Measuring Local RF Heating in MRI: Simulating Perfusion in a Perfusionless Phantom

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Introduction: In the MR literature on the evaluation of RF heating safety, conflicting methods for measuring local RF heating are proposed. Each method suffers from some non-ideality. In vivo temperature measurements are difficult to conduct reproducibly. Measurements conducted using fluid phantoms introduce convection, which is not physiological. Gel phantom measurements lack the cooling effect of physiological perfusion. Earlier, we proposed a Green's function averaging technique [1], but this method requires extensive computer simulations. In this study, we propose a method for predicting in vivo temperature increase by using a straightforward gel phantom experiment.

Methods: The Fourier transform of the Green's function (GF) of the Bioheat equation was determined analytically in the presence and absence of the perfusion effect. Analysis of the two functions revealed that the steady-state GF with perfusion approximates the time dependent GF without perfusion at the specific time, t=τ, where $\tau = c_t / (m\rho_b c_b)$ is the tissue perfusion time constant. The match is exact at very low and high spatial frequencies and deviates less than 20% for all spatial frequencies. This means that a good estimate of steady-state in vivo temperature can be made by measuring the temperature change in a phantom after heating for a duration τ. Figure 1 is the graphical representation of this theory, assuming that the phantom has the same electrical and thermal characteristics as the tissue. Perfusion time constants for distinct tissues are readily available in literature [2, 3, 4, 5].

Ideally, one should make a gel phantom that matches the geometric, electrical, and thermal properties (except perfusion) of the tissue in which the metallic device will be used. In our case, to test the theory, we used a cylindrical homogeneous gel phantom. The relative electrical permittivity (70) and conductivity (0.7S/m) of the gel were deduced from measurements with a Network Analyzer (HP5763).

A thermal ablation probe, fed via a signal generator and amplifier, was used as the power source as shown in Figure 2. An oscilloscope was used to measure the voltage on a serially connected resistor to measure the power (0.45-1.75 W) delivered to the rabbit. Fiberoptic temperature data (FISO Technologies, Ste. Foy, Quebec, Canada) from four tissues (fat, kidney, liver, and brain) were collected and compared with the data taken from phantom experiment. Our live rabbit experimental protocol was approved by our local IRB.

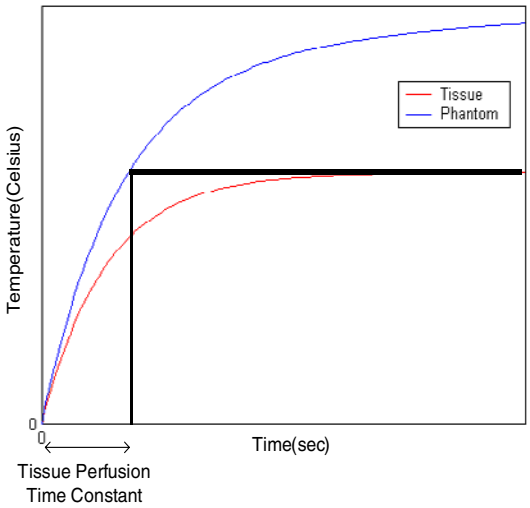

Figure 1. Predicting in vivo steady state temperature increase

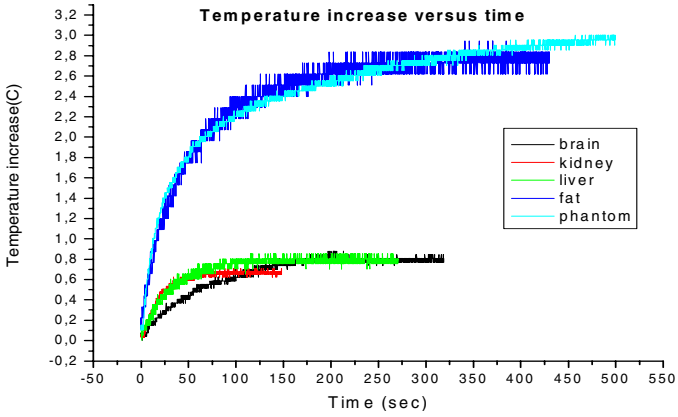

Figure 3. Temperature increase versus time graphs of phantom and the tissues

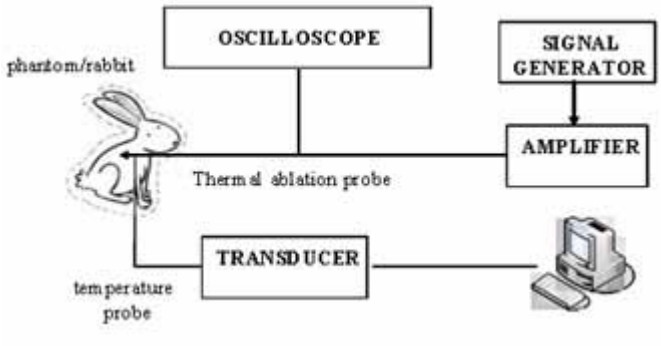

Figure 2. Experimental Setup

Results: The results are shown in Figure 3 and Table 1. As can be seen from Table 1, the experimental data mostly supports the theoretical calculations, which demonstrates that theoretical temperature increase calculations are suitable for actual predictions. In each case, the calculated temperature overestimates the actual in vivo temperature.

Tissue	Temperature increase(°C)	Estimated Temperature increase(°C) (calculated from phantom data at t=τ)	Time constant, τ (sec)	Error(%)
Kidney [4,5]	0.617	0.8-0.9	11.4-14	22.8-31
Fat [2,5]	2.5	3.2-3.3	1128-1620	21.8-24
Brain [3,4,5]	0.84	2.1-2.2	88.2-108	60-61.8
Liver [4,5]	0.75	1.75-2.2	48-102	57-66

Table1: Comparison of predicted and measured temperature increases for different tissues.

Discussion: Temperature prediction error up to 66 percent was observed. One reason for the error is the mismatch between thermal and electrical properties of the phantom and the tissues. More accurate results may be obtained if different gels were used for simulating each tissue. Electrical properties of the gel and all the tissues (except fat) are similar[6] but we did not attempt to match the thermal conductivity of the phantom to the tissues Also, perfusion time constants of humans were taken as a reference. Using corresponding rabbit values would reduce the error. Perhaps more importantly, local thermoregulation was not modeled with the bioheat equation. As such, the predicted in vivo temperature changes have an intrinsic overestimation that builds in a safety factor with this approach.

Conclusion: Using this method, in vivo heating can be predicted by simple perfusionless phantom experiments. This method can be used for setting local RF heating safety limits with implants in MRI.

References: **[1]** Yeung CJ, Atalar E. A Green's function approach to local RF heating in interventional MRI. Med Phys 2001; 28:826-832.**[2]** B. I. Tropea and R. C. Lee, ''Thermal injury kinetics in electrical trauma,'' J. Biomech. Eng. **114**, 241–250 ~1992. **[3] .** A. W. Guy, J. F. Lehmann, and J. B. Stonebridge, ''Therapeutic applications. of electromagnetic power,'' Proc. IEEE **62**, 55–75 ~1974 **[4]** A. Shitzer and R. C. Eberhart, ''Heat generation, storage, and transport processes,'' in *Heat Transfer in Medicine and Biology*, New York, 1985, p. 141. **[5]** Adapted from M. Rowland, T. N. Tozer, & R. Rowland, *Clinical Pharmacokinetics: Concepts and Applications*, copyright 1995, with permission of Lippincott Williams & Wilkins. **[6]** http://www.fcc.gov/fcc-bin/dielec.sh

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