Features of diffuse photon migration in soft biological tissue

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Abstract. Specific features of photon density normalized maximum movement in soft biological tissue under the influence of deforming force are described. Soft biological tissue is simulated as a turbid and linear isotropic pseudo-incompressible medium. It has been shown that the deforming effects do not influence the direction of photon density normalized maximum (PDNM) movement, but affect the speed of its movement. The decay of diffuse transmittance intensity is 8 to 10\% faster in undeformed state than in those with deformations. The established features can be useful for the generation of an initial approximation to the spatial distributions of optical parameters during solution of the diffuse optical tomography (DOT) inverse problem.

1. Introduction

Optical radiation scattering through biological object gives useful information about tissue structure. However, due to strong scattering and significant nonlinearity of photon trajectories there are difficulties in interpretation of experimental results (acquired TPSF – time point spread function) and challenges in image reconstruction in diffuse optical tomography (DOT). Overcoming the difficulties is possible by using numerical simulation of radiation passing through human-tissue-like phantom which migration is of extreme importance [1-3]. Among features the solution of the inverse problem of DOT (image reconstruction) is most often reduced to a multiple iterative solution of the forward problem (numerical simulation of photon migration) with the calculation and minimization of the discrepancy between the TPSFs obtained as a result of numerical simulation and the TPSFs from detectors of DOT system [4, 5].

It is known, that tissue external shape can significantly affect the quality of DOT image reconstruction [4, 6]. Therefore, it is necessary to provide good tight contact of optical fibers to inject and to detect diffusely transmitted photons in the investigated area. It can significantly reduce the loss of useful signal, and thus to improve the accuracy of absorption and scattering properties mapping [5, 7]. However, in some cases, this also leads to the deformation of tissue shape. In this regard, compression plates are usually appears as the cause of the deformation [5, 6]. In authors research is used an elastic bracelet to fixture the irradiation source and detector fibers, which allows achieving good contact of optical ports with the skin at acceptable deformations of individual parts of the investigated object. Application leads to the image artifacts caused by the discrepancy between actual boundaries of investigated area and its mathematical model representation [6-8]. Due to the susceptibility to deformation these artifacts can complicate examination of a soft tissue like breast [3, 7].

The purpose of this work is to identify specific features of the photon density normalized maximum (PDNM) movement in biological soft tissues with considering possible tissue deformation. The
influence of tissue deformations on photon density distribution is numerically simulated using diffusion approximation to the radiative transfer equation (optical properties simulation) and basic equation of the elasticity theory for quasi-static deformation (stress-related properties simulation).

2. Experimental part

2.1. Light propagation model

To describe the diffusion of photons in biological tissue we used particular case of diffusion approximation of the radiative transfer equation. It was the Model of a Dropus – radiation pulse containing fixed initial number of photons that appeared in the object near its surface and diffused within the object [8]. Such approach allows sufficiently precise description of experimental data for homogeneous and inhomogeneous cases and it is based on the solution of the radiative transfer equation for a light pulse, containing a finite number of photons. According to the diffusion equation, the density of diffusing photons is described as [9-11]:

\[ c^{-1} \frac{\partial \phi(x, y, z, t)}{\partial t} - D(x, y, z) \nabla^2 \phi(x, y, z, t) + \mu_s(x, y, z) \phi(x, y, z, t) = S(x, y, z, t), \quad \forall x, y, z \in \Omega. \] (1)

where \( c = c_0/v_{\text{object}} \) is light speed in the medium; \( c_0 \) – light speed in vacuum; \( v_{\text{object}} \) – relative refractive index of the simulated object \((\Omega)\) and its boundaries \((\partial\Omega)\); \( x, y, z \) are coordinates of all points of the final simulated area consisting of inner part of simulated object \(\Omega\), its boundaries \(\partial\Omega\), radiation source \((q)\), detectors, and the medium surrounding the object;

\[ D(x, y, z) = \left\{ \left[ \mu_s(x, y, z) + \left( 1 - g(x, y, z) \right) \mu_a(x, y, z) \right] \right\}^{-1} \] (2)

where \( D \) is the diffusion coefficient at the position \(x, y, z\); \( \mu_s(x, y, z) \) is diffusion and absorption coefficients; \( \mu_a(x, y, z) \) – scattering coefficient; – anisotropy factor (the average cosine of the scattering angle); \( \phi(x, y, z, t) \) – photon density at the point with coordinates \(x, y, z\) at a time \(t\); and \( S(x, y, z, t) \) – photon source function.

The boundary condition of the third kind (the Robin condition) was used to describe the photon flux on the object boundary. The flux of photons, leaving the finite domain through the boundary, was equal to the flux at the boundary, multiplied by the coefficient including the light back reflection into the object [11-13]:

\[ \phi(x, y, z, t) + 2D(x, y, z)F \frac{\partial \phi(x, y, z, t)}{\partial n_{\lambda, y, z, -1}} = 0, \quad \forall x, y, z \in \partial\Omega, x, y, z \in q, \] (3)

where \( n(x, y, z) \) is direction of the outer normal to the boundary \(\partial\Omega\) at the point with coordinates \(x, y, z\). \( F \) – Fresnel reflection coefficient [4, 14] was calculated as follows:

\[ F = \frac{2/(1 - R_0) - 1 + \cos(Q_c)^2}{1 - \cos(Q_c)^2}, \] (4)

where \( R_0 \) and \( Q_c \) – coefficients, respectively equal to:

\[ R_0 = \left( \frac{v_{\text{object}}}{v_{\text{medium}}} - 1 \right)^2 \left( \frac{v_{\text{object}}}{v_{\text{medium}}} + 1 \right)^{-2} \quad \text{and} \quad Q_c = \arcsin\left( \frac{v_{\text{medium}}}{v_{\text{object}}} \right), \] (5)

where \( v_{\text{medium}} \) – the relative refractive index of the medium surrounding the object (for the air \( v_{\text{medium}} = 1 \)).

After completion of the simulation iterative process PDNM function, \( \phi(x, y, z, t) \), is normalized [3] with respect to its maximum \( \phi_{\text{max}}(x, y, z, t) \):
\[
\varphi_{\text{norm}}(x, y, z, t) = \frac{\varphi(x, y, z, t)}{\varphi_{\text{max}}(x, y, z, t)},
\]
and represented as follows:
\[
\varphi_{\text{PDNM}}(x, y, z, t) = \begin{cases} 
1, & \varphi_{\text{norm}}(x, y, z, t) \geq P \\
\varphi_{\text{norm}}(x, y, z, t), & \text{else}
\end{cases},
\]
where \( P \) is the experimentally determined minimum of photon density level \( 0 < P \leq 1 \).

2.2. Breast deformation model

Soft biological tissues are nonlinear elastic media. Nevertheless, in some cases, for example when the deforming forces cause small bending (\( \leq 5\% \)), the soft tissues can still be considered as a media with linear properties. For example in DOT, the breast tissue should be presented as a linear isotropic pseudo-incompressible medium [3, 5-7].

In this case, the basic equation of the elasticity theory for quasi-static deformation on the internal nodes of the simulated area is given by [3, 15]:
\[
(\lambda + \mu)\nabla(\nabla u) + \mu \nabla^2 u = 0,
\]
and the same equation for nodes on the boundary \( \partial \Omega \) of the studied area is represented as follows [6]:
\[
((\lambda + \mu)\nabla(\nabla u) + \mu \nabla^2 u) \cdot \mathbf{n} - n = 0,
\]
where \( \mathbf{n} \) is a unit vector directed outwards from \( \Omega \); \( h \) represents the tension on the surface and boundary of the simulated area; \( u = (u_1, u_2, u_3) \) – the displacement vector components at the axes \( x, y, z \) in the Cartesian coordinate system; \( \mu \) and \( \lambda \) – Lame’s elastic constants. These constants for isotropic medium (first and second Lame’s elastic constants) are associated with the Young’s modulus \( E \) and Poisson ratio, \( \nu \), as follows [7, 16]:
\[
\mu = 0.5 \cdot E \cdot (1 + \nu)^{-1}
\]
and
\[
\lambda = \nu E \cdot [(1 + \nu)(1 - 2\nu)]^{-1}.
\]

Thus, the simulated object was considered to be free of any initial deformations and an influence of the internal deforming forces (for example, muscle activity). All deformations were considered to be caused by external loads as source and detection fibers and their fixtures [3].

To simplify the calculations it is assumed, that these fibers are located on slightly reflective elastic band with adjustable diameter consisting of two identical halves. It allows fixing painlessly the source and detection fibers on the investigated biomedical object. Only one fiber was used for the injection of photons, meanwhile detection fibers surrounded the investigated object, and were located at the equal angles to each other to the right and left from the source fiber. Thus, the investigated object will be deformed in a plane of the fibers only. Cross-section made at this height will appear as an ellipse rather than a circle. In the simulation of elastic properties of biological tissue it will be assumed that Poisson ratio \( \nu = 0.495 \), Young’s modulus \( E = 20 \) kPa [3].

It should be noted that scattering and absorption coefficients of the tissue under the deformation may insignificantly change [15, 16], therefore suggested model does not consider these changes.

The described approach was numerically implemented using LabVIEW software package. The numerical solution of the equations (1) - (11) was performed using finite difference method (FDM), using the implicit difference scheme built on seven-point grid pattern [17].

It is important to note that in contrast to the finite element method (FEM) FDM is characterized by a regular grid in all directions, that is more convenient for determining specific features of radiation propagation. Further, compared to Monte-Carlo simulation the diffusion approximation of radiative transfer equation has higher computing speed, that is beneficial for serial experiment.
We also used numerical simulation to determine the effect of bioobject deformation on the fitures of PDNM movement.

3. Results and discussion
Photon density distribution was consequently simulated (\( P=0.995-0.999 \)) for the time-resolved cases using pulsed irradiation in homogeneous and inhomogeneous, deformed and undeformed conical objects with optical properties of breast tissue. Figure 1 shows the position of spherical inhomogeneity in a conical object. The optical properties of the single inhomogeneity and the homogeneity part of phantom are chosen in such a way as to correspond to real clinical data. The radius of the heterogeneity corresponds to the average radius of the tumor, taking into account the region of angiogenesis [8, 11]. The slice of the object is taken at the half height of the object and in the plane of the fibers [3, 4, 8].

![Figure 1. Spatial distribution of optical properties in the plane taken at the half height of heterogeneous conical object (inhomogeneity is considered to be spherical).](image1)

Figure 2 shows the simulation results of the photon distribution in the plane (made at the level of the source and detector fibers) of homogeneous undeformed object at different moments of time \( t \). Figure 2a corresponds to \( t=0.7 \) ns, and Figure 2b to \( t=3 \) ns. Optical properties, absorption, \( \mu_a(x,y,z) \), and reduced scattering, \( \mu'_s(x,y,z) = (1 - g(x,y,z))\mu_a(x,y,z) \), coefficients for \( \forall(x,y,z) \in \Omega \) are equal to 0.004 mm\(^{-1}\) and 0.5 mm\(^{-1}\), respectively [18]. The same simulation results for the optically homogeneous deformed object are shown in Figure 3a and Figure 3b.

![Figure 2. Photon density distributions in the slice of the homogeneous conical object at the following times after the injection of the pulse: 0.7 ns (a) and 3 ns (b).](image2)
Figure 3. Photon density distributions in the slice of the deformed conical homogeneous object at the following times after the injection of the pulse: 0.7 ns (a) and 3 ns (b). PDNM moves to the center.

Similar simulations have been performed in a same conical object with a spherical absorbing inhomogeneity inside of it [13, 14]. Inhomogeneity diameter is 0.4 of the object radius, located at the angle of 135° with respect to the axis of the incident irradiation, at the depth of 0.25 of the plane radius, R, \( \mu_a(x, y, z) = 0.01 \text{ mm}^{-3} \) [18]. Photon density distributions in the plane at the half height of the inhomogeneous undeformed object at time points 0.7 ns and 3.5 ns are presented in Figure 4a and Figure 4b, respectively. As it can be seen in the case of absorbing inhomogeneity, PDNM moves toward the point, which is symmetrical to geometrical center of the heterogeneity relative to the center of the investigated object. Separate investigations have been shown that increasing inhomogeneity absorbing coefficient and size of it leads to increase of PDNM motion speed [14].

Figure 4. Photon density distributions in the slice of the inhomogeneous conical object at the following times after the injection of the pulse: 0.7 ns (a) and 3.5 ns (b).

For the object in the deformed state the similar results were obtained. Figure 5a (t=0.7 ns) and Figure 5b (t=3.5 ns) shows that PDNM moves similarly, hence the object deformation and its
embedded inhomogeneity have no significant influence on the photon density and its normalized maximum [18].

![Figure 5. Photon density distributions in the slice of the deformed conical inhomogeneous object at the following times after the injection of the pulse: 0.7 ns (a) and 3.5 ns (b).](image)

However, it has been found that rate of the diffuse transmittance intensity decay in the undeformed objects is slightly higher (8-10%) than in those with deformations. The revealed specific features were verified using tissue-like phantoms [3].

Thus, the deforming effects do not influence the direction of PDNM movement, but they affect the speed of its movement. This means that soft tissue deformations should be taken into account in DOT mammography system design [19] and in solution of the inverse problem based on the acquired data.

In the design of an effective DOT system, glass compression plates (which cause severe deformations of tissue), or the cup-shaped applicator (which has minimum deformation, but poor contact with the investigated biological tissue), should be replaced with an elastic bracelet with injection and detection ports. This approach provides a combination of acceptable deformations and acceptable loss of a useful signal. In addition, an elastic reflective bracelet-holder of injection and detection ports makes it possible to simplify DOT system design.

The software improvements in reconstruction algorithm should consist of more effective initial approximations to the solution of the inverse problem. If the radiation intensity registered by all detectors of time-resolved DOT system in the region of late arriving photons (LAP) is approximately the same, this should be interpreted as the relative homogeneity of the investigated biological tissue (PDNM reaches the geometric center of the object). If LAP intensity decreases more rapidly in one or more detectors of a region, this can be interpreted as evidence of the presence of an absorbing heterogeneity near to them (PDNM at the point symmetric to the center of the absorbing pathological structure). If the intensity of radiation on one or several detectors decreases slower, then a scattering heterogeneity near these detectors (PDNM in the center of the scattering pathological structure) is highly likely [20].

**Conclusions**

Numerical models of optical and mechanical properties of biological tissues were described. A series of numerical simulations considering effect of deformations of biological tissues on the trajectory of the PDNM motion are presented. It has been revealed that independently on the values of the absorption and scattering coefficients PDNM moves toward the geometric center of the investigated optically homogeneous object in the deformed and undeformed states. Then, in presence of an
absorbing inhomogeneity, PDNM moves toward the point being symmetric to its geometric center in relation to the center of conical object regardless with or without deformation. Finally, it has been shown that in the undeformed object radiation intensity decay is about 8-10% higher than in the deformed one.

The significance of the described specific features favour better understanding the influence of different factors on optical radiation propagation and decay in biological tissues. The specific features of PDNM movement in a deformed and undeformed biological object or tissue-like phantom are useful in developing effective methods for solving DOT inverse problem described in [3, 17, 20], since they have more precise initial approximation to spatial distribution of the optical properties in investigated tissue.

An important direction of further development is more accurate accounting for the decrease in average absorbing coefficient of breast tissue in the deformed case. This effect is quite important for mammography DOT applications.

Acknowledgements
This work is supported by The Russian Science Foundation (RSF Project 16-15-10327).

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