Computational modeling of microwave ablation and hyperthermia: applications to device design and treatment planning

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http://ece.ksu.edu/bcdl/
• Relevant features for tissue heating:
  – Radiative applicators, direct tissue contact not required
  – Microwaves can propagate through charred tissue
  – Rapid heating of large tissue volumes
Tumor ablation

Ablation target: tumor + margin

- Microwave ablation in use for treatment of tumors in liver, kidney, lung, bone, and other organs
- Non-oncologic applications: cardiac arrhythmias; benign prostate enlargement; menorrhagia

(Dodd et al., Radiographics, 20(1), 2000)
Components of MW ablation systems

- **Energy source:** signal generator + amplifier
  - 915 MHz and 2.45 GHz most widely used
  - > 50 W
- **Applicator:**
  - Feedline cable and radiating antenna
  - Cooling circuit
- **Treatment time:** ~5-15 min
- **Image-guidance**
  - X-ray CT, ultrasound are the most widely used
  - Post-treatment verification: contrast-enhanced imaging
Role of computer modeling

• Device design

• Energy delivery strategies:
  – Applied power levels, treatment time
  – Pulsed vs. continuous operation
  – Feedback control

• Patient-specific models
  – Treatment planning and optimization
  – Post-treatment assessment
Microwave energy deposition

• Maxwell’s equations describe the propagation of microwave energy radiated within tissue
• Electric field profile and tissue conductivity determine the energy deposited within tissue

\[ Q_s [W/m^3] = \frac{1}{2} |E|^2 \]

\[ SAR [W/kg] = \frac{|E|^2}{2} \]
Microwave heating

Complex permittivity: \* = \j

Effective conductivity: \eff = s + 0

(Brace, Curr Probl Diagn Radiol, 38(2), 2009)

An applied microwave field causes water molecules to rotate billions of times per second, which leads to heat generation.
Bio-heat transfer

Transient temperature

Thermal conduction term

\( c \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + Q_S \)

Heat-sink due to blood perfusion

\( \tilde{m}_{bl} c_{bl} (T - T_{bl}) \)

Energy deposited by heat source

Inner zone: Direct energy deposition

Outer zone: Thermal conduction

Ablation device

Blood perfusion
Modeling workflow

Specify geometry/import from CAD

Create mesh

Assign material properties, boundary conditions

Solve for electric field

Solve for temperature

Update material properties and treatment parameters

Ablation duration complete?

Yes

Compute thermal damage maps

No
Electrical properties of tumors and diseased tissue (e.g. cirrhotic liver) may be considerably different than those of healthy tissue.
Temperature dependent electrical properties

(Ji and Brace, Phys Med Biol, 56(16), 2011)
Blood perfusion and thermal properties

- Considerable differences in perfusion rates between tumor and healthy tissue
- Large changes in thermal properties due to water loss at elevated temperatures

(Schutt and Haemmerich, Med Phys, 2008)
Interstitital antenna characterization

Ideal ablation applicator:

- Low reflected power at operating frequency; large bandwidth
- Low sensitivity to tissue electrical properties
- Creates large volume ablation zones in short duration
- Minimal thermal damage to non-targeted tissue

(McWilliams et al., IEEE TBME, 2015)
Selecting coaxial cables

- Percutaneous use, 13 G (~2.4 mm) or smaller is desirable
- Flexible vs. rigid
- Smaller coaxial cable:
  - Less invasive; lower power handling capacity; more lossy
- Larger cables:
  - More invasive; greater power handling capacity; less lossy

\[ R = \frac{R_i}{2} \left( \frac{1}{a} + \frac{1}{b} \right) \]
Interstitial antennas for ablation

- Monopole antenna
- Asymmetric dipole
- Slot antenna
- Monopole antenna with choke
Modeling tips

- Simplify geometries by identifying planes of symmetry
- Employ appropriate boundary conditions to limit extents of simulation region
- Ensure adequate mesh resolution and quality in critical regions
- Define desired accuracy and check for convergence
- Validate with experiments
Model definition

- MW input
- Device cooling (e.g. convective flux)
- \( T = T_{\text{inf}} \) or adiabatic boundary condition
- Axis of symmetry
- Absorbing boundary conditions (e.g. PML)
Impedance matching vs. monopole length

Reflection coefficient vs. frequency

$S_{11}$ [dB] vs. Frequency [GHz]

- 10 mm
- 12 mm
- 14 mm
- 16 mm
- 18 mm
- 20 mm

Monopole length
Monopole SAR and temperature profiles

30 W, 450 s ablation @ 2.45 GHz
Liver tissue
Perfusion = 10 kg/m$^3$/s

Diameter = 2.5 cm
Axial ratio = 0.45
Monopole antenna with choke

Diameter = 2.9 cm
Axial ratio = 0.53

log (SAR [W/kg])

\[ \lambda/4 \]

15 mm

20 mm

T [°C]

\[ \Delta 113 \]

\[ \Delta 6.36 \]

\[ \Delta -26.7 \]

\[ \Delta 110 \]

\[ \Delta 100 \]

\[ \Delta 90 \]

\[ \Delta 80 \]

\[ \Delta 70 \]

\[ \Delta 60 \]

\[ \Delta 50 \]

\[ \Delta 40 \]
Cooled monopole with choke

- Diameter = 2.8 cm
- Axial ratio = 0.70
- $T_{\text{cool}} = 25 \, \text{C}$

Uncooled

Diameter = 2.8 cm
Axial ratio = 0.70
Dynamic tissue properties during heating affect SAR pattern and antenna impedance match.
Dynamic tissue properties

Discrepancies in temperature profiles simulated with models employing static vs. dynamic tissue properties
Models and experiments indicate more rapid heating to greater tissue temperatures with 2.45 GHz dipole antennas compared to 915 MHz.
Sensitivity to tissue properties

- Assess sensitivity of model outputs to tissue physical and physiological properties
- Facilitates design of robust devices
Model of prostate ablation with a directional interstitial ultrasound ablation device

(Prakash and Diederich., Int J Hyperthermia, 28(1), 2012)
Patient-specific model: GYN hyperthermia

Model-based planning for ultrasound hyperthermia treatment following HDR brachytherapy

(Wootton et al, Phys Med Biol, 56(13), 2011)
Patient-specific modeling considerations

• Segmentation
  – Large blood vessels

• Assignment of tissue properties

• Meshing at interfaces

• Uncertainty in model predictions
  – Identify regions at risk of inadequate treatment

• Use time-temperature relationships to compute cell death
Challenges and opportunities

• Broadband temperature dependent changes in dielectric properties
• Thermal properties at $T > 100 \, ^\circ C$
• Characterization of heat transfer due to water vapor movement
• Characterization of tissue volume changes during ablation
• Identification of patient-specific tissue physical and physiological properties
Hyperthermia

• Moderate tissue heating (40-44 °C), for long duration (~mins – hours)
  – Increased metabolism, oxygenation, radiosensitization, chemosensitization, rupture cell membranes, hyperemia
  – Stimulation of anti-tumor immune response
  – Localized, heat-triggered release of therapeutic agents
  – Cellular repair mechanisms can’t keep up with accumulating damage
Wearable breast hyperthermia applicator

- Treatment of intact breast
- Conformal and wearable
- Patient customizable
- Antennas may be implemented with printed ink
- Improved energy deposition
  - 10 times larger treatment volume
  - 28% reduced power requirements
- Reduced scattered E-field

Breast hyperthermia system concept
2D view of the proposed antenna with flared groundplane and breast model

Antenna Optimized dimensions:
$W = 3.9 \text{ mm}$, $L = 13.7 \text{ mm}$
$h_1 = 2.65 \text{ mm}$, $h_2 = 5 \text{ mm}$
$BD = 40 \text{ mm}$, $TD = 123 \text{ mm}$

(Curto and Prakash., Int J Hyperthermia, In Press, 2015)
Simulated E-field and temperature profiles

E-field profile Single applicator

Temperature profile Single applicator

(Curto and Prakash., Int J Hyperthermia, In Press, 2015)
$S_{11}$ vs. frequency: model and measurement

(Curto and Prakash., Int J Hyperthermia, In Press, 2015)
13 element antenna array

Model-based array design

\[ AP = \frac{\text{Power absorbed in target}}{\text{Power absorbed in all tissues}} \]
Four-element hyperthermia array

Multi-antenna systems afford some spatial control of treatment zone

Uniform amplitude and phase

Phase adjustments to shift heated zone:
- Antenna 1: 84.5°
- Antenna 2: 84.5°
- Antenna 3: 123°
- Antenna 4: 0° (reference)
Experimental characterization

T3 located in the breast center
T2 at 10 mm from center
T1 and T4 at 22.5 mm from center

Antennas with equal phase Focusing in the center
Antennas with phases optimized to focus in the right lobe of the breast

T3 located in the breast center
T2 at 10 mm from center
T1 and T4 at 22.5 mm from center
Summary

• Computer models provide a powerful and convenient tool for design, optimization, and evaluation of microwave tissue heating devices and treatment strategies

• Important to validate modeling results with experiments

• Knowledge of patient-specific properties and dynamic changes during heating will yield improved results
• Review papers on modeling MW tissue heating:
  – *Int J Hyperthermia* (special issue), Vol. 29, No. 4, 2013; papers on modeling MW hyperthermia and ablation


• Web-based tool for tissue dielectric properties: http://niremf.ifac.cnr.it/tissprop/htmlclie/htmlclie.php