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Contribution of Direct Heating, Thermal Conduction and Perfusion during Radiofrequency and Microwave Ablation

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Abstract- **Heat based tumor ablation methods such as radiofrequency (RF) and microwave (MW) ablation are increasingly accepted treatment methods for tumors not treatable by traditional surgery. Typically, an interstitial applicator is introduced under imaging guidance into the tumor, and tissue is destroyed by heating to above ~50 ºC, with maximum tissue temperatures over 100 ºC. Since high thermal gradients occur during the procedure, thermal conduction contributes significantly towards tissue heating. We created finite element method (FEM) computer models of RF and MW applicators, and determined the thermal conduction term, the resistive (for RF) or dielectric (for MW) loss term, and perfusion term. We integrated these terms over the heating period to obtain relative contribution towards tissue temperature rise (in ºC) as a function of distance from the applicator. We performed simulations without and with perfusion, where perfusion was assumed to stop above 50 ºC. During the first 6 minutes, direct heating by RF and MW were dominating throughout the tissue. Over the treatment period (12 min for RF, and 6 min for MW), thermal conduction was dominating at distances between than 12 and 19 mm from the RF electrode, while for MW ablation direct heating dominated everywhere. Even though thermal conduction significantly contributes towards tissue heating during ablative therapies, direct heating by RF or MW is dominating throughout most of the tissue volume. Tissue cooling due to perfusion is more significant during RF heating, in part due to the longer treatment times.**

Keywords **- radiofrequency ablation; RF ablation; tumor ablation; cancer**

I. Introduction

Radiofrequency (RF) ablation and Microwave (MW) ablation are clinically used for treatment of inoperable tumors of liver, as well as other organs such as lung, kidney and bone [1]. Differences between RF and MW ablation are much higher tissue temperatures obtained with MW with typically shorter application times.

High temperature gradients with temperatures up to ~100 ºC for RF, and up to ~160 ºC for MW are obtained; the high thermal gradients result in significant thermal conduction. Knowledge of regions where direct heating and thermal conduction are dominant is especially important when regions close to large vasculature are heated, where thermal conduction alone may not be sufficient to create temperatures in the therapeutic range (>50 ºC). In this study we examined the heating of liver tissue and determined the tissue regions where direct heating and where thermal conduction are dominating for clinical RF and MW ablation devices.

II. Methodology

We used Abaqus 6.5-1 for the RF model, FEMLab for the MW model, and Matlab 7.0.1 for further analysis of the generated results. The analysis was performed on a PC with 2GB RAM and a 3.2GHz Intel Pentium 4 CPU. Both models were designed axi-symmetric due to the symmetry of electrode and antenna. Initial tissue temperature was 37ºC, and this temperature was also applied to the model boundaries. For the model including perfusion according to Pennes' Bioheat Equation we assumed perfusion to stop when tissue coagulation occurs above ~50 ºC.

Description of the RF-Model (RFM):

We simulated a cooled needle electrode used currently clinically (Cool-Tip, Valleylab, Boulder, CO). The diameter of the exposed electrode is 1.5 mm and the length is 3 cm. This electrode uses internal cooling by circulating water. Applied power is controlled by tissue impedance such that maximum tissue temperatures obtained are ~100 ºC. In the RF model we controlled applied voltage such that maximum tissue temperature was 100 ºC during the 12 min ablation.

RFM Tissue properties:

ρ, kgm^{-3}	$c, J(kg*K)^{-1}$	$k, W(m*K)^{-1}$	σ, Sm^{-1} at 500kHz
1060	3600	0.512	0.333

Description of the Microwave-Model (MWM):

We simulated a dipole antenna (2.3 mm diameter, 10 mm dipole length) similar to antennas used clinically. The antenna was inserted 90mm into the tissue. 75W of Power was applied for 6 min at a frequency of 2.45 GHz. The SAR (specific absorption rate) is significantly affected by changing dielectric tissue properties as tissue water evaporates. Therefore in the electromagnetic (EM) model, tissue water related phenomena, including evaporation, diffusion and condensation, are simulated. The thermal model is based on expanded Bioheat equation which includes

tissue water evaporation at higher temperature. Tissue properties are adjusted depending on changes in water content. This model generates results significant closer to experimental results than previous static antenna EM models and basic thermal models.

MWM Tissue properties:

In contrast to the RFM, c (specific heat) and ρ (tissue density) is calculated during the simulation by the following equations:

Density:

$$\rho = v_w \times 1000 + 0.222 \times 1300 \quad (1)$$

v_W is the tissue water volume per unit volume of tissue, $0 \leq v_W \leq 0.778$. It is unit less. For normal tissue, $v_W = 0.778$. Proteins account for 22.2% of the total volume. Density of solid material is 1300 *kg/m³* Density of water is 1000 *kg/m³*.

Specific Heat:

$$C = \sum_n w_n C_n \quad (2)$$

For liver tissue, we assume that tissue is composed by water and proteins. The equation for normal and desiccated liver tissue could be expressed as:

$$C = 4200 \times w_w + 1560 \times 0.27 \quad (3)$$

where C is the heat capacity [*J/g•C*], w_W is the remained tissue water mass per unit mass of tissue, $0 \leq w_W \leq 0.73$. The equation is based on that heat capacity of solid tissue materials (proteins) is *1560 J/g•C* and heat capacity of water is 4200 J/g•C.

Bioheat Equation:

$$\rho c \frac{\partial T}{\partial t} = \nabla \cdot k \nabla T + Q_A - Q_p \quad (4)$$

ρ denotes the tissue density.
c denotes the specific heat of the tissue.

Energy $Q_A (W/m^3)$ is applied to the tissue by the applicator (electrode or antenna), resulting in heating of the tissue. Some energy Q_p is carried away by blood perfusion.

$$Q_p = \rho_{bl} c_{bl} w_{bl} (T - T_{bl}) \quad (5)$$

Where $\rho_{bl}[kg/m^3], c_{bl}[J/(kg \cdot K)]$ and T_{bl} are density, specific heat and temperature of the blood, respectively T is the tissue temperature, and w_{bl} is the blood perfusion (l/s).

The temperature increase due to thermal conduction was calculated using the following equation:

$$\Delta T_{cond} = \int_t \frac{\nabla \cdot k \nabla T}{\rho c} \delta t \quad (6)$$

In the same way, we determined temperature increase due to direct heating (SAR):

$$\Delta Q_A = \int_t \frac{Q_A}{\rho c} \delta t \quad (7)$$

and perfusion:

$$\Delta Q_p = \int_t \frac{Q_p}{\rho c} \delta t \quad (8)$$

Ablation Zone Boundary:

Even though tissue damage depends both on temperature and time, we found in previous studies that the 50 ºC isotherm correlates with ablation zone boundary within acceptable accuracy [2]. Therefore we used the 50 ºC isotherm to determine ablation zone boundaries.

III. RESULTS

Maximum temperatures were ~100 ºC in the RF model, and 177 ºC in the MW model; tissue charring and tissue vapor does not limit propagation of microwaves, so a higher temperature can be achieved. These temperature values are comparable to clinical results [3, 4]. The ablation zone diameters were 26 mm for the perfused RFM, and 36 mm for the unperfused RFM; diameters were 44 mm for the perfused MWM and 54 mm for the unperfused MWM.

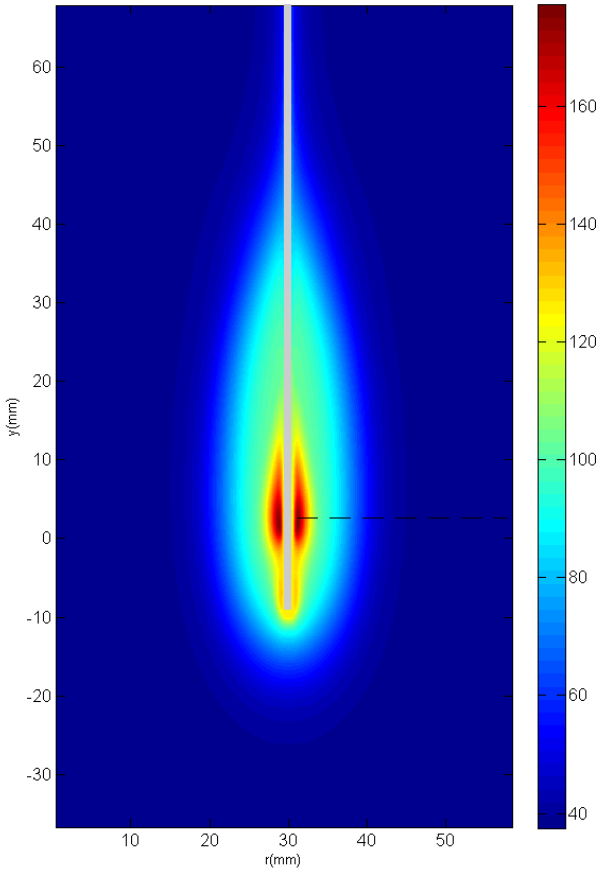

Figure 1. Temperature profile after 6min of MW-Ablation. The dotted line shows the location where analysis was performed for Fig 5. The area with temperatures >50 ºC is the ablation zone.

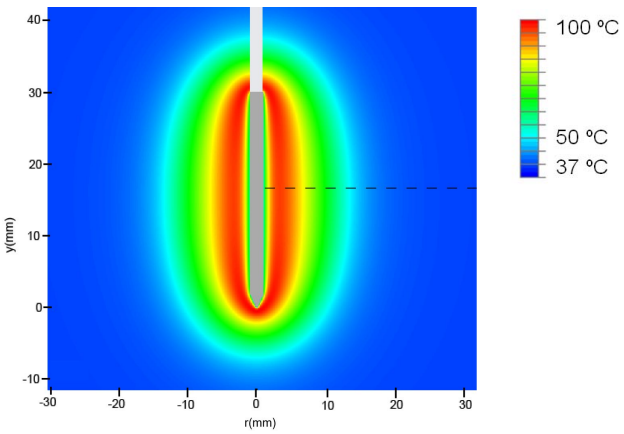

Figure 2. Temperature profile after 12 min of RF-Ablation. The dotted line shows the location where analysis was performed for Fig 3. The area with temperatures >50 ºC is the ablation zone.

IV. Discussion

Initially direct heating is dominating everywhere for both RF and MW ablation since tissue temperature is uniform (i.e. no temperature gradient and no heat flux). The integrals of the different heating terms show that for MW ablation, direct heating due to dielectric losses is dominating up to a radius of 20 mm over the 6 min MW ablation. Further away thermal conduction and direct heating have similar contributions (Figure 5). For RF ablation thermal conduction is dominating in the range from 12 mm to 19 mm radially over the 12 min procedure while direct heating due to resistive losses is dominating elsewhere (Figure 3). The increase of thermal conduction can be attributed to high temperature gradients occurring at the border of the perfusion zone (50 ºC). Since this discontinuity of thermal gradients is missing in the unperfused models, the thermal conduction term is significantly smaller near the ablation zone boundary (Figures 4, 6). Tissue cooling due to perfusion is highest just below 50 ºC, and zero above 50 ºC; this discontinuity in perfusion promotes high thermal gradients, and increased thermal flux near the ablation zone boundary.

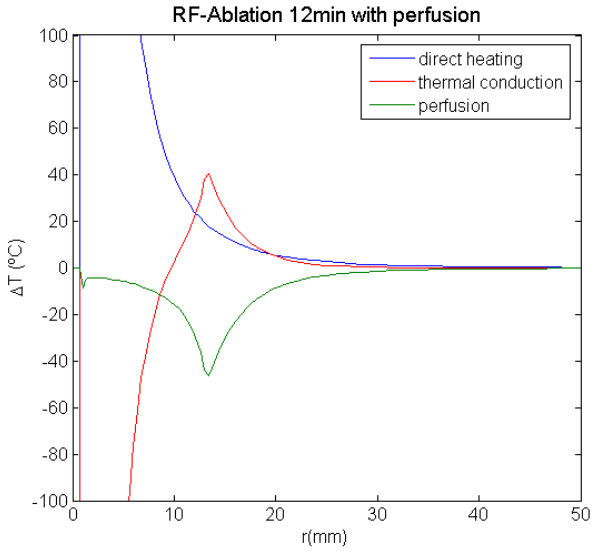

Figure 3. Temperature increase ΔT due to direct heating (blue), thermal conduction (red), and perfusion (green). Over the 12 min ablation procedure, thermal conduction dominates in the range from 12 to 19 mm radially.

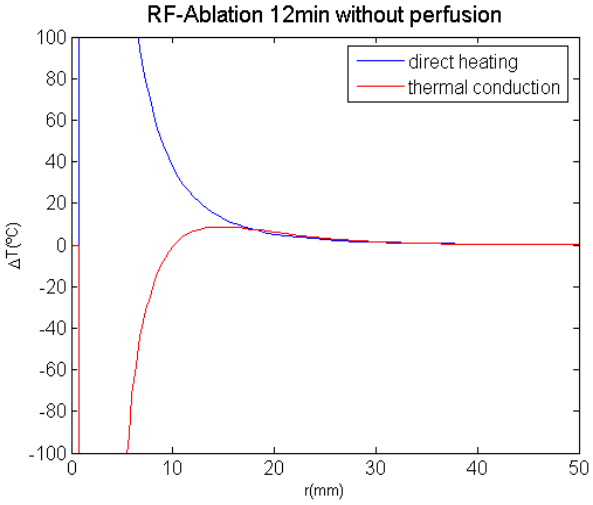

Figure 4. Temperature increase ΔT due to direct heating (blue), thermal conduction (red).

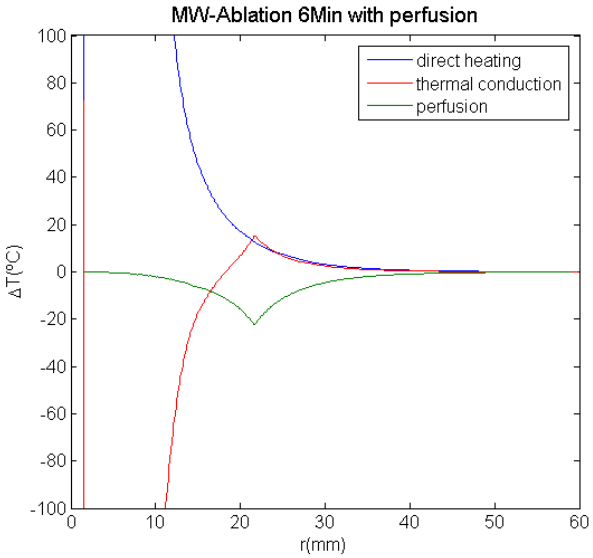

Figure 5. Temperature increase ΔT due to direct heating (blue), thermal conduction (red), and perfusion (green). Over the 6 min ablation procedure, direct heating is dominating the range up to 20 mm radially.

V. Conclusion

During RF ablation thermal conduction contributes more towards tissue heating compared to MW ablation. In addition, tissue cooling due to perfusion is more significant during RF ablation in part due to the longer treatment times. This may in part explain the superior performance of MW heating close to large vessels compared to RF [3].

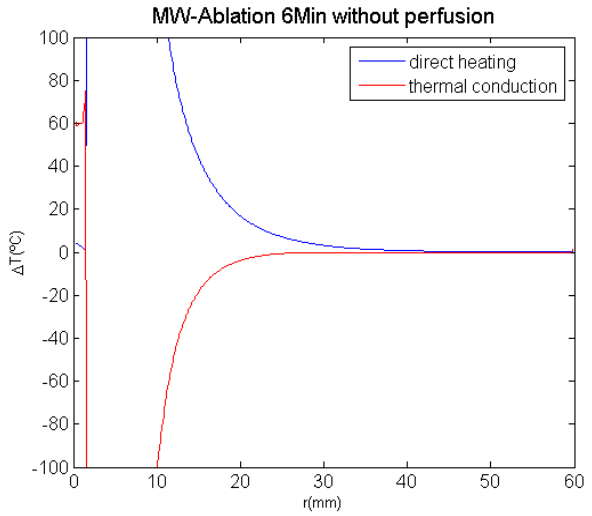

Figure 6. Temperature increase ΔT due to direct heating (blue), thermal conduction (red).

VI. Acknowledgements

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VII. References

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