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NOTE

Laser thermal therapy: utility of interstitial fluence monitoring for locating optical sensors

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Abstract

Multipoint optical fluence measurements can potentially be used to detect coagulation-induced changes in optical propagation during interstitial laser thermal therapy. Estimating the dimensions of coagulation using on-line optical monitoring, which is applicable to treatments where the tip of the source fibre is not precharred, may be limited by the accuracy of the placement of optical sensors with respect to source fibres. A strategy has been developed to determine accurately the position of a four-sensor linear array, prior to treatment, using optical fluence data obtained from the sensors for low-power ($\leqslant$0.5 W) irradiation. A minimum of four sensors in an array was required in order to develop a mathematical formulation for position determination that did not require tissue optical properties or laser power as input. Optical propagation was based on diffusion theory for homogeneous tissues in spherical geometry. Low input laser power is needed to ensure that there are no thermally induced changes in tissue optical properties not accounted for in the mathematical description. Experimental evaluation was performed in a tissue-equivalent liquid phantom using 0.5 W of 805 nm optical energy and a translatable isotropic optical sensor. For sensor locations with 2 mm spacing, placement accuracy of 0.67 mm was achieved. The accuracy improved to 0.13 mm as the sensor spacing increased to 5 mm.

1. Introduction

Interstitial laser thermal therapy is a minimally invasive technique used to destroy thermally solid tumours. Investigations have demonstrated the feasibility of this technique in a number of sites including the brain (Roux *et al* 1992) and the liver (Amin *et al* 1993). However, induced thermal lesions are variable and unpredictable due to variations in blood perfusion and thermally induced changes in the optical properties of tissue. Coagulation-induced increases in

tissue optical scattering of two- to eight-fold have been reported (Splinter *et al* 1991, Pickering *et al* 1994). It is therefore important to include on-line monitoring and feedback control to compensate for these factors. Most on-line monitoring systems for thermal therapy generally rely on point temperature measurements or volumetric imaging in tissue. The effectiveness of point temperature measurements to control heating is in question, due to the time delay in the arrival of thermal energy to a temperature sensor coupled with the large temperature gradients induced during laser heating. Furthermore, temperature sensors such as microthermocouples can be inaccurate when placed in large temperature gradients due to thermal conduction smearing along metallic leads (Gerig *et al* 1992).

Alternatively, due to the speed of light in tissues, optical fluence monitoring may provide almost instantaneous and accurate detection of coagulation-induced changes in optical propagation. However, this method would only be applicable to treatments where the tip of the source fibre is not precharred, since tissue char is opaque. The utility of optical fluence monitoring during laser heating has not been thoroughly investigated. For example, it is recognized that accurate knowledge of the position of optical sensors with respect to sources is critical for on-line prediction of coagulation dimensions. Therefore, a strategy has been developed to determine the position of a four-sensor linear array, prior to treatment, using optical fluence data obtained from the sensors for low-power ($\leqslant$0.5 W) irradiation. The developed strategy does not require prior knowledge of tissue optical properties or input optical power.

2. Theory

Diffusion theory has been used extensively to model optical propagation in laser heating of tissues (Beacco *et al* 1994, Anvari *et al* 1994, Whelan and Wyman 1999). In this work, the determination of sensor position is achieved prior to treatment, using a low input laser power to ensure that there are no thermally induced changes in tissue optical properties. Therefore it is assumed that the tissue is optically homogeneous; the validity of this assumption is discussed later. The optical fluence from a point isotropic source of input power, P, in an infinite homogeneous medium can be expressed in spherical coordinates as

$$\phi(r) = \frac{P}{4\pi Dr} \exp\left(-\sqrt{\frac{\mu_a}{D}}r\right) \tag{1}$$

where the diffusion coefficient, $D = [3(\mu_a + \mu_s')]^{-1}$, is expressed as a function of the optical absorption coefficient, μ_a, and reduced optical scattering coefficient, μ_s'.

There are four unknowns in equation (1): μ_a, μ_s', P and r. The goal of the strategy is to predict the sensor location without prior knowledge of the tissue optical properties or input optical power, since these quantities are difficult to obtain and will have associated uncertainties. If we consider using a linear sensor array with known intersensor spacing, then a minimum of four sensors would be required to uniquely determine the position of the array. A diagram of the model geometry (figure 1(*a*)) shows the four sensors at different radial distances, with intersensor spacing S. Thus four fluence equations similar to equation (1) at four radial positions, r_1, r_2, r_3 and r_4, are generated. Solving the four fluence equations simultaneously eliminates μ_a, μ_s' and P and gives

$$\frac{r_2 r_3}{r_1 r_4} = \frac{\phi(r_4)\phi(r_1)}{\phi(r_3)\phi(r_2)}. \tag{2}$$

Furthermore, radial distances r_1, r_2, r_3 and r_4 can be expressed in terms of the distance from the source to the centre of the array, R, and half the intersensor distance, $S/2$ (see figure 1(*a*)).

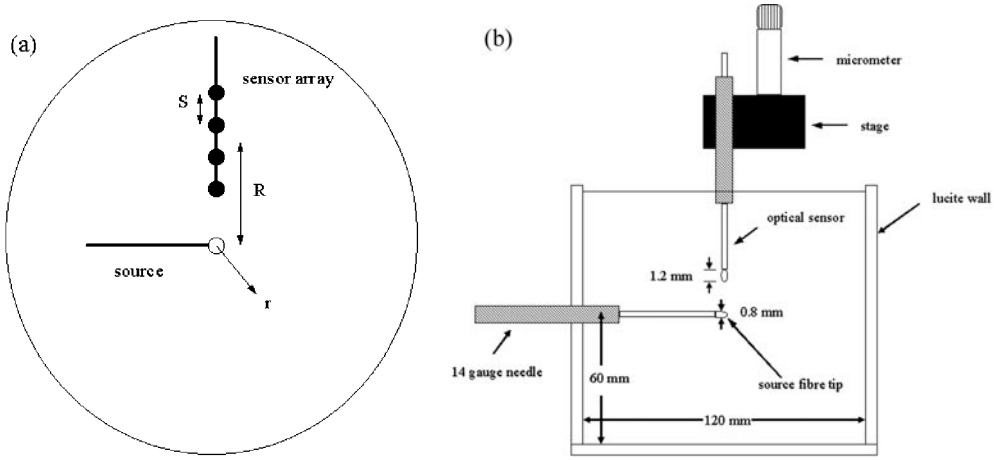

Figure 1. (*a*) Model geometry for position determination of a four-sensor linear array. (*b*) Schematic of the tissue-equivalent liquid phantom experiment.

Then equation (2) can be rearranged to obtain

$$\frac{R^2 - (S/2)^2}{R^2 - 9(S/2)^2} = \frac{\phi(r_4)\phi(r_1)}{\phi(r_3)\phi(r_2)}. \tag{3}$$

Solving equation (3) for R gives

$$R = \frac{S}{2}\sqrt{9 + \frac{8}{\frac{\phi(r_4)\phi(r_1)}{\phi(r_3)\phi(r_2)} - 1}}. \tag{4}$$

An optical sensor typically measures a photovoltage, which can be scaled by a calibration factor to obtain the fluence. The calibration factor generally depends on the design of the sensor (Van Staveren *et al* 1995) and is difficult to determine accurately. These experimental factors cancel using our methodology, so measured photovoltages can be used directly in equation (4).

3. Materials and methods

A liquid tissue-equivalent phantom was constructed to evaluate the sensor positioning strategy. The phantom consisted of Intralipid-10% and naphthol green dye. Intralipid-10% is a weakly absorbing, highly scattering liquid emulsion (10% solution of soybean oil in water) that can be diluted with water to yield tissue-like scattering in the near infrared (van Staveren *et al* 1991). Naphthol green dye is an organic powder that can be dissolved in water and added to Intralipid-10% suspension to yield absorption properties of tissues in the near infrared. These materials have been used extensively to construct tissue-equivalent optical phantoms (Iizuka *et al* 1999, Vulcan *et al* 2000). In this work, a liquid phantom with nominal optical properties of $\mu_a = 0.13\ \text{cm}^{-1}$ and $\mu_s' = 13.0\ \text{cm}^{-1}$ was prepared, following the methods of Iizuka *et al* (1999). These properties were not explicitly measured, since one of the advantages of the current positioning strategy is that the optical property values are *not* required as input.

True isotropic fluence sensors are not commercially available. Furthermore, optical sensor arrays are not yet available for illuminations at 805 nm. Therefore an array was approximated by translating a single sensor using a micrometer-driven stage. The optical sensor was fabricated

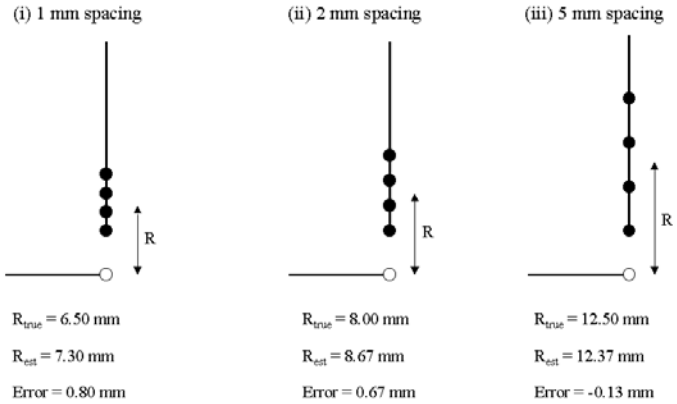

Figure 2. Error between the estimated (R_{est}) and true (R_{true}) distances to the centre of a simulated array in a tissue-equivalent liquid phantom, as a function of sensor spacing (first sensor at 5 mm).

by first removing approximately 2 mm of the cladding from the end of a 600 μm core plane-cut fibre. The end was then dipped in a mixture of epoxy and titanium dioxide, creating an approximate sphere of diffusing material at the end of the fibre. The tip was irradiated with ultraviolet light for 10 min to fix the epoxy. The sensor fibre emission was not truly isotropic, with small null regions in the backward direction. A diagram of the experimental set-up is shown in figure 1(*b*). A spherical diffusing source fibre (Surgimedics, Orb 5601), coupled to a 15 W diode laser (Diomed, UK) emitting at 805 nm, was placed at the centre of a 120 mm by 120 mm Lucite box. The dimensions of the box were chosen to ensure optically infinite geometry for a range of tissue optical properties, consistent with the mathematical formulation. The sensor, coupled to a photodiode (Thorlabs, PDA-55) connected to a data acquisition system (Labmate, Sciemetrics), was initially placed orthogonal to and in contact with the tip of the source fibre. The micrometer was then referenced and the liquid phantom material was poured into the Lucite box. The lowest power setting available on the Diomed-15 (0.5 W) was used and the sensor was withdrawn to 20 mm, in 0.5 mm increments. A four-sensor array at different positions and with different intersensor spacing was simulated using this photovoltage data set.

4. Results and discussion

Photovoltages reported by the sensor at each location were stable. For an input power of 0.5 W, the photodiode saturated (greater than 10 V) when the sensor was within 1 mm of the source. This suggests that at therapeutic powers of several watts it is likely that sensors should not be placed within 4–5 mm from a source for on-line monitoring. Furthermore, diffusion theory is inaccurate within approximately 1–2 mm from sources. Therefore, in this work the first sensor was 5 mm from the tip of the source fibre. The range of measured photovoltages, from 5 mm to 20 mm, was 1.2 V to 0.044 V respectively. The error between the estimated (R_{est}) and true (R_{true}) distance to the centre of a simulated array as a function of intersensor spacing is shown in figure 2. For intersensor spacings of 1 mm, 2 mm and 5 mm, the errors in R are 0.80 mm, 0.67 mm and -0.13 mm respectively. This demonstrates the utility of using photovoltage data to predict the position of an optical sensor array under conditions of spherical geometry, provided that the intersensor spacing is known. This would generally be under the experimenter's control prior to treatment. In addition, the results indicate that

the positioning strategy may not be that strongly dependent on the sensor fibres being truly spherically isotropic.

The assumption of optical homogeneity of the medium, while valid for the phantom experiments described here, may prove less accurate in tissues. The magnitude of the resulting inaccuracies in sensor positioning is not clear due to limited knowledge of the (heterogeneity of) optical properties of normal and diseased tissues, although the assumption of a homogeneous medium is probably more valid for bulk normal tissues. It should also be emphasized that thermally induced heterogeneities are not important here, as the developed methodology uses low-power irradiation prior to treatment, hence avoiding any thermal denaturation effects.

For clinical applications, it is anticipated that multiple data sets at each sensor location may be required due to instabilities in optical fields *in vivo*. This could increase the irradiation time and may result in undesirable thermally induced changes in tissue optical properties during the measurement period. This can be minimized using the lowest power setting on the laser system. For example, the lowest power setting on the Diomed 60 (Diomed, UK) is 0.5 W. Alternatively, if power below $\sim$1 W is not available, faster sampling (kHz to MHz rates) is easily achievable using commercially available data acquisition systems.

The current mathematical formulation assumes that the linear sensor array is positioned radially from the tip of the source fibre (i.e. pointing straight at the emitter). This may not always be known or achievable. For example, in figure 1(a), if the sensor array was shifted to the right, the centre of the array can be specified by R and a new unknown orthogonal offset distance from the tip of the source fibre to the line of the array. A mathematical approach in Cartesian coordinates is currently being developed to handle this configuration. An additional sensor may be required to determine the extra unknown. Furthermore, this Cartesian method will address the general problem of the determination of the position of multiple sensor arrays illuminated by multiple sources for the clinically relevant geometry of parallel insertion of devices.

Acknowledgments

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