Optical method using fluence or radiance measurements to monitor thermal therapy

Lee Chin[a] and Mihaela Pop
Ontario Cancer Institute, Princess Margaret Hospital, University Health Network, Toronto, Ontario M5G 2M9, Canada and Department of Medical Biophysics, University of Toronto, 610 University Avenue, Toronto, Ontario M5G 2M9, Canada

William Whelan
Department of Medical Biophysics, University of Toronto, 610 University Avenue, Toronto, Ontario M5G 2M9, Canada and Department of Mathematics, Physics, and Computer Science, Ryerson Polytechnic University, Toronto, Ontario M5B 2K3, Canada

Michael Sherar and Alex Vitkin
Ontario Cancer Institute, Princess Margaret Hospital, University Health Network, Toronto, Ontario M5G 2M9, Canada and Department of Medical Biophysics and Department of Radiation Oncology, University of Toronto, 610 University Avenue, Toronto, Ontario M5G 2M9, Canada

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We present data from a two-region (coagulated and native tissue) Monte Carlo simulation in spherical geometry to assess the feasibility of using optical fluence and radiance information for determining the boundary of tissue coagulation during thermal therapy. Our results demonstrate that radiance offers directional sensitivity that might be useful in monitoring asymmetric coagulation growth. Preliminary experimental data from fluence monitoring of radio-frequency kidney thermal therapy further indicates that fluence information might be insufficient for determining the coagulation boundary. Therefore, in addition to optical fluence, we are also exploring the potential of radiance monitoring as a method for establishing the position of the coagulation boundary during radio-frequency thermal therapy. © *2003 American Institute of Physics.*
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I. INTRODUCTION

Interstitial thermal therapy (ITT) is currently under investigation for the treatment of solid tumors. Thin needle applicators are inserted into the tumor site and energized with lasers, microwaves, radio frequency (rf), or ultrasound. The energy is absorbed heating the tumor site resulting in coagulative necrosis. Due to the complex and dynamic nature of tissue properties of the tumor tissue during ITT, the development of real-time monitoring systems that can accurately detect tissue damage during the procedure and preserve critical normal structures is vital. Currently, temperature sensors are commonly used to monitor and control ITT treatments. However, due to slow thermal conduction, patient-specific variability in thermal damage, and the inability to directly sense tissue coagulation, temperature measurements alone, might be insufficient for assessing treatment efficacy. Optical point monitoring, where light intensity information is measured using interstitial optical sensors, provides near instantaneous response to structural changes that occur during treatment.[1,2] This is due to the fact that the scattering coefficient of tissue increases significantly (between 2 and 7 $\times$) upon coagulation. The sharp contrast between coagulated and native tissue led Iizuka[3] to suggest that ITT might be amenable to optical monitoring.

Continuous-wave optical information can be measured either in the form of direction-dependent intensity (radiance), $L(r,\hat{\Omega})$ where r is the radius from the source and Ω is the solid angle relative to the radius normal, or direction-independent intensity (fluence) $\Phi(r)$. The fluence (Wm^{-2}) can be obtained from the radiance ($\mathrm{Wm}^{-2}\,\mathrm{sr}^{-1}$) by integrating over all solid angles:

$$\Phi(r)=\int_{4\pi} L(r,\hat{\Omega})\,d\Omega.$$

Radiance measurements offer the advantage of directional information but due to the lack of readily available radiance probes, we and others have primarily employed fluence probes for ITT monitoring. However, recently Dickey *et al.*[4] reported that by attaching a small right-angle prism to a cleaved fiber, radiance information can be retrieved by performing a 180° rotation of the fiber. These authors employed radiance measurements for photodynamic therapy dosimetry. The potential of utilizing radiance for monitoring of ITT is yet to be explored. Optical monitoring is well suited to laser interstitial thermal therapies (LITT) because the laser source may act to simultaneously treat and provide optical energy that can be detected for optical sensing.

We previously tested the potential of Monte Carlo (MC) models to relate fluence changes to coagulation radius during LITT with a single spherical source.[5] Since the coagulation growth around the fiber is spherically symmetric, the determination of coagulation volume can be modeled one-

[a]Electronic mail: chinl@oci.utoronto.ca

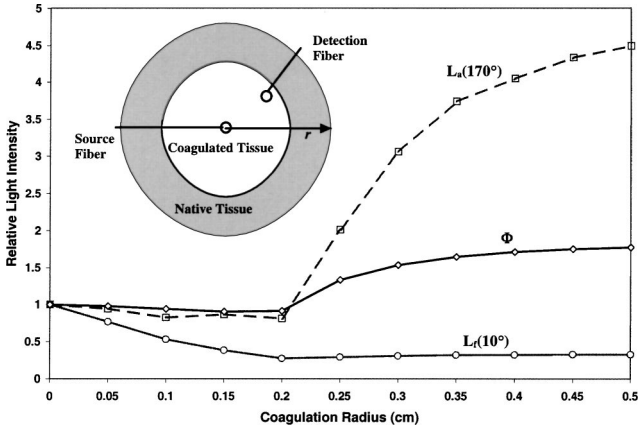

FIG. 1. Simulation geometry (inset) and Monte Carlo results showing predicted changes in fluence, Φ, forward radiance, $L_f(10°)$, and backward radiance, $L_a(170°)$, at 2 mm away from the source during LITT employing a spherically emitting source fiber. The fluence behavior has been verified experimentally (see Ref. 5). Symbols are Monte Carlo results while the lines are fitted.

dimensionally (spherically). This geometry allows for the basic physics of the optical monitoring problem to be investigated. However, for practical application the method needs to be extended to other geometries such as cylindrical sources source which are more commonly used in ITT.

This article extends our previous MC simulation to include radiance information. We also present preliminary experimental data demonstrating fluence monitoring of rf thermal therapy of *ex vivo* porcine kidney. Based on our MC results, we discuss why radiance might prove advantageous over fluence data for this complex geometry.

II. PRINCIPLES OF OPTICAL MONITORING

For a spherical source of optical energy positioned within homogeneous tissue, our MC algorithm models thermal damage as concentric spheres of coagulated and native optical properties centered at the source (inset of Fig. 1). It is assumed that the index of refraction of coagulated tissue is the same as native tissue. To mimic the ''growth'' of the damaged region during heating, separate MC simulations are run for increasing radii of the inner ''coagulated'' sphere. Next, changes in light intensity at a given detector position are plotted versus the inner coagulation radius. Since the extent of damage increases as a function of heating time, the *x* axis (damage radius) is effectively a measure of heating time with the horizontal scaling of the graph expanding or contracting depending on the rate of heating for a given treatment. Additional details of the MC simulation can be found in Ref. 5.

Figure 1 shows MC simulation results showing relative changes at 2 mm away from a spherical source in fluence, radiance facing the source, $L_f = L(10°)$, and radiance facing away from the source, $L_a = L(170°)$. Native ($\mu_a = 0.5\ \mathrm{cm}^{-1}$, $\mu_s' = 2.67\ \mathrm{cm}^{-1}$) and coagulated ($\mu_a = 0.7\ \mathrm{cm}^{-1}$, $\mu_s' = 13.1\ \mathrm{cm}^{-1}$) optical properties were chosen to simulate albumen phantoms, which have similar properties as soft tissue.[6] Prior to the coagulation boundary reaching the 2 mm point, there is a significant decrease (−70%) in L_f. In contrast, the fluence and L_a decrease less noticeably indicating that at 2 mm, the majority of interstitial light intensity is traveling forward away from the source fiber. As the coagulation radius passes 2 mm, a light trapping effect causes an increase in the fluence (+150%), and a more significant change in the L_a value (+450%). Here, L_f remains almost constant as opposed to L_a in the previous case. Here, the implication is that the increase in fluence is due to photons traveling back towards the source fiber.

The differences between the two radiances are a result of the highly forward scattering nature of light in tissue. For small coagulation radii, the increased scattering properties of coagulated tissue will have minimal effect on the bulk radiance distribution since photons scatter forward (away from the source fiber) and readily escape the coagulation volume. Still, there is sufficient scattering to cause collimated photons to diverge from their original straight-line path. Hence, as the coagulation front approaches 2 mm, it is primarily L_f that is affected. After passing 2 mm, the increase in fluence is caused by photons that have scattered numerous times, enough to reverse their initial forward direction. Consequently, L_a increases significantly while L_f remains relatively constant following passing of the coagulation front. The MC results show that compared to fluence, L_f and L_a offer increased sensitivity to an approaching and passing coagulation boundary, respectively.

The fluence results of the MC model have been validated experimentally in well-characterized tissue mimicking albumen phantoms[6] and demonstrated excellent qualitative and quantitative agreement (±20%),[5] experiments are underway to validate the new radiance results.

III. MONITORING RADIOFREQUENCY THERMAL THERAPY

Rf ablation has shown promise for the minimally invasive treatment of small solid renal masses.[7,8] We are currently investigating a system produced by Radio Therapeutics Corp. (Sunnyvale, CA), which delivers rf energy at 460 kHz via a LeVeen electrode (Radio Therapeutics Corp.) consisting of an array of evenly spaced wire electrodes enclosed by a metal cannula. When deployed, the array appears similar in shape to an umbrella (Fig. 2) and produces a spherical thermal lesion in tissue. A recent study in patients that underwent radical nephrectomy following rf ablation showed that areas of viable tumor were present at the margins of the treated volume.[8] The untreated areas were attributed to heat sink effects, abnormal tumor vasculature, and uneven rf energy distribution.[8] We envision that optical methods might play a role in rf thermal therapy monitoring to ensure complete coagulation of the tumor volume.

We performed fluence monitoring of the coagulation front in fresh *ex vivo* porcine kidney during rf thermal therapy using a previously developed protocol that delivered continuous rf energy until a coagulation-induced impedance of 400 Ω was reached, which caused a cessation of power delivery.[7] Figure 2 shows a schematic of our experimental geometry. The optical fibers and rf applicator were fixed in an acrylic jig for accurate positioning. A 2 mm diam diag-

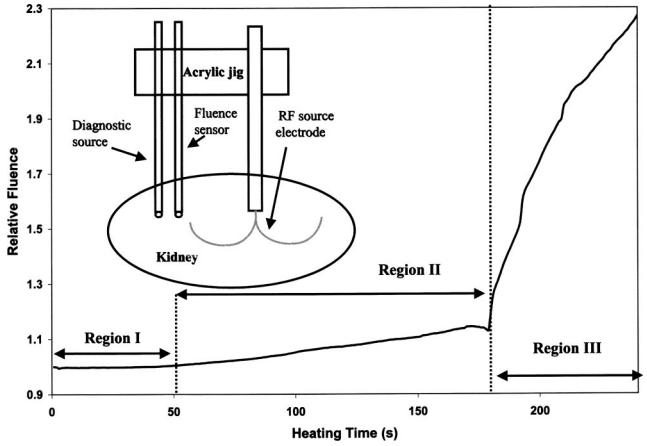

FIG. 2. Experimental geometry (inset) and measured changes in optical fluence during rf thermal therapy of *ex vivo* kidney.

nostic source fiber (Resonance Optics, Dennis, MA) delivering energy from a diode laser (Diomed, England) at 805 nm and 0.5 W was positioned 5 mm from a 800 μm diam fluence sensor (Resonance Optics) positioned at 2 mm from the tip of the rf electrode. The light source was positioned 7 mm from the electrode tip. The fluence sensor was attached to a PDA55 photodiode (Thorlabs, Newton, NJ) that converted the light signal to a photovoltage reading. The average of 30 photovoltage signals was read each second and displayed in real time on a PC. Following the treatment, the kidney was cut open in the plane containing the optical fibers.

Figure 2 shows the changes in fluence measured during rf thermal therapy of the kidney. The curve can be divided into three regions: (I) $0<t<50$ s, fluence remains constant; (II) $50<t<180$ s, fluence begins to rise slowly increasing by $+12\%$; and (III) $180<t<240$ s, a sudden and sharp fluence increase with a relative change of $\sim+225\%$ was observed. It is likely that the constant light intensity of region (I) indicates that coagulation has yet to occur. The gradual fluence increase observed in region (II) suggests the onset of coagulation around the source tines. The sudden increase in light intensity in region (III) possibly signifies that the coagulation front has passed or is close to the fluence sensor location. Visual examination of the thermal lesion after the treatment showed that the resulting coagulation boundary passed the fluence probe.

Unlike the LITT case, the complex evolution of the coagulated region during rf ablation (relative to the diagnostic optical source) makes prediction of the coagulation location difficult. During LITT, a significant increase in fluence occurs when the coagulation radius passes the sensor.[1] However, for our chosen rf monitoring geometry, the coagulation radius grows towards the diagnostic source (as opposed to around it), which could potentially cause an increase in fluence even before the front passes the optical sensor. Hence, for optical monitoring of rf thermal therapy, fluence alone might be insufficient to predict the extent of coagulation. In such asymmetric problems, measurements of directional light intensity might prove more useful.

IV. APPLICATION OF RADIANCE TO MONITORING OF RF THERMAL THERAPY

Based on the MC results of Sec. II, we hypothesize that employing a radiance probe rotating between 0° and 180° might provide the additional information necessary to determine the location of the coagulation radius during rf thermal therapy. We envision that prior to the coagulation boundary passing the radiance sensor, L_a would increase steadily as the coagulation front approached the sensor, while L_f would remain relatively constant. Using this approach, one would strategically place the radiance probe at a desired location of the final treatment boundary, and wait for a change in the forward flux caused by coagulated tissue between the sensor and diagnostic source. We have recently constructed radiance probes and are in the process of designing an automated rotation system for the application of radiance monitoring of rf kidney thermal therapy.

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