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Abstract. Most methods for measuring light-tissue interaction focus on volume reflectance, while very few measure light transmission. In a previous work, we suggested investigating the influence of blood vessel diameter on photons exiting the tissue at all exit angles to receive the full scattering profile. By this method, we have shown that there is a central angle, i.e., the isobaric point, independent of blood vessel diameter. The vessel diameter changes the effective reduced scattering coefficient. However, both the scattering profile and the value of the isobaric point strongly depend on optical properties and the exact geometry of the tissue. In this study, we investigate the dependency of the isobaric point on tissue diameter and scattering coefficient in both two-dimensional and three-dimensional simulations. We show that the value of this point linearly depends on tissue diameter. The findings of this work solve the dilemma of whether to measure transmission or reflection since the isobaric point reduces by half the total amount of exiting photons. Furthermore, the full scattering profile is sensitive to changes in the scattering properties, but a single isobaric point to these changes is expected. If this point is not found, it is a diagnostic indication of an unexpected change in the tissue. © The Authors. Published by SPIE under a Creative Commons Attribution License 3.0 Unported License. Distribution or reproduction of this work in whole or in part requires full attribution of the original publication, including its DOI. [DOI: 10.1117/1.JBO.19.2.026007]

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1 Introduction

Human tissue is one of the most complex optical media since it is turbid and nonhomogeneous. Hence, optical properties, such as absorption coefficient $\mu_a$, scattering coefficient $\mu_s$, and anisotropy factor $g$, are unknown and vary in different areas and physiological states. Furthermore, in respiration, blood vessels, which are the main cause for light absorption and also cause strong scattering, vary in size.

Measurement of light-tissue interaction uses either direct methods that need no theoretical models of light propagation or indirect methods. Direct methods include attenuation measurements in calibration solutions or integrating spheres for measurements of total transmitted light through a blood sample and total reflected light from it. Indirect methods are based on a mathematical simulation model of photon migration. Photon migration simulations are commonly described by the diffusion theory. However, in the close region near the light source, diffusion theory does not accurately describe the light distribution. The Monte Carlo (MC) simulation method assumes that all photons begin as ballistic photons and that change in their direction is due to the scattering coefficient of the medium.

Most optical-physiological diagnostic methods are based on the insertion of light with known parameters to a tested tissue, followed by the measurement of the re-emitted light. When investigating light–tissue interaction, human tissue is usually dealt with as a semi-infinite surface. Hence, the investigation is commonly based on the light reflected from the tissue (volume reflection). Very few methods test the transmitted light or ballistic photons. It has been previously suggested looking for the first time at the full profile of light that is scattered (transmitted and reflected) at all possible exit angles from a circle of tissue, such as a fingertip joint, ear lobe, or pinched tissue. In this work, we found that the full scattering profile is affected by blood vessel diameter. This effect can be explained by the change in the effective scattering coefficient. This parameter is influenced by both tissue and blood vessels, where larger blood vessels are less effective in light absorption because of the shielding effect. However, we have also made a unique discovery that there is an angle (i.e., the isobaric point) that is indifferent to these changes. We have mentioned that the value of this isobaric point depends on optical properties and exact geometry; however, further investigation is necessary.

In this work, we establish the meaning and usefulness of this isobaric point and calculate the change in its value in both two-dimensional (2-D) and three-dimensional (3-D) simulations in a simplified tissue model. In this model, we replaced the change in blood vessel diameter with the change in the effective reduced scattering coefficient. Since absorption due to blood vessels does not change the shape of the full scattering profile, it can influence the total scattered intensity and is neglected in the current simulations. We calculate the full scattering profile via MC simulation of a circular cross-section of tissue in the near-infrared (NIR) wavelength of 850 nm since it is common in photoplethysmography (PPG) measurements due to its high penetration depth and minimal scattering. We investigate the dependency of the isobaric point in case of different tissue diameters and show that this phenomenon is independent of the value of the anisotropy factor. The new isobaric point can be implemented in NIR spectroscopy, PPG experiments, and analyzing exact oxygen saturation values.
2 Materials and Methods

2.1 2-D Model of Circular Tissue

To understand the influence of tissue diameter on the isobaric point, several simulations were done on a 2-D model of a cross-section of circular tissue (Fig. 1). A beam of photons with a waist of \( w_0 = 1.5 \text{ mm} \) enters the circle of tissue parallel to the \( z \) direction. The propagation path of each photon is calculated from the scattering constant, assuming the absorption is negligible. An MC simulation of photon migration within irradiated tissues was built\(^1\)\(^8\) in order to calculate the full scattering profile at all possible exit angles. This simulation is based on the assumption that all photons reaching the tissue begin as ballistic photons. Given the current photon’s direction (\( \theta_{\text{old}} \)), and the probability of a photon to scatter \( \frac{1 - \exp(-\mu_s \text{dr})}{\text{C}^{13}} \), if the photon scattered, its new direction (\( \theta_{\text{new}} \)) was calculated using

\[
\theta_{\text{new}} = \theta_{\text{old}} + s \times \cos(g),
\]

where \( s \) is a random number from the group \( \{-1, 1\} \).

This process is repeated until the photon exits the tissue, and then its location is saved.

Several repetitions of the simulation for each tissue diameter were held to determine the variations in transmission due to the change in the reduced scattering coefficient \( \mu_0' = \mu_s(1-g) \). The values of the reduced scattering coefficient (\( \mu'_s \)) were in the range of the human skin values of 2 to 26 cm\(^{-1}\).\(^2\)\(^1\)

2.2 3-D Model of Cylindrical Tissue

A 3-D model of a cylinder of tissue was created with the same method and properties as described in the 2-D model (Fig. 2). Ten repetitions of the 3-D MC simulation\(^1\)\(^3\) were built in order to calculate the full scattering profile at all possible exit angles. Simulations for each tissue diameter (4, 5, and 6 mm) were held to determine the variations in transmission due to the change in reduced scattering coefficient. The illumination was set to be a square of \( 1.5 \times 1.5 \text{ mm} \), and the transmitted photons were collected from a 1-mm-thick slice (gray area in Fig. 2) along the cylinder surface.

\[\text{Fig. 3 The full scattering profile in 2-D simulation for various tissue diameters: (a) } D_t = 5 \text{ mm}; \text{ (b) } D_t = 10 \text{ mm}; \text{ and (c) } D_t = 15 \text{ mm. Reduced scattering coefficients (in cm}^{-1}\text{) are noted in the figure legend on the right.}\]

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3 Results

3.1 Isobaric Point for Different Tissue Diameters in 2-D

The 2-D simulation described in Sec. 2.1 was held for several tissue diameters. The mean fraction of photons that exited the tissue at several central angles ($\theta$ in Fig. 1) from 0 to 180 deg is presented in Figs. 3(a), 3(b), and 3(c) for tissue diameters of 5, 10, and 15 mm, respectively. The mean of 10 different simulations was calculated. The value of the isobaric point, as extracted from these graphs for 5, 10, and 15 mm, is 115, 135, and 165 deg, respectively. As expected, as tissue diameter increases, so does the value of the isobaric point. In order to facilitate the presentation and to compare it to known results from the literature,9,18 reflected photons are defined as those that exit the tissue near the entrance point at a central angle of $|\theta| > 135$ deg, and forward transmitted photons are defined as those that exit the tissue at a central angle of $|\theta| < 45$ deg. Figures 4(a), 4(b), and 4(c) present the average fraction of the reflected (squares) and forward transmitted photons (diamonds), respectively, versus scattering coefficients, for several tissue diameters.

3.2 Isobaric Point for Different Tissue Diameters in 3-D

The 3-D simulation described in Sec. 2.2 was held for several tissue diameters. The mean fraction of photons that exited the tissue at several central angles ($\theta$ in Fig. 2) from 0 to 180 deg is presented in Figs. 5(a), 5(b), and 5(c) for tissue diameters of 4, 5, and 6 mm, respectively. The value of the isobaric point, as extracted from these graphs for 4, 5, and 6 mm, is 155, 158, and 160 deg, respectively.

Since the photons are collected from a 1-mm-thick slice of tissue, there is no point at looking at the percent of forward transmitted photons since it is too low. Hence, a comparison of the reflection as defined in Sec. 3.1 is presented in Fig. 6. Note that although the change in tissue diameter hardly affects (<5%) the fixed angle reflection in Fig. 6, it significantly changes the maximal value in the full scattering profile (>30%) for the same scattering coefficient in Fig. 5.
In order to verify that the value of the isobaric point does not change via anisotropy factor ($g$), which is also a variable in photon migration, and changes between different tissue areas (for example, fate as opposed to blood), three different simulations were held. In these simulations for a 4-mm diameter of tissue, the reduced scattering coefficient was the same but $g$ changed, and the scattering coefficient was set to compensate for this change using the relation $\mu'_s = \mu_s(1 - g)$. As presented in Fig. 7, this change significantly influences the fixed angle reflection (up to 10%), but does not change the value of the isobaric point (data not shown).

Dependency of the isobaric point on the tissue diameter in both 2-D and 3-D simulations is presented in Fig. 8. As expected, the value of the isobaric point in the 3-D simulation is different from that obtained in 2-D due to the scattering in the additional dimension. According to these simulations, we have found a linear dependency (with a 98% fitting) to tissue diameter.

4 Discussion

In our current work, we have investigated the full scattering profile in different tissue diameters in 2-D, as well as 3-D simulations. From these profiles, we have extracted the value of the isobaric point to reduced scattering coefficient. Although the isobaric point value was different in the 2-D as well as 3-D simulations, a linear dependency was found to the tissue diameter.

When measuring light-tissue interactions, the first question that arises is whether to measure reflected or transmitted light. The common method for measuring blood saturation is from volume reflection since there is a very high amount of transmitted light from thick tissues. Furthermore, diffusion reflection methods which are used for the investigation of tissue’s optical parameters, are based on the assumption that the tissue is semi-infinite.

Very few methods test the transmitted light or ballistic photons because of low light intensity. By measuring the isobaric point, we solve this dilemma since this is the point that truly divides by half the total amount of exiting photons. For example, in our published results, the dependency between the reflectance and blood vessel diameter in a circular tissue illuminated at 850 nm was found to be opposite to that found by Jacques from a semi-infinite tissue illuminated at 585 nm since the tissue scattering is much lower at this wavelength.

In order to demonstrate the effect of lower scattering, the 2-D simulation where $D_t = 5$ mm (Fig. 9) was repeated for a wider range of reduced scattering coefficients (diamond, triangle, asterisk, cross, circle, x, and square in respect to 2, 6, 10, 14, 18, 22, and 26 cm$^{-1}$) in a wider range may completely alter the profile.
range of reduced scattering coefficients (diamond, triangle, asterisk, cross, circle, and square in respect to 2, 6, 10, 14, 18, 22, and 26 cm$^{-1}$). While in a lower reduced scattering coefficient (for example, 2 cm$^{-1}$ marked by diamonds), the light is mainly transmitted; in a higher reduced scattering coefficient (for example, 26 cm$^{-1}$ marked by squares), the light is mainly reflected. Furthermore, the isobaric point is only relevant in higher reduced scattering coefficients.

In conclusion, this work allows predicting the isobaric point, which is optimal for different types of NIR spectroscopy measurements for a given tissue with known optical properties. Furthermore, we have presented a method that is sensitive to the reduced scattering coefficient. Therefore, if experimental measurements will show a different full scattering profile, the reduced scattering coefficient. Therefore, if experimental measurements for a given tissue with known optical properties.

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References

1. A. N. Yaroslavsky et al., *Optics of Blood*, SPIE Press, Bellingham, Washington (2002).
2. M. Friebel et al., “Influence of oxygen saturation on the optical scattering properties of human red blood cells in the spectral range 250 to 2000 nm,” *J. Biomed. Opt.* 14(3), 034001 (2009).
3. M. Friebel et al., “Determination of optical properties of human blood in the spectral range 250 to 1100 nm using Monte Carlo simulations with hematocrit-dependent effective scattering phase functions,” *J. Biomed. Opt.* 11(3), 034021 (2006).
4. A. Roggan et al., “Optical properties of culcatin human blood in the wavelength range 400–2500 nm,” *J. Biomed. Opt.* 4(1), 36–46 (1999).
5. G. Zaccanti, S. Del Bianco, and F. Martelli, “Measurements of optical properties of high-density media,” *Appl. Opt.* 42(19), 4023–4030 (2003).
6. S. L. Jacques and B. W. Pogue, “Tutorial on diffuse light transport,” *J. Biomed. Opt.* 13(4), 041302 (2008).
7. R. Ankri et al., “In-vivo tumor detection using diffusion reflection measurements of targeted gold nanorods—a quantitative study,” *J. Biophotonics* 5(3), 263–273 (2012).
8. R. Vered, S. Havlin, and H. Taitelbaum, “Optical detection of hidden tumors,” *Proc. SPIE* 2389, 851–858 (1995).
9. S. L. Jacques, “Optical assessment of cutaneous blood volume depends on the vessel size distribution: a computer simulation study,” *J. Biophotonics* 3(1–2), 75–81 (2010).
10. L. Wang and S. L. Jacques, “Hybrid model of Monte Carlo simulation and diffusion theory for light reflectance by turbid media,” *J. Opt. Soc. Am. A* 10(8), 1746–1752 (1993).
11. L. Wang, S. L. Jacques, and L. Zheng, “MCML—Monte Carlo modeling of light transport in multi-layered tissues,” *Comput. Methods Programs Biomed.* 47(2), 131–146 (1995).
12. R. Ankri, H. Taitelbaum, and D. Fixler, “Reflected light intensity profile of two-layer tissues: phantom experiments,” *J. Biomed. Opt.* 16(8), 085001 (2011).
13. D. Fixler and R. Ankri, “Subcutaneous gold nanorod detection with diffusion reflection measurement,” *J. Biomed. Opt.* 18(6), 061226 (2013).
14. R. A. J. Groenhuis, A. H. Ferweda, and J. J. Ten Bosch, “Scattering and absorption of turbid materials determined from reflection measurements. 1: Theory,” *Appl. Opt.* 22(16), 2456–2462 (1983).
15. D. Jakubowski et al., “Quantitative absorption and scattering spectra in thick tissues using broadband diffuse optical spectroscopy,” Chapter 12 in *Biomedical Optical Imaging*, J. G. Fujimoto and D. L. Farkas, Eds., pp. 330–355, Oxford University Press, New York (2009).
16. T. H. Pham et al., “Broad bandwidth frequency domain instrument for quantitative tissue optical spectroscopy,” *Rev. Sci. Instrum.* 71(6), 2500–2513 (2000).
17. L. Zhang, A. Shi, and H. Lu, “Determination of optical coefficients of biological tissue from a single integrating-sphere,” *J. Mod. Opt.* 59(2), 121–125 (2012).
18. H. Duadi, D. Fixlerand, and R. Popovtsjer, “Dependence of light scattering profile in tissue on blood vessel diameter and distribution: a computer simulation study,” *J. Biomed. Opt.* 18(11), 111408 (2013).
19. M. Nitzan and S. Engelberg, “Three-wavelength technique for the measurement of oxygen saturation in arterial blood and in venous blood,” *J. Biomed. Opt.* 14(2), 024046 (2009).
20. S. Del Bianco, F. Martelli, and G. Zaccanti, “Penetration depth of light re-emitted by a diffusive medium: theoretical and experimental investigation,” *Phys. Med. Biol.* 47(23), 4131–4144 (2002).
21. T. Lister, P. A. Wright, and P. H. Chappell, “Optical properties of human skin,” *J. Biomed. Opt.* 17(9), 090901 (2012).
22. A. N. Bashkatov et al., “Optical properties of human skin, subcutaneous and mucous tissues in the wavelength range from 400 to 2000 nm,” *J. Phys. D: Appl. Phys.* 38(15), 2543–2555 (2005).

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