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Hamootal Duadi
Idit Feder
Dror Fixler

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Hamootal Duadi, Idit Feder, and Dror Fixler*

Bar Ilan University, Faculty of Engineering and the Institute of Nanotechnology and Advanced Materials, Ramat Gan 5290002, Israel

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 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](https://doi.org/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 μ_a , scattering coefficient μ_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 solutions¹ or integrating spheres for the measurement of total transmitted light through a blood sample and total reflected light from it.²⁻⁴ Indirect methods are based on a mathematical simulation model of photon migration. Photon migration simulations are commonly described by the diffusion theory.⁵ However, in the close region near the light source, diffusion theory does not accurately describe the light distribution.⁶ 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.⁷⁻¹¹

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).^{7,12-16} Very few methods test the transmitted light^{2-4,17} or ballistic photons.⁸ It has been previously

suggested¹⁸ 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.¹⁷ 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^{13,18} 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.^{19,20} 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.

*Address all correspondence to: Dror Fixler, E-mail: Dror.Fixler@biu.ac.il

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 $w_0 = 1.5$ 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 built¹⁸ 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 (θ_{old}), and the probability of a photon to scatter [$1 - \exp(-\mu_s dr)$], if the photon scattered, its new direction (θ_{new}) was calculated using

$$\theta_{new} = \theta_{old} + s \times \cos(g),$$

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 $\mu'_s = \mu_s(1 - g)$. The values of the reduced scattering coefficient (μ'_s) were in the range of the human skin values of 2 to 26 cm^{-1} .²¹

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 simulation¹³ 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 1.5×1.5 mm, and the transmitted photons were collected from a 1-mm-thick slice (gray area in Fig. 2) along the cylinder surface.

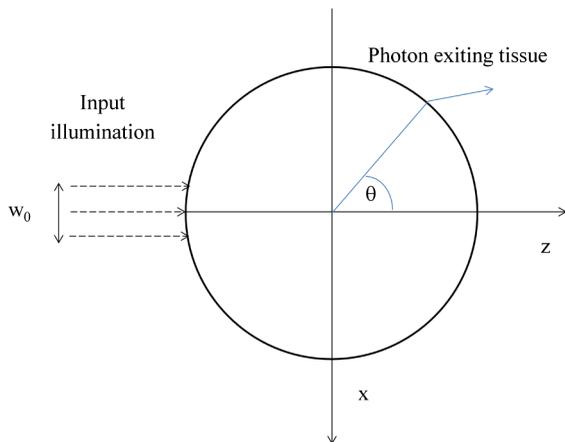


Fig. 1 Scheme of simulated two-dimensional (2-D) model. A beam of photons (dashed arrows) with a waist of w_0 was introduced at one end and collected when exiting the tissue at different central angles (θ).

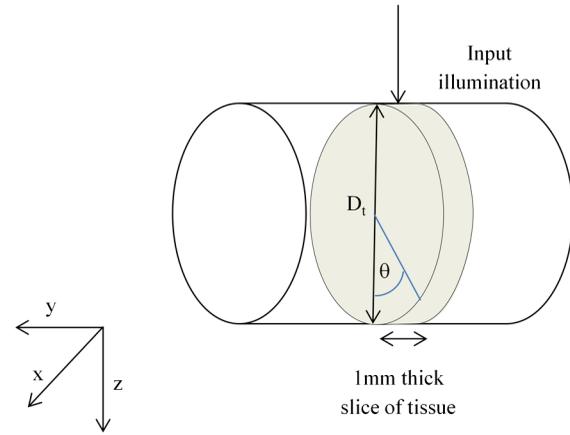


Fig. 2 The cross-section of simulated three-dimensional (3-D) tissue model. Photons are collected from the surface of a 1-mm-thick area of interest.

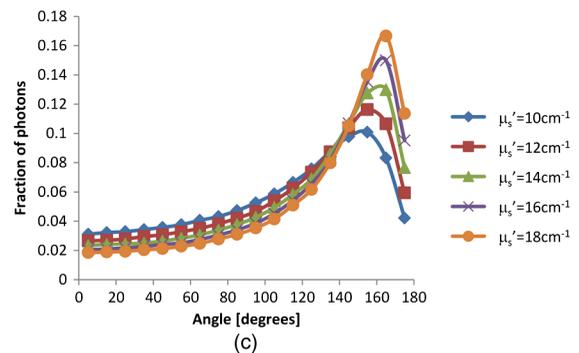
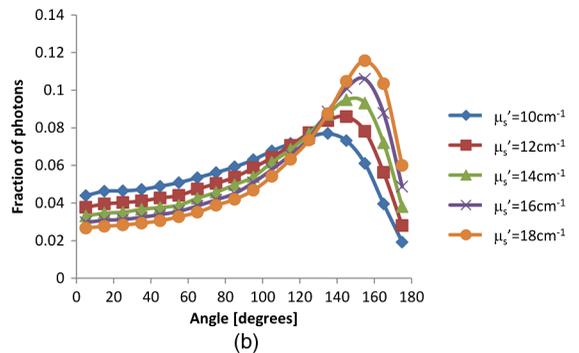
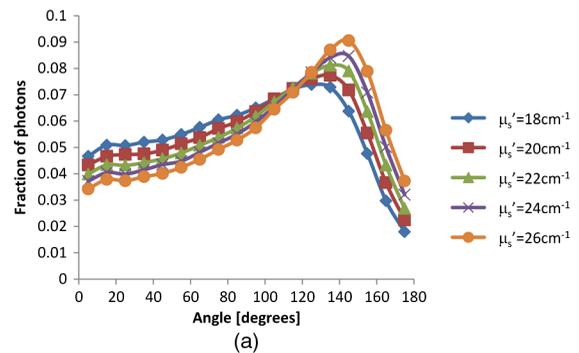


Fig. 3 The full scattering profile in 2-D simulation for various tissue diameters: (a) $D_t = 5$ mm; (b) $D_t = 10$ mm; and (c) $D_t = 15$ mm. Reduced scattering coefficients (in cm^{-1}) are noted in the figure legend on the right.

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 (θ 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

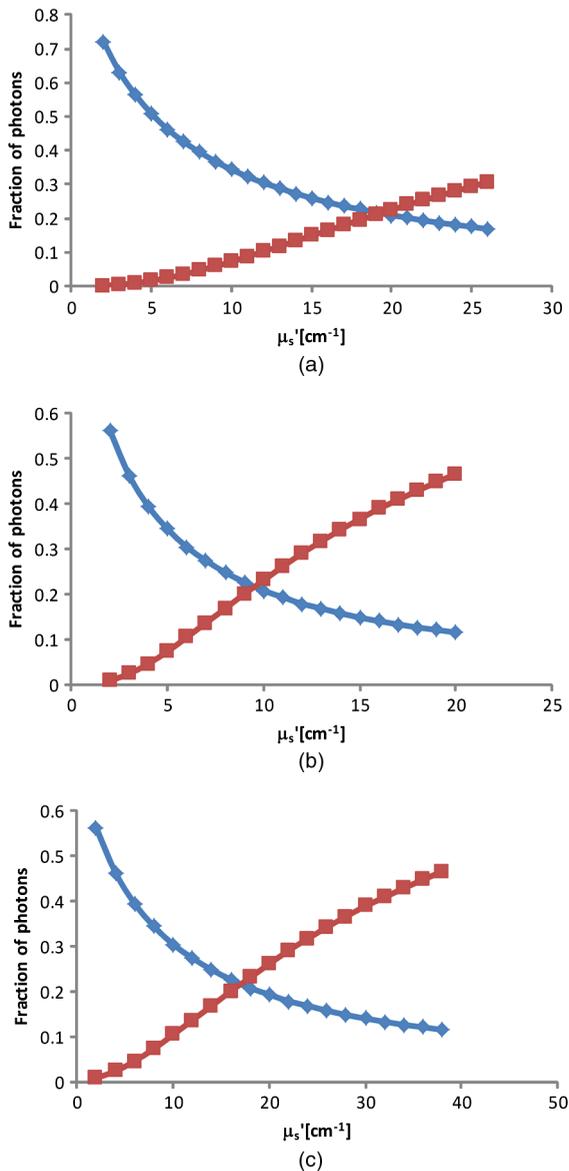


Fig. 4 The fixed angle transmission (diamonds) and reflection (squares) in 2-D simulation for different tissue diameters: (a) Dt = 5 mm; (b) Dt = 10 mm; and (c) Dt = 15 mm.

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 (θ 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.

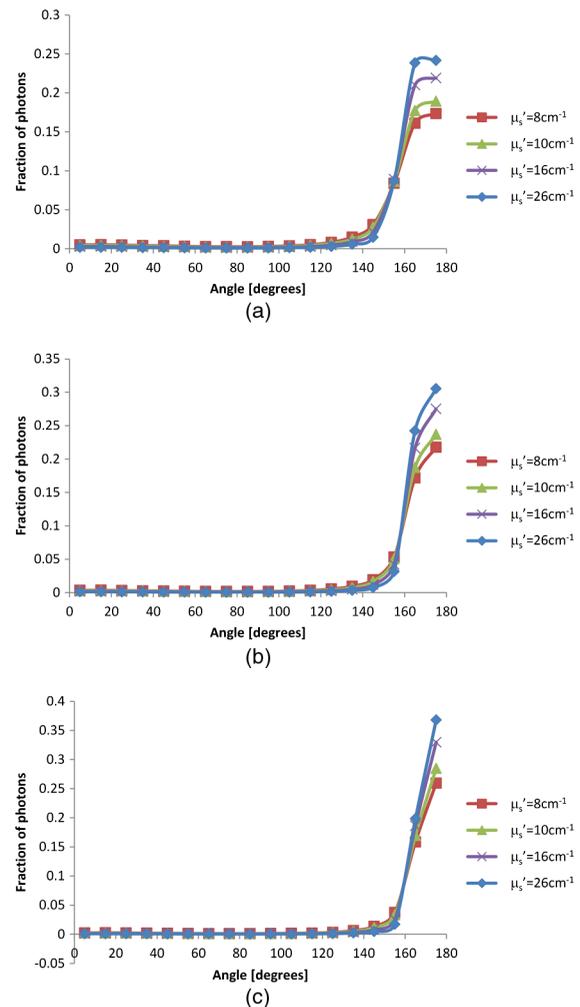


Fig. 5 The full scattering profile for various reduced scattering coefficients (square, triangle, x, and diamond in respect to 8, 10, 16, and 26 cm^{-1}). Tissue diameter in 3-D simulation is (a) Dt = 4 mm; (b) Dt = 5 mm; and (c) Dt = 6 mm.

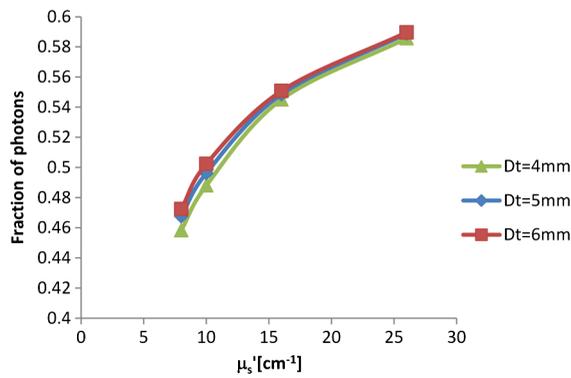


Fig. 6 Reflection versus reduced scattering coefficient for different tissue diameters (triangle, diamond, and square in respect to 4, 5, and 6 mm).

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, fat 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.

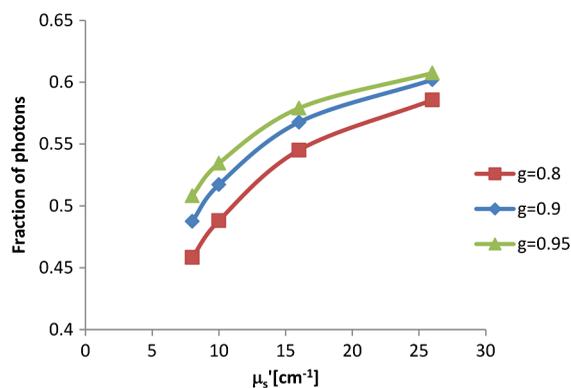


Fig. 7 Reflection versus reduced scattering coefficient for different anisotropy factors (square, diamond, and triangle in respect to 0.8, 0.9, and 0.95) for a 3-D simulation with a 4-mm tissue diameter.

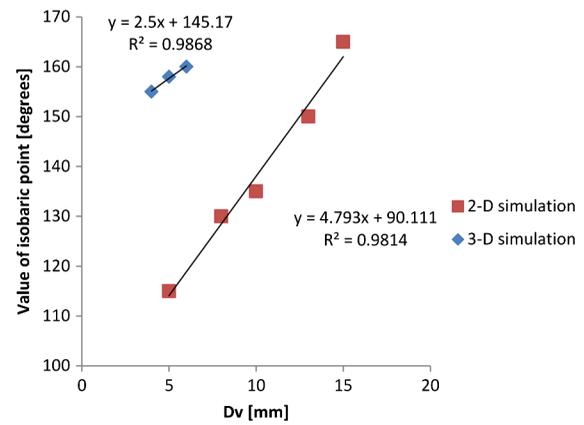


Fig. 8 Value of the isobaric point versus tissue diameter in 2-D (square) and 3-D (diamond) simulation.

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^{7,12-16} since there is a very high amount of transmitted light from thick tissues. Furthermore, diffusion reflection methods,⁶⁻¹⁴ 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^{2-4,17} or ballistic photons⁸ 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.

In our current work, we have chosen an NIR wavelength of 850 nm since it is commonly used in PPG measurements^{19,22} and because the range of scattering coefficients was according to known ranges in this wavelength.^{4,21,22} However, if one wants to investigate the full scattering coefficient in a lower reduced scattering coefficient, one may generate a completely opposite profile. For example, in our published results,¹⁸ 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 Jacques⁹ 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 $Dt = 5$ mm (Fig. 9) was repeated for a wider

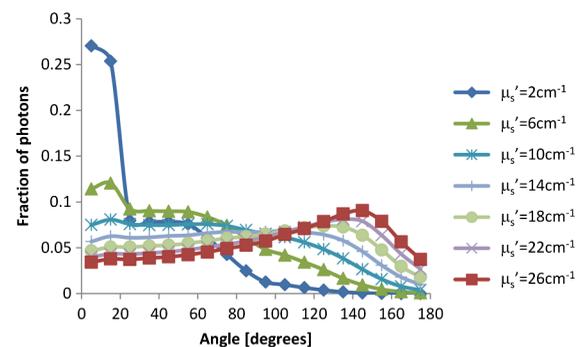


Fig. 9 The full scattering profile in 2-D simulation for a wide range of reduced scattering coefficients. For a tissue diameter of $Dt = 5$ mm reduced scattering coefficients (diamond, triangle, asterisk, cross, circle, x, and square in respect to 2, 6, 10, 14, 18, 22, and 26 cm^{-1}) in a wider range may completely alter the profile.

range of reduced scattering coefficients (diamond, triangle, asterisk, cross, circle, x , 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, which does not cross the isobaric point, this indicates an abnormal change in the optical tissue properties.

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Hamootal Duadi received her PhD degree in 2011 from the Engineering Faculty, Bar-Ilan University, Israel. She specialized in optical information processing, super resolution, diffractive optical elements and beam shaping, 3D and range estimation. Her current research fields are pulse oximetry, tissue light interaction, and live cell imaging.

Idit Feder received her BSc degree in 2013 from the Faculty of Engineering of Bar-Ilan University, and currently she is working toward the Msc degree and researching the dependency of the scattering profile of tissues.

Dror Fixler received his PhD degree in 2003 from the Department of Physics, Bar-Ilan University, Israel. He is a member of the Faculty of Engineering and the Nano Center of Bar-Ilan University. He has published over 50 original research papers and holds over 9 issued patents. His research interests include fluorescence measurements (FLIM and anisotropy decay), optical super resolution, high-end electro-optical system engineering, and light-tissue interaction. He received several international awards and organized and presented at over 20 international conferences.