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Anisotropic modeling for IR navigation-based PDT dosimetry

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Abstract

An infrared (IR) camera system has been developed for use in pleural photodynamic therapy (PDT). This system was introduced to pleural PDT to provide uniform light dose distribution to ensure predictable PDT outcome. Light is delivered through a fiber that is in an endotracheal (ET) tube filled with Intralipid as scattering media. A tracking tool is attached to the ET tube to monitor the position of the optical fiber based point source. An anisotropic light distribution model is introduced to correct the angle dependent light distribution due to a capped end by design of the ET tube, which scatters light differently than the sides. In this study, the anisotropic nature of the balloon was characterized and incorporated into the calculation for light fluence during treatment. This model is verified by the light dose calculation from a phantom study. Furthermore, a new tracking tool was designed with multiple faces to increase the angular field of view and thus collect more viable data during treatment. The new tracking tool is directly entered into the ET tube with the light delivering fiber, thus eliminating the need to calibrate the laser source position prior to treatment via an optimization method. With this improved system, the calculated light fluence and the measured isotropic detector readings are more accurately matched.

Keywords

Photodynamic therapy; IR navigation; dosimetry

INTRODUCTION

Photodynamic therapy (PDT) is an effective treatment modality for cancer and other localized diseases. PDT uses light, photosensitizer, and oxygen to create activated singlet oxygen, which then causes cell death and tissue necrosis. Accurate dosimetry calculation in PDT is critical in determining the treatment efficacy. To improve the light dose calculation, an anisotropic correction function was determined to characterize asymmetric nature of the treatment delivery wand. An infrared tracking system has been developed to passively determine the light fluence rate distribution [1, 2, 3, 4]. This calculation was improved with the anisotropy correction from this study.

METHODS AND MATERIALS

Light is delivered through a fiber that is in an endotracheal (ET) tube filled with Intralipid as scattering media. A tracking tool is attached to the ET tube to monitor the position of the optical fiber-based point source. An infrared camera with real-time tracking abilities is used to collect location data during a phantom treatment.

Using an isotropic detector, the anisotropy of the balloon was measured. Assuming azimuthal symmetry about the wand, the measurements were fitted with a polynomial of order 2. As seen in figure 1 (a), the least attenuated light mostly results from the capped end of the balloon (defined as $\theta = 180^\circ$). The anisotropy function (eqn. (1)) was fitted so that the assumptions were satisfied.

$$A(\theta) = p_1\theta^2 + p_2\theta + p_3 \quad (1)$$

A treatment was done on a chest phantom of known geometry. Four isotropic detectors were placed in the chest cavity and treated for 10 J/cm^2 . The measured fluence values were compared with the calculations from the IR camera position data. Previously light calculation to the surface of treatment was done using equation (2), where S is the source strength, and r is the distance from the light source to a point on the surface.

$$\phi = \frac{S}{4\pi r^2} \quad (2)$$

$$\phi = \frac{S}{4\pi r^2} \cdot A(\theta) \quad (3)$$

From the IR navigation system, the position of the treatment delivery point source is given in Cartesian coordinates along with a rotation coordinates (Q_1, Q_2, Q_3, Q_4). The raw data, $R = (x, y, z)$, is said to be in the “camera coordinates,” as seen in figure 2. There also exists coordinates that are called “wand coordinates,” $R' = (x', y', z')$, which are described in figure 1 (c). Coordinate transformations are performed with the inverse of the rotation matrix.

$$R' = M^{-1}R \quad (4)$$

where the rotation matrix, M , is given by

$$M = \begin{bmatrix} Q_1Q_1 + Q_2Q_2 - Q_3Q_3 - Q_4Q_4 & 2(Q_2Q_3 - Q_1Q_4) & 2(Q_2Q_4 + Q_1Q_3) \\ 2(Q_2Q_3 + Q_1Q_4) & Q_1Q_1 - Q_2Q_2 + Q_3Q_3 - Q_4Q_4 & 2(Q_3Q_4 - Q_1Q_2) \\ 2(Q_2Q_4 - Q_1Q_3) & 2(Q_3Q_4 + Q_1Q_2) & Q_1Q_1 - Q_2Q_2 - Q_3Q_3 + Q_4Q_4 \end{bmatrix} \quad (5)$$

The laser position (lpos), $R_L = (x_L, y_L, z_L)$, is predetermined by a laser position calibration process previously described [3]. The data collected can finally be converted to the laser-wand coordinate system

$$R' = R - R_L \quad (7)$$

From the wand coordinates, the desired angle, θ , can be found by the following equation:

$$\theta = \arctan \left(\frac{\sqrt{x'^2 + y'^2}}{z'} \right) \quad (6)$$

This angle is then used in the anisotropy function $A(\theta)$, and applied in the light dose calculation.

RESULTS

The anisotropy function fitted was a polynomial of order 2 with the following values:

$$\begin{aligned} p_1 &= 2.764 \times 10^{-5} \\ p_2 &= 0.01027 \\ p_3 &= 0.03685 \end{aligned}$$

The new light fluence calculation method did account for some of the over-estimated light calculations made with the original formula. Figure 4 shows four detectors used for a treatment with the original calculation method (eqn (2)), as well as the same four detectors used for the same treatment with the calculation method that incorporates balloon anisotropy (eqn (3)). While the data is better matched, it is still difficult to account for the larger variation after a long treatment time. Any attenuation of the light from the wand was not accounted for.

A large factor affecting the anisotropy modeling is the variation in ET tube balloons. Each balloon has its own unique shape and anisotropy, which is difficult to model. Further studies would have to be done with a number of different treatment wands and better angular resolution. Angular measurements should be done at various distances from the wand as well.

CONCLUSION AND FUTURE WORK

Anisotropy will be averaged over many different treatment wands. Furthermore, phantom treatment conditions will be modeled to better imitate clinical treatment cases, such as filling the treatment cavity with scattering media, using more detector locations, and treating for long periods of time.

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Figure 1. Treatment delivery system. (a) The tracking wand attached to the treatment delivery wand. Light anisotropy is emphasized with low laser output. (b) Anisotropy measurement set-up. An isotropic detector was rotated around the light scattering balloon at small angular intervals. (c) Schematic of treatment wand anisotropy angle definition.

Treatment positions, detector positions and contour

Figure 2. Phantom contour with detector positions. The chest phantom was treated with two detectors along the anterior wall (B and C) and detectors placed on the sides (A and D).

Figure 3.

Treatment wand balloon anisotropy. Relative light fluence measurements (circles) were obtained every 5 degrees. The two sets of circles represent different azimuthal measurements. Assuming azimuthal symmetry, an anisotropy function (solid line) was fitted according to equation (1). The solid flat line represents the function $A(\theta)$ without the anisotropic model (constant).

Figure 4. Comparison of measured and calculated doses. The thinnest line represents the measured data. The next thickest line represents the original calculation method. The thickest line represents the calculation with the anisotropy function. The detector locations are described in Figure 2.

Table 1

Error between original calculation and new anisotropy calculation compared with the measurement for the 4 detectors. Detectors A and D are located on the left and right sides of the chest phantom, and detectors B and C are located on the anterior wall of the chest phantom.

	A	B	C	D
Normal	9.5%	20.7%	66.2%	34.5%
Anisotropy	2.2%	9.5%	39.4%	12.9%