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Mathematical method for diffuse optical tomography imaging: A Research Study

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Abstract: In this work, depth normalization algorithm (DNA), which normalizes elements of weight matrix for the reflectance diffuse optical tomography system has been introduced. Construction of weight matrix plays important role for diffuse optical tomography systems. Voxel elements of tissue are key factors for image reconstruction process. Value of the voxel elements are mapped to color codes. As well as try to solve inverse problem of photon measurements related to voxels for diffuse optical tomography systems weight matrixes are very important parts of image reconstruction. Weight matrix could be constructed according to theoretical physics diffuse equation of photon transport model for homogenous and heterogeneous medium or Monte Carlo (MC) simulation method. In this work, while the weight matrix has been constructed, imaging space has been divided into each equal sublayers. For each layer weight matrix elements have been normalized separately. Thus, importance matrix has been created for the most popular photon trajectories.

Keywords: Depth normalization algorithm (DNA), reflectance diffuse optical tomography (DOT), monte carlo (MC) simulation, inverse problem, weight matrix.

1. Introduction

Interaction between light and the biological system consists of the absorption of light by a molecular species and the redirection of incidental light over a range of angles due to refractive index mismatch in tissue microstructure. Measurement of tissue optical properties, like absorption, scattering, and reduced scattering coefficient, has gained importance for therapeutic and diagnostic applications. The spectral information of tissue images depend both on tissue absorption and scattering. This requires developing analysis software that accounts for these events. In this work the algorithms which get benefit from Monte Carlo (MC) simulations are developed for both scattering and absorption. Construction and normalization of weight matrix has been represented. Elements of weight matrix characterize photon energy absorption and scattering properties of tissue. Photon absorption or energy transfer occurs when photon frequency must match with the frequency associated with molecular

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energy transition. In general absorption of light in tissue plays a dynamic role in imaging and spectroscopic applications [1]. According to absorption and scattering parameters of tissue, photon trajectories are shaped. Photon paths from the view of detector point to source inside the tissue is most likely banana shape. Depth normalization algorithm (DNA) uses different depth layer normalization for weight matrix, which amplifies importance of banana path. For reflectance diffuse optical tomography system weight matrix elements for each source and detector couplings for separate depth layers has been normalized to their maximum value. Banana shape functions have been used for different source detector pairs which have been created by multiplying individual fluence rates of each source and detectors. The weight matrix has been used in inverse problem solution to get absorption and scattering coefficients for each voxel of tissue phantoms. Tikhonov regularization method has been applied as an inverse problem solution to reconstruct the tissue phantom image. Imaging with the non-invasive, non-ionised low power laser light is most competitive and most incapable radiology modality for medical imaging field. Back reflected diffuse optical tomography (DOT) systems run with the laser light which uses mostly hemoglobin and deoxy-hemoglobin wavelength band. With low power laser light, imaging is hard to succeed because light penetrated and absorbed in tissue. Depending on the tissue properties incoming light scattered and absorbed. The photon paths are determined by absorption and scattering characteristics of tissue. Photon trajectories are very important for DOT systems. When the light has penetrated inside the tissue, nothing is known about where the photons tend to go. This is all about the stored photon energy flux inside the tissue. Tissue properties define the correct way for the photon path. Different approaches had been used to create weight matrix. DOT systems have been started to be used especially for breast imaging as an auxiliary tool. The reason why it is being selected as a new method is to investigate non-invasive, non-ionized, low cost, easy transport, low energy consumption. The image is always blurry and the spatial resolution is not good. Breast tissue mostly consists of fat and tissue properties are like homogenous structure. These are the reasons for selecting as an imaging tissue. On the other hand, there is also functional brain imaging works have been done [2-5]. Multiple detectors, multiple sources systems have been applied in several different geometries to image brain activity and spatial resolution. Different methods are used to reconstruct the perturbation in the absorption distribution within a region of interest (ROI). One of the most common method is the linear Rytov approximation to the photon diffusion equation [6, 7]. The intensity of the detected light decreases rapidly by increasing depth for large source-detector (s-d) separations, and therefore, deep tissue localization is often not accurate as it shows a shift toward the tissue surface. A depth compensation algorithm (DCA) was developed to counterbalance this negative effect [8] and has been used in simulation studies and in vivo experiments to accurately localize an absorber within tissue [9, 10]. Back reflected DOT systems have also been used to quantify breast tumors [11, 12].

2. Method

For breast imaging, ratio of consecutive phantom experiments have been used. Particle data has been divided over non-particle data. This is known as perturbation data and is being used for inverse problem solution. Perturbation data is representing measured data ratio. While photons are absorbed and scattered they draw trajectories. Absorbed and scattered photons are going through their pathways. These trajectories have been defined for special equation system. For breast tissue weight matrix has been constructed to image the tissue. Inverse problem of linearized equation system has been solved and linked to the concentration of tissue voxel coefficients and source detector coupling photon measurement data for imaging the tissue. This is forward model of photon penetration method. The matrix representation of perturbation data is shown in Equation 1, the weight matrix and chromophore concentrations are given symbolically.

$$Perturbation = TT * \Delta\mu_a * Volume$$

$$\begin{bmatrix} pert_s_1d_1 \\ . \\ . \\ . \\ pert_s_{49}d_{49} \end{bmatrix} = \begin{bmatrix} TT_{1,1} & ... & TT_{1,10186} \\ . & . & . \\ . & . & . \\ . & . & . \\ TT_{2401,1} & & TT_{2401,10186} \end{bmatrix} * \begin{bmatrix} \Delta\mu a_1 \\ . \\ . \\ . \\ \Delta\mu a_{10186} \end{bmatrix} \quad (1)$$

In this equation system, the first column is the perturbation vector which is the of particle data over non-particle data for homogeneous in-vivo phantom experiments. In the right hand side of the equation system, first matrix is the weight matrix. And the final column is the absorption chromophore concentrations. The goal is to find the unknown $\Delta\mu_a$ absorption in the inverse problem solution method. The most common method to construct the weight matrix is the linear Rytov approximation theoretical physical method to the photon diffusion equation for the specific tissue [9].

3. Inverse Problem Solution

For DOT systems, iterative, subspace and regularization mathematical algorithmic methods are being used. For iterative methods, the most famous one is the Algebraic Reconstruction Technique (ART). In ART technique, at the beginning the pre-assumption of unknown space is selected and then by using iteration techniques the unknown spaces are calculated. The sub-space methods are also used. Conjugated Gradient (CG), Truncated Singular Value Decomposition (TSVD) methods are two of the subspace techniques. In addition, most famous regularization method is Tikhonov regularization method. In this work, diffuse optical images are reconstructed by using Tikhonov regularization inverse problem solution method. For better image resolution different source and detector geometries had been studied. Because of the nature of penetrated light into the diffusing tissue medium, light is absorbed and the photon fluence rate exponentially decreases when the light goes through deeper position of the tissue. This means that superficial voxels have bigger weight coefficients than deeper voxels. The inverse problem solution uses different weight matrix coefficients and this makes superficial voxels more important than the deeper voxels By this way image reconstruction of inverse problem solution approaches to superficial. To overcome this negative effect, DCA method [10] has been used in simulation studies and in vivo experiments to accurately localize an absorber within tissue [11-13]. In the algorithm, the depth compensation parameter requires to estimate actual depth of an inclusion. Therefore, an adjuvant system like Magnetic Resonance Imaging (MRI) is required to estimate depth compensation parameter. In this study, Monte Carlo (MC) simulation has been run to define fluence rates for each voxel within a volume of 3.9x3.9x3.9 cm^3 which is defined by μ_a=0.04 cm^{-1}, μ_s'=10 cm^{-1} for tissue background according to literature in [14]. In addition, the absorption probabilities and the Henyey-Greenstein scattering phase function has been used in simulations. The MC simulation technique can be used to model a specific problem, generate random numbers, probability distribution functions and perform output analysis. The radiation transport problem has been applied into the MC simulation technique. In this technique, photons are sent to tissue, and its step distance are computed. The back reflection coefficients, photon absorption and generation parameters are calculated. By using the photon scattering phase function the photon fluence rates are computed and used in these simulations. For each source and detector positions photon energy flux rates have been computed from MC simulation codes. By this way photon energy flux rates inside the tissue for radial distance from one source position and through the different depths have been obtained. Then one source position photon energy flux rates have been translated through each source and the detector positions of our specific probe geometry of the back reflected DOT system to acquire weight matrix coefficients. Photon flux rates for different source and detector positions have been multiplied for each source and detector coupling. Every product of multiplication is the weight coefficient for each voxel position. Then novel method has been used which normalizes weight matrix

coefficients in the same depth layer for specific source and detector coupling by dividing the all voxel values of the same layer to their maximum value.

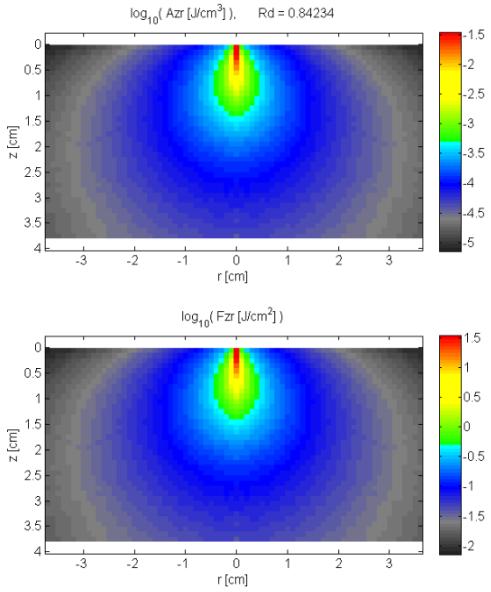


Fig. 1. Fluence rates for 3.9 cm radial (r) and depth (z) distance.

In Figure 1. the fluence rates are shown on logarithmic scales. Fluence rates have been computed for specific radial r distance and for specific depth z distance. The 3.9 cm r radial distance and 3.9 cm z depth distance have been used. Fluence rates are transferred into real problem. In Figure 2 first real banana shape has been shown. This is the real banana shape of photon trajectories from one source position to other detector position. In this figure banana shape has been created for 30th source and 26th detector and one of the x slice has been cut out to show the banana shape of source-detector coupling. For imaging medium 26*26*16 voxels have been used. After getting banana shape, novel method has been applied. This is normalization method for each one of the depth layer. For specific source and detector coupling each depth layer has been normalized to their maximum weight value. By this method the Figure 3. has been created.

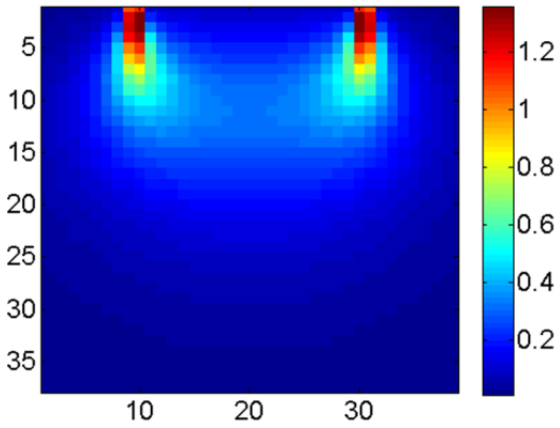


Fig. 2. Banana shape

As it could be seen this picture is different than Figure 2. All color codes are same inside the banana shape. In the literature most effective method that has been accomplished is DCA (Depth

Compensation Algorithm). But in this method extra adjuvant imaging modality is needed according to literature. This method has not been used here. Each voxel weights have been normalized for each one of the source-detector coupling. DNA has been used. In Figure 3, after DNA algorithm has been applied and weight function has been represented for one of the specific source-detector coupling. This novel method mostly consists of normalization and after normalization processes. In Figure 4, real geometric problem has been demonstrated. Sources and detectors have been placed as a chess board. This geometric approach has been thought to acquire maximum intersections for source and detector couplings. After normalization hot spot areas which are close to center of banana shape could be seen. But also the other colors could be seen. After DNA weight function has been modified. Weight function has been put into cage. After DNA process for one of the source-detector coupling 16th neighborhood has been shown for demonstration purpose in Figure 5. This is one of the banana shape after all of the normalization processes.

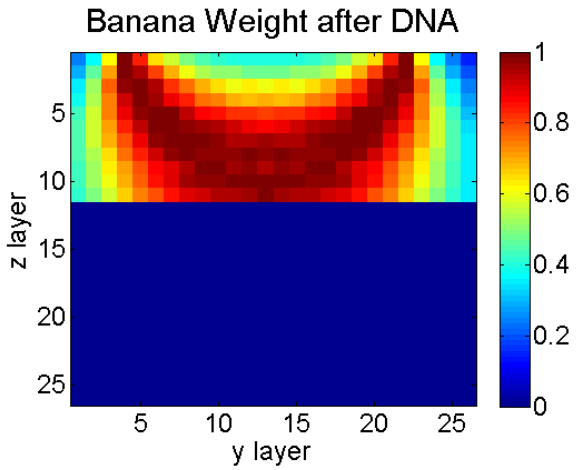


Fig. 3. Depth Normalization Algorithm (DNA) weight function.

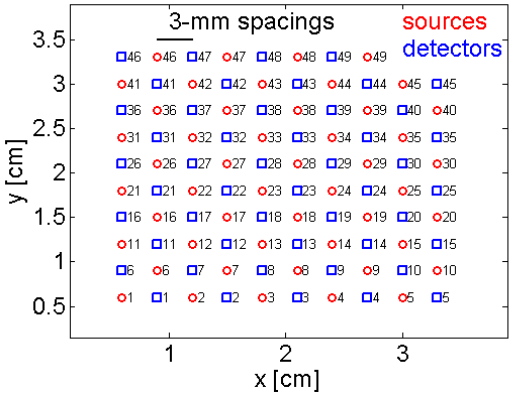


Fig. 4. Source and detector locations on the probe surface.

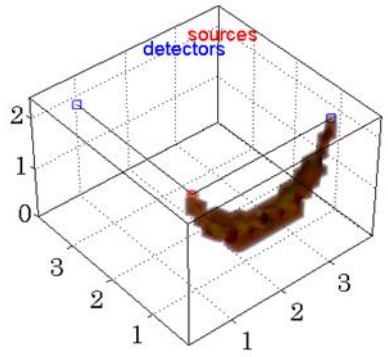


Fig. 5. 3D Depth Normalization Algorithm (DNA) weight function.

The weight matrix is very important for inverse problem solution system and constructed by using final weight coefficients. Tikhonov regularization method has been used for inverse problem solution. The traditional methods such as iterative Algebraic Reconstruction Technique (ART) or any other iterative, subspace or regularization methods have also been used. Iterative methods are mostly used by Computerized Tomography (CT), X-Ray digital tomography, Magnetic Resonance Imaging, Ultrasound imaging, Positron Emission Tomography (PET) and other high energy imaging modalities. In X-ray imaging modality, photons have high energy and they are always going directly to detector position. Weights have been used in forward diffusion model equation. Tikhonov regularization inverse problem solution is given in Equation 2:

$$\Delta\mu a = (TT^{T}TT + \lambda I)^{-1}TT^{T} \Delta P \qquad (2)$$

In equation 2, $\Delta\mu_a$ is an absorption coefficient of the related voxels and ΔP is perturbation vector. By using Continuous Wave Reflectance DOT system with the same fiber probe geometry which is designed as a chess board except one source and one detector fiber position is missing. For the 49 sources and 49 detectors, data have been acquired from a tissue phantom with the similar optical properties of the MC simulation. The tissue phantom Continuous Wave Reflectance DOT system measurements have been done with and without inclusions inside the intralipid tank. Perturbation vector has been calculated by division over the consecutive measurement data which is with particle and without particle data [15]. This is the first type of perturbation vector which has been used for inverse problem solution. In this work, back reflection DOT system has been used which has been originally designed to image breast tissue phantom. Generally the images are reconstructed using two different ways: The first and gold standard one is division of particle data over non particle data. The second one is subtraction between average and original data.

4. Experimental Materials

Intralipid tank has been used as an in vitro medium. This device is designed and implemented to use for breast imaging. Absorption and scattering properties of background medium is close to real breast tissue. 49 sources and 49 detectors back reflection geometry cylindrically fiber probe has been used as an imaging probe. Photons which have been sent to tissue have also been collected from same reflection geometry. The probe structure has been shown in Figure 4. Totally there are 49*49=2401 continuous wave measurement data. Perturbation data have been used in forward model equation system. Then Tikhonov regularization inverse problem solution method has been used. Finally unknown absorption and scattering chromophore properties for specifically designed medium have

been solved. For weight matrix DNA method has been used which has been mentioned earlier before this section. 1 cm diameter inclusion has been put inside the intralipid tank and two different depths have been used. The unknown parameters are the real absorption and scattering parameters for specifically used laser. 798 nm wavelength laser diode has been used as a light source. Photodetectors are OSI Optoelectronics OSD35-LRD semiconductor photodiodes. All reflected photons are collected by the photodetectors and 6 integration times have been used for better resolution. Current input analog to digital converter has been used to convert the analog voltage into digital voltage which is generated by integrators inside the chip. For this purpose DDC232 Burr-Brown and Texas Instruments 2x32 current input analog to digital converters have been used. All the synchronization has been controlled by PIC18f2550 microcontroller. The microcontroller has also controlled DICON VX500 optical switch which is sending laser light inside to the tissue. PIC microcontroller has handshaked with PC and data have been acquired to the continuous wave reflectance DOT system. This is all about the data acquisition process. After acquiring the data inside to the computer Tikhonov regularization inverse problem solution method has been used and unknowns have been reconstructed for tissue.

5. Results

For intralipid tank $\mu_s'=10cm^{-1}$ and $\mu_a=0.04\ cm^{-1}$ has been used as in-vitro tissue-like intralipid medium. Indocyanine Green (ICG) spherical inclusion material $\mu_a=0.16\ cm^{-1}$ which has 1 cm diameter has been put inside the intralipid tank in two different depths. In experiment, inclusions are placed into the tissue of 1 cm and 1.5 cm depths then data are acquired. After both data acquisition processes, Tikhonov regularization inverse problem solution method has been used. This mathematical method is used for diffuse optical tomography imaging modality. In reconstruction process, images are created and simulated in xyz, xy and xz intersections. In Figure 6 and 7, inclusion images have been shown for 1 cm and 1.5 cm depth, respectively. As shown in figures, the images at different depth lengths are realized. The images are reconstructed also without using DNA method. In Figure 8, reconstructed image for 1.5 cm depth could be seen with basic Tikhonov method without using DNA method. Reconstructed images are moving through the superficial surface.

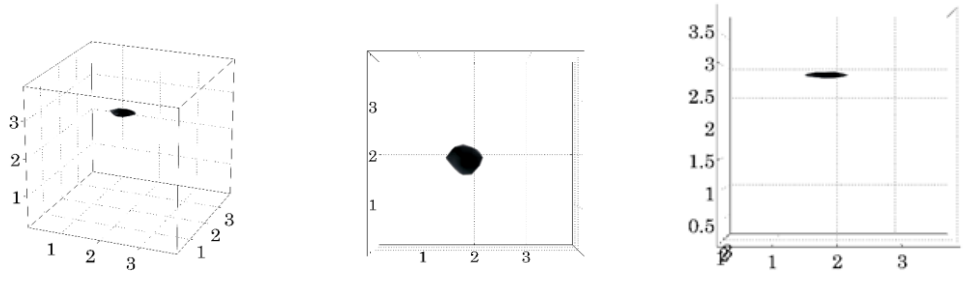


Fig. 6. Reconstructed inclusion for xyz, xy and xz intersections at 1cm depth with DNA.

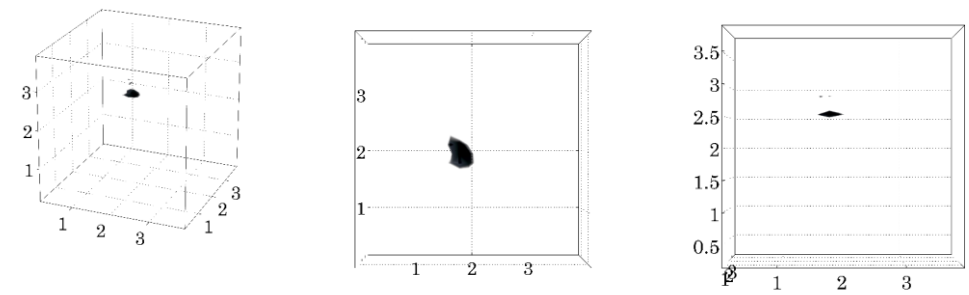


Fig. 7. Reconstructed inclusion for xyz, xy and xz intersections at 1.5 cm depth with DNA.

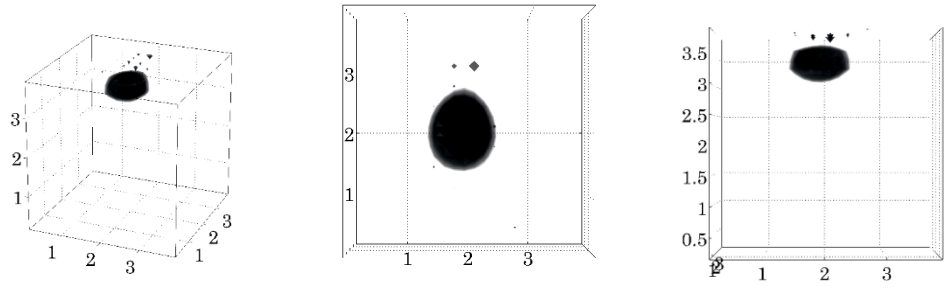


Fig. 8. Reconstructed inclusion for xyz, xy and xz intersections at 1.5 cm depth without DNA.

6. Conclusion

MC simulations have been run and weight function has been constructed. The MC simulation and mathematical inverse problem solution algorithms have been integrated together. The constructed weight matrix coefficients are important for inverse problem solution algorithms. The simulation outputs are more convenient and close to real photon distribution in real tissues in inverse problem solution space. Continuous wave reflectance DOT device and reconstruction algorithm is ready to be used for real breast tissue. In this work, next step is to acquire real data from real tissue and to get an image by reconstruction method which we have designed, and implemented. This method is novel and it would be used in new research purposes. This research work has given new ideas that could be used in future.

References

[1] S. L. Jacques, "Spectral Imaging and Analysis to yield tissue Optical properties," *Journal of Innovation Optical Health Sciences*, 123-129, (2009).

[2] D. A. Boas, T. Gaudette, and S. R. Arridge, "Simultaneous imaging and optode calibration with diffuse optical tomography," *Optical Express* 8, 263-270 (2001).

[3] T. Yamamoto, A. Maki, T. Kadoya, Y. Tanikawa, Y. Yamad, E. Okada, and H. Koizumi, "Arranging optical fibers for the spatial resolution improvement of topographical images," *Phys. Med. Biol.* 47, 3429-3440 (2002).

[4] X. Song, B. W. Pogue, S. Jiang, M. M. Doyley, H. Dehgani, T. D. Tosteson, and K. D. Paulsen, "Automated region detection based on the contrast to noise ratio in near infrared tomography," *App. Opt.* 43, 1053-1062 (2004).

[5] W. Cui, C. Kumar, B. Chance, "Experimental study of migration depth for the photons measured at sample surface," *Proc. SPIE* 1431, 180-191 (1991).

[6] S. Feng, F. Zeng, B. Chance, "Monte Carlo simulations of photon migration path distributions in multiple scattering media," *Proc. SPIE* 1888, 78-89 (1993).

[7] S. Feng, F. Zeng, and B. Chance, "Photon migration in the presence of a single defect a perturbation analysis," *Appl. Opt.* 34, 3826-3837 (1995).

[8] A.Yodh and B. Chance, "Spectroscopy and imaging with diffusion light", *Phys. Today*, 48, 34-45 (1995).

[9] X. D. Zhu, S. Wei, S. C. Feng, B. Chance, "Analysis of a diffuse photon density wave in multiple scattering media in the presence of a small spherical object", *J. Opt. Soc. Am. A*, 13, 494-499 (1996).

[10] H. J. Niu, F. Tian, Z. J. Lin, and H. Liu, "Development of a compensation algorithm for accurate depth localization in diffuse optical tomography," *Opt. Lett*. 35, 429-431 (2010).

[11] F. H. Tian, H. Niu, S. Khadka, Z. J. Lin, and H. Liu, "Algorithmic depth compensation improves quantification and noise suppresion in functional diffuse optical tomography," Biomed. *Opt. Express* 1, 441-452 (2010).

[12] H. J. Niu, Z. J. Lin, F. Tian, S. Dhamne, and H. Liu, "Comprehensive investigation of three-dimensional diffuse optical tomography with depth compensation algorithm," *Biomed. Opt.* 15, 046005 (2010).

[13] F. Liu, M. Li, B. Zhang, J. Luo, J. Bai, "Weighted depth compensation algorithm for Fluorescence molecular tomography reconstruction," *Appl. Opt.* 36, 518883 (2012).

[14] H. Wang, S. L. Jacquess, L. Q. Zheng, "MCML – Monte Carlo modeling of photon transport in multi-layered tissues," *Comput. Methods Program. Biomed.* 47, 131-146 (1995).

[15] H.Ö. Kazancı, T. Mercan, M. Canpolat, "Design and evaluation of a reflectance diffuse optical tomography system," *Opt. Quant. Electron*. 46/4, (2014).