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Effective optode configuration for the image reconstruction in diffuse optical tomography

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Abstract

The influence of the configuration of the optodes on the images of diffuse optical tomography (DOT) was investigated using 3D numerical simulations. 3D distributions of the absorption coefficients in a spherical object were reconstructed from the numerically simulated measurement data for various configurations of the optodes. When the optodes were placed in a plane containing a target which strongly absorbs light, the target could be reconstructed with good localization. For good reconstruction of the target, it was found to be very important that the optode configuration was optimized in order to detect light propagating through the target effectively. The simulations also showed that the optode configuration affects the quality of the reconstructed images and that some prior information about the measured object improved the DOT images. Finally the simulation results were verified by a phantom experiment.

Keywords: Diffuse optical tomography; Near-infrared light; Inverse problem

Introduction

Diffuse optical tomography (DOT) [1–3] is only one step away from practical medical application and will soon play a very important role in medical diagnoses and treatments, such as the monitoring of the brains of newborn infants [4–7] and for the screening of the breast cancers [8–11]. DOT utilizes the nature of the near-infrared (NIR) light which can be easily transmitted in biological media, and provides information about concentrations of the biological constituents, such as oxy- and deoxyhemoglobin, with spectroscopic techniques. By reconstructing the distribution of the optical properties such as the scattering and absorption coefficients inside the measured object from light detected non-invasively at the surface of the object, DOT allows us to diagnose the object because changes in the optical properties reflect the condition of the biological tissues.

One of the main issues with DOT is the spatial resolution. Unlike X-ray computed tomography (X-ray CT), and magnetic resonance imaging (MRI), it is not easy for DOT to achieve the spatial resolution in the millimeter range. Although the images often look blurred, DOT can provide physiological information, whereas X-ray CT and MRI provide anatomical information. DOT technology is required to improve the spatial resolution in order to diagnose the early stages of breast cancer and/or make visible the localized disorders in newborn infant's brains. The low resolution can be attributed to the ill-posedness of the inverse problem appearing in the DOT image reconstruction. Due to the diffusive nature of light propagation in biological media, the inverse

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problem of DOT is accompanied by a difficult aspect which is completely different from that of the X-ray CT.

A large amount of measurement data is essential for good image reconstruction. To improve the image quality, such as spatial resolution, increasing the amount of measurement data is effective in alleviating the ill-posedness of the inverse problem. Using the time-resolved measurement with an ultra-short pulsed laser and the time-correlated single-photon counting method is one of a number of powerful approaches. Another approach is to increase the number of detectors. The effect of increasing the number of the source–detector pairs has been precisely reported for optical topography [12], although the optical topography usually does not involve the inverse problem. However, for DOT, especially with the time-resolved measurement, it is not easy to add the source–detector pairs because it is costly. The addition of the detectors also leads to an increase in the time required for data acquisition and for reconstruction, which is also problematic.

Therefore, for the efficient application of DOT, it is very important to investigate the optimum arrangement of the detectors. Within the instrumental limitation, the performance of the DOT should be maximized. The influence of the positions of the detectors on the DOT images is intuitively or experimentally known but has hardly ever been reported before now. In this paper, we carried out some numerical simulations to compare the reconstructed images for various arrangements of the optodes, i.e. the sources and detectors. The modified generalized pulse spectrum technique (mGPST) which recovers the image from the time-resolved data was used for the image reconstruction algorithm [13,14]. Some conditions of the optode configuration are discussed with regard to improving the quality of the reconstructed image. Finally, a phantom experiment was carried out, and the result is discussed, taking into account the results of the numerical simulations.

Methods of the image reconstruction of DOT

DOT in this study employs a time-domain technique which uses an ultra-short pulsed laser for illuminating the object and picosecond (ps) time-resolved detectors for measuring the transmitted light pulses. The incident NIR light is propagated through the object and is subject to scattering and absorption by the biological tissues. A fraction of the light is reemitted from the surface of the object. The light measured by detectors, such as the photomultiplier tubes, is used to reconstruct the optical properties inside the object.

Light propagation in biological media is rigorously described by the radiative transfer equation (RTE). This integropartial differential equation can be solved numerically by deterministic or stochastic methods. However, these methods are computationally intensive which is problematic for image reconstruction, as it usually requires iterative calculation. Therefore, the photon diffusion equation (PDE) which is the approximation of RTE and valid for biological objects larger than 10 mm is frequently employed in the DOT technique [1–3]:

$$-\nabla \cdot (D(r)\nabla\phi(r,t)) + \mu_a(r)\phi(r,t) + \frac{1}{c}\frac{\partial\phi(r,t)}{\partial t} = +q(r,t), \qquad (1)$$

where $\phi(\boldsymbol{r}, t)$ is the fluence rate at the position $\boldsymbol{r}$ and time t, $D(\boldsymbol{r}) = 1/(3\mu_s'(\boldsymbol{r}))$ the diffusion coefficient, $\mu_s'(\boldsymbol{r})$ the reduced scattering coefficient, $\mu_a(\boldsymbol{r})$ the absorption coefficient, $q(\boldsymbol{r}, t)$ the light source in the medium, and c the speed of light.

Based on the forward model of PDE, the mGPST [13,14] is employed to reconstruct the distributions of the optical properties in the medium. The mGPST uses the Laplace transform to extract the features of the time-resolved measurement data. The transformed "featured data" which effectively characterize the measurement data are used for the reconstruction. After the Laplace transform, the PDE is described as follows:

$$-\nabla \cdot (D(r)\nabla\phi(r,p)) + \mu_a(r)\phi(r,p) + \frac{p}{c}\phi(r,p) = +q(r,p), \qquad (2)$$

where $\phi(\boldsymbol{r}, p) = \int_0^\infty \phi(\boldsymbol{r}, t)e^{-pt}dt$ is the Laplace-transformed fluence rate with the power of the exponential, p, and $q(\boldsymbol{r}, p)$ the Laplace-transformed light source.

When the light source of impulse light illuminates the medium from the surface at the position $\boldsymbol{r}_s$, Eq. (2) is subject to the following initial condition:

$$\phi(\boldsymbol{r}, p) = \delta(\boldsymbol{r} - \boldsymbol{r}_s), \qquad (3)$$

where δ is the delta function. Also Eq. (2) is subject to the following boundary condition describing that no light flux is coming in from the boundary, $\boldsymbol{r}_b$, except the illuminating position:

$$D(\boldsymbol{r}_b)\nabla_n\phi(\boldsymbol{r}_b, p) = -\frac{1}{2A(\boldsymbol{r}_b)}\phi(\boldsymbol{r}_b, p), \qquad (4)$$

where ∇_n means the gradient normal to the surface, $A(\boldsymbol{r}_b) = [1 + R_d(\boldsymbol{r}_b)]/[1 - R_d(\boldsymbol{r}_b)]$ is the parameter depending on the internal reflection ratio, R_d [16].

The light intensities measured by the detectors located at the boundary, $\Gamma(\boldsymbol{r}_b, p)$, are the fluxes of the fluence rate outgoing from the boundary (Fick's law) and are given by the following:

$$\Gamma(\boldsymbol{r}_b, p) = -D(\boldsymbol{r}_b)\nabla_n\phi(\boldsymbol{r}_b, p), \qquad (5)$$

The forward problem of Eq. (2) is solved numerically by the finite element method (FEM) [12] under the initial and boundary conditions Eqs. (3) and (4), and the measured light intensities are obtained by Eq. (5).

Fundamentally, the reconstruction algorithm minimizes the residual errors between the set of the measured featured data, $\boldsymbol{\Gamma}_m$, and that of the calculated featured data, $\boldsymbol{\Gamma}$. The vectors, $\boldsymbol{\Gamma}_m$ and $\boldsymbol{\Gamma}$, have M components corresponding to