontal axis ($z$ axis) corresponds to the longitudinal axis of the antenna. The aspect ratio used in this diagram is nonphysical in order to show the details in the radial direction.

The dielectric insulator of the coaxial cable, the antenna slot and the outer coating materials are all Teflon with relative permittivity equal to 2.1. The dielectric constant and conductivity of liver tissue at $37^\circ$C are $\varepsilon_r = 43.03$ and $\sigma = 1.69$ S/m, respectively, at 2.45 GHz. These values were computed according to equation 2.57 and parameters in Table 1 in chapter 2. The effective wavelength in liver tissue is approximately 18.5 mm.

The SAR [W/kg] in tissue is calculated as a function of position as follows:

$$\text{SAR} = \frac{\sigma |E|^2}{2\rho}$$

(7.1)

where $\sigma$ is the tissue conductivity [S/m] at the excitation frequency, $\rho$ is the tissue density [kg/m$^3$] and $E$ is the spatially dependent time-harmonic electric field vector [V/m].
Figure 3: Plot of normalized SAR on a dB scale. The SAR values are normalized to the maximum SAR value in the simulation region. For reference, the probe tip is at 0 mm, the slot is centered at 12 mm, and the sleeve begins at 22 mm and extends to 41 mm. The region of the simulated liver tissue is from –23 to 83 mm horizontally and 0 to 60 mm vertically. The boundary of the region of the liver tissue is not shown in the figure so SAR pattern near the antenna slot can be shown in better details. The antenna is inserted 70 mm deep into the liver, from the center of the antenna slot at \( z = 23 \) mm to liver tissue right side boundary at \( z = 83 \) mm.

Figure 3 shows the SAR pattern, normalized to the maximum value in the simulation region, in dB. Figure 4 shows normalized SAR values along the longitudinal direction at a number of different radial positions. Figure 3 and Figure 4 indicate that the antenna SAR pattern is completely constrained by the sleeve and is localized in the region from the antenna tip to the end of the floating sleeve. Figure 5 shows the SAR pattern with less antenna insertion depth. It shows that the localization of the SAR pattern is independent of the insertion depth as long as the sleeve is completely immersed in the liver.
Figure 4: Plot of normalized SAR on linear scale as a function of $z$, at constant values of $r = 2.5$, 5.0, 10.0, and 20.0 mm. The SAR values are normalized to the maximum value in the plot.

Figure 5: Plot of normalized SAR on a dB scale. Comparing to Figure 3, the antenna is inserted 40 mm deep into the liver, from the center of the antenna slot at $z = 43$ mm to liver tissue right side boundary at $z = 83$ mm. Boundaries of simulated liver tissue region are marked in dashed lines.

**B.2 Frequency sweep for antenna power reflection**

To validate our model as well as examine the power reflection characteristics of this antenna design, an antenna was fabricated and its reflection coefficient ($S_{11}$) spectrum was measured from 0.5 to 10 GHz using a vector network analyzer (Agilent E8364A). For this measurement, the open end of the antenna was immersed in fresh
bovine liver tissue. The simulated antenna dimensions were revised to match the exact fabricated dimensions and the $S_{11}$ spectrum was computed at discrete frequencies from 0.5 GHz to 10 GHz. At each discrete frequency, the dielectric properties of bovine liver tissue were adjusted in the model to account for the frequency dependence of the dielectric properties. Liver tissue dielectric properties were computed over the frequency range of interest according to equation 2.57.

![Figure 6: Input reflection coefficient ($S_{11}$) for the floating sleeve antenna versus frequency.](image)

Figure 6 shows both the measured and computed results. Note that the computed and measured results agree quite well. Figure 6 shows that the antenna’s minimum reflection is near 2 GHz, off from the desired frequency of 2.45 GHz. We expected this, as the antenna was not designed to minimize the reflected power, but to obtain a good SAR pattern while maintaining a reasonably low reflection coefficient. The measured $S_{11}$ is $-17.1$ dB and the simulated $S_{11}$ is $-18$ dB at 2.45 GHz and we believe these are acceptably low for this initial design. Further optimization of the antenna could reduce
this reflection further and permit tuning the null to 2.45 GHz. We note, however, that a small degree of detuning is expected in practice as the dielectric properties change during ablation.

C Experimental validation

Ultimately it is the coagulated region produced by an antenna that determines its effectiveness. To test the new design we performed ex-vivo ablations using the new antenna. The floating sleeve coaxial antenna was connected to a CoberMuegge MG0300D 300 W, 2.45 GHz microwave generator through a 1 m long flexible coaxial cable. It was then carefully inserted into peripheral regions of fresh bovine liver to avoid heating near the largest blood vessels. MW generator output power level and antenna power reflection levels were carefully monitored and recorded. The liver was then heated using 120 W of power for 150 s. Initial liver temperature was 37 °C. Note 120 W, while at the upper limit, is within the power handling capabilities of the UT-085 coaxial cable used.

After each experiment the liver tissues were sliced into either longitudinal cross sections or transverse cross sections. For longitudinal slicing, a probe was placed into the track created by the antenna and a longitudinal transection of the lesion was made close to the inserted probe. Tissue slices were photographed using a ruler for reference and scanned using an HP ScanJet 3970 scanner at 200 dpi or higher resolution. Figure 7 shows one of the ex-vivo experiment results. Here the lesion size is approximately 5.6 × 3.7 cm, when measured to the periphery of the “white zone”. This is a slightly different shape than predicted by the SAR pattern. However, the procedure of the thermal lesion
formation is a complex combination of microwave energy absorption, heating conduction as well as possible tissue water evaporation, condensation and movement during the ablation period. Therefore the SAR pattern is only expected to be a guide to the final lesion shape. The lesion does show a very well constrained tail as predicted by the SAR pattern. 16 lesions were created using 120 W of power for 150 s, to examine repeatability, mean and std deviation of the lesions were $5.87 \pm 0.32$ cm by $3.64 \pm 0.33$ cm.

![Figure 7: A photo of ablated bovine liver tissue. The lesion was created by applying 120 W for 150 s. The initial temperature of liver tissue was about 37 °C. The spacing between the markers on the antenna body was 1 cm. The lesion was about 5.6 × 3.7 cm and clearly localized to the active radiation region of the antenna. The lesion size was comparable to large lesion size reported by Strickland [21].](image)
D  The theoretical analysis

D.1 The theory of two field interference

Besides the original floating sleeve antenna design as shown in Figure 1, we have an alternative design of the floating sleeve antenna, shown in Figure 8. There are two major differences between this design and the original design. In the original design, the sleeve is covered by a Teflon layer of optimized thickness. There is no Teflon layer covering the sleeve in this alternative design. The sleeve length is much shorter in the alternative design. It is easier for us to explain our theory with this new design.

Even though there are many things to be considered to design the floating sleeve antenna, the key is that the sleeve is electrically isolated from the outer conductor of the coaxial cable. Electrical isolation of the sleeve differentiates the sleeve antenna from the cap-choke antenna and the simple dipole antenna.

Because of the isolation of the sleeve, the EM wave propagates in two ways into the liver tissue. After the wave comes out from the antenna slot, the major part of it propagates directly into the liver tissue through the opening of the antenna slot and the Teflon coating at the antenna tip. As shown in Figure 8, the major power flow is marked by the big arrows. The EM wave also propagates through the Teflon gap between the floating sleeve and the outer conductor. Having passed the sleeve through the gap, the power flows into the liver tissue through the Teflon coating after the sleeve. We marked this power flow as the minor power flow in Figure 8.
Figure 8: Demonstration of separated power flows and their interference for the floating sleeve antenna. The sizes of the arrows show the relative magnitudes of the power flow but they are not scaled accordingly. The total of the minor power flow through the gap of the sleeve is about 5% of the total of major power flow.

The magnitude of the minor power flow depends on the width of the gap and the reflection between the Teflon coating and the liver tissue. With computer simulation of our optimized design, the total power through the Teflon gap is only about 5% of the total power of the major power flow. The major power flow accounts for 95% of the total power, so it is called the major one.

Liver tissue is a lossy medium. At 2.45 GHz, its electric conductivity is about 1.67 S/m. As the wave propagates in the liver tissue, microwave power is absorbed and converted to heat energy. The wave is attenuated during propagation. Teflon is a lossless medium. There is no attenuation and absorption for the wave to propagate through it. Liver tissue and Teflon have different permittivities. At 2.45 GHz, the relative permittivity of liver tissue is about 43 and the relative permittivity of Teflon is 2.1. The wavelength of the EM wave is determined mainly by permittivity of the medium. The wavelength is about 18.1 mm in liver tissue and is about 84.4 mm in Teflon at 2.45 GHz.
Both waves propagate to the direction of the antenna tail. The major wave propagates in the liver tissue and the minor wave propagates in the Teflon medium. The major wave attenuates during its propagation and the minor wave does not attenuate. The two waves also have different phase shifting because the wavelengths in the two media are different.

By carefully selecting the length of the sleeve, it is possible for the two waves to have similar amplitude and phase difference of $\pi$ in the area towards to the tail of the antenna. When both waves reach the area marked as “Area of interference” in Figure 8, the two waves have similar amplitude and reversed phase. In such a situation, the two waves interfere and cancel each other and result in complete attenuation of SAR in the area.

**D.2 Verification with computer simulation**

Computer simulations of the floating sleeve antenna were performed using the electromagnetic modeling capabilities of FEMLAB version 3.1a. Simulations were carried out using an axially symmetric model which minimized the computation time while maintaining good resolution and the full 3D nature of the coaxial-based antenna.

From the FEMLAB simulation we can plot the power flow of the antenna EM field and directly show that the power of the fields pass both outside the floating sleeve and through the gap between the sleeve and the coaxial cable outer conductor. As both the wave on the outside surface of the sleeve and the wave through the gap propagate backward from the antenna slot to the antenna tail direction, phase difference between the two waves increases as the propagation distance increases.
\[
\phi = \left( \frac{2\pi}{\lambda_{\text{liver}}} - \frac{2\pi}{\lambda_{\text{Teflon}}} \right) x 
\]  
(7.2)

where \( \phi \) is the phase difference, \( \lambda_{\text{liver}} \) is the wavelength in liver tissue, \( \lambda_{\text{Teflon}} \) is the wavelength in Teflon. \( x \) is the propagation distance.

If we assume both waves have the same phase at the beginning of the sleeve near the antenna slot, \( x \) is equal to the length of the sleeve when both waves meet together at the end of the sleeve. In order for the phase difference to equal \( \pi \), the length of the sleeve needs to be about 11.5 mm. In another words, if the sleeve is 11.5 mm long, both waves will meet and cancel each other at the end of it. In order for both waves to have a phase difference of \( \pi \) to cancel each other well at some distance after the end of the sleeve, the length of the sleeve needs to be shorter than 11.5 mm. The length of the sleeve is optimized to give the best cancellation for the two waves. The major wave through the liver tissue does not just propagate along the surface of the sleeve, but in the whole liver tissue. The minor wave does not stop at the surface of the Teflon coating. It propagates into the liver tissue from the Teflon coating. We do not seek the best cancellation at a particular position, but an overall good cancellation in the whole area we marked as “Area of interference” in Figure 8. By selecting the sleeve length as 9.3 mm, roughly the half wavelength in liver tissue, we obtained the best overall cancellation in the computer simulation.

Since we believe the cancellation of the two waves is the reason for the sleeve antenna to yield a very localized SAR pattern, we present a way to separate the two waves. We study the EM field created by each wave and show the two separated EM
fields cancel each in the area of interference. Due to the complex structure of the sleeve antenna, the EM field created by the antenna cannot be solved mathematically with a closed form solution. We use computer simulation of FEMLAB for the analysis of the field separation.

With the sleeve antenna EM computer model shown in Figure 9 (a), we calculated the power from the power source $P_{src}$, the power passing through the sleeve gap $P_{gap}$ and the power passing into the liver from the antenna tip to the first end of the slot $P_{open}$. $P_{src}$ was calculated by linear integration of the power flow on the boundary marked by “Power source #1”, which is actually the boundary of the dielectric of the coaxial cable. $P_{gap}$ was calculated by linear integration of the power flow along the line marked as “Integration boundary #1”. The boundary is along the radial direction. It is 7 mm away from the near edge of the antenna slot. $P_{open}$ was calculated by linear integration of the power flow along the boundaries marked as “Integration boundary #2 and #3”. The sum of $P_{gap}$ and $P_{open}$ should equal $P_{src}$ for the reason of energy conservation. There could be a small mismatch because of the limited accuracy of the computer FEM simulation. Phases of the magnetic field $H_{phi}$ at three spots were recorded. $\phi_{src}$ is the phase at the power source #1, and was checked at the boundary of power source #1. $\phi_{gap}$ is the phase in the gap of the sleeve at the spot of the integration boundary #1. $\phi_{open}$ is the reference phase for the antenna slot opening, and was checked at the spot marked as A. The spot is where integration boundary #2 joins the floating sleeve. We called the EM field of the sleeve antenna $F_{sleeve}$. Figure 10 (a) plots the SAR of the EM field.
Figure 9: Power measurement and two field separation. The antenna is not drawn exactly to scale.

(a) The normal floating sleeve antenna. (b) Adding low reflection boundaries and the higher permittivity block to (a).

We added a wall inside the sleeve gap. As shown in Figure 9 (b), two sides of the wall are marked as “Absorption boundary” and “Integration boundary #1 and power source #2”. Both sides had low-reflection boundaries. We also created a block of material with higher permittivity before the absorption boundary.

The higher permittivity block is set to have a permittivity value higher than Teflon, the dielectric material in the coaxial cable and the antenna slot. It causes reflection of the power coming from power source #1. We can control the amount of power reflection by selecting the value of the permittivity of the block. The absorption boundary absorbs all the remained power. The absorption boundary and the high permittivity block work together to block the wave from power source #1 from passing through the gap of the sleeve and controlling the amount of power entering the liver via the opening of the antenna slot.
For the sleeve antenna model in Figure 9 (b), we calculated $P_{open}$ in the same way that we did for Figure 9 (a). We set the correct permittivity value for the higher permittivity block in order for $P_{open}$ to be the same as $P_{open}$ in Figure 9 (a). Since there is no power passing through the gap, the EM field in the liver is created only by $P_{open}$, which is the power passing into the liver tissue from the slot opening. We call this EM field as $F_{open}$. Figure 10 (c) plots the SAR of this EM field. The reference phase for the antenna slot opening $\phi_{open2}$ was checked at the same spot as $\phi_{open}$ for Figure 9 (a).

We add a second power source at the other end of the wall, which is a low reflection boundary, marked as “Integration boundary #1 and power source #2”. Both power source #1 and #2 are magnetic power sources. The magnitude and the phase of magnetic field of the second power source are controlled. The phase $\phi_{src2}$ requires that $\phi_{src2} - \phi_{open2} = \phi_{gap} - \phi_{open}$. The magnitude requires that the power output of the power source #2 is equal to $P_{gap}$ from Figure 9 (a) when the power source #1 is on. After setting the power source #2, we can set the power source #1 to 0. We call the EM field created by power source #2 only as $F_{gap}$. Figure 10 (d) plots the SAR of this EM field.

With the antenna model in Figure 9 (b), we have two separated power sources. Each power source can be set to 0 individually and we will obtain the two separated EM fields respectively. If we have both power sources on, the EM field we obtain is the interfered sum of the two separated EM fields $F_{sum}$. Figure 10 (b) plots the SAR of $F_{sum}$.
Figure 10: Plot of SAR in LOG scale. The contours are equal-dB lines. Power differences between two contour lines are 5 dB. The most outer contour line is –60 dB. The next one is –55 dB and so on.
(a) SAR of the floating sleeve antenna. (b) SAR with both power sources on. (c) SAR with only power source #1 on. (d) SAR with only power source #2 on. The fields of SAR in (a) and (b) are practically the same.

We are interested in the field $F_{\text{sum}}$. We expect that $F_{\text{sum}}$ is practically equal to $F_{\text{sleeve}}$ from Figure 9 (a). From the contour of SAR plots in Figure 10 (a) and Figure 10 (b) we see that $F_{\text{sum}}$ is virtually equal to $F_{\text{sleeve}}$.

We understand that the method of field separation we have just presented isn’t perfected. The low reflection boundary wall does not incorporate the secondary reflection on the Teflon–liver tissue interface before and after the sleeve. It does follow the special
structures of the sleeve antenna, and basic principle of field vector summation. It also follows the principle of energy conservation. The accuracy of the method is still questionable, but the interfered field $F_{\text{sum}}$ turns out to be quite a good approximation of the field $F_{\text{sleeve}}$.

From the result of field separation we can tell with strong confidence that our theory is correct. We can then use the hypothesis to guide our antenna design and optimization.

E Analysis of the antenna design

We analyze critical design aspects for the sleeve antenna. With the theory we have presented, we can explain very well the critical aspects and other important observations for which we could not provide a good explanation before.

Because the key aspect of the sleeve antenna is to get the best cancellation for the two separated waves, most critical design aspects are related to it. In order to have the best cancellation, the two waves need to have similar magnitude and required phase difference near $\pi$. Any antenna structure which affects either magnitude or the phase difference is important to be considered in the designing and optimization.

E.1 The length of the sleeve

The length of the sleeve is apparently critical because it affects the phase difference between the two power flows. In fact it also affects the magnitude of the major power flow because the liver tissue is a lossy medium.
Figure 11: Demonstration of different sleeve lengths. The sleeve is not covered by a Teflon layer. Antenna SAR patterns are plotted in dB scale. The most outer contour is –40 dB. (a) The sleeve is 6 mm long, shorter than the optimized length. (b) The sleeve is at the optimized length, 9.3 mm long. (c) The sleeve is 14 mm long, longer than the optimized length. If the sleeve is either too long or too short, it fails to create a localized SAR pattern.

Because we desire that the sleeve create a phase difference of $\pi$ between the two power flows, the optimal length of it is determined by the wavelengths, which are in turn determined by effective dielectric properties of the materials on both sides of sleeve. We use the frequency at 2.45 GHz and we use Teflon and liver tissue. The optimal length is about 9.3 mm. If a different material is used to replace Teflon or the antenna is to be used in a different tissue than liver, the optimal sleeve length would change accordingly.

If the material on one side of the sleeve is not homogeneous, its effective dielectric properties could change from its native dielectric properties. A good example is
the difference from the original sleeve antenna design to the new alternative antenna
design. In the original design, the sleeve is covered by a 0.15 mm thick protective Teflon
layer. Both the Teflon coating layer and liver tissue outside of it have to be considered for
the effective dielectric properties. The resultant optimal length of the sleeve is 19 mm
instead of 9.3 mm in the alternative design. The optimal sleeve length is not often
obvious and simple. Computer simulations or experimental measurements are often
required in order to determine it.

**E.2 The thickness of the Teflon gap layer**

The sleeve is isolated from the outer conductor of the coaxial cable by a layer of
Teflon. The thickness of this layer controls the magnitude of the minor power flow which
propagates through the gap layer. We used 0.15 mm thick Teflon for both designs. As we
determined from computer simulation, the SAR pattern was less localized if the layer was
either too thin or too thick. A value between 0.1 mm to 0.25 mm was good.

![Figure 12: Demonstration of the thickness of the Teflon gap. (a) the gap is 0.05 mm thick. (b) the gap is 0.35 mm thick. From Figure 11 (b), neither too thick nor too thin a gap layer would create a well localized SAR pattern.](image)
E.3 The Teflon layer covering the sleeve and its thickness

With or without a Teflon coating layer covering the sleeve is the difference between the original design and the new alternative design. If the Teflon coating layer is used, the effective dielectric properties for the materials outside the sleeve are affected. The thicker the Teflon layer, the more the effective dielectric properties change from the values of liver tissue in the direction of Teflon. From computer simulations we can see that the optimal sleeve length increases if the thickness of the Teflon coating layer increases. When the optimal sleeve length changes, the shape of the SAR pattern changes accordingly and the SAR patterns are always localized very well.

There are other considerations about the Teflon protective layer, including mechanical stability and sterilization.

E.4 The sleeve position

The floating sleeve can slide along the coaxial antenna body, which is covered by the Teflon isolation layer. As long as the sleeve does not cover the antenna slot and is kept at its optimal length, it can always constrain the SAR pattern despite its placement position. The SAR pattern is always constrained from the antenna distal tip to the back end of the floating sleeve. If the sleeve position is changed, the antenna SAR pattern changes accordingly. We can effectively control the SAR pattern for a MWA operation to best cover a liver tumor’s size and shape by selecting the placement of the sleeve, shown in Figure 13.
Figure 13: Demonstration of changing the shape of the SAR pattern by sliding the floating sleeve. (a) The sleeve is aligned to the edge of the antenna slot. (b) The sleeve is 9 mm away. (c) The sleeve is 18 mm away. At each position, the sleeve always works well to constrain the SAR pattern.

E.5 Other observations:

The Teflon wrapping outside the antenna is critical for the antenna performance. It enlarges the active radiation region of the antenna, reduces the peak SAR near the antenna slot, and allows the wave to propagate further into the tissue. It therefore enhances the effects of MW tissue heating and creates a larger thermal lesion.
F Alternative designs

F.1 With and without Teflon coverage on the sleeve

The Teflon coating layer outside the sleeve directly affects the optimal length of the sleeve. I have presented two different designs for the situations with and without the Teflon coating layer on the sleeve in Figure 1 and Figure 8. We usually desire to have a short sleeve in order to generate thermal lesions as spherical as possible because most liver tumors have a spherical shape and sleeve antennas with longer sleeves generate thermal lesions longer in the longitudinal direction than the transverse direction, so a shorter sleeve is usually more desirable than a longer sleeve.

In order to achieve sleeves with shorter length, we could reduce the thickness of the Teflon coating layer as much as possible, or completely remove the Teflon coating layer and expose the floating sleeve to the tissue. From the consideration of mechanical stabilities and clinical safety, a Teflon layer coating the sleeve is a better design.

F.2 With larger and smaller coaxial cable

I used a UT-085 coaxial cable as the basis for my current antenna. The normally allowed power transmission for UT-085 coaxial cable is 120 W. Strickland used 180 W with a larger antenna. Using higher power is desirable to generate larger thermal lesions for tumors of larger sizes. At the meantime, the outer diameter of my current antenna is about 3.5 mm. The antenna is slightly larger than the size for regular percutaneous treatment, which requires probes from 2 to 2.5 mm in diameter.
We can have the alternative design for the sleeve antenna based on larger or smaller coaxial cables. By selecting a sleeve of appropriate float size and coaxial cable size, we can have the final antenna size large enough to allow higher power throughput, or small enough for percutaneous treatments.

F.3 Dipole sleeve antenna versus slot sleeve antenna

I used a dipole antenna as the basis in my sleeve antenna design. In fact, the idea of a floating sleeve can also be applied to other basis antenna designs, including the slot antenna, or a monopole antenna. I have performed computer simulation for such ideas and the results showed that the floating sleeve can always constrain the SAR pattern effectively, regardless of the basis antenna.

Further study can be performed in order to evaluate the performance differences of the floating sleeve on different basis antennas, to decide which basis antenna works the best with the floating sleeve for the overall requirement of MWA applications.

G Further study

G.1 Using high permittivity material for coverage layer

I used Teflon as the coating material outside the sleeve. The fact that the sleeve length depends on the thickness of the coating layer makes the antenna performance sensitive to the thickness. If the coating layer is slightly off the designed thickness, the antenna could perform less optimally.
One way to improve this problem is to use some other coating material with higher permittivity. Teflon tape is very easy to use in fabrication, but its relative permittivity is only 2.1.

According to equation 7.3, the sleeve needs to be 11.5 mm long in order to have the phase difference $\pi$ for the two waves. So the uncoated sleeve length should be slightly shorter than 11.5 mm.

\[
x = \pi \left( \frac{2\pi}{\lambda_{\text{medium}}} - \frac{2\pi}{\lambda_{\text{Teflon}}} \right)
\]  

(7.3)

According to equation 2.45, the wavelength in a lossless medium is $\lambda \propto 1/\sqrt{\varepsilon_{\text{medium}}}$, then the equation can be transformed into:

\[
x = \frac{1}{2\sqrt{\varepsilon_{\text{medium}}} - 2\sqrt{\varepsilon_{\text{Teflon}}}}
\]  

(7.4)

$\varepsilon_{\text{medium}}$ is the effective permittivity of the medium outside the sleeve. If the sleeve is not coated, $\varepsilon_{\text{medium}}$ is the permittivity of liver tissue. If the sleeve is coated, $\varepsilon_{\text{medium}}$ is the permittivity of the combined material of the coating material and liver tissue, and $\varepsilon_{\text{medium}}$ is largely affected by the coating material. For all cases,

\[
\varepsilon_{\text{coating}} < \varepsilon_{\text{medium}} < \varepsilon_{\text{liver}}
\]  

(7.5)

$\varepsilon_{\text{medium}}$ and $\lambda_{\text{medium}}$ can be determined by using computer simulation, their values apparently depend on the thickness and permittivity of the coating material. In my sleeve antenna design, I used a Teflon coating layer 0.15 mm thick. $\lambda_{\text{medium}}$ is determined from the CEM model as 2.95 mm and $\varepsilon_{\text{medium}}$ can be calculated from equation 2.45 as 17.2.
Using another coating material with higher relative permittivity could significantly increase $\varepsilon_{\text{medium}}$ and reduce $\lambda_{\text{medium}}$. An increased $\varepsilon_{\text{medium}}$ will largely reduce the half wavelength sleeve length according to equation 7.4. I have looked for a suitable coating material and found both Nylon and aluminum oxide have relative permittivity of about 8. Without considering the thermal and mechanical stability, both Nylon and aluminum oxide could be very good choices for the coating material for the sleeve antenna.

I have checked the CEM models by changing the relative permittivity of the coating layer to 8. The optimal sleeve length will reduce from 16 mm to 10 mm and the optimal length is much less dependent on the thickness of the coating layer.

So there are a few advantages to use coating material with higher permittivity.

- Shorter optimal sleeve length
- Thermal lesion shape is more spherical
- Optimal sleeve length and antenna performance are less sensitive to the thickness of the coating layer

**G.2 Optimization for changed tissue dielectric properties**

The floating sleeve antenna is designed and optimized for the dielectric properties of normal liver tissue. It is not optimized for ablated liver tissue with changed dielectric properties.

After I have a better understanding of the tissue physical property changes during MWA, I could optimize the floating sleeve antenna accordingly.


H References


Chapter 8

Comprehensive computer simulation for MWA
This paper presents a comprehensive computer model for microwave tissue ablation. Besides an antenna EM model and thermal model, it simulates tissue water related phenomena, including evaporation, diffusion and condensation. The thermal model is based on an expanded Bio-heat equation, which includes tissue water evaporation at higher temperatures. Both the EM model and the thermal model are improved to utilize water content dependent tissue physical properties. The comprehensive model generates results significantly closer to experimental results than results from previous static antenna EM models plus basic thermal models.

Major contents of this chapter are also presented in [1], which will be submitted to IEEE TBME for publication in 2006.

A Introduction

A good computer model of MWA is very essential. It would help to improve the design of a MWA system by giving us the ability to evaluate and optimize all the system parameters in fast and cost effective ways. It could also help to us to better understand the fundamentals of the physical phenomena that are happening during a MWA procedure.

I have shown the basic EM model and thermal model in chapter 3. Both the basic EM model and the basic thermal model are built on static physical properties of normal liver tissue. They don’t support temperature and tissue water dependences of tissue physical properties. The basic thermal model is only valid for temperatures lower than 80 °C. It only simulates the basic thermal conduction and approximate blood perfusion. It does not simulate critical tissue water related physical phenomena at higher temperatures. I have demonstrated the entire MWA in Figure 1 of chapter 4 and shown why it is critical
to consider tissue water evaporation, movement and water vapor condensation in chapter 4.

Because of the incompleteness of the basic models, the results from the current computer models, especially the basic thermal model, do not match the experimental results obtained from either ex-vivo or in-vivo experiments. I have gained more quantitative knowledge from our experimental measurements on tissue water content and tissue temperature, presented in chapter 5 and 6. I have also expanded the bioheat equation in chapter 4 to extend its working region into temperatures high enough to evaporate tissue water. With all the new knowledge, we can develop a better computer model. The new and better computer model integrates both the EM and thermal simulation, and covers or approximates other important physical phenomena, especially tissue water content related mass and thermal movements, for MWA procedures.

B Tissue water dependencies of tissue properties

I have mentioned tissue physical properties and their dependencies on tissue temperature and tissue water content. Dependencies of tissue physical properties on tissue water content will be discussed in detail in this section.

During microwave heating, tissue water evaporates and water vapor is generated. Tissue is mixed with solid material, tissue water and tissue water vapor. The mass and volume of tissue solid material does not change. The mass and volume of tissue water reduces as tissue water evaporates. The volume left over is filled by water vapor and air. In fact, the total volume of tissue may expand or shrink slightly depending on volume of
generated water vapor and speed of water vapor diffusion. Tissue volume changing cannot be analyzed quantitatively and will be ignored in this study.

According to the Maxwell Garnett’s mixture theory, all tissue physical properties depend on the tissue composition. We consider that tissue is composed of solid material (protein) and tissue water, as well as water vapor or gas if evaporation occurs. Since volumes of tissue water and water vapor content change during the MWA procedure, all tissue properties change accordingly, depending on the relative volume or mass of all tissue compositions.

**B.1 Tissue density**

![Diagram showing the relationship between tissue density and tissue water volume fraction.](image)

Figure 1: Converting the remaining tissue water volume per unit volume of tissue to tissue density

\[
\rho = \sum_n v_n \times \rho_n
\]  

(8.1)
where $\rho$ is the tissue density [kg/m$^3$], $v_n$, $\rho_n$ are the volume fraction and density of the $n$th component. For ablated tissue water,

$$\rho = v_w \times 1000 + 0.222 \times 1300$$  \hspace{1cm} (8.2)$$

$v_w$ is the tissue water volume per unit volume of tissue, $0 \leq v_w \leq 0.778$. It is unitless. For normal tissue, $v_w = 0.778$. Proteins account for 22.2% of the total volume. The density of solid material is 1300 kg/m$^3$ [2]. The density of water is 1000 kg/m$^3$. When tissue water is lost due to evaporation, the total tissue volume and the amount of solid tissue materials do not change. The density of water vapor and air is small and is safely ignored from the equation. For 0.778 and 0.222 volume fraction of water and protein, the respective mass fractions are 0.73 and 0.27. We obtained the number of 0.73 with bulk measurements with cow liver tissue. Figure 1 plots tissue density versus tissue water content volume fraction.

**B.2 Tissue heat capacity**

The heat capacity in composite materials can be estimated as the sum of heat capacities of each fractional component [2].

$$C = \sum_n w_n C_n$$  \hspace{1cm} (8.3)$$

where $w_n$ is the mass fraction of the $n$th component and $C_n$ is the specific heat.

For the $n$th component, its volume fraction can be converted to mass fraction as:

$v_n$ can be converted to mass fraction $w_n$ as:

$$w_n = \frac{v_n \times \rho_n}{\rho}$$  \hspace{1cm} (8.4)$$
For tissue water content:

\[ w_w = \frac{v_w \times 1000}{v_w \times 1000 + 0.222 \times 1300} \]  \hspace{1cm} (8.5)

Figure 2: Tissue heat capacity versus tissue water volume fraction

For liver tissue, we assume that tissue is composed of water and proteins. The equation for normal and desiccated liver tissue is:

\[ C = 4200 \times w_w + 1560 \times 0.27 \]  \hspace{1cm} (8.6)

where \( C \) is the heat capacity [J/kg-K], \( w_w \) is the remaining tissue water mass per unit mass of tissue, \( 0 \leq w_w \leq 0.73 \). The equation is based on the heat capacity of solid tissue materials (proteins), which is 1560 J/kg-K [2] and the heat capacity of water, which is 4200 J/kg-K. Gas is again safely ignored because the mass fraction of gas is very small. Figure 2 shows the plot of calculated heat capacity versus tissue water volume fraction.
B.3 Tissue thermal conductivity

Calculation of tissue thermal conductivity is also based on the mixing theory equation [2]:

\[ k = \rho \times \sum \frac{k_n w_n}{\rho_n} \quad (8.7) \]

where \( k \) is the thermal conductivity of the composite material, \( k_n \) is the thermal conductivity of the \( n \)th component respectively. \( k_{\text{water}} = 0.603 \), \( k_{\text{protein}} = 0.195 \) [2].

Using equation 8.4, equation 8.7 can be written as:

\[ k = \sum k_n v_n \quad (8.8) \]

The thermal conductivity of gas \( k_{\text{gas}} \) varies with temperature \( T \) [K], it can be calculated by using the empirical equation [ref]:

\[ k_{\text{gas}} = 1.52 \times 10^{-11} T^3 - 4.8574 \times 10^{-8} T^2 + 1.0184 \times 10^{-4} T - 3.9933 \times 10^{-4} \quad (8.9) \]

Tissue temperature could be 10 to 200 °C during ablation and the calculated \( k_{\text{gas}} \) value is in the range of 0.0249 to 0.0385. For liver tissue in ablation, we have:

\[ k = 0.603 \times v_w + 0.222 \times 0.195 + (0.778 - v_w) \times k_{\text{gas}} \quad (8.10) \]

Because there is no significant tissue water evaporation less than 100 °C, the gas volume in tissue and contribution of gas in equation 8.10 are insignificant at lower temperatures. We can safely use the constant value of \( k_{\text{gas}} = 0.0316 \) at 100 °C in the equation. Figure 3 plots tissue thermal conductivity versus tissue water volume fraction.
B.4 Tissue dielectric properties

We use the empirical equations by Jonathan Schepps and Kenneth R Foster to calculate tissue relative permittivity and conductivity at microwave frequencies from tissue water volume fraction, as shown in equations 8.11 and 8.12 [3, 4]. Parameters $\varepsilon_s^m$ and $\sigma_{0.1}$ are extrapolated from Figure 4 and Figure 5, with $f = 2.45$ GHz [4].

\[ \varepsilon' = 1.71 f^{-1.13} + \frac{\varepsilon_s^m - 4}{1 + (f/25)^2} + 4 \]  
(8.11)

\[ \sigma = 1.35 f^{0.13} \varepsilon_{0.1} + \frac{0.0222(\varepsilon_s^m - 4)f^2}{1 + (f/25)^2} \]  
(8.12)
Figure 4: Relationship of tissue electric conductivity vs. tissue water volume fraction at 0.1 GHz [4].

Figure 5: Relationship of tissue relative permittivity vs. tissue water volume fraction at 25 GHz [4]

C Overview
The new comprehensive MWA computer model is based on an enhanced EM model, an enhanced thermal model and the hypothesis about tissue water and thermal energy movement. Figure 6 shows the flowchart of the data interactions between the different parts of the entire computer model. Compared to Figure 28 in chapter 3, which shows the data flow from the basic EM model to the basic thermal model, parts in the new advanced model are interactively linked and data flows are looped to form complex relationships between different tissue physical properties.

I try to cover most of the physical phenomena happening in the liver tissue during a MWA procedure. I understand that it is not possible to cover all phenomena completely and accurately. Many of the phenomena cannot be described mathematically with good accuracy. Some of such physical phenomena have to be approximated with simpler mathematical descriptions and others have to be ignored in the current implementation.

**Simulated or approximated physical phenomena:**

- Microwave radiation by the antenna
- Tissue SAR absorption and microwave heating
- Conductivity heat transfer
- Heat loss due to blood perfusion
- Tissue water evaporation
- Water vapor condensation
- Movement of tissue water content
- System energy conservation
- Heat loss due to leaking of water vapor
Not covered physical phenomena:

- Convective heat transfer
- Blood flow in larger blood vessels

Important results of the comprehensive computer model are tissue temperature, tissue water content, antenna SAR pattern and tissue physical properties, dependent on tissue water content.

C.1 Explanation of important interactions

In the interactions shown in Figure 6, antenna radiation is the main heating source. All heat energy of the system is from EM energy delivered by the antenna. Heat loss because of blood perfusion only occurs with in-vivo experiments. There is no blood
perfusion with ex-vivo experiments. Nevertheless some water vapor could escape from the liver tissue during ex-vivo experiments, and such water vapor escaping does not occur with in-vivo experiments. Tissue temperature is determined by the heat source, heat loss, tissue thermal properties, and tissue water evaporation and condensation. Conductivity heat transfer is not the only thermal process occurring in the tissue. Water evaporation and condensation are also critical. Tissue temperature determines tissue water evaporation and water vapor condensation. The relationship between temperature and tissue water evaporation follows the previous measurement results in chapters 4 to 6. Tissue temperature determines tissue blood perfusion. Blood perfusion only occurs when tissue temperature is below 50 °C. When tissue temperature is higher than 80 °C, evaporation starts to happen and tissue physical properties are affected by tissue water content. Water vapor pressure and condensation determine the tissue water redistribution after evaporation. After tissue water starts to evaporate, vapor pressure builds up and vapor diffuses from high pressure to low pressure. The low pressure part of the liver is also at low temperature. Water vapor condenses at lower temperatures. The overall effect of water evaporation, water vapor diffusion and condensation is the redistribution of tissue water. Tissue water determines both tissue dielectric properties and thermal properties through water dependencies of the tissue properties. Tissue dielectric properties determine the antennas SAR pattern. EM fields and antenna SAR patterns directly depended on the tissue dielectric properties.
C.2 Implementation and work flow

The entire computer simulation is implemented in 4 modules: the main module and 3 individual modules for EM, thermal and tissue water. Figure 7 shows that modules are interactive and circularly executed to form a loop. The entire computer model is built with MATLAB and FEMLAB 2.3. It fully utilizes the integration of FEMLAB into MATLAB.

![Diagram showing modular implementation of the comprehensive computer model for MWA]

Figure 7: Modular implementation of the comprehensive computer model for MWA

The main module is responsible for initialization of the entire simulation, and for procedural control and convergence control. Procedural control provides the ability to start, pause and stop the simulation. The convergence control mechanism adjusts the magnitude of the time step after dynamically checking the convergence of the solutions between the consequent time steps. The purpose of convergence control is to generate accurate results using the least computation time.
The enhanced EM module is written in MATLAB for FEMLAB. It accepts tissue dielectric properties from the main module. The tissue dielectric properties are calculated by the main module according to the tissue temperature and tissue water content distribution. It calls FEMLAB from MATLAB to solve the EM model and returns calculated antenna SAR back to the main module.

The major enhancement of the EM module is to use tissue dielectric properties provided by the main module of the new EM model instead of constant tissue dielectric properties. The main module is responsible to calculate tissue dielectric properties for each meshing node according to the current tissue temperature and tissue water content on each meshing node.

The thermal module is written in MATLAB for FEMLAB. It accepts tissue thermal properties from the main module. Tissue thermal properties are calculated by the main module according to the tissue temperature and tissue water content distribution. It accepts solving parameters from the main module. It calls FEMLAB from MATLAB to solve the thermal model at selected time steps for the selected duration and returns the calculated tissue temperature back to the main module.

One major enhancement in the thermal module is to use tissue thermal properties provided by the main module instead of constant tissue thermal properties. Another major enhancement is the usage of the expanded bioheat equation instead of the original bioheat equation. Both thermal conduction and tissue water evaporation are covered in the expanded bioheat equation.

The tissue water module calculates the evaporated tissue water during the time step, and approximates tissue water condensation as described in the previous section. It
also calls the tissue water diffusion model and adjusts the tissue water content and tissue
temperature after diffusion as described in the previous sections. The diffusion model,
similar to the thermal module, is written in MATLAB for FEMLAB.

**C.3 Finite difference method**

The expanded Bioheat equation can be solved numerically in a sequence of minor
time steps by the finite difference method. All values in the equation are time and spatial
variant. \( T \) is the independent variable to be solved and all other values are parameters for
the equation. All parameters are dependent on the tissue temperature or tissue water
content. For each time step, we use tissue temperature and tissue water content of the
previous time step to calculate the parameters and assume them to be constant within the
new time step. In such a way, the parameters are constant within each minor time step,
but varying on the sequential results of one step earlier for the entire sequence of time
steps. If time steps are selected to be small so that the assumption is valid, the final
results will be stable and relatively accurate.

FEMLAB is a FEM solver from Comsol. We use its multiple physical capability
to solve the expanded bioheat equation for the thermal model, as well as antenna EM
model and tissue water diffusion model. All parameters in the models change at each
time step. Such step-by-step changing parameters can not be set directly in the FEMLAB
model. FEMLAB integrates with MATLAB, by calling MATLAB functions in order to
retrieve the current value of the parameters at each meshing node of the FEM models.
Initial values for each time step are also set by calling a MATLAB callback function.
Because MATLAB functions are called from FEMLAB to retrieve the parameter values, we have freedom to perform all necessary calculations in these functions. MATLAB also keeps tracking tissue temperature and tissue water content at each time step, passed from FEMLAB in the function calls.

**C.4 Goals of the advanced computer model**

The goal of the advanced computer model is to generate results that match the measured results from experiments. Lesion shape and size prediction are the most important results. Besides them, the computer model generates other results including tissue temperature time and spatial distribution, tissue water time and spatial distribution, changes of antenna heating pattern and changes of tissue physical properties.

**D Implementation**

**D.1 Finite element models**

![Figure 8: The axial-symmetrical computer model geometry shows that the coaxial slot antenna is in liver tissue. Only the tip portion of the antenna is shown. The horizontal axis is the $z$ axis. The vertical axis is the $r$ axis. The aspect ratios are not equal for both axes. The entire geometry is](image)
symmetrical along the $z$ axis, so only half of the geometry is drawn in the model. Radius of the coaxial antenna is 1.75 mm. It is inserted 70 mm deep into liver tissue.

We create the FEM computer models according to our ex-vivo experiment setup where the floating sleeve slot antenna was inserted 70 mm into the liver tissue. Figure 8 shows the geometry of the models. We used an axially symmetric model [5], which minimized the computation time, and improved resolution while maintaining the full 3D nature of the fields.

EM, thermal simulation and tissue water diffusion models use the same geometric model. Our validation experiments are performed ex-vivo with cow liver tissue. To be consistent, blood perfusion is not included in the computer model.

The antenna was inserted into the tissue deep enough and liver tissue pieces were large enough. Boundaries of liver tissue were set to be low reflection boundaries in the EM model and isolation boundaries in the thermal and water diffusion models.

**D.2 Special considerations**

**D.2.1 Approximation of water vapor condensation**

We have assumed that water vapor condensation only happen in region of temperature less than 80 °C, but we haven’t discussed how to approximate it without considering vapor diffusion and partial pressure in detail.
Figure 9 shows that we approximate water vapor condensation in the following procedure. After each time step, we calculate the amount of tissue water vaporized during the time step. Water vapor diffuses to tissue regions of lower temperature and condenses uniformly in tissue regions having temperatures from 60 to 80 °C. During condensation, water vapor releases latent heat energy and heats the surrounding tissue where condensation occurs. Temperature change of water vapor and corresponding heat energy transferred are both ignored because the amount of heat energy involved is minimal comparing to the latent heat. After water vapor condenses back to water liquid, the amount of tissue water increases in the surrounding tissue. Driven water vapor pressure from higher temperature regions and water concentration gradient, excessive tissue diffuses further away into the low temperature region of tissue and leak out of the system. The amount of tissue water in regions at temperatures from 60 – 80 °C does not change. The ratio of the amount of water vapor condensed to the amount of tissue water evaporated is configurable from 0 to 1. We usually configure the ratio to 0.5.
D.2.2 Tissue water diffusion and adjustments

Tissue water diffuses from high concentration area (lower temperature) to low concentration area (higher temperature). The diffusion procedure changes the distribution of tissue water. Tissue $T > 80 \degree C$, in which tissue has been desiccated due to evaporation, gains tissue water content during diffusion and tissue at temperature lower than 80 \degree C loses water content. However, after the high temperature area of the tissue has gained more tissue water content through diffusion, excessive tissue water in the high temperature area will be vaporized at cost of heat energy and reduction of tissue temperature.

Tissue water diffusion and its effects were simulated in two steps in the tissue water module—diffusion simulation in FEM model and adjustment of temperature and tissue water after diffusion. Figure 8 shows that the geometry of the water diffusion in the FEM model is the same as that in the EM and thermal model. The diffusion FEM model solves the diffusion equation 8.13.

$$\frac{\partial W}{\partial t} = \nabla \cdot (D \nabla W)$$

(8.13)

where $W$ is tissue water content volume fraction and $D$ is the diffusion constant. We use $D = 5 \times 10^{-10}$ in our simulations [6].

Tissue water is redistributed by the diffusion simulation. Because we expect tissue water and tissue temperature to follow their normal relationship defined by the mapping curve, shown in Figures 2 to 4 in Chapter 4, we need to perform adjustment on them according to equation 8.14, which is based on the principle of energy conservation. The
adjustment, as shown in Figure 10, is only performed in the high temperature region where water evaporates.

\[ \rho C \Delta T = \alpha \Delta W \]  \hspace{1cm} (8.14)

where \( \rho \) is tissue density, \( C \) is tissue specific heat, \( \alpha \) is water latent heat, \( \rho_w \) is the density of water, \( \Delta T \) is the change of tissue temperature and \( \Delta W \) is the volume change of tissue water content. The equation is based on the fact that the thermal energy lost is equal to the water latent energy required for evaporation.

Tissue is at \((T_0, W_0)\) after diffusion. Please note that \( W_0 > W(T_0) \) because of the gain of excessive tissue water from diffusion process, where \( W(T) \) is the tissue temperature to tissue water mapping function. After an adjustment, we expect tissue at \((T_1, W_1)\), where \( W_1 = W(T_1) \). Change of \( W \) is \( \Delta W \) and change of \( T \) is \( \Delta T \). Both \( \Delta W \) and \( \Delta T \) are negative.

\[ W_1 = W(T_0) + (T_1 - T_0) \frac{\partial W}{\partial T} = W(T_0) + \Delta T \frac{\partial W}{\partial T} \]  \hspace{1cm} (8.15)

\[ \Delta W = W_1 - W_0 = W(T_0) + \Delta T \frac{\partial W}{\partial T} - W_0 = \frac{\rho C \Delta T}{\alpha} \]  \hspace{1cm} (8.16)

\[ W(T_0) - W_0 = \Delta T \left( \frac{\rho C}{\alpha} - \frac{\partial W}{\partial T} \right) = \frac{\rho C \Delta T}{\alpha} \left( C - \frac{\alpha}{\rho} \frac{\partial W}{\partial T} \right) = \frac{\rho C' \Delta T}{\alpha} \]  \hspace{1cm} (8.17)

\[ \Delta T = \frac{\alpha}{\rho C'} (W(T_0) - W_0) \]  \hspace{1cm} (8.18)

\[ W_1 = W(T_0) + \Delta T \frac{\partial W}{\partial T} = W(T_0) + \frac{\alpha}{\rho C'} (W(T_0) - W_0) \frac{\partial W}{\partial T} \bigg|_{T=T_0} \]  \hspace{1cm} (8.19)

\[ T_1 = T_0 + \Delta T = T_0 + \frac{\alpha}{\rho C'} (W(T_0) - W_0) \]  \hspace{1cm} (8.20)
Figure 10: Demonstration of adjustment of tissue temperature $T$ and tissue water $W$ after diffusion simulation

\section*{D.3 Computer simulation procedures and configurations}

Figure 11 shows the entire computer simulation. For time step $n$, we use tissue dielectric properties calculated from results of the last time step, antenna heating pattern $Q_n$ is calculated by EM simulation. $Q_n$ is assumed not to change during the time step $n$. Using current tissue temperature $T_n$ and tissue water content $W_n$ as initial condition, thermal and water evaporation simulation is performed with $Q_n$ and current tissue thermal properties $\rho_n$, $\kappa_n$ and $C_n$, which are the tissue density, thermal conductivity and effective specific heat respectively. The results are tissue temperature $T_{n2}$. Tissue water content $W_{n2}$ can be calculated according to equation 8.21. Water vapor condensation and tissue water diffusion are simulated by the tissue water module. The resultant $T$ and $W$ are then adjusted to $T_{n+1}$ and $W_{n+1}$. Tissue properties for the next step are calculated by using $T_{n+1}$
and $W_{n+1}$. Time $t$ is increased by $\Delta t$ for the next step. The procedure is repeated until the desired ablation duration is finished.

$$W_{n2} = W_n + (T_{n2} - T_n) \times \left. \frac{\partial W}{\partial T} \right|_{T - T_n}$$

Time step $\Delta t$ is adjustable during the simulation, from 0.2 s to 1 s. It is controlled by the convergence control mechanism in the main control module so that maximal tissue temperature change and maximal tissue water content change are within allowed values. Table 1 lists simulation configurations and the values we used.

With similar configurations, it takes 300 to 400 steps in 5 to 8 h to simulate for a heating duration of 360 s. Average computation time for each step was about 1 min. Our computer is a Dell desktop with 2.8 GHz P4 processor and 1.5 GB memory. We are using FEMLAB version 2.3 and MATLAB version 6.5.
Figure 11: Flowchart of the entire computer simulation procedures. $T_n$ and $W_n$ are tissue temperature and tissue water content at step $n$. $\varepsilon_r$ is tissue relative permittivity, $\rho$ is tissue density, $\sigma$ is electric conductivity, $\kappa$ is thermal conductivity, $C$ is tissue effective specific heat and $Q$ is the antenna heating pattern. The entire problem is solved using minor time steps according to the finite difference method.

Table 1: Simulation settings

<table>
<thead>
<tr>
<th>Simulation parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Initial tissue temperature:</td>
<td>8 °C</td>
</tr>
<tr>
<td>Initial tissue water volume fraction</td>
<td>0.778</td>
</tr>
<tr>
<td>Equivalent tissue water mass fraction</td>
<td>0.73</td>
</tr>
<tr>
<td>--------------------------------------</td>
<td>--------------</td>
</tr>
<tr>
<td>Microwave power</td>
<td>75 W</td>
</tr>
<tr>
<td>Duration</td>
<td>0 to 360 s</td>
</tr>
<tr>
<td>Blood perfusion</td>
<td>Off</td>
</tr>
<tr>
<td>Tissue water evaporation</td>
<td>On</td>
</tr>
<tr>
<td>Water vapor condensation</td>
<td>50%</td>
</tr>
<tr>
<td>Water diffusion</td>
<td>On</td>
</tr>
<tr>
<td>Water diffusion constant</td>
<td>$5 \times 10^{-10}$</td>
</tr>
<tr>
<td>Maximal tissue water change in 1 step</td>
<td>0.02</td>
</tr>
<tr>
<td>Maximal tissue temperature change in 1 step</td>
<td>10 °C</td>
</tr>
<tr>
<td>EM simulation interval</td>
<td>Every 10 s</td>
</tr>
<tr>
<td>Microwave frequency</td>
<td>2.45 GHz</td>
</tr>
<tr>
<td>Copper thermal conductivity</td>
<td>400</td>
</tr>
<tr>
<td>Copper density</td>
<td>8700</td>
</tr>
<tr>
<td>Copper specific heat</td>
<td>385</td>
</tr>
<tr>
<td>Teflon relative permittivity</td>
<td>2.1</td>
</tr>
<tr>
<td>Teflon electric conductivity</td>
<td>0</td>
</tr>
<tr>
<td>Teflon thermal conductivity</td>
<td>0.24</td>
</tr>
<tr>
<td>Teflon density</td>
<td>1200</td>
</tr>
<tr>
<td>Teflon specific heat</td>
<td>1050</td>
</tr>
</tbody>
</table>

**E  Experiment verifications**

All experiments were performed ex-vivo with bovine liver, which was obtained from a local slaughter house. Liver tissue was refrigerated overnight. The tissue initial temperature was approximately 6 to 9 °C before experiments. Homogeneous blocks of bovine liver were used for the experiment. They were big and thick enough to accommodate the lesion for a 6 min microwave heating.

We used Luxtron Fluoroptic thermometer model 3100 to measure tissue temperature during ex-vivo bovine liver tissue microwave ablation. Figure 12 shows placement of the microwave coaxial antenna and fluoroptic thermal sensors. Fiber-optic temperature sensors were inserted into the tissue guided by biopsy needles. Positions of the sensors were measured after ablation by measuring the positions of the thin plastic
fibers that were inserted together with fiber-optic temperature sensors. The temperature measurement experiment setup is described in [7].

Figure 12: The fiber-optic temperature sensors were placed 5 mm away from the antenna slot in longitudinal direction, and 5 mm, 7 mm and 9 mm respectively away from the antenna body in the radial direction.

Tissue water content measurements were performed with a similar setup, shown in Figure 13 (a) and (b). A 5 mm thick slice was cut from one of the two pieces, at a distance 5 mm from the slot, perpendicular to the antenna insertion direction as shown in Figure 13 (a). A strip of tissue was then cut off the slice and cut into 1 mm thick pieces, shown in Figure 13 (b). Weights of all tissue pieces were measured with an Ohaus Explorer E11140 digital analytical balance with 0.1 mg resolution. Tissue pieces were
vacuum-dried overnight with an Edwards ETD4 Tissue Dryer, and weighed the next morning. Water content for each tissue pieces was then calculated from the dry weight and wet weight of the piece. The tissue water content measurements are described in [8].

![Diagram](image)

**Figure 13**: (a) Ex-vivo experiment setup. The master slice of 5 mm thick was cut at 5 mm above the antenna slot. (b) Cutting 1 mm thick pieces from the master slice.
F Results

F.1 Lesion size and shape comparison

Figure 14: Ablated bovine liver tissue. The lesion was created by applying 75 W for 360 s. Tissue initial temperature was 8 °C. The lesion size was about 5 × 3.6 cm.

Figure 14 shows a scanned photo of ablated tissue. Figure 15 shows calculated tissue temperature contours from the new computer model. Figure 16 shows calculated tissue temperature contours from the previous simple computer model. According to previous measurements, tissue temperature was about 60 °C at the lesion boundary [7]. We can use the 60 °C temperature contour from computer simulation as the expected lesion boundary.
Figure 15: Tissue temperature contours from computer simulation

Figure 16: Tissue temperature contours from simple computer simulation
Figure 17: Calculated lesion sizes and measured lesion sizes versus ablation duration. Values in the plot were the lesion maximal diameters in the radial direction of the antenna. 60 °C was used as the lesion boundary in the computer simulations [7].

Figure 17 compares calculated lesion maximal radial diameters versus ablation duration to measured results. It shows that the predicted lesion radial diameter matches fairly well to experimental results for both the new computer model and the previous simple computer model.

**F.2 Temperature history comparison**

Figure 18: Measured tissue temperature during ablation. Measurement positions were 5 mm above the antenna slot longitudinally, shown in Figure 12. Distances from the thermal sensors to the antenna body were labeled for each curve.

Figure 18 shows plots of measured tissue temperature versus time at a few positions. These are the same results as presented in [7]. Figure 19 and Figure 20 show
tissue temperature versus time at the same positions from the new computer model and from previous simple model respectively.

Figure 19: Tissue temperature history from simulation. The positions are one-to-one matched to measurement positions in Figure 18.

The possible effects of tissue expansion and shrinkage were analyzed for temperature measurement [7]. The tissue temperature calculated by the new model could match closer to the measured temperature if these problems are taken into consideration.
Figure 20: Tissue temperature history from simulation of the simple model. The positions are one-to-one matched to measurement positions in Figure 18.

**F.3 Temperature distribution comparison**

Figure 21 shows measured tissue temperature versus positions at different ablation durations. Figure 22 and Figure 23 show simulation tissue temperature at the same ablation durations from the new model and the previous simple model respectively. Results from new model are fairly well matched to the measured results. If tissue shrinkage effects are considered, the simulated results could match better.
Figure 21: Measured tissue temperature versus the measurement positions for different ablation durations. Measurement positions were 5 mm above the antenna slot longitudinally, shown in Figure 12. Markers on the curves are the data points.

Figure 22: Simulated tissue temperature versus the positions for different ablation durations.
Figure 23: From the simple models, simulated tissue temperature versus position for different ablation durations.

**F.4 Tissue water distribution comparison**

Figure 24 shows the plots of tissue water content after ablation versus ablation duration and position. Figure 25 shows the plots of corresponding results generated by the new computer model. Both figures have demonstrated tissue water content movement during a MWA procedure. Figure 26 shows how tissue water content is spatially distributed at the end of 75 W, 360 s ablation.
Figure 24: Mean values of water content mass fraction, the ratio of remaining tissue water mass after ablation to the original tissue mass before ablation. See [8] for the error ranges and further information.

Figure 25: Water content mass fraction from computer simulation.
Figure 26: Tissue water content mass fraction contours at the end of 75 W, 360 s ablation, from computer simulation. Water content mass fraction of normal tissue is 0.73.

G Discussion and conclusion

Because of its complexity and comprehensiveness, the presented computer model has many advantages over the previous simple EM model plus regular bioheat equation based simple thermal model. The major advantages of the new model are the inclusion of tissue water and energy movement, and the usage of water content dependent tissue physical properties.

Compared to previous simple EM and thermal models, the new model is able to generate results that match much better to experimental results. Not only do results match lesion size, but also lesion shape, temperature history, temperature spatial distribution, tissue water content history and spatial distributions. With such improved results, the new model could provide better guidance for clinical procedures.

The new model could also be an important tool to aid antenna design and optimization. Figure 27 shows plots of normalized antenna heating patterns for before and after MWA. Figure 28 shows the calculated antenna reflection during the ablation
period. Similar changes of antenna power reflection have been observed in experiments. The figures have clearly shown that antenna performance has changed dramatically during MWA procedures. One antenna, optimized for normal tissue dielectric properties, will become less optimized after a MWA procedure because of the changes in tissue dielectric properties due to the changes of tissue water content. Such changes provide additional challenges to antenna design and optimization. MWA antennas have to be optimized not for static dielectric properties of normal tissue, but for the entire MWA duration during which tissue properties change.
Figure 27: Antenna heating pattern. (a) Before MWA. (b) After MWA with 75 W for 360 s. Both plots are normalized to the maximal values in plot (a).

Figure 28: Antenna power reflection versus time from computer simulation. I have observed similar trend of changes with power reflection reading on the microwave power generator that was used in the experiments.

Note that the computer simulation results do not match the experimental results exactly. The differences could be caused by many reasons, including the limitations of computer simulation, the inhomogeneity of liver tissue, imperfectness of antenna construction, as well as system and measurement errors in the experiments.

There are limitations with the new computer model. The methods presented in this paper were based on many currently unproved speculations, many aspects of it are questionable and are need to be validated and improved. The way to approximate water vapor condensation in the computer simulation is questionable. We approximate it by letting water vapor uniformly condense in tissue at temperatures from 60 to 80 °C. We
understand that such an approximation is definitely not the best. We have claimed that such an energy movement in a heated tissue system is a secondary heating source to the low temperature region. The approximation we have applied supports such a secondary heating process. We expect the way to simulate water vapor condensation will be improved in our future research. It is well known that tissue physical properties are dependent on both tissue temperature and tissue water content. Only dependencies on tissue water content are considered in the computer model and the method of implementation is yet to be validated through experiments. It is critical to have correct tissue properties in the computer model, but accurate values of such tissue properties are unfortunately unavailable and very difficult to measure by experiment.

Duck indicated that tissue volume changes with tissue temperature. In our temperature measuring experiments, plastic threads were inserted together with fiber-optic temperature sensors into the tissue. By cutting the ablated tissue open longitudinally and checking the placement of the plastic threads after ablation, we observed slight tissue volume shrinkage in the center of the ablated region. Such slight shrinkage could lead to 0.5 to 1 mm position shifting of the temperature sensors during temperature measurement, depending on the distance of the sensors. Because positions of temperature measurement sensors were measured after ablation, actual positions of the thermal sensors could be farther away from the antenna body than measured.

Nevertheless, just like computer models for all other applications, computer models for MWA are only able to approximate the real physical phenomena using the best available knowledge, and are probably never able to simulate the phenomena exactly and entirely. The most important goal of MWA computer models is to accurately predict
the lesion size and shape with a given MWA procedure setup, so as to help select the setup for a clinical procedure. For this purpose, the methods we present in this paper have shown improvements over previous simpler methods.

To summarize, we have introduced new computer models for microwave ablation. The computer model proposed by this paper is comprehensive and practical. Compared to the previous computer model, the new model is able to generate results much closer to experimental measurement results. The results match not only lesion size and shape, but also tissue temperature distribution and history, as well as tissue water content distribution and history. The new model would serve well as guidance to clinical MWA procedures.

H References


Appendix 1
A Floating Sleeve Antenna Yields Localized Hepatic Microwave Ablation

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Index Terms—ablation, coaxial aperture antennas, finite element methods, microwave heating, floating sleeve, interstitial

I. ABSTRACT

We report a novel coaxial antenna for hepatic microwave ablation. This device uses a
floating sleeve, that is, a metal conductor electrically isolated from the outer connector of the antenna coaxial body, to achieve a highly localized SAR pattern that is independent of insertion depth. This floating sleeve coaxial dipole antenna has low power reflection in the 2.4 GHz IMS band. Ex-vivo experiments confirm our numeric simulation results.

II. INTRODUCTION

Microwave ablation (MWA) is a promising technology for the treatment of hepatic tumors. The goal of MWA is to destroy the tumor along with a 1 cm margin of normal hepatic tissue. This technology has been used in both intra-operative and percutaneous approaches for primary hepatocellular carcinoma and hepatic metastasis of colorectal carcinoma [1]–[3]. Hepatic tumors are commonly spherically shaped and range in size from less than 1 cm to larger than 10 cm in diameter of various sizes. However, current technology limits a single ablated region to approximately 3 cm [4]–[8], not large enough to treat large tumors in a single pass. When the input power is increased beyond this point, undesired heating occurs along the coaxial feedline of the antenna. This detrimental heating causes damage to the liver outside the desired treatment region and can lead to burning of the skin during percutaneous treatment.

There are three potential causes of detrimental heating along the coaxial feedline. First, any impedance mismatch between the antenna and the surrounding medium will create reflections that set up standing waves within the coaxial feedline. Under such conditions, the local currents on the inside of the outer conductor can become large enough to cause local heating. If the wall of the outer conductor is thin, the heat may transfer to the surrounding tissue. Second, an impedance mismatch between the antenna and surrounding medium may also result
in unbalanced currents on the inner and outer conductors of the coaxial feed. In this case, a remainder current flows along the outside of the outer conductor of the coaxial feedline. The ‘tail’ seen in many of the specific absorption rate (SAR) patterns computed from simulations of MWA antennas is attributed to this current flow. Finally, most antenna designs are based upon copper coaxial cables. Since copper is a good thermal conductor, heat generated near the distal tip may be conducted along the feedline.

Several types of coaxial-based antennas, including the coaxial slot antenna [9], coaxial dipole antenna [10], coaxial monopole antenna [11], coaxial cap-choke antennas [12], [13] and others [14]–[17], have been designed for MWA or microwave hyperthermia therapies in an attempt to prevent this backward heating while creating as large an ablation radius as possible. The cap-choke antenna seems to most efficiently prevent backward heating [12], [18]. Cap-choke antennas offer a localized SAR pattern and the smallest SAR tail. A cap-choke antenna uses a metal sleeve, usually one-fourth of a wavelength long, soldered to the outer connector as a choke, and an extra metal ring at the distal tip of the antenna as a cap. This antenna works well at lower power levels, but the residual SAR tail of the antenna increases in size and causes backward heating problems at higher power levels or during extended ablations. At high microwave input power levels, even low level backward heating is detrimental enough to cause damage to normal hepatic tissue, as well as serious burning during percutaneous ablation.

This paper presents a novel coaxial antenna design—the floating sleeve antenna—that addresses many of the critical problems with current MWA antennas.
III. DESIGN OF THE FLOATING SLEEVE ANTENNA

Our goal was to design a coaxial antenna with a highly localized SAR pattern and low reflectivity for higher power transmission. Fig. 1 shows the design of the floating sleeve antenna. The antenna is based on a 50 Ω UT-085 semirigid copper-Teflon coaxial cable. A standard coaxial dipole antenna [10] is constructed from the coaxial cable and tightly wrapped with thin layers of Teflon tape. The metal sleeve, which is comprised of a section of copper tube (3.2 mm outer diameter, 2.5 mm inner diameter), is slid onto the Teflon-coated coaxial dipole antenna and positioned behind the antenna slot. The whole antenna assembly is then tightly wrapped with Teflon tape. The Teflon tape is heated during and after wrapping in order to prevent air from being trapped in the tape layers. Fig. 1(a) shows the longitudinal dimensions of each section of the antenna along with the overall diameter, while Fig. 1(b) shows the interior diameters in the region of the sleeve.

The floating sleeve antenna differs from existing laboratory and clinical devices (such as the cap-choke antenna) in that the sleeve is electrically isolated from the outer conductor of the coaxial feedline. This floating sleeve is similar to the open sleeve antenna [17] which also uses a floating sleeve. However the floating sleeve of the open sleeve antenna in [17] is quite long.

We designed the floating sleeve using computer simulations once we gained a qualitative understanding of the importance and effect of the following parameters.

We determined that the SAR pattern is affected by both the length of the sleeve and the thickness of the Teflon layer. If the sleeve is not covered by a Teflon layer, it needs to be approximately half of the effective wavelength in liver tissue. If the sleeve is covered by a Teflon layer, the sleeve needs to be longer, with the length depending on the thickness of the
Teflon layer. This length is critical, and if significantly longer or shorter than the ideal length, the sleeve can be less effective than other reported antenna designs. As long as the sleeve is half a wavelength (adjusted for the presence of Teflon) in length and is not covering the antenna slot, it seems able to constrain the tail of the SAR pattern. In fact the edge of the SAR pattern seems tied to the termination of the sleeve, to the extent that by sliding the floating sleeve, we can control and change the SAR pattern from a spherical shape to an elliptical shape. This may make it possible to control the shape of the lesion to fit different tumor shapes. The thickness of the Teflon isolation layer (the Teflon layer between the floating sleeve and the outer conductor of the coaxial cable) does not seem to affect the SAR pattern as long as it is at least 0.1 mm.

Our design uses a 2 mm wide slot, which is easily fabricated, and also gives good power reflection. The length of the antenna tip also slightly affects the power reflection and shape of the SAR pattern. The length in the design was adjusted to give the good trade off between the power reflection and a spherical SAR pattern.

The antenna design reported here is based on commercially available coaxial cable and utilizes readily available and inexpensive construction materials. This easily fabricated antenna is suitable for open or laparoscopic operative therapies. However, the diameter of 3.5 mm precludes this antenna from being useful for percutaneous therapies.

IV. COMPUTER SIMULATION AND EXPERIMENTAL RESULTS

We used computational electromagnetics simulations to compute the SAR distribution and input reflection coefficient, or $S_{11}$, as a function of frequency for the proposed antenna design. We compared simulated $S_{11}$ to experimentally measured $S_{11}$ to validate the model, and
evaluated lesion size and shape after ex-vivo ablation.

A. The computational electromagnetics (CEM) model

We performed simulations of the floating sleeve antenna using the electromagnetic modeling capabilities of FEMLAB™ version 2.3. We used an axially symmetric model [19], which minimized the computation time while maintaining good resolution and the full 3D nature of the fields. The 2D axisymmetric model requires 180 MB memory and 50 s of CPU time for each FEMLAB simulation on an Intel P4 2.8 GHz desktop computer. Fig. 2 shows the structure of the floating sleeve antenna near the tip in our CEM model. The model assumes that the floating sleeve antenna is immersed in homogeneous bovine liver tissue. The horizontal $z$ axis is oriented along the longitudinal axis of the antenna and the vertical $r$ axis is oriented along the radial direction. The liver is assumed to be infinite in extent, which is accomplished using low-reflection boundary conditions provided in FEMLAB™. The computational domain corresponds to a physical domain size of 60 mm in radius and 110 mm ($z = -30$ mm to $z = 80$ mm) in length.

The dielectric insulator of the coaxial cable, the antenna slot and the outer coating materials are all Teflon with relative permittivity equal to 2.1. The dielectric constant and conductivity of liver tissue at 37 °C are $\varepsilon_r = 43.03$ and $\sigma = 1.69$ S/m, respectively, at 2.45 GHz. We computed these values were computed using equation (1) and parameters in Table 1 [20]. The effective wavelength in liver tissue is approximately 18.5 mm.

$$\hat{\varepsilon}(\omega) = \varepsilon_n + \sum_n \frac{\Delta \varepsilon_n}{1 + (j\omega\tau_n)^{\varepsilon_n}} + \frac{\sigma_n}{j\omega\varepsilon_0}$$

(1)

where $\omega$ is the angular frequency [rad], $\hat{\varepsilon}$ is the resultant complex relative permittivity and $\varepsilon_0 = 8.854 \times 10^{-12}$ [F/m] is the permittivity of empty space. Table 1 gives all other
parameters of equation (1).

The SAR [W/kg] in tissue is calculated as a function of position as follows:

\[ \text{SAR} = \sigma |\overline{E}|^2 / (2\rho) \]  \hspace{1cm} (2)

where \( \sigma \) is the tissue conductivity [S/m] at the excitation frequency, \( \rho \) is the tissue density [kg/m\(^3\)] and \( \overline{E} \) is the spatially dependent time-harmonic electric field vector [V/m].

Fig. 3 shows the SAR pattern, normalized to the maximum value in the simulation region, in dB. Fig. 4 shows normalized SAR values along the longitudinal direction at a number of different radial positions. Figs. 3 and 4 indicate that the antenna SAR pattern is completely constrained by the sleeve and is localized in the region from the antenna tip to the end of the floating sleeve. Fig. 5 shows the SAR pattern with less antenna insertion depth. It shows that the localization of the SAR pattern is independent of the insertion depth as long as the sleeve is completely immersed in the liver.

\textit{B. Frequency sweep for antenna power reflection}

To validate our model as well as examine the power reflection characteristics of this antenna design, we fabricated an antenna and measured its reflection coefficient (\( S_{11} \)) spectrum from 0.5 to 10 GHz using a vector network analyzer (Agilent E8364A). We immersed the open end of the antenna in fresh bovine liver tissue. We revised the simulated antenna dimensions to match the exact fabricated dimensions and computed the \( S_{11} \) spectrum at discrete frequencies from 0.5 GHz to 10 GHz. At each discrete frequency, we adjusted the dielectric properties of bovine liver tissue in the model to account for the frequency dependence of the dielectric properties. We computed liver tissue dielectric properties over the frequency range of interest.
using equation (1)

Fig. 6 shows that the measured and computed results agree quite well. Fig. 6 shows that the antenna’s minimum reflection is near 2 GHz, off from the desired frequency of 2.45 GHz. We expected this, as the antenna was not designed to minimize the reflected power, but to obtain a good SAR pattern while maintaining a reasonably low reflection coefficient. The measured $S_{11}$ is –17.1 dB and the simulated $S_{11}$ is –18 dB at 2.45 GHz and we believe these are acceptably low for this initial design. Further optimization of the antenna could reduce this reflection further and permit tuning the null to 2.45 GHz. We note, however, that a small degree of detuning is expected in practice as the dielectric properties change during ablation.

C. Ex-vivo experiments and results

Ultimately it is the coagulated region produced by an antenna that determines its effectiveness so we performed ex-vivo ablations. We connected the floating sleeve coaxial antenna to a CoberMuegge MG0300D 300 W, 2.45 GHz microwave generator through a 1 m long flexible coaxial cable. We then carefully inserted the antenna into peripheral regions of fresh bovine liver to avoid heating near the largest blood vessels. We carefully monitored and recorded MW generator output power level and antenna power reflection levels. We then heated the liver using 120 W of power for 150 s. Initial liver temperature was 37 ºC. Note 120 W, while at the upper limit, is within the power handling capabilities of the UT-085 coaxial cable used.

After each experiment we sliced the liver tissues into either longitudinal cross sections or transverse cross sections. For longitudinal slicing, we placed a probe into the track created by the antenna and made a longitudinal transection of the lesion close to the inserted probe. We photographed tissue slices using a ruler for reference and scanned using an HP ScanJet 3970.
scanner at 200 dpi or higher resolution. Fig. 7 shows one of the ex-vivo experiment results. Here the lesion size is approximately $5.6 \times 3.7$ cm, when measured to the periphery of the “white zone”. This is a slightly different shape than predicted by the SAR pattern. However, the procedure of the thermal lesion formation is a complex combination of microwave energy absorption, heating conduction as well as possible tissue water evaporation, condensation and movement during the ablation period. Therefore the SAR pattern is only expected to be a guide to final lesion shape. The lesion does show a very well constrained tail as predicted by the SAR pattern. We created 16 lesions using 120 W of power for 150 s, to examine repeatability, mean and std deviation. Size of the lesions were $5.87 \pm 0.32$ cm by $3.64 \pm 0.33$ cm.

V. DISCUSSION AND CONCLUSION

Currently, the major limitation of MWA in treating larger liver tumors is the inability to deliver sufficient power to the tumor while minimizing detrimental heating to normal liver tissue outside the treatment region. We have developed a novel floating sleeve coaxial antenna for microwave liver tumor ablation, using both simulation and experimental measurements. This antenna is capable of providing very localized power deposition in hepatic tissue with minimal backward heating; as well as low power reflection, and high power throughput. The significant new feature of this antenna is the floating sleeve used to prevent the flow of electromagnetic energy along the coaxial applicator. The critical parameter seems to be the length of the sleeve. One other feature of this sleeve is its apparent ability to constrain the SAR pattern to the end of the sleeve allowing a certain amount of control over the shape of the SAR pattern.

While this study has yielded a qualitative understanding of the importance and effect of
the parameters of the sleeve. More studies are currently being performed to identify the reason for the superior performance of the floating sleeve in constraining the tail of the SAR pattern.

ACKNOWLEDGMENT

We would like to thank Dieter Haemmerich for his contribution in this study.

REFERENCES


Fig. 1: (a) schematic of the floating sleeve antenna and (b) cross section of the antenna at the sleeve.

dielectric, 1.676 mm
inner conductor, 0.512 mm
outer conductor, 2.2 mm
Teflon isolation layer, 2.5 mm
Floating sleeve, 3.2 mm
Teflon coating, 3.5 mm

(b)

Fig. 2: The axi-symmetric CEM model in the vicinity of the tip of the floating sleeve antenna. The vertical axis (r axis) corresponds to the radial direction while the horizontal axis (z axis) corresponds to the longitudinal axis of the antenna. The aspect ratio used in this diagram is nonphysical in order to show the details in the radial direction.
Fig. 3: Plot of normalized SAR on a dB scale. The SAR values are normalized to the maximum SAR value in the simulation region. For reference, the probe tip is at 0 mm, the slot is centered at 12 mm, and the sleeve begins at 22 mm and extends to 41 mm. The region of the simulated liver tissue is from -23 to 83 mm horizontally and 0 to 60 mm vertically. The boundary of the region of the liver tissue is not shown in the figure so SAR pattern near the antenna slot can be shown in better details. The antenna is inserted 70 mm deep into the liver, from the center of the antenna slot at $z = 23$ mm to liver tissue right side boundary at $z = 83$ mm.
Fig. 4: Plot of normalized SAR on linear scale as a function of $z$, at constant values of $r = 2.5, 5.0, 10.0, \text{ and } 20.0 \text{ mm}$. The SAR values are normalized to the maximum value in the plot.

Fig. 5: Plot of normalized SAR on a dB scale. Comparing to Fig. 3, the antenna is inserted 40 mm deep into the liver, from the center of the antenna slot at $z = 43 \text{ mm}$ to liver tissue right side boundary at $z = 83 \text{ mm}$. Boundaries of simulated liver tissue region are marked in dashed lines.
Fig. 6: Input reflection coefficient ($S_{11}$) for the floating sleeve antenna versus frequency.
Fig. 7: A photo of ablated bovine liver tissue. The lesion was created by applying 120 W for 150 s. The initial temperature of liver tissue was about 37 °C. The spacing between the markers on the antenna body was 1 cm. The lesion was about 5.6 × 3.7 cm and clearly localized to the active radiation region of the antenna. The lesion size was comparable to large lesion size reported by Strickland [21].

Table 1: Parameters for liver tissue for equation #1 [20].

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<th>Parameter</th>
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<td>$\tau_2$ (ns)</td>
<td>530.52</td>
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<tr>
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<td>$\alpha_2$</td>
<td>0.20</td>
</tr>
<tr>
<td>$\Delta\varepsilon_3$</td>
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</table>
Appendix 2

Expanding the bioheat equation to include tissue internal water evaporation during heating

1 Abstract

We propose a new method to study tissue water evaporation using an expanded bioheat diffusion equation. An extra term added to the bioheat equation is incorporated with the specific heat into an effective (temperature dependent) specific heat. It replaces the normal specific heat term in the modified bioheat equation, which can then be used even at temperatures where water evaporation is expected to occur. This new equation is used to numerically simulate the microwave ablation of bovine liver and is compared to experimental ex vivo results.

2 Introduction

Pennes’ bioheat equation, based on the heat diffusion equation, is a much used approximation for heat transfer in biological tissue [1-3]. Many publications have shown it is a valuable approximation [4, 5]. However, it does not include a method to account for the evaporation of tissue water which is expected to occur at tissue temperatures greater than 90 °C. Evaporation is one of a number of water related processes which are expected to be significant at higher temperatures. These include tissue water evaporation, diffusion, water vapor diffusion and condensation. Most tissue physical properties change with changes of tissue temperature and tissue water concentration. These processes may
be significant at high temperatures. In fact, they may become the dominant heat transfer processes in the system when tissue temperature approaches 100 °C. Without considering these processes, results from the bio-heat equation may significantly differ from experimental results at high temperatures.

The study of tissue heating processes at high temperatures is relevant to therapeutic applications (such as RF, microwave, and laser ablation and hyperthermia) and food processing applications (such as baking and frying).

Research on water evaporation related processes is sparse. Coupled heat transfer and liquid water transfer in porous material have been studied [6, 7]. Some work has been done for heat and mass transfer under laser radiation by considering tissue water evaporation on the heating surface [8-11]. Other studies include food processing, such as baking and frying, using external heating sources [12, 13].

Most studies of ablative procedures do not consider tissue water related processes, or have mentioned tissue water effects but do not include them in models, or only consider surface evaporation [14-16]. As not much study has been done, tissue water related processes are largely unquantified, and it is difficult to measure the physical effects or water movement and state changes.

Our expectation during heating to high temperatures is that tissue loses water through an evaporation process. Generated water vapor increases the gas pressure within the area of water evaporation. This water vapor diffuses to lower pressure areas in which tissue temperature is also lower. In this lower temperature region, water vapor condenses to water liquid and releases its latent heat. Released latent heat energy heats the surrounding tissue and increases tissue temperature. Tissue in this region also gains water
content during the condensation process. The entire process of water evaporation, water vapor diffusion and condensation is a process of water movement and energy movement and is as significant as direct thermal conduction.

None of the procedures can be easily studied quantitatively. Evaporation or condensation could be analyzed alone with partial pressure rules if the phenomenon occurred at a free interface of water liquid and air [17]. Under such a condition, the maximal amount of water vapor allowed in the air is equal to the water vapor saturated partial pressure at the current temperature and current air pressure. This rule does not apply directly to the situation of heating inside the tissue because the air pressure, water vapor diffusion rate and water liquid diffusion rate are largely unknown.

We present a new method to study tissue water related processes together with heat transfer processes for tissue at high temperature. We first map tissue temperature to changes of tissue water content caused by heating and evaporation. With such a mapping from temperature to water content, we define a new term—tissue effective specific heat—and use it instead of the normal tissue specific heat in the bioheat equation. The modified bioheat equation can be solved in the same way as the normal bioheat equation.

The rest of this paper proceeds as follows. A new term is added to the bioheat equation and a new effective specific heat term is presented in section 3 along with a description of the numeric simulation and experiment. Section 4 presents the results of the comparison between simulation and experiment and is followed by the conclusion.
3 Methods

3.1 Theoretical solution of tissue water evaporation with the bioheat equation

Below is the Pennes bioheat diffusion equation

\[ \rho C \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + Q + Q_b + A \]  

where \( \rho \) is density [kg/m³], \( C \) is specific heat [J/kg°C], \( T \) is temperature [°C], \( k \) is thermal conductivity [W/m°C], \( Q \) is the microwave power density [W/m³], \( Q_b \) is a term which accounts for the effects of perfusion (see equation 2) [W/m³], and \( A \) is the metabolic heat generation term [W/m³] which is considered insignificant with respect to the heating term and will be ignored for the purposes of this study.

\[ Q_b = \rho_b C_b \omega_b (T - T_b) \]  

where \( \rho_b \) is the blood mass density (kg/m³), \( C_b \) is the blood specific heat [J/kg°C], \( \omega_b \) is the blood perfusion rate [1/s] and \( T_b \) is the ambient blood temperature [°C] before entering the ablation region.

Note, all variables but t are spatially dependent. For purposes of clarity the spatial dependence is left out of the equations and is to be implied.

Evaporation requires energy, specifically termed the latent heat. To account for the energy needed to vaporize water we add a term to the bioheat equation, \( Q_E \) [W/m³], yielding a modified bioheat equation

\[ \rho C \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + Q + Q_b - Q_E \]  

(3)
Note, here we have dropped the metabolic heat generation term.

The power density used for evaporation is related to the change in water content of tissue as a function of time.

\[ Q_E = -\alpha \frac{dW}{dt} \]  \hspace{1cm} (4)

where \( \alpha \) is the water latent heat constant, which is 2260 [kJ/kg] and \( W \) is the tissue water density [kg/m\(^3\)] which is assumed to be only a function of temperature.

From the chain rule the derivative of \( W \) with respect to time is

\[ \frac{dW}{dt} = \frac{\partial W}{\partial T} \frac{\partial T}{\partial t} \]  \hspace{1cm} (5)

Substituting this into Equation 3, yields

\[ Q_E = -\alpha \frac{\partial W}{\partial T} \frac{\partial T}{\partial t} \]  \hspace{1cm} (6)

The modified bioheat equation then becomes

\[ \rho C \frac{\partial T}{\partial t} = \nabla \cdot k\nabla T + Q + Q_B + \alpha \left( \frac{\partial W}{\partial T} \frac{\partial T}{\partial t} \right) \]  \hspace{1cm} (7)

Pulling the last term in the above equation to the left hand side,

\[ \left( \rho C - \alpha \frac{\partial W}{\partial T} \right) \frac{\partial T}{\partial t} = \nabla \cdot k\nabla T + Q + Q_B \]  \hspace{1cm} (8)

Examining the above equation we can define an effective specific heat,

\[ C' = C - \frac{\alpha}{\rho} \frac{\partial W}{\partial T} = C - \frac{\alpha W_t}{\rho} \]  \hspace{1cm} (9)

which yields a new modified Pennes bioheat equation.
Equation 10 is in the same format as the original bioheat Equation 1, with an effective specific heat used instead of the normal specific heat. Since \( W'_T(T) \) is 0 when evaporation doesn’t occur and is negative when evaporation occurs, effective specific heat \( C' \) is never less than normal specific heat value \( C \) which is consistent with it requiring more energy to raise the temperature during a phase change.

Tissue effective specific heat [J/kg°C] is the only new term in equation 10. It has a similar physical meaning as the normal specific heat. It is defined for a unit mass of tissue as the amount of energy required to increase temperature by 1 °C, including the water latent heat energy required if tissue water evaporation occurs, which regular specific heat does not account for.

### 3.2 Numeric Simulation

To examine the effectiveness of this new specific heat we attempt to model microwave ablation of ex-vivo bovine liver. Fig. 1 shows the model geometry based upon the experimental setup (discussed subsequently). We used an axially symmetric model [15], which minimized computation time and allowed improved resolution while yielding a full 3D solution.

Both electromagnetic (EM) and thermal solutions were obtained. The EM solution was solved once and used as the heat source of the thermal model. While we expect the dielectric properties (and therefore the heating pattern) to change with temperature as well as loss of tissue water [18], for ease of computation, we assume that
the initial heating pattern remains unchanged during the course of ablation. Normal thermal properties (i.e. $c_p$, $k$, $\rho$) are assumed to be temperature and water content independent. Also, since we are simulating an ex-vivo case, blood perfusion is not included in the computer model. Boundary conditions are set to be convective boundary conditions with the heat transfer coefficient set to 12 and the ambient air temperature set to 25. Microwave and thermal properties for the various tissues and materials are listed in Table 1.

Fig. 1: The axially-symmetrical computer model shows that the coaxial slot antenna is in liver tissue. The liver tissue piece was placed on a plastic cutting board, having a plastic template on the top for the antenna and thermal sensor placement, and air on the outside. The horizontal axis is the $r$ axis. The vertical axis is the $z$ axis. The entire geometry is symmetrical along the $z$ axis, so only half of the geometry is drawn in the model. The radius of the coaxial antenna is 1.25 mm. It is inserted 20 mm deep into liver tissue.
To solve the modified bioheat equation we need the functional form of the temperature effective specific heat and therefore the temperature dependence of the water content. Based upon experiments which measured water content as a function of temperature, we have developed an equation, as equation 11, to define the water content and thereby the effective specific heat as a function of temperature [19]. The equation and its derivative are plotted in Fig. 2 (a) and (b).

\[
W(T) = \begin{cases} 
1 - \exp\left(\frac{T - 106}{3.42}\right) & \text{if } T \leq 103 \\
0.03713T^3 - 11.477T^2 + 1182T - 40582 & \text{if } 103 < T \leq 104 \\
\exp\left(\frac{T - 80}{34.37}\right) & \text{if } T > 104
\end{cases}
\]  

(11)

![Fig. 2: (a) Mapping from tissue temperature $T$ to $W$ [kg/m$^3$], the mass of tissue water per unit volume of tissue, (b) Derivative of mapping function from tissue temperature $T$ to $W$.](image)

Table 1: Parameters used in the model

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<th>Parameter name</th>
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<th>Parameter name</th>
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<td>Input power</td>
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<tr>
<td>------------------------</td>
<td>----------------------------</td>
<td></td>
<td></td>
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<tr>
<td>Specific heat</td>
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<td>volume percentage</td>
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**Copper**

<table>
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<tr>
<td>Specific heat</td>
<td>385 [J/kg°C]</td>
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**Teflon, cutting board and plastic template**

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<tr>
<td>Specific heat</td>
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The dynamics of water vapor movement and the issue of condensation are not well understood at this point. So to incorporate the transfer of energy that occurs when water vapor is generated then condenses in a new location, we have developed a simple mechanism to account for condensation effects. In this method:

- After each time step, we calculate the amount of tissue water vaporized in that time step.
- We assume no water vapor escapes from the system.
- We assume all water vapor diffuses to tissue region of lower temperature and condenses uniformly in tissue region of temperature from 50 to 80 °C.
- During condensation, water vapor releases latent heat energy and heats the surrounding tissue where condensation happens.
- We ignored the effects of thermal transfer between tissue and water vapor, assuming it was insignificant compared to the latent heat energy returned to the tissue during condensation.
After water vapor condenses back to water liquid, the amount of tissue water in the surrounding tissue increases. We assume that the amount of tissue water in region at temperature from 60 – 80 °C does not change as a function of condensation.

![Diagram](image)

**Fig. 3: Approximation of water vapor diffusion and condensation.**

To simulate the model we used the multiphysics simulation tool FEMLAB in conjunction with MATLAB. The functional form of the effective specific heat term is too complex for FEMLAB to incorporate into its calculations, so a call to a MATLAB function is used in its place. Therefore the solution proceeds as follows: \( T_n \) is used to calculate \( W_n \) and the effective specific heat. The bioheat equation is then solved in the solution region for \( T_{n+1} \). The entire computer simulation is illustrated in Fig. 4. For this model it took about 500 steps and 3.75 h to simulate a heating duration of 180 s. The average time step was about 0.4 s.
Tn

Calculate \( W_n \) and effective specific heat with MATLAB

Solving the heat equation in FEMLAB

Approximate water vapor condensation in MATLAB

\( T_{n+1} \)

To next cycle

Fig. 4: Schematic of solution procedure. Thermal simulation is performed in minor time steps. Time step \( \Delta t \) is adjustable during the simulation, from 0.2 s to 1 s. It is controlled so that maximal tissue temperature change is less than 10 °C and maximal tissue water content change is less than 1% of the liver tissue volume.

### 3.3 Experiment setup and procedures

We performed ex-vivo experiments on bovine liver to validate the computer simulation results. Whole cow livers were obtained from a local slaughter house and were kept refrigerated over night. Initial tissue temperature was approximately 8 °C. We selected the most uniform liver tissue with the fewest blood vessels for experiments. Liver tissue samples were a minimum of 4x12x12 cm. Fig. 5 shows that the coaxial slot antenna was inserted straight down from the top, together with four optical fiber thermal sensors feeding a Luxtron 1000 fluoroptic thermometer. We connected the antenna to a CoberMuegge MG0300D 300 W, 2.45 GHz microwave generator through a 1 m long flexible coaxial cable. The Luxtron thermometer was connected to a desktop PC via its
serial port for data collection. Temperatures were measured 4 times per second for the entire duration of microwave heating. Data were stored on the PC for further processing.

![Diagram](image)

**Fig. 5**: Ex-vivo experiment setup. The beef liver sample was approximately 4 x 12 x 12 cm. Through a plastic template, the antenna was inserted from the top to a position where the antenna slot was 2 cm deep. Four thermal sensors were all inserted 2 cm deep.

Liver tissue was heated with using 75 W of power for 180 s. Antenna and thermal sensors were withdrawn after the procedure. Actual positions of thermal sensors to the antenna slot were then measured after ablation. Tissue was cut open longitudinally into two halves along the trace of the antenna to expose the lesion. The lesion was scanned with a HP scanner at a resolution of 300 DPI to measure lesion size and shape.

### 4 Results and Discussion

Fig. 6 shows the antenna heating pattern plot for this configuration.

Fig. 7 compares computer simulation results of tissue temperature versus time, to ex-vivo experimental measurement results. The results of the simulation compare
reasonable well to the experimental results with similar trends in temperature profiles over the same approximate time range. A certain amount of mismatch between experiment and simulation is expected as the temperature dependence of the normal thermal properties is not considered nor is the change to the SAR pattern from changes in the complex permittivity due to temperature and tissue water changes.

Fig. 6: Plot of antenna heating pattern in dB scale. It is normalized by the maximal value.
Fig. 7: Tissue temperature at four thermal sensor positions. Curves without markers are measurement results from one of the ex-vivo experiments. Curves with markers are corresponding results from computer simulation. Sensor positions were measured as 2.5, 4.5, 7 and 9.5 mm away from the antenna respectively. All sensors were longitudinally aligned with the antenna slot.

Fig. 8 compares the lesion shape and size from experiments to temperature distribution from the computer simulation. The lesion in the experiment roughly conforms to the 80 °C contour line in the simulation. For scanned lesion image from in-vivo experiments, lesion boundary can be easily identified as a clear dark ring which is caused by increased blood perfusion at the boundaries. Lesion size is often associated with the 55 °C temperature contour of computer simulation. However in the ex-vivo case, it is usually very difficult to identify the lesion boundary because of the smooth tissue color changes. The relationship from the ex-vivo lesion color change to tissue temperature is not well known so lesion color cannot be translated to tissue temperature directly.
Fig. 8: (a) A lesion created in an ex-vivo experiment at the end of a 180 s MWA procedure. Lesion size is about 3.9 cm $\times$ 2.7 cm (b) Temperature contours from computer simulation at 180 s. Width of 2.7 cm roughly corresponds to 60 °C boundary.

Fig. 9 compares results from two different computer simulations, to results from experiments. Temperature results from only two sensor positions are used for clarity. Simulation #1 uses the original bioheat equation, which does not consider tissue water evaporation or condensation. Evaporation and condensation are considered in simulation #2 with the modified bioheat equation. As can be seen from the results, using the original bioheat equation (simulation #1), the temperature continues to grow, at 4.5 mm, even though experimentally and for simulation #2 it levels off. Also, at 9.5 mm the modified bioheat equation seems to match the experimental results better. This seems to indicate an improvement in accuracy when incorporating the effect of water evaporation and condensation in the form of effective specific heat over the original bioheat equation. However, it is also possible that one of the temperature effects such as changing normal thermal properties or change SAR pattern would also improve the simulation to more
closely match experimental results if they were added in place of the evaporation effects. This question is currently under investigation as we incorporate these effects into our model.

![Comparison of measurement and simulation results](image)

Fig. 9: Comparison among the measurement results and two different computer simulation results. Simulation #1 is uses the original Bioheat equation, which does not consider tissue water evaporation and condensation. Simulation #2 uses the modified heat equation to consider tissue water evaporation and includes condensation. The comparison is done for thermal sensors which were 4.5 mm and 9.5 mm away from the antenna.

## 5 Conclusion

We have presented a new modified bioheat equation which incorporates the effect of water evaporation from the tissue in the form of an effective specific heat. We have performed experiments to validate a model of microwave ablation of bovine liver using this new bioheat equation. Comparing the simulation results to the experimental results,
the new method has shown promise in generating a more accurate prediction of temperature profiles. Future studies will focus more basic research to understand how water changes state and moves in tissue during heating. This will allow a better understanding of the evaporation processes in tissue during heating to high temperatures as well as yielding a more complete understanding of the other processes which may be important during tissue heating such as water vapor diffusion, and condensation.

6 References


Appendix 3

Measurement and Analysis of Tissue Water Content
during Microwave Tissue Ablation

Deshan Yang, Mark C. Converse, David M. Mahvi, MD, John G. Webster

1. Abstract

Due to high temperatures in the liver during microwave ablation (MWA), one expects a significant amount of tissue water movement and corresponding energy movement during the process of ablation. This process is a function of tissue water related phenomena including evaporation, water vapor diffusion and condensation. Understanding such mass and energy movements and their effects on heat transfer, normal thermal parameters (i.e. thermal conductivity, density, etc.), and SAR patterns may be critical to improving control of the MWA process. To begin to understand the dynamics of water movement during ablation, we measured the water content of ablated tissue lesions. The results demonstrate significant tissue water content changes and lead to a better understanding of tissue water movement.

2. Introduction

Microwave ablation (MWA) is a promising technology for the treatment of hepatic tumors. The goal of MWA is to destroy the tumor along with a 1 cm margin of normal hepatic tissue. This technology has been used in both intra-operative and
percutaneous approaches for primary hepatocellular carcinoma and hepatic metastasis of colorectal carcinoma [1-3].

Because the heating mechanism of MWA is different from radiofrequency ablation (RFA), higher temperatures within the liver are achievable during MWA. Preliminary studies have shown that MWA was able to heat liver tissue to temperatures over 125 °C, compared to approximately 95 °C with RFA. As tissue water evaporation can start as low as 70 °C [4], this means that water movement and evaporation may be a significant issue during MWA.

During the course of heating, tissue will lose water content, due to the generation of water vapor and/or by the diffusion of liquid water from the treated cells. If a significant amount of evaporation takes place, gas pressure will increase and the vapor will diffuse into lower pressure areas where the temperature is lower. One would expect this water vapor to condense in this lower temperature region adding energy and increasing water content. Some tissues thus gain water content and heat energy during the condensation process while others lose water content and heat energy during the evaporation process. The entire process of water evaporation, water vapor diffusion and condensation is a process of water movement and energy movement and may be as significant as direct thermal conduction [5-7].

As well as adding a new heat transfer mechanism, the movement of water and specifically the loss of water in some volumes of tissue are expected to affect other tissue properties. Both tissue dielectric properties and thermal properties are directly related to tissue water content [8-10]. Changes in tissue thermal properties will directly affect the heat conduction within tissue and changes in tissue dielectric properties are expected to
cause the antenna heating pattern to change.

We have found no existing study that provides quantitative measurement or analysis of tissue water content during the course of MWA of tissue. To examine how the water content changes, we measured the water content distribution in ablated tissue. Our initial results confirmed our expectation that there are significant changes in tissue water content during MWA.

3. Experimental setup

All experiments were performed ex-vivo with bovine liver, which was obtained from a local slaughter house. Liver tissue was refrigerated overnight and the initial temperature at the beginning of the experiments was approximately 6 °C. Initial water content was about 73% by weight, measured with a wet-dry procedure on larger tissue pieces to reduce measurement errors.

A homogeneous block of liver tissue, free of major blood vessels, was selected. The coaxial antenna was inserted into the tissue and connected to a MW generator through a 1 m long flexible coaxial cable. A lesion was created after 75 W of MW power was applied to the liver tissue for a controlled duration. After the antenna was smoothly drawn out, the liver tissue was cut into 2 pieces along the trace of the antenna insertion. A 5 mm thick slice was cut from one of the two pieces, at a distance 5 mm from the slot, perpendicular to the antenna insertion direction, as shown in Fig. 1(a). A strip of tissue was then cut from the slice and cut into 1 mm thick pieces, as shown in Fig. 1 (b). The plane 5 mm above the slot was chosen as this is the position along the axis of the antenna with the largest lesion radius, based upon simulation and experimental observation.
Weights of all tissue pieces were measured with an Ohaus Explorer E11140 digital analytical balance with 0.1 mg resolution. Tissue pieces were vacuum-dried overnight with an Edwards ETD4 Tissue Dryer, and weighed the next morning. All measurements were performed as quickly as possible to minimize the effects of evaporation from the surfaces of the tissue.

The mass fraction of tissue water remaining after ablation, \(w_t\), is calculated as

\[
    w_t = \frac{M_w - M_d}{M_B} = \frac{0.27(M_w - M_d)}{M_d}
\]

where, \(M_w\) is the measured tissue weight before drying, \(M_d\) is the measured dry tissue weight, and \(M_B\) is the weight before ablation. \(M_B\) is calculated as

\[
    M_B = \frac{M_d}{0.27}
\]

where we assume that the mass fraction of water and solid material of normal tissue was 73% and 27%, respectively.

4. Results

Fig. 2 shows a cross section of a lesion created using 75 W power for 6 min. The lesion is smaller than usually seen for this antenna and power due to the low initial temperature. A slice transverse to this plane, cut approximately 5 mm from the slot, is used in the measurement of tissue water content.

We performed water content measurements on lesions created by MWA with durations of 1, 2, 3, 4, 5, 6, 9 and 12 min calculating \(w_t\) for each case. We performed at least 4 measurements for each duration. Fig. 3 shows the mean results and standard
deviations.

5. Discussion and conclusion

We used the ANCOVA (Analysis of Covariance) statistical model to analyze the measured data and show statistical significance. The results show that tissue water content changes significantly \((p < 0.001)\) with ablation duration by controlling the tissue piece position. Tissue water content also changes significantly \((p < 0.001)\) with tissue piece position by controlling the ablation duration. The statistical results are consistent with the plots.

The results suggest that microwave ablation has several effects on tissue water properties. First, tissue is desiccated to a greater extent near the antenna. This is to be expected both from an observation of the lesion and from a basic understanding of the SAR pattern (i.e. where energy is deposited) and heat transfer. Second, even after only 1 min there is significant water loss in the tissue up to 3 mm away from the edge of the antenna. This illustrates the volume heating effect we expect from MWA. Microwaves emitted from the antenna propagate through the tissue, resistively heating the tissue as the wave attenuates. This leads to a direct deposition of energy in a larger volume of tissue than seen during RF ablation. Third, the change in water content seen early and over a significant volume can be expected to affect both the thermal and electromagnetic properties. This may affect them enough to significantly alter the SAR pattern and/or the dynamics of heat flow that occurs in conjunction with the direct deposition of energy due to the microwaves. Finally, water content drops to near 0\% for tissue very close to the antenna for durations of greater than 6 min. This indicates charring of the tissue near the
Regarding measurement errors: Results of the water content measurement above 50% mass fraction are not presented here due to water drip from cut tissue. During the cutting stage of the experiment we noticed a significant loss of water from the higher water content tissues pieces. This led to a discrepancy between the measured values of the high water content (normal) tissue outside the lesion and our bulk water content measurements taken for calibration purposes. Water loss was only observed in tissue pieces with higher than 50% of water content, therefore measurements on such tissue pieces were unreliable and are not presented in Fig. 3. For tissue pieces of lower water content, below 0.5, no visible water loss was detected, nor was expected, so the measurement results are considered acceptable. For the purposes of understanding water related phenomena near the antenna these are clearly the most significant.

Cutting and weighing of the tissue after ablation took approximately 1 to 3 min. During this delay, tissue water may have diffused from a high water content area to low water content area—the highly ablated region. The overall effect would be to artificially increase the tissue water content near the antenna body, and artificially decrease the water content near the lesion boundary. The estimated water diffusion constant in normal tissue is approximately $10^{-9}$ to $10^{-10}$ m$^2$/s [11]. Diffusion rates may be even higher in the highly ablated area because the tissue is porous. Nevertheless our data do not suggest that the amount of diffusion was quantitatively significant and initial computer simulations utilizing the base water diffusion constant support this hypothesis.

This study utilized one antenna structure with one set of operating parameters. We expect that for different antennas structures (and therefore different heating patterns) and
for different operating parameters, the water content profiles may be different. But, we expect the levels of tissue water content to be comparable.

We have presented a measurement of water content in liver during the course of a MWA procedure. We see significant changes in the water content within the lesion created and expect these changes to have significant effects on how energy is deposited in the tissue and how heat transfers within that tissue during MWA. These results will improve the ability to model MWA and subsequently develop more effective cancer treatment devices.

6. References:

Fig. 1: (a) Schematic of the ex-vivo experimental setup. The master slice of 5 mm thick was cut at 5 mm above the antenna slot. (b) Illustration of the distribution of 1 mm thick samples cut from the master slice.

Fig. 2: Ablated bovine liver tissue. The lesion was created by applying 75 W for 6 min. The maximum lesion radius is located at about 5 mm proximal from the slot in the longitudinal direction.
Fig. 3: Plots of post ablation water content mass fraction $w_r$, calculated according to equation 1. Mean results from multiple measurements for the same duration are plotted as solid or dashed lines. The vertical bars represent the standard errors at the corresponding data points. The data points are plotted at the position where the center of each tissue piece is located on the lesion strip. Position value is the distance from the edge of the antenna insertion hole to the center of the tissue piece. 0 is at the edge or the surface of the antenna when the antenna is in the tissue.
Appendix 4

Measurement and Analysis of Tissue Temperature during Microwave Liver Ablation

Deshan Yang, Mark C. Converse, David M. Mahvi, John G. Webster

1 Abstract

We measured tissue temperature changes during ex-vivo microwave ablation (MWA) procedures for bovine liver tissue. Tissue temperature increased rapidly at the beginning of the MW power application. It came to a plateau at 100 to 104 ºC before it increased again. We divided the changes of tissue temperature versus time into four phases. They suggested that tissue temperature changes may be directly related to tissue water related phenomena during MWA, including evaporation, diffusion, condensation and tissue water composition. Changes of tissue temperature versus position in lesion showed a similar plateau at 100 to 104 ºC. We also studied the relationship of tissue water evaporation and tissue temperature by mapping temperature to remaining tissue water after ablation. An additional analysis indicated that lesion boundary was at about 50 - 60 ºC temperature.

2 Introduction

Microwave ablation (MWA) is a promising technology for the treatment of hepatic tumors. The goal of MWA is to destroy the tumor along with a 1 cm margin of normal hepatic tissue. This technology has been used in both intra-operative and
percutaneous approaches for primary hepatocellular carcinoma and hepatic metastasis of colorectal carcinoma [1-3].

MWA technologies have been investigated for many years. Much of the previous research has focused on the design and optimization of the antenna applicator utilizing predominantly electromagnetic (EM) computer simulations and experimental measurement of EM fields and final lesion size. Less prevalent in the literature are temperature measurements and computer thermal models. This may be due to the complexity of heating mechanisms and difficulties with measuring tissue temperature and other relevant variables during MWA procedures. While more effort has been put into thermal modeling of radiofrequency ablation (RFA), the heating mechanisms of MWA are different from RFA and are likely more complex due to higher temperatures seen with MWA. Preliminary studies have shown that MWA was able to heat tissue to temperatures higher than 125 °C, much higher temperatures than RFA. As tissue water evaporation can start as low as 60 to 70 °C [4], tissue water movement and evaporation may be more significant during MWA because of the higher temperature.

We present results of temperature measurements during MWA and attempt to explain gross tissue changes based on temperature profiles and the results of a previous water content study [5]. Our initial results have revealed a few interesting features that were related to heating mechanisms and expected tissue water processes.

3 Experiment setup

We used a Luxtron Fluoroptic thermometer model 3100 to measure tissue temperature during MWA of ex-vivo bovine liver tissue. A fluoroptic thermometer was
selected because its fiberoptic temperature sensors are unaffected by microwave radiation and have minimal affects on the SAR pattern of the antenna.

All experiments were performed ex-vivo with bovine liver which was obtained from a local slaughter house. Liver tissue was refrigerated overnight and the initial temperature at the beginning of the experiments was approximately 6 - 9 °C. Homogeneous blocks of bovine liver, at least 8x8x5 cm, were used for the experiment. Pieces of such size were large enough to accommodate the lesion for a 6 min microwave treatment.

Fig. 1 shows the placement of the microwave coaxial antenna and fluoroptic temperature sensors. In addition to the antenna, fiberoptic sensors were inserted into the liver tissue guided by 14 gauge (2.1 mm diameter) biopsy needles through holes in a 1 cm thick plastic template. This ensured that the antenna and the needles were placed parallel to each other in the tissue. The antenna was inserted 7 cm deep into the tissue, referenced to the antenna tip. The needles were 15 cm long, with a needle guide and a 15 gauge (1.83 mm in diameter) stylet. The needle guide and the stylets were pushed together through the liver block, and then the stylets were pulled out. Temperature sensors, each paired with a section of thin plastic fiber (0.5 mm), 12 mm long, were introduced into the needle guides. The temperature sensor fibers were inserted to a position 5 mm proximal to the antenna slot. This is the axial location of maximum power deposition and lesion radius, (Fig. 2 and Fig. 6 a). The plastic fibers were inserted 2 cm beyond the temperature sensor fibers. The needle guides were then withdrawn from the tissue on the template side while the fiberoptic temperature sensors and the plastic fibers were left in the liver tissue.
Fig. 1: The fiberoptic temperature sensors are placed 5 mm away from the antenna slot in the longitudinal direction, and various distances away from the antenna body in the radial direction.

The temperature sensors were positioned at various distances away from the surface of the antenna body. Each was connected to a Luxtron 3100 thermometer, which was connected to a PC via its serial port for data collection. Temperature values from all sensors were updated by the thermometer 4 times a second and were recorded by the PC for the entire ablation procedure. After ablation, the temperature sensors and the antenna were removed, leaving the plastic fiber sections behind to mark the temperature sensor locations. Tissue was then cut open with scissors, perpendicular to the antenna insertion direction, along the marked line in Fig. 1. During cutting, the plastic fibers were kept together with the tissue. The distances from each fiber position to the near edge of antenna hole were measured on the cross section. Temperature sensors were assumed to be at the same positions as the plastic fibers.
Fig. 2: Antenna SAR plotted in dB scale from EM computer simulation. We use a coaxial-based sleeve antenna introduced in [6], which creates a localized SAR pattern. The longitudinal position of the antenna slot is at 10 mm. The longitudinal position of the maximum SAR is at approximately 17 mm.

4 Results

Fig. 3 shows our measurement results. The shape of the curves in Fig. 3 shows that we can separate the temperature curve into four regions, labeled 1, 2, 3, 4 on the graph. What separates the regions are the temperature ranges and shape of the curve. Region 1 of the temperature curve is slightly concave, starting at the initial temperature and ending between 70 to 90 ºC. Region 2 is convex and appears to be a transition region between regions 1 and 3. In region 3 the temperature curve becomes almost flat and slowly increases from 100 ºC to 104 ºC. After the temperature reaches 104 ºC, temperature increases quickly again in region 4. Most curves showed these different regions although the slopes changes as distance from the antenna increased and the transition region 2 was less sharp.
Fig. 3: Plots of measured temperature versus time. The MWA procedure was performed with 75 W for 360 s. Sensors were 5 mm above the antenna slot longitudinally. Temperature curves are labeled by the distance of the temperature sensors to the antenna surface, measured after heating.

Fig. 4: Plot of measured tissue temperature versus the measurement positions for different ablation durations. Data points are derived from the same measurement results used in Fig. 1.

Fig. 4 shows measured temperature data as a function of position for various times during ablation. The interesting feature here is the flat spatial region, corresponding to the 100 to 104 ºC plateau seen in the time domain plots. After 3 min this plateau is
seen from 4 to 6 mm away from the antenna.

Tissue water content versus ablation duration was measured in previous experiments [5] and is combined with the temperature measurements in Fig. 5. In Fig. 5, values of tissue water content are the ratios of the remaining tissue water mass, after ablation, to the normal tissue mass before ablation. According to bulk measurements, normal tissue had about 73% water content. Fig. 5 shows that, for all locations shown, as tissue temperature increases to 100 °C, water content drops to 45% to 50%. For locations near the antenna (2, 3, and 3.5 mm), as the temperature reaches 104 °C, water content drops to about 30% to 35%. Note that for locations 4 and 5 mm, although the temperature does not reach 104 °C, water content at those locations eventually drops to 35%.

![Fig. 5: Tissue water fraction and measured temperature versus ablation duration. Curves are labeled by the distance of the temperature sensors to the antenna surface. Curves without markers correspond to plots of tissue temperature. Curves with markers are plots of tissue water content with the markers being data points.](image-url)
Fig. 6: Ablated bovine liver tissue. (a) Longitudinal section, the maximum lesion radius is located at about 5 mm from the slot along the longitudinal direction. (b) Cross section. Both lesions were created separately by applying 75 W for 360 s. The lesion sizes were smaller than reported for this antenna due to a low initial tissue temperature of 8 °C. Lesion size is approximately 26 to 28 mm in diameter.

As part of the measurement of tissue temperature during ablation we measured the temperature at the lesion edge. Fig. 6 shows a scanned photo of ablated tissue. The lesion is identified as the region in which tissue color changes from dark red color to light red color. In Fig. 6 this is the transition from the darker to lighter region. The antenna hole is approximately 3.5 mm in diameter. Based upon 9 experiments, the length from lesion boundary to the edge of antenna hole had a mean value of 14.5 mm and standard deviation of 0.9 mm. The mean lesion diameter was 32.5 mm. Fig. 4 shows that tissue temperature was 60 °C at 13.5 mm after 6 min ablation. This suggests that ex-vivo liver lesion boundary had a temperature of 55 to 60 °C. This is very similar to the 50 °C, determined in [7] for RF ablation.
5 Discussion

5.1 Differences between regions

Tissue temperature during microwave ablation is a complex function of the direct deposition of microwave energy, surrounding spatial temperature gradients, thermal properties (which are a function of water content), evaporation, and water mass transfer. From plots of temperature during ablation it is difficult to determine which phenomena contribute most to the change in temperature. However from the temperature curves and associated water content we can make some observations and forward some hypotheses.

Considering how the temperature changes as a function of time, we see initially, a steep increase in temperature with time. In this region the temperature increases more slowly than expected if we only consider the heating due to direct deposition of microwave energy. This could imply that more energy is lost due to heat conduction away from the region than gained due to heat conduction into the region, which may be a function of the cylindrical nature of the problem.

For points closer to the antenna we see a slight increase in slope of the temperature plot from 30 to 90 s. Comparing plots of temperature at different positions in Fig. 3, we see that at the time when the slope changes, points closer to the antenna have reached temperatures approaching 100 ºC. Fig. 5 shows some loss of water in those regions. This water is likely moving outward radially and could be contributing to radial heat transfer, due to diffusion of high temperature liquid water or vapor. Another possibility is that as the water moves outward, the dielectric and thermal properties of the tissue close to the antenna change. This would affect thermal conductivity or increase the
direct deposition of energy to the lower temperature regions as less energy is absorbed in the higher temperature regions due to loss of tissue water.

As the curve approaches 90 to 100 ºC, the rate of temperature change decreases. Fig. 5 shows that water content begins decreasing. Either water is diffusing away from this region or likely some evaporation is beginning to occur. Ultrasound measurement during ablation has shown anomalies attributed to air bubbles [8, 9]. As energy is required to convert liquid water to vapor, less is available to increase the temperature. The temperature curve eventually flattens out and slowly increases from 100 to 104 ºC, in conjunction with this plateau the water content begins to drop significantly. This lends credence to the idea of water evaporation. Note, due to the complex structure of tissue and the way that water is distributed, as well as what is likely a relatively slow diffusion rate of water vapor from the region, we would not expect the temperature to remain at a ‘boiling’ temperature until all water is evaporated.

After the plateau at approximately 100 ºC, the temperature again begins to rise. At this point the mass fraction of water has dropped to approximately 30 to 35% of the original tissue mass. Thus approximately half of the water is left in the tissue. This may be intracellular water or water that is somehow bound to the tissue and requires significantly more energy to evaporate. Temperature continues to rise until it is beyond the range of our measurement tools.

5.2 Tissue temperature versus tissue water content

We expect tissue water content to play a significant role in MWA. To include the effects of water evaporation and movement into a thermal model of MWA we have
attempted to create a function that correlates tissue temperature to water content.

Ramachandran et al. measured water evaporation from heated tissue [4]. For liver tissue, their results showed that tissue water evaporation started when tissue temperature reached 70 °C and approximately half of the tissue water content was lost by the time the temperature reached 104 °C [4]. Haemmerich et al. introduced a new method to measure specific heat, which reduced the effects of surface evaporation [10]. This method utilized larger liver tissue pieces and is more consistent with the environment seen by the liver during MWA. Their results indicated that the liver tissue began losing water at 80 °C. The results also showed that there is a 17% increase in tissue specific heat by 83.5 °C and 15% tissue mass reduction by 92 °C. In [11] we introduced an effective specific heat, in which the latent heat of evaporation is incorporated into a temperature-dependent specific heat. The change in specific heat at high temperature Haemmerich et al. observed, is consistent with our effective specific heat, which could imply that this change is due to water vaporization. Their observation of 15% tissue mass reduction seems too high. We think such mass reduction was mainly caused by loss of tissue water from the blood vessels under the gas pressure of water vapor, rather than caused by tissue water evaporation.

As the method proposed by Haemmerich et al. is more consistent with the environment seen by the liver during ablation, we incorporated his conclusions and our experimental measurements into a water content function based upon tissue temperature as below.
where $T$ is tissue temperature [°C] and $W$ is ratio of remaining tissue water mass to the mass of normal tissue before heating.

Fig. 7 shows the proposed functional relationship between tissue temperature and tissue water evaporation, along with our measurements of water content as a function of temperature from [5]. Fig. 7 shows that for temperatures less than 80 °C, tissue water content is 73%, which is the value we measured for normal liver. At 80 °C the water content begins to drop quickly to about one-half by 104 °C, and then exponentially decays as the temperature get higher. This exponential decay is relatively arbitrary as we have little data above 120 °C and have found nothing in the literature.

\[
W(T) = \begin{cases} 
0.778 - 0.778 \times \exp\left(\frac{T - 106}{3.42}\right) & T \leq 103 \\
0.0289T^3 - 8.924T^2 + 919.6 \times T - 31573 & 103 < T \leq 104 \\
0.778 \times \exp\left(\frac{T - 80}{34.37}\right) & T > 104 
\end{cases}
\]

(1)
Fig. 7: Remaining mass of tissue water versus tissue temperature after ablation. Tissue water and tissue temperature from matched ablation durations and positions in lesion are plotted as thin solid lines with asterisks. Only data points lower than 0.5 were used to create the approximation line because data points higher than 0.5 were not reliable [5]. The dotted line is the approximation.

Normal tissue has about 73% water content in mass.

6 Conclusion

This study presents results of tissue temperature measurement during MWA. Combined with previously obtained tissue water content measurement data, our initial results indicate a relationship between tissue temperature and tissue water content. Based on this perceived relationship a function to related water content to tissue temperature is presented. Although performed for a specific set of conditions and antenna geometry useful insights were obtained and may be applicable to a broader range of conditions.
7 References:


Appendix 5

Computer Simulation of Microwave Liver Ablation

Deshan Yang, Mark Converse, David Mahvi, John Webster

This paper is still in preparation.

The content is covered entirely by chapter 8.
Appendix 6

MATLAB and FEMLAB programs for the comprehensive computer models

The MATLAB and FEMLAB programs are available on CD-ROM media.