MODELING TISSUE ABLATION WITH RADIO FREQUENCY HEATING: EFFECT OF A COOLING RF PROBE

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ABSTRACT

Thermal ablation of a solid tumor in tissue with radio-frequency (RF) energy can be accomplished using a probe inserted into the tissue under guidance with magnetic resonance imaging (MRI). The extent of the ablation can be significantly increased by cooling the RF probe with a saline solution that can leak into the surrounding tissue. The flow of solution into the tissue is considered as very slow flow through a porous medium. We examine these effects using a 3-D bio-heat transport model describing temperature distribution dynamics in tissue and in the cooling solution.

MATHEMATICAL MODEL.

We model the dynamic temperature distribution in a cube of tissue in which the tumor is centrally located (Figure 1).

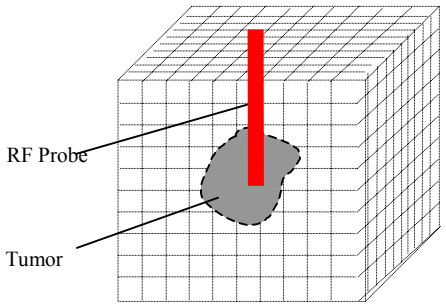

Figure 1. RF heating probe in tissue

Within the tumor, current from the RF probe (2cm long, 3 mm in diameter) produces a distributed heat source. The heat is dissipated by thermal conduction through the tissue and uptake by blood flowing through primarily small vessels that perfuse the tissue. A special case of this model has been examined previously [1].

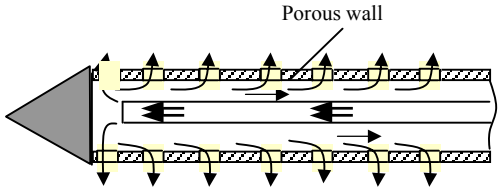

Figure 2.
RF heating probe with saline flowing
through the tube wall and into surrounding tissue

We can quantify and predict the temperature distribution dynamics in tissue as a consequence of RF thermal ablation based on a local heat balance:

$$\frac{\partial T}{\partial t} = \alpha \nabla^2 T + \sigma(r,t) - \omega(r,t)[T - T_b] - k_S[T - T_S] \qquad (1)$$

where the temperature T(r,t) is a function of r, the three-dimensional position vector, and t, time. The first term on the right side of equation (1) represents the rate of heat transfer in tissue by thermal diffusion. The second term represents the heat source produced in tissue by the RF-generated current.The third term represents the heat loss to randomly distributed capillaries perfused by blood at the basal temperature,T_b.The fourth term represents heat transfer from the tissue to the solution that leaks from a heating probe internally cooled by flowing saline (Figure 2). The solution-tissue heat transfer coefficient k_S is assumed constant.

Through a cooled porous probe wall, we consider a slow, small leak of saline solution occurs. The effect of this leak is expected to be dissipated within 1 or 2 mm from the probe surface because the velocity decreases as the inverse of distance from the probe. This comes from the continuity equation assuming radial symmetry, steady

state and neglecting end effects. We regard the flow as if it were occurring in the interstitial spaces of a porous tissue medium. The solution temperature $T_S(r,t)$ varies in time and space within the tissue. The governing equation includes thermal convection, thermal diffusion, and thermal exchange with the tissue medium:

$$\frac{\partial T_S}{\partial t} = -v \cdot \nabla T_S + \alpha \nabla^2 T_S + k_S[T - T_S] \qquad (2)$$

where the velocity is $v(r)= Q/2\pi Lr$, the probe length L, and volume flow Q are known.

The RF probe electrode ($r=r_p$) considered as a uniform line source whose temperature variation with time is known by independent measurement, $T(r_p,t)=T_p$. The solution leaks at the same temperature, $T_s(r_p,t)=T_p$ Since the boundaries of the theoretical cube surrounding the tumor are far enough from the heat source ($r=r_\infty$), the distant temperature remains at the basal level: $T(r_\infty,t)=T_s(r_\infty,t)=T_b$.

The RF heat source $\sigma(r, t)$ is a function of input power $P(r_p,t)$ and varies inversely with the square of the distance from the RF electrode probe, $r-r_p$:

$$\sigma(r,t) = \frac{h_0 P(r_{p,}t)}{[r - r_p]^2} \quad (r > r_p) \qquad (3)$$

Perfusion heat loss is proportional to the local temperature difference and perfusion coefficient $\omega(r,t)$, which is reduced by ablation causing blood coagulation. The perfusion coefficient is dependent on the local tissue temperature history:

$$\omega(r,t) = \omega_0 \exp\{\int_0^t \beta[T(t,v)]dv\} \qquad (4)$$

To describe the rate β at which cells become irreversibly damaged, we consider a simple piecewise linear model. The damage rate increases linearly with temperature above a critical temperature T_0=43°C.

$$\beta(T) = \beta_0 (T - T_0) U(T - T_0) \qquad (5)$$

where $U(T-T_0)$ is a unit step function.

MODEL SIMULATION

With internal probe cooling, a larger tissue lesion can be produced because the heat input power can be increased without exceeding 100°C at which charring and vaporization can occur and lead to reduced heat transfer. In preliminary simulations, we examined the effects of a cooled probe surface without any solution leaking into the surrounding tissue. Without internal probe cooling, we set the input power at 40 watts and the probe temperature at 90°C. For comparison, internal cooling began after 3 min. In the latter case, when the probe temperature was reduced to 20°C, the input power could be increased to 90 watts.

The effect of internal cooling on the perfusion distribution in tissue is shown in Figure 3. Initially, the perfusion is uniform and constant. Essentially, when perfusion is zero, the tissue is ablated and the cells are irreversibly damaged. After heating for 10 min without internal probe cooling, the perfusion was zero in a radius of about 5 mm around the probe. With internal probe cooling, the radius of ablated tissue increased to about 12 mm. These simulation results indicate that with internal probe cooling the lesion size can be significantly increased by using a higher input power without exceeding 100°C. (Figure 4).

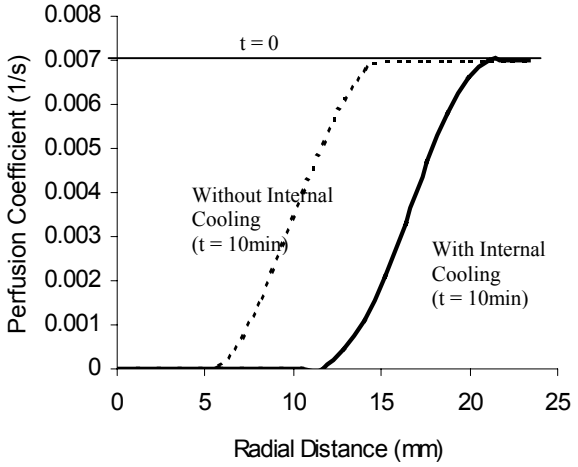

Figure 3. Internal Probe Cooling Allows Higher Heat Input To Produce a Larger Tissue Lesion

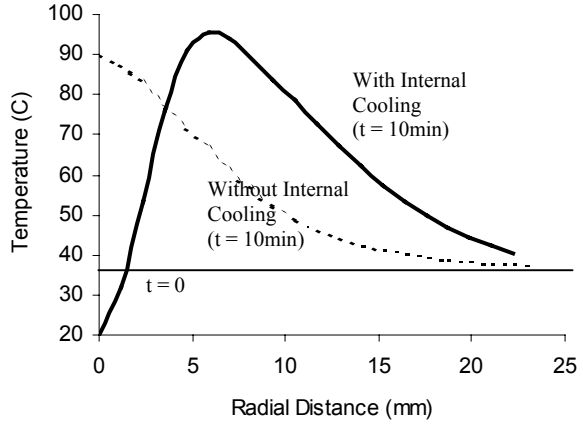

Figure 4. Tissue Temperature Distribution After Heating With and Without Internal Probe Cooling

ACKNOWLEDGEMENT

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REFERENCE

1. Johnson, P.C. and Saidel, G.M. (2002) "Thermal Model for RF Tumor Ablation under MRI Guidance. Ann. Biomed. Eng. 30, pp.1152-1161.