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Light Dosimetry at Tissue Surfaces for Small Circular Fields

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Abstract

Small circular light fields (2 cm diameter) are sometimes used for photodynamic therapy of skin and recurrent breast cancers on the chest wall. These fields have lateral dimensions comparable to the effective mean free path of photons in the turbid medium, which causes reduced light fluence rate compared to that of a broad beam of uniform incident irradiance. We have compared Monte-Carlo simulation with in-vivo dosimetry for circular fields ($R = 0.25, 0.35, 0.5, 0.75, 1, 2, 3$, and 8 cm) in a liquid phantom composed of intralipid and ink ($\mu_s' = 4 - 20 \text{ cm}^{-1}$ and $\mu_a = 0.1 \text{ cm}^{-1}$) for wavelengths between 532 and 730 nm. We used anisotropy $g = 0.9$ and the index of refraction $n = 1.4$ for all Monte-Carlo simulations. The measured light fluence rate agrees with Monte-Carlo simulation to within 10%, with the measured value lower than that of the Monte-Carlo simulation on tissue surface. The ratio of the peak fluence rates between a circular beam and a broad beam under tissue is $0.58 - 0.96$ or $0.84 - 1.00$ for R between $0.5 - 2$ cm and $\mu_{\text{eff}} = 1.1$ or 2.0 cm^{-1} , respectively. The ratio of peak fluence rate and incident irradiance for the broad beam is 5.9 and 6.4 for $\mu_{\text{eff}} = 1.1$ and 2.0 cm^{-1} , respectively. The optical penetration depth δ varies from $0.34 - 0.48$ cm for R between 0.5 and 2 cm, with the corresponding $\delta = 0.51$ cm for a broad beam. The ratio of fluence rate and incident irradiance above tissue surface is $1.4 - 1.8$ or $1.9 - 2.2$ for R between $0.5 - 2$ cm and $\mu_{\text{eff}} = 1.1$ or 2.0 cm^{-1} , respectively. At depth of 0.2 cm inside tissue, Off-axis ratio OAR , defined as the ratio of fluence rate at off-axis distance r to that on the central axis, varies between $0.91 - 0.54$ or $0.93 - 0.52$ for off-axis distances r between 0.6 and 1.0 cm and $\mu_{\text{eff}} = 1.1$ or 2.1 cm^{-1} , respectively. In conclusion, in-vivo light dosimetry agrees with Monte-Carlo simulation for small field dosimetry provided the isotropic detector is corrected for the blind spot. The light fluence rates for small circular fields are substantially lower than that of the broad beam of the same incident irradiance.

Keywords

Small circular fields; in-vivo light dosimetry; tissue optical properties; Monte Carlo; diffusion theory; diffuse reflectance

1. INTRODUCTION

Photodynamic therapy (PDT) is a cancer treatment that involves the use of a photosensitizer, oxygen, and light of a wavelength specific to the absorption characteristics of the photosensitizer¹. Several clinical trials are on-going at University of Pennsylvania for

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patients with surface malignancies. A Phase II clinical trial of Motexafin lutetium-mediated PDT for patients with recurrent breast cancers using the 732 nm light has been completed². Motexafin lutetium (MLu) is a pentadentate aromatic metalloporphyrin manufactured by Pharmacyclics, Inc.³. Photofrin-mediated PDT is also being used for skin cancer using the 630 nm light. Photofrin® is a purified mixture of porphyrin monomers and oligomers manufactured by QLT, Ltd. Vancouver, Canada⁴. These photosensitizing drugs are not sufficiently tumor specific; hence, light dosimetry is required in order to control light fluence and thereby restrict cell kill to the target tissue to avoid damage to healthy tissue.

When treating the surface lesions with PDT, it is often necessary to treat a lesion as small as 2-cm in diameter. For these small fields, the light fluence rate reduces significantly from that for an infinite large field because of the reduced photon scattering by tissue. The clinical questions to ask are (1) how does the light fluence rate change with field size, given similar treatment conditions? (2) What lateral extent of the light field is sufficient to cover the tumor at depth in tissue? For the theoretical calculations, we used a Monte-Carlo simulation because the diffusion theory is not valid when the lateral dimension of beam geometry becomes comparable to the mean-free-path of the photons or when the region of interest is near the air-tissue interface. We also measured light fluence rate as a function of field size at tissue surface and at several depths in an optical phantom for several wavelengths to confirm the theoretical calculation.

In a previous study, we determined the ratio between the light fluence rate and the incident irradiance ϕ/ϕ_{air} as a function of the tissue optical properties for a broad laser beam incident on the skin surface, both inside and outside tissue⁵. The current work extends the relationship to a light field with a finite circular dimension. In this study, we concentrate on light dosimetry for the subcutaneous soft tissue, whereby tissue is treated as a homogeneous thick layer. The effect of multiple layers of tissue types⁶ are beyond the scope of the current study.

2. METHODS AND MATERIALS

1. Monte-Carlo Simulation

The experimental setup to be calculated by Monte-Carlo simulation for a semi-infinite medium with uniform optical properties, i.e. the absorption coefficient μ_a , the scattering coefficient μ_s , the scattering anisotropy g ($= 0.9$), and the index of refraction n is shown in Fig. 1a. The outside medium is air ($n_0 = 1$). The light field is parallel and uniform inside a circle with radius R and incident toward the air-tissue interface. Most tissues have a preferential scattering in the forward direction with a scattering anisotropy $g = 0.90$ ⁷. In the diffusion approximation, the tissue scattering is converted to isotropic conditions ($g = 0$) and the tissue scattering property is described by a reduced scattering coefficient $\mu_s' = \mu_s (1-g)$.

To improve the scoring efficiency, Monte-Carlo simulation is performed for a pencil-beam incident on a semi-infinite tissue phantom (Fig. 1b). The photon density (ρ_n) is scored in a cylindrical geometric relationship consisting of rings of 0.01 cm thickness (dz) and 0.02 cm width (dr). A ring voxel is shown in Fig. 1b. The light fluence rate (ϕ) per unit incident light power of the pencil beam (in unit of $1/\text{cm}^2$) is calculated as $\rho_n N_{inc} / \mu_a$, where ρ_n is the

photon density (in unit of $1/\text{cm}^3$), N_{inc} is the number of incident photons, and μ_a is the absorption coefficient (in unit of $1/\text{cm}$). Reciprocity theorem⁸ is used to calculate ratio of the light fluence rate ϕ to the incident irradiance (or light fluence rate in air ϕ_{air}) on the central axis of a circular field with radius R by an area integral:

$$\phi/\phi_{air} = \int_0^R \rho_n / N_{inc} / \mu_a \cdot 2\pi r dr. \quad (1)$$

We have also scored the fractional photons escaping the air-tissue interface into the air as a function of lateral radius r (with a resolution of 0.02 cm) and emitting angle θ (with a resolution of 0.18°). The reciprocity theorem is used to calculate the angular distribution of differential diffuse reflectance r_d for circular fields as a function of radius and emitting angle θ . The light fluence rate above the tissue surface is calculated using the diffuse reflectance R_d ($R_d = \int_0^{90^\circ} r_d d\theta$) as previously described^{5,9}

$$\phi/\phi_{air} = 1 + 2R_d. \quad (2)$$

Here we have assumed that the tissue surface is a Lambertian surface. Under this assumption, the radiant intensity (also called radiance¹⁰) emerging from the tissue surface follows $\cos\theta$ for an azimuth angle θ ¹¹. This assumption will be examined from the MC calculated r_d as a function of emitting angle. Equation (2) states that the light fluence at a tissue surface is composed of 1 part from the incident photons and $2R_d$ part from the backscattered photons, according to the theory described by Marijnissen and Star¹¹.

The Monte-Carlo algorithm is similar to approaches used previously in the literature¹²⁻¹⁵. The effect of reflection at the air-tissue interface, resulting from the refractive index mismatch, was modeled according to the Fresnel reflection coefficient for unpolarized light¹². Photons are propagated through the medium using a step size $s = mfp \cdot \ln\xi$, where ξ is a uniform distributed random number and $mfp = 1/(\mu_s + \mu_a)$ is the mean free path and a scattering angle selected from the phase function determined by the Henyey-Greenstein phase function¹³ with anisotropy $g = 0.90$. Each incident photon (normal to the interface along the z direction) is followed until it is escaped from the turbid medium at the interface or when its weight is insignificant. We followed 1 million incident photons for each Monte-Carlo simulation. Each simulation takes about 20 hours on a 1.8 GHz Pentium 4 PC, depending on the optical properties used. The code is written in matlab® language. The MC result has a typical statistical uncertainty of less than 0.1%. This small standard variation for only 10^6 incident photons is possible because a convolution of pencil-beam result is used to calculate the broad beam parameters.

2. Phantom Experiments

The phantom measurements were performed in a liquid phantom made of intralipid (0.25, 1, 1.5%) and ink to verify the Monte-Carlo results, as shown in Fig. 1. Series of circular fields (radius $R = 0.25, 0.35, 0.5, 0.75, 1, 1.5, 2, 3$ cm), defined at the phantom surface, were produced by a circular block made of blackened paper. The laser was connected to an optical fiber fitted with a microlens at the exit tip, producing a circular field on the phantom

surface of 8 cm radius. The laser system used for measurement was a Nd:Yag KTP laser (LaserScope, Model 820) operating at 30 W power at 532 nm, and a dye laser module (Laser Scope, 600 Series) operating at 7W at 630 nm. A 3W diode laser (DioMed, Inc, Cambridge, UK) provided light source for 730 nm. The output of the laser was verified by a Coherent power meter with a calibration traceable to the National Institute of Technology and Standards. The phantom has known optical properties determined independently⁵. The phantom has a reduced scattering coefficients of ($\mu_s' = 3.7, 13, \text{ and } 21 \text{ cm}^{-1}$) and an absorption coefficient of $\mu_a = 0.1 \text{ cm}^{-1}$. It was necessary to change the concentration and the types of ink to keep μ_a constant for different wavelengths: 0.01% black ink for 532 nm, 0.015% green ink for 630 nm and 730 nm.. A 1 mm-diameter isotropic detector¹⁶ was used to measure the fluence rate on the phantom surface (0.2 cm above phantom) and inside the phantom at various depths ($z = 0.2, 0.5, 1, 1.5 \text{ cm}$). The isotropic detector was manufactured by CardioFocus, Inc. (west Yarmouth, MA) with an anisotropy of less than 10% and should measure the fluence rate accurately. The isotropic detector was calibrated in a 15.2 cm diameter integrating sphere (Labsphere, North Sutton, NH) using the method previously described¹⁶.

Since the isotropic detector has different responses in air and in water due to the mismatch of the index of refraction¹¹, we also made measurements of the isotropic detector response in pure water, without any scattering medium. The ratio of the isotropic detector response in liquid tissue-simulating phantom and pure water, under similar conditions, was used to determine the true light fluence rate in tissue. The measured fluence rate was then normalized to the incident light fluence rate for a broad beam in free air, at the same distance from the light source. An additional correction was made to correct for the under-response of the isotropic detector to scattered light due to a blind spot near the stem where light was not detectable¹¹. This effect was about 15% for the isotropic detector used in this study and was determined by measuring the ratio ϕ/ϕ_{air} of the isotropic detector with and without a perfectly diffused reflecting surface with a diffused reflectance of $R_d = 0.99$.

3. RESULTS AND DISCUSSIONS

The comparison between measurements and Monte-Carlo simulations on a semi-infinite medium is shown in Figs. 2 and 3. Figure 2 shows the radius dependence for three different wavelengths (532, 630, 730 nm). The optical properties of the phantom are $\mu_a = 0.1 \text{ cm}^{-1}$ and $\mu_s' = 13 \text{ cm}^{-1}$. Figure 3 shows the depth dependence. The optical properties used for the MC simulation are $\mu_a = 0.1 \text{ cm}^{-1}$ and $\mu_s' = 4, 13, \text{ and } 20 \text{ cm}^{-1}$.

Figure 2 shows that once the optical properties (μ_a and μ_s') are fixed, the light fluence rate becomes independent of wavelength. For 1% Intralipid, μ_s' does not substantially vary with the wavelength: $\mu_s' = 13.5, 14.1, 13.1 \text{ cm}^{-1}$ for 532 nm, 630 nm, and 730 nm, respectively. The absorption coefficient is the same for all wavelengths ($\mu_a = 0.1 \text{ cm}^{-1}$).

The measured ϕ/ϕ_{air} agreed with Monte-Carlo (MC) simulation inside tissue to within 10%, as shown in Fig. 3, for μ_s' between 4 and 20 cm^{-1} . The measured value tended to be larger than the MC calculation for small radii. On the tissue surface, the opposite became true, i.e. the measured ϕ/ϕ_{air} was smaller than that from the MC simulation. The difference between

the measurement and the MC simulation was typically less than 20%, except for the measurement taken above the phantom surface under conditions with $\mu_a = 0.1 \text{ cm}^{-1}$ and $\mu_s' = 20 \text{ cm}^{-1}$, where a slightly larger deviation was observed (26%).

One potential cause for this difference between the measurement and the Monte-Carlo simulation is the difference of index of refraction between water ($n = 1.33$) and the tissue ($n = 1.4$). Figure 4 shows the influence of n on the light fluence rate for otherwise the same optical properties (μ_a and μ_s'). When the index of refraction is increased, the light fluence rate inside the tissue also increases but the light fluence rate outside of tissue decreases. One would not expect a change of curve shape since the index of refraction n affects how many photons are transmitted inside the phantom and how many photons escape from the tissue surface. This effect, however, does not seem to be the cause of the difference we observed in this study because the trend was opposite from what was observed.

The other possible cause of the difference between the measurement and the MC simulation is an overcorrection of the isotropic detector response inside tissue. The isotropic detector is estimated to underestimate light fluence rate by 10 % in a water-based turbid medium¹⁷. However, we applied a correction factor on the order of 15% based on the detector underresponse measured on the phantom surface. This over correction may result in the detector measuring a higher light fluence rate than the actual fluence rate.

On the whole, the relative good agreement between our measurement and the MC simulation suggests that the MC results is useful to predict ϕ/ϕ_{air} for circular fields inside and above tissue surface.

For a fixed absorption coefficient μ_a , the peak light fluence rate increases with increasing μ_s' (or μ_{eff}), as shown in Fig. 3. However, the optical penetration depth ($\delta = 1/\mu_{eff}$) decreases with increasing μ_s' because the effective attenuation coefficient $\mu_{eff} = \sqrt{3\mu_a\mu_s'}$ increases. Table 1 shows the range of the ratio of the peak fluence rate, relative to that of a broad beam ($R = 8 \text{ cm}$), as a function of the radius R of the circular field in tissue. The ratio of the peak fluence rates between a circular beam and a broad beam under tissue is 0.58 - 0.96 or 0.84 - 1.00 for r between 0.5 - 2 cm and $\mu_{eff} = 1.1$ or 2.0 cm^{-1} , respectively.

Figure 3 also shows that the penetration depth of light fluence rate ($\delta = 1/\mu_{eff}$) decreases with decreasing radius due to the reduced scattering of photons for smaller fields. This value is obtained by an exponential fit to the depth dependence of ϕ/ϕ_{air} well within the tissue. For a larger field radius with sufficient photon scattering inside the irradiated area of the turbid phantom, the penetration depth becomes a constant, independent of the field radius. Table 2 summarizes the radius dependence of the penetration depth. δ starts to reduce significantly when the radius of circular field is less than 2 cm. The reduction of optical depth is less for smaller μ_{eff} than for larger μ_{eff} . The optical penetration depth δ varies from 0.34 - 0.48 cm for r between 0.5 and 2 cm, with the corresponding $\delta = 0.51 \text{ cm}$ for a broad beam.

The radius dependence of ϕ/ϕ_{air} in tissue is shown in Fig. 5 for μ_s' between 4 and 20 cm^{-1} and $\mu_a = 0.1 \text{ cm}^{-1}$. Table 3 summarizes the values of ϕ/ϕ_{air} for different beam radii and depth in tissue. ϕ/ϕ_{air} is always larger than 1 above the tissue surface, as predicted by Eq.

(2). The ratio of fluence rate and incident irradiance above the tissue surface is 1.4 - 1.8 or 1.9 - 2.2 for R between 0.5 - 2 cm and $\mu_{\text{eff}} = 1.1$ or 2.0 cm^{-1} , respectively. Just underneath the tissue surface, ϕ/ϕ_{air} increases significantly after an initial buildup (Fig. 3). At depth of maximum fluence rate, ϕ/ϕ_{air} can be as high as 6. The ratio of peak fluence rate and incident irradiance for the broad beam is 5.9 and 6.4 for $\mu_{\text{eff}} = 1.1$ and 2.0 cm^{-1} , respectively. This phenomena is caused by strong photon scattering. The total energy is still conserved by the fact that the effective attenuation of the light (μ_{eff}) is much larger than that caused by the absorption (μ_a) alone.

To show the relative change of the fluence rate with beam radius, Table 4 summarizes the ratio of fluence rate to that of a broad beam ($R = 8 \text{ cm}$) for the same depth and optical properties. The reduction of this ratio is significant for $R < 2 \text{ cm}$ (or a 4-cm-diameter field). The degree of ratio reduction increases with increasing depth, reflecting reduced optical penetration depth as discussed above (Table 2). The radius dependence of the ratio $\phi(R)/\phi(R=8\text{cm})$ above tissue does not change as much as that inside the tissue. For $\mu_a = 0.1 \text{ cm}^{-1}$ and $\mu_s' = 13 \text{ cm}^{-1}$, the ratio changed between 0.73 - 0.98 above tissue surface for radius R between 0.5 - 2.1 cm, while it changed between 0.14 - 0.96 at depth of 0.2 cm for the same radius range.

ϕ/ϕ_{air} above the tissue surface was calculated according to Eq. (2). This requires that the radiance of exiting photons follows a $\cos\theta$ law, where θ is the angle between the axis of the air-tissue interface and exit direction. Figure 6 examines the angular distribution of the differential diffuse reflectance, $r_d = dR_d/d\theta$, for different beam radii and optical properties. Based on the conservation of the number of photons, r_d is related to radiance L by $r_d d\theta = L d\Omega/\phi_{\text{air}}$, where $d\Omega = 2\pi\sin\theta d\theta$ and Ω is the solid angle. Thus $L = r_d \phi_{\text{air}}/2\pi\sin\theta$. Figure 6 shows that the angular dependence of the MC calculated r_d follows a function $\sin 2\theta = 2\sin\theta \cos\theta$ for all beam radius and optical properties examined. As a result, we conclude that the exit radiance follows the Lambert law, i.e., $L \propto \cos\theta$. Other people have confirmed the same result¹⁸.

The radius dependence of ϕ/ϕ_{air} is not a function of μ_{eff} alone, rather it is a function of both μ_a and μ_s' . Figure 7 shows that even if one kept μ_{eff} constant ($\mu_{\text{eff}} = 1.1 \text{ cm}^{-1}$), the beam radius as well as the depth dependence of ϕ/ϕ_{air} is very different if both μ_a and μ_s' are very different. Diffusion theory¹⁹ has proven that the light fluence rate on tissue surface is only a function of transport albedo, $a' = \mu_s' / (\mu_a + \mu_s')$. For the two cases examined in Fig. 7, the values of a' are $a' = 0.714$ and 0.976 for $\mu_a = 0.4 \text{ cm}^{-1}$, $\mu_s' = 1 \text{ cm}^{-1}$ and $\mu_a = 0.1 \text{ cm}^{-1}$, $\mu_s' = 4 \text{ cm}^{-1}$, respectively.

The lateral distribution of light fluence rate can be calculated using a convolution of the pencil-beam kernel and the uniform unit incident light:

$$\phi/\phi_{\text{air}} = k \otimes I = \iint k(x' - x, y' - y, z) \cdot I(x', y') dx' dy', \quad (3)$$

where I is 1 inside and 0 outside the light field radius and k is the pencil beam kernel calculated by MC simulation: $k = \rho_n / N_{ino} / \mu_a$. Note that the pencil-beam kernel is cylindrically symmetrical, i.e. it is a function of radius $\sqrt{x^2 + y^2}$ and depth z only.

Figure 8 shows the lateral ϕ/ϕ_{air} distribution at different depths for a 2-cm diameter beam for $\mu_a = 0.1 \text{ cm}^{-1}$ and four different μ_s 's. The beam penumbra width is approximately 2 cm regardless of the optical properties. If the beam radius is smaller than 1 cm, then it is possible for the penumbra of the light field to cut into each other, resulting in ϕ/ϕ_{air} on the central axis to decrease. This might be the reason why the radius dependence increases significantly for $R < 1 \text{ cm}$ (or 2-cm-diameter beam).

To characterize the variation of the fluence rate as a function of lateral distance r , the off-axis ratio $OAR(r, z)$ was defined as the ratio of fluence rate at off-axis distance r to the fluence rate on the central axis, at the same depth z . Table 5 summarizes OAR for various optical properties. At depth of 0.2 cm inside tissue, OAR varied from 0.91 – 0.54 or 0.93 – 0.52 for off-axis distances r between 0.6 and 1.0 cm and for $\mu_{eff} = 1.1$ or 2.0 cm^{-1} , respectively.

4. CONCLUSION

We have studied the ratio ϕ/ϕ_{air} between the light fluence rate and the incident irradiance using both Monte Carlo calculations and phantom measurements. Previously we have demonstrated that it is possible to estimate the tissue optical properties using the ratio between fluence rate and incident irradiance using the diffusion theory⁴. However, the diffusion theory used is only valid when the lateral dimension is so large that the medium can be treated as infinite laterally. Our Monte-Carlo study predicts that the light fluence rate in tissue will drop substantially for small light fields. We have shown in this study the variation of ϕ/ϕ_{air} as a function of beam radius and depth.

5. REFERENCES

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Fig. 1.

(a) Experimental setup for Monte-Carlo simulation for a semi-infinite phantom. The light fluence rate are calculated along the central-axis on the center of the field. (b) Pencil beam setup is actually used for the Monte Carlo simulation. The ring scoring voxel has radius width dr and thickness dz at depth z and radius r in a cylindrical geometry.

Fig. 2.

Comparison of measured (symbols) and Monte Carlo calculated (lines) ϕ/ϕ_{air} in a semiinfinite turbid medium (1% Intralipid and $\mu_a = 0.1 \text{ cm}^{-1}$) as a function circular field diameter, $Dia = 2R$. Symbols are for different wavelengths: o – 532 nm, + – 630 nm, and x – 730 nm. The MC simulation is made for $\mu_a = 0.1 \text{ cm}^{-1}$, $\mu_s = 130 \text{ cm}^{-1}$, $g = 0.9$, and $n = 1.4$.

Fig. 3.

Comparison of measured (symbols) and Monte Carlo calculated (lines) ϕ/ϕ_{air} in a semi-infinite turbid medium as a function of depth for different intralipid concentrations: (a) 0.25%, (b) 1%, (c) 1.5%. $\mu_a = 0.1 \text{ cm}^{-1}$ for all plots. Curves are for different field radius: from bottom to top, $R = 0.25, 0.35, 0.5, 0.75, 1, 1.5, 2, 3, 8 \text{ cm}$.

Figure 4.

Effect of the index of refraction on the depth dependence of light fluence rate ϕ/ϕ_{air} for (a) depth and (b) radius dependence. The solid lines are for $n = 1.4$ and the dashed lines for $n = 1.33$, both are calculated by Monte-Carlo simulation. The curves in (a) are for different beam radius, from bottom to top, $R = 0.1, 0.25, 0.35, 0.5, 0.75, 1, 1.5, 2, 3, 8$ cm.

Figure 5.

Radius dependence of light fluence rate under uniform irradiation of circular light beam at various depths in tissue. For comparison, ϕ/ϕ_{air} above the tissue surface in air and at very shallow depth inside tissue (at depth of 0.01 cm) are shown. The optical properties for each subplot are $\mu_a = 0.1 \text{ cm}^{-1}$ and μ_s' are (a) 4 cm^{-1} , (b) 10 cm^{-1} , (c) 13 cm^{-1} , and (d) 20 cm^{-1} . Error bars of MC results are plotted for selected points (symbols).

Figure 6.

Angular distribution of differential diffuse reflectance, $r_d = dR_d/d\theta$, as a function of beam radius (from bottom to top, $R = 0.1, 0.25, 0.35, 0.5, 1, 2, 3, 8$ cm) and optical properties: μ_s' = (a) 4 cm^{-1} , (b) 10 cm^{-1} , (c) 13 cm^{-1} , (d) 20 cm^{-1} and $\mu_a = 0.1 \text{ cm}^{-1}$. Error bars are plotted for each point.

Figure 7.

Comparison of ϕ/ϕ_{air} for two tissue phantoms with the same $\mu_{eff} (=1.1 \text{ cm}^{-1})$ but different μ_a : (a) depth dependence for $\mu_a = 0.4 \text{ cm}^{-1}$, $\mu_s' = 1 \text{ cm}^{-1}$; (b) depth dependence for $\mu_a = 0.1 \text{ cm}^{-1}$, $\mu_s' = 4 \text{ cm}^{-1}$; (c) radius dependence for $\mu_a = 0.4 \text{ cm}^{-1}$, $\mu_s' = 1 \text{ cm}^{-1}$; (d) radius dependence for $\mu_a = 0.1 \text{ cm}^{-1}$, $\mu_s' = 4 \text{ cm}^{-1}$. Error bars of MC results are plotted for selected points (symbols).

Figure 8.

The lateral distribution of light fluence rate for a 2-cm diameter circular beam at various depths (from top to bottom, 0.2, 0.5, 1.0, 1.5 cm) for different optical properties.

Table 1

Ratio of peak fluence rate in tissue between a circular beam and a broad beam ($R = 8$ cm). The absorption coefficient is fixed ($\mu_a = 0.1 \text{ cm}^{-1}$) and the reduced scattering coefficient μ_s' varies between 4 and 20 cm^{-1}

$\mu_{\text{eff}} (\text{cm}^{-1}) \setminus R (\text{cm})$	0.1	0.25	0.35	0.5	0.75	1.0	1.5	2	8
1.1	0.273	0.395	0.474	0.582	0.696	0.808	0.910	0.956	1.000
1.7	0.322	0.564	0.677	0.793	0.869	0.944	0.983	0.994	1.000
2.0	0.356	0.630	0.741	0.844	0.928	0.966	0.991	0.997	1.000
2.4	0.436	0.737	0.834	0.911	0.966	0.986	0.997	0.999	1.000

Table 2

Optical penetration depth δ (in unit of cm) as a function of the radius of circular beams, R . The absorption coefficient is fixed: $\mu_a = 0.1 \text{ cm}^{-1}$ and the reduced scattering coefficient μ_s' varies between 4 and 20 cm^{-1}

$\mu_{\text{eff}} (\text{cm}^{-1}) \setminus R (\text{cm})$	0.25	0.35	0.5	0.75	1.0	1.5	2	3	8
1.1	0.45	0.49	0.51	0.54	0.63	0.70	0.91	0.91	0.91
2.0	0.32	0.34	0.35	0.36	0.41	0.45	0.48	0.50	0.51
2.4	0.31	0.32	0.32	0.33	0.36	0.38	0.39	0.40	0.41

Table 3 ϕ/ϕ_{air} at various depths for different optical properties (μ_a and μ_s')

$\mu_a, \mu_s' \text{ (cm}^{-1}\text{)}$	Depth \ R (cm)	0.1	0.3	0.5	0.7	1.1	1.5	2.1	8
$\mu_a = 0.1$ $\mu_s' = 4$	Above surface	1.067	1.252	1.423	1.558	1.732	1.825	1.891	1.939
	0.2	0.605	1.613	2.341	2.893	3.594	3.969	4.234	4.429
	0.5	0.074	0.560	1.124	1.617	2.324	2.741	3.059	3.307
	1.0	0.014	0.132	0.332	0.567	1.008	1.335	1.627	1.902
	1.5	0.005	0.045	0.122	0.222	0.445	0.645	0.856	1.100
$\mu_a = 0.1$ $\mu_s' = 13$	Above surface	1.272	1.789	2.031	2.147	2.237	2.263	2.275	2.278
	0.2	0.433	2.136	3.241	3.839	4.343	4.503	4.570	4.590
	0.5	0.066	0.558	1.168	1.650	2.186	2.397	2.496	2.528
	1.0	0.010	0.097	0.243	0.404	0.662	0.809	0.899	0.936
	1.5	0.002	0.023	0.061	0.107	0.200	0.268	0.321	0.349
$\mu_a = 0.1$ $\mu_s' = 20$	Above surface	1.437	2.033	2.234	2.313	2.364	2.376	2.380	2.381
	0.2	0.451	2.286	3.362	3.872	4.239	4.333	4.365	4.371
	0.5	0.063	0.532	1.086	1.496	1.894	2.025	2.075	2.087
	1.0	0.008	0.076	0.185	0.301	0.473	0.558	0.600	0.613
	1.5	0.001	0.014	0.037	0.065	0.116	0.149	0.171	0.179

Table 4

Ratio of fluence rate in tissue between a circular beam and a broad beam at various depths for different optical properties (μ_a and μ_s')

μ_a, μ_s' (cm^{-1})	Depth \ R (cm)	0.1	0.3	0.5	0.7	1.1	1.5	2.1	8
$\mu_a = 0.1$ $\mu_s' = 4$	Above surface	0.550	0.646	0.734	0.803	0.893	0.941	0.975	1.000
	0.2	0.136	0.364	0.529	0.653	0.811	0.896	0.956	1.000
	0.5	0.022	0.169	0.340	0.489	0.703	0.829	0.925	1.000
	1.0	0.007	0.069	0.175	0.298	0.530	0.702	0.856	1.000
	1.5	0.004	0.041	0.111	0.202	0.404	0.587	0.778	1.000
$\mu_a = 0.1$ $\mu_s' = 13$	Above surface	0.558	0.785	0.892	0.942	0.982	0.994	0.998	1.000
	0.2	0.094	0.465	0.706	0.836	0.946	0.981	0.996	1.000
	0.5	0.026	0.221	0.462	0.652	0.865	0.948	0.987	1.000
	1.0	0.011	0.104	0.260	0.432	0.708	0.865	0.961	1.000
	1.5	0.006	0.066	0.174	0.308	0.574	0.768	0.919	1.000
$\mu_a = 0.1$ $\mu_s' = 20$	Above surface	0.604	0.854	0.938	0.972	0.993	0.998	1.000	1.000
	0.2	0.103	0.523	0.769	0.886	0.970	0.991	0.999	1.000
	0.5	0.030	0.255	0.520	0.717	0.908	0.970	0.994	1.000
	1.0	0.013	0.124	0.302	0.491	0.772	0.910	0.979	1.000
	1.5	0.008	0.079	0.207	0.362	0.648	0.833	0.955	1.000

Table 5Off-axis Ratio for a 2-cm diameter circular beam at various depths for different optical properties (μ_a and μ_s')

$\mu_a, \mu_s' \text{ (cm}^{-1}\text{)}$	Depth \ r (cm)	0	0.2	0.4	0.6	0.8	1.0	1.2	1.4
$\mu_a = 0.1$ $\mu_s' = 4$	0.2	1.000	0.992	0.964	0.909	0.806	0.538	0.269	0.163
	0.5	1.000	0.987	0.946	0.869	0.743	0.569	0.393	0.261
	1.0	1.000	0.984	0.936	0.858	0.752	0.631	0.506	0.389
	1.5	1.000	0.986	0.944	0.876	0.789	0.692	0.589	0.486
$\mu_a = 0.1$ $\mu_s' = 13$	0.2	1.000	0.994	0.973	0.920	0.794	0.519	0.239	0.107
	0.5	1.000	0.988	0.946	0.863	0.723	0.537	0.349	0.204
	1.0	1.000	0.983	0.930	0.841	0.720	0.582	0.442	0.313
	1.5	1.000	0.983	0.932	0.852	0.748	0.631	0.511	0.396
$\mu_a = 0.1$ $\mu_s' = 20$	0.2	1.000	0.996	0.978	0.932	0.806	0.515	0.218	0.085
	0.5	1.000	0.989	0.951	0.869	0.724	0.527	0.328	0.179
	1.0	1.000	0.983	0.929	0.837	0.710	0.563	0.415	0.283
	1.5	1.000	0.981	0.929	0.842	0.731	0.606	0.479	0.359