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Juan Felipe Perez-Juste Abascal,*a+; Juan Aguirre,ab; Judit Chamorro-Servent,a; Martin Schweiger, b Simon Arridge, b Jorge Ripoll,ab; Juan J. Vaquero,a and Manuel Desco,a,d

Abstract. Reconstruction algorithms for imaging fluorescence in near infrared ranges usually normalize fluorescence light with respect to excitation light. Using this approach, we investigated the influence of absorption and scattering heterogeneities on quantification accuracy when assuming a homogeneous model and explored possible reconstruction improvements by using a heterogeneous model. To do so, we created several computer-simulated phantoms: a homogeneous slab phantom (P1), slab phantoms including a region with a two-to-six fold increase in scattering (P2) and in absorption (P3), and an atlas-based mouse phantom that modeled different liver and lung scattering (P4). For P1, reconstruction with the wrong optical properties yielded quantification errors that increased almost linearly with the scattering coefficient while they were mostly negligible regarding the absorption coefficient. This observation agreed with the theoretical results. Taking the quantification of a homogeneous phantom as a reference, relative quantification errors obtained when wrongly assuming homogeneous media were in the range +41 to +94% (P2), 0.1 to −7% (P3), and −39 to +44% (P4). Using a heterogeneous model, the overall error ranged from −7 to 7%. In conclusion, this work demonstrates that assuming homogeneous media leads to noticeable quantification errors that can be improved by adopting heterogeneous models. © 2012 Society of Photo-Optical Instrumentation Engineers (SPIE). [DOI: 10.1117/1.JBO.17.3.036013]

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1 Introduction

Diffuse optical tomography in fluorescent mode (fDOT), also called fluorescent molecular tomography when fluorescent probes are used for molecular imaging, is becoming an important pre-clinical non-invasive method for small animal imaging. fDOT reconstruction algorithms model the propagation of light through the subject under study. The model is usually described by the diffusion approximation, since biological tissues present high scatter for wavelengths in the visible and near infrared spectrum ranges. Under this approximation, the optical properties of tissues are described statistically by the scattering coefficient \( \mu_s \) and the absorption coefficient \( \mu_a \). While analytical solutions exist for media with homogeneous optical properties and simple geometries, more complex numerical solutions are required for media with heterogeneous optical properties or more general spatial domains.

The existence of different scattering and absorption coefficients for different tissue types is a realistic assumption when facing the problem of modeling light transport in living subjects. One solution is to estimate these values using DOT and then perform fDOT. However, one cannot simultaneously recover absorption and scattering coefficients in constant-wave mode. Additionally, prior knowledge of the optical parameters in the frequency domain is necessary for accurate fDOT quantification. Some authors neglect the different values of the scattering parameter and only estimate the absorption coefficient with DOT. This coefficient is then used to reconstruct the fluorophore distribution. Another approach involves combining previously published values for tissue-specific optical parameters with high-resolution anatomical information provided by CT or MRI; however, there is no consensus in the literature on the actual values of tissue-specific optical parameters, and considerable inter-subject variability is reported.

An alternative approach simplifies the problem by assuming homogeneous optical properties and normalizing fluorescence photon density to excitation photon density following the so-called normalized Born approximation. This approach is an adaptation of previous DOT algorithms and facilitates the use of fDOT for small animal imaging, thanks to its low sensitivity to inhomogeneities and straightforward implementation.

However, only a few studies have assessed the quantification accuracy of the method in the presence of different levels of absorption and, to our knowledge, no studies using either simulations or experiments have assessed the effect of scattering...
heterogeneities on quantification accuracy. In addition, the effects of tissue-specific optical properties reported in previous DOT-fDOT studies without data normalization cannot be extrapolated to normalized data.11,12

The goal of our work was to assess the effect of absorption and scattering coefficients on quantification accuracy. In order to provide a theoretical framework, we obtained analytical expressions to explain the dependence of normalized data on changes in background absorption and scattering for homogeneous media using Green’s function for an infinite space and an infinite slab. In principle, if normalized data are not affected by inaccuracies in optical property modeling, they should be invariant to changes in the scattering or absorption of the medium. We then studied the quantification error using four computer-simulated phantoms based on finite element methods. The first phantom was a homogeneous slab with different background absorption and scattering coefficients (Sec. 2.6.1) that was used to validate the previously obtained theoretical result. The second was a homogeneous slab with a fluorescent cylindrical region in which absorption and scattering coefficients were defined as two, four and six times the background value (Sec. 2.6.2) to cover the wide range of absorption and scattering coefficients reported in the literature.13-16 The third was a computer-simulated mouse model that included different scattering coefficients for lung and liver and was used to assess the relationship between quantification accuracy and the tissue-specific values of the scattering coefficient (Sec. 2.6.3). Image reconstructions were performed with the algebraic reconstruction technique based on two different models: 1) assuming homogeneous media and 2) realistically modeling the heterogeneous optical properties using a Jacobian matrix for which heterogeneous optical properties can be explicitly modeled by the finite element method. The resulting reconstructed images were compared quantitatively.

We hypothesized that the assumption of homogeneous optical properties for modeling diffuse media with a highly heterogeneous scattering coefficient could lead to noticeable quantification errors despite data normalization. Accordingly, we also assessed whether a reconstruction method controlling for heterogeneous scattering coefficients could lead to an improvement in quantification accuracy.

2 Methods

2.1 Forward Problem

2.1.1 Diffusion approximation

Let $\mu_a(r, \lambda)$ be the absorption, $\mu'_a(r, \lambda)$ the reduced scattering, and $k(r, \lambda) = \{3\mu_a(r, \lambda) + \mu'_a(r, \lambda)\}^{-1}$ the diffusion coefficient for a wavelength $\lambda$ at position $r$ in a domain $\Omega$. In a diffuse range, which is valid for highly scattering ($\mu'_a \gg \mu_a$) and weakly anisotropic media, photon density is approximated by the diffusion equation.17 Hence, in fDOT, the excitation photon density $\Phi_{ex}(r, \lambda_{ex})$ at the excitation wavelength $\lambda_{ex}$ and emission photon density $\Phi_{em}(r, \lambda_{em})$ at the emission wavelength $\lambda_{em}$ are given by the solution of a pair of coupled diffusion equations.15 The excitation photon density is emitted by an external source $q_0(r_0)$ at a location $r_0 \in \Omega$, and the emission photon intensity comes from a fluorescent region characterized by a fluorescent yield $F(r)$, which accounts for its quantum efficiency, absorption parameter and the concentration of fluorophore. Assuming a time-stationary problem (frequency $\nu = 0$) and that the absorption parameter is not affected by the presence of the fluorophore, excitation and emission photon densities are given by

$$-\nabla \cdot k(r, \lambda_{ex}) \nabla \Phi_{ex}(r, \lambda_{ex}) + \mu_a(r, \lambda_{ex}) \Phi_{ex}(r, \lambda_{ex}) = q_0(r_0)$$

(1)

$$-\nabla \cdot k(r, \lambda_{em}) \nabla \Phi_{em}(r, \lambda_{em}) + \mu_a(r, \lambda_{em}) \Phi_{em}(r, \lambda_{em}) = F(r) \Phi_{ex}(r, \lambda_{ex})$$

(2)

In Eqs. (1) and (2), optical coefficients are different at excitation and emission wavelengths, and should be modeled differently in practice. However, since our main goal was to assess the impact of scattering, we believe that including additional problems such as optical properties mismatch at emission and excitation wavelengths would introduce additional variations. Hence, we opted for this simplified expression

$$-\nabla \cdot k(r) \nabla \Phi_{ex}(r) + \mu_a(r) \Phi_{ex}(r) = q_0(r_0)$$

(3)

$$-\nabla \cdot k(r) \nabla \Phi_{em}(r) + \mu_a(r) \Phi_{em}(r) = F(r) \Phi_{ex}(r)$$

(4)

The diffusion equations [Eqs. (1) and (2)] can be solved analytically using Green’s function for homogeneous media and canonical geometries17 and by applying numerical methods, such as the finite element method, for heterogeneous media and/or general geometries.15-16

From a general perspective, let us define a Green function $G(r, r')$ that solves the heterogeneous problem

$$-\nabla \cdot k(r) \nabla G(r, r') = \delta(r - r').$$

(5)

The photon densities that solve Eqs. (3) and (4) are therefore given by

$$\Phi_{ex}(r) = \int dr' G(r, r') q_0(r')$$

(6)

$$\Phi_{em}(r) = \int dr' G(r, r') F(r') \Phi_{ex}(r').$$

(7)

In this work, the solutions to Eqs. (3) and (4) are given for homogeneous canonical geometries by Green’s function and for both homogeneous and heterogeneous media with general geometries by a Galerkin finite element approach applied using TOAST. In the finite element implementation, photon density and concentration of fluorophore were approximated by piecewise-linear nodal basis functions. Sources [RHS on Eq. (3)] were modeled as isotropic point sources (located at a depth $1/\mu'_a$ below the surface), which resemble a collimated laser, as described in Ref. 13. Measurements were modeled by a Gaussian kernel centered at the detector location and computed as a linear operator $\mathcal{M}$ acting on the photon density at the boundary of the domain. Thus, measured excitation and emission photon densities at the detector position $r_d$ are given by

$$\Phi_{ex}^{meas}(r_d) = \mathcal{M} \Phi_{ex}(r) \quad \text{and} \quad \Phi_{em}^{meas}(r_d) = \mathcal{M} \Phi_{em}(r).$$

(8)
2.1.2 Normalized data: general case

Normalized data for IDOT, that is, the incorrectly termed Born ratio, are given by

\[
U_h(r_d) = \frac{\Phi_{\text{meas}}(r_d)}{\Phi_{\text{ex}}(r_d)} = \frac{\int \mathrm{d}^3r' G(r_d; r') F(r') \Phi_{\text{ex}}(r')}{\int \mathrm{d}^3r' G(r_d; r') q(r')}
= \int \mathrm{d}^3r' G(r_d; r') F(r') \int \mathrm{d}^3r'' G(r'; r'') q(r''),
\]

(9)

where the \( M \times 1 \) vector \( U_h \) accounts for all source-detector measurement pairs.

2.2 Normalized Data: Homogeneous Media

In this section, we used Green’s function for homogeneous media to derive analytical expressions that describe the dependence on normalized data with background absorption and scattering coefficients. We theoretically solved two cases of interest: infinite space and infinite slab.

Assuming a point source \( q(r) = \delta(r - r_s) \) and a fluorescent point \( F(r) = \delta(r - r_f) \), after substitution in Eq. 4, leads to a simplified form, as follows:

\[
u(r_d; r_s, r_f) = \frac{\int \mathrm{d}^3r' G(r_d; r') \delta(r' - r_s) \int \mathrm{d}^3r'' G(r'; r'') \delta(r'' - r_f)}{\int \mathrm{d}^3r' G(r_d; r') \delta(r' - r_s)} \frac{G(r_d; r_r) G(r_r; r_s)}{G(r_d; r_s)}.
\]

(10)

2.2.1 Infinite space

The Green function \( G(r, r') \) for an infinite space is

\[
G(r, r') = \frac{1}{(2\pi)^{3/2} 2\kappa} \exp \left( -\sqrt{3\mu_a + \mu'_a} |r - r'| \right) \frac{1}{|r - r'|},
\]

and the normalized data are

\[
u(r_d; r_s, r_f) = \frac{1}{(2\pi)^{3/2} 2\kappa} \exp \left( -\sqrt{3\mu_a + \mu'_a} |r_s - r_f| \right)
\times \frac{\int \mathrm{d}^3r' G(r_d; r') \delta(r' - r_s) \int \mathrm{d}^3r'' G(r'; r'') \delta(r'' - r_f)}{\int \mathrm{d}^3r' G(r_d; r') \delta(r' - r_s)} \frac{G(r_d; r_r) G(r_r; r_s)}{G(r_d; r_s)}.
\]

(11)

2.2.2 Infinite slab

The Green function \( G(r, r') \) for an infinite slab is

\[
G_{\text{slab}}(r, r') = \frac{1}{(2\pi)^2 2\kappa} \sum_{n = -\infty}^{\infty} \exp \left( -\sqrt{3\mu_a + \mu'_a} |r - r' + \rho_{12 - n'} \right) \rho_{12 - n'}
\]

(14)

and \( \rho \) is the slab height.

Thus, the normalized data are given by

\[
u(r_d; r_s, r_f) = \frac{1}{(2\pi)^{3/2} 2\kappa} \exp \left( -\sqrt{3\mu_a + \mu'_a} |r_s - r_f| \right) \frac{G(r_d; r_r) G(r_r; r_s)}{G(r_d; r_s)}.
\]

(15)

which for \( \mu'_a \gg \mu_a \) is simplified as

\[
u(r_d; r_s, r_f) = \frac{3\mu'_a}{2(2\pi)^2} \exp \left( -\sqrt{3\mu_a + \mu'_a} |r_s - r_f| \right) \frac{G(r_d; r_r) G(r_r; r_s)}{G(r_d; r_s)}.
\]

(16)

Hence, we obtain a relationship similar to that described in the case of infinite space.

2.3 Jacobian Matrix

The emission photon density \( \Phi_{\text{ex}} \) is linearly dependent on the fluorophore concentration \( F \) under certain assumptions. This can be seen by taking its Fréchet derivative with respect to the concentration of fluorophore and by assuming that a variation of fluorophore \( \delta F \) induces a variation of emission photon density \( \delta \Phi_{\text{ex}} \), while the excitation photon density \( \Phi_{\text{ex}} \) (and implicitly the optical parameters) remains invariant and there is no re-emission by the fluorophore,

\[
\frac{\partial}{\partial F} \left[ \frac{\Phi_{\text{ex}}(F + \tau \delta F)}{\Phi_{\text{ex}}} \right] \Big|_{\tau = 0} = V U_h \delta F.
\]

(20)

By differentiating both sides of Eq. 2 and discretizing the integral as the sum of all finite elements \( \Omega_h \), the variation of emission photon density is given by
\[ \delta \Phi_{\text{em}}(r_d) = \sum_j \int_{\Omega_j} \text{d}r_j G(r_d, r_j) \Phi_{\text{ex}}(r_j, r_s) \delta F_j \]
\[ = \sum_j \int_{\Omega_j} \text{d}r_j \tilde{\Phi}(r_d, r_j) \Phi_{\text{ex}}(r_j, r_s) \delta F_j, \quad (21) \]
where \( \Phi_{\text{ex}}(r_j, r_s) \) is the excitation photon density at \( r_j \) due to a source at \( r_s \), reciprocity \( G(r_j, r_d) = G(r_d, r_j) \) is used, and \( \Phi(r_j, r_d) \) is the adjoint field at \( r_j \) due to a source \( \tilde{q}_0 \) located at the detector position \( r_d \), which solves an adjoint equation
\[ [-\nabla \cdot \kappa(r) \nabla + \mu_a(r)] \tilde{\Phi}(r) = \tilde{q}_0(r_d). \quad (22) \]
Comparing Eqs. (21) and (22), the Jacobian matrix entry \( J_{ij} \) that relates each measurement \( (U_b)_i \) (where the subindex \( i \) accounts for each source-detector pair) to the concentration of fluorophore at the element \( \Omega_j \) is
\[ J_{ij} = \frac{\partial (U_b)_j}{\partial F_j} = \frac{1}{\Phi_{\text{em}}(r_d, r_j)} \int_{\Omega_j} \text{d}r_j \tilde{\Phi}(r_d, r_j) \Phi_{\text{ex}}(r_j, r_s). \quad (23) \]
Let \( J \) be an \( M \times N \)-matrix after it has been mapped to a uniform grid and \( F \) an \( N \times 1 \)-vector of fluorophore coefficients, where \( N \) is the number of voxels. Then, the iDOT linear system can be expressed as
\[ U_b = JF. \quad (24) \]

The Jacobian matrix in Eq. (24) was computed using the finite element method, which enables modeling of heterogeneous optical properties.

### 2.4 Inverse Problem and Image Analysis

Reconstruction was performed using a randomized algebraic reconstruction technique which has been previously applied in fDOT\cite{Abascal2008}. A relaxation parameter of 0.1 and 20 iterations was applied to all the reconstructions.

Image quantification was performed by defining a Region of Interest (RoI) that corresponded to the known target distribution and by taking the average of the voxels within this RoI. The final concentration of fluorophore is expressed as a percentage of the corresponding value in the homogeneous phantom, which was used as a reference. Thus, concentrations become expressed in a relative scale as
\[ \frac{F}{F_{\text{homo}}} \]
where \( F \) and \( F_{\text{homo}} \) are the recovered concentrations of fluorophore corresponding to the heterogeneous and homogeneous model, respectively. The measurement error is also expressed as a percentage, given by \( 100(F - F_{\text{homo}}) / F_{\text{homo}} \).

In addition, profiles were drawn along the reconstructed images as a test of image quality and positional accuracy.

Five percent additive white noise was incorporated into normalized data before reconstruction.

### 2.5 Phantoms and Acquisition Protocol

#### 2.5.1 Slab phantom

We defined a heterogeneous slab-geometry phantom with two regions, a cylindrical fluorescent region in which optical parameters could be varied and a background region. The slab was 10 mm thick and the cylinder diameter was 5 mm (Fig. 1). A homogeneous phantom was used as a reference.

The acquisition protocol used a configuration of 9 \times 9 sources located on the lower plane covering a surface of 12 by 12 mm² (Fig. 1), together with an equivalent configuration of detectors located at the corresponding positions on the upper plane.

Computer-simulated data for each experimental phantom were generated using a fine tetrahedral finite element mesh (115 350 nodes and 540 000 tetrahedra). A coarser mesh (55 061 nodes and 253 125 tetrahedra) was used for calculation of the Jacobian matrix, which was resampled into a uniform mesh of 20 \times 20 \times 20 for the reconstruction.

#### 2.5.2 Mouse model phantom

A heterogeneous model of a mouse that simulates differences in lung and liver scattering was designed based on the Digimouse atlas.\cite{Gouverneur2006} Resembling the experimental set-up proposed in Ref. \cite{Gouverneur2006}, we simulated a parallel-plate imaging chamber, where the mouse is gently compressed by moving plates filled with intralipid. The model comprised four tissue types: lung, liver, surrounding mouse tissue, and intralipid. It was created by selecting lung and liver regions on the Digimouse model and mapping them into a slab-geometry finite element mesh measuring 12.5 mm in thickness (Fig. 1). The heterogeneous model was created by assigning different \( \mu'_s \) for these four regions: 2.12 mm⁻¹ for the lung and 0.65 mm⁻¹ for the...
liver as reported in Ref. 14 (for a wavelength of 700 nm); 1.2 mm\(^{-1}\) for the surrounding tissue, which was also used in Ref. 19; and 1.5 mm\(^{-1}\) for the intralipid region selected as a middle value. The absorption parameter was set to \(\mu_a = 0.01\) mm\(^{-1}\) and kept constant for all models.

The simulated acquisition configuration consisted of 10 \(\times\) 10 sources located at the inferior plane covering a surface of 15 \(\times\) 15 mm\(^2\), and the same configuration of detectors was used at the corresponding positions on the superior plane.

Finally, two spherical fluorophore spots measuring 2 mm in diameter were simulated and placed inside the right lung and the liver at the same \(z\)-level (\(z = 6.5\) mm) (Fig. 2).

2.6 Quantification Analysis

2.6.1 Homogeneous slab phantom

For the homogeneous slab phantom described in Sec. 2.5.1, adopting the parameters \(\mu_a = 0.01\) mm\(^{-1}\) and \(\mu_s' = 0.8\) mm\(^{-1}\) as a reference, quantification errors were studied as a function of an \(n\)-fold change in the background absorption coefficient and an \(n\)-fold change in the scattering coefficient (\(n = 0.5, 1.6, 2.7, 3.8, 4.9, 6\)). All data sets corresponded to identical spherical fluorophore distributions measuring 2.5 mm in diameter and were reconstructed assuming the reference coefficients.

2.6.2 Heterogeneous slab phantom

Heterogeneous slab-geometry phantoms (Fig. 1) were simulated by increasing the scattering coefficient (2\(\mu_s', 4\mu_s', 6\mu_s'\)) or the absorption coefficient (2\(\mu_a, 4\mu_a, 6\mu_a\)) in the cylindrical region with respect to the background optical parameters (\(\mu_a = 0.01\) mm\(^{-1}\) and \(\mu_s' = 0.8\) mm\(^{-1}\)). All data sets were reconstructed 1) by assuming a homogeneous medium and 2) by modeling the differences in scattering and absorption coefficients. The reconstructed images were quantitatively compared with the reconstruction corresponding to the homogeneous phantom as detailed above.

2.6.3 Mouse model phantom

Four reconstructions of this phantom were generated. The first one (reference phantom) made use of the homogeneous model with \(\mu_s' = 1\) mm\(^{-1}\), for both the simulation and the reconstruction. The remaining three were as follows: a) a heterogeneous model reconstructed using the heterogeneous model, b) a...
homogeneous model with optical parameters chosen as in Ref. 19 (\(\mu_0 s = 1\text{ mm}^{-1}\)), and c) a homogeneous model with scattering computed as the volume-weighted average of the heterogeneous tissue values (\(\mu_0 s = 1.27\text{ mm}^{-1}\)). The reconstructed images were quantitatively compared with the reconstruction corresponding to the homogeneous phantom, as detailed above.

3 Results

3.1 Normalized Data: Homogeneous Media

The dependence of the normalized data on background absorption and scattering was studied for the homogeneous infinite space solution [Eq. (12)]. Normalized data were generated for a point source located at (0,0,0), a point fluorophore located at (0,0,5) and a moving position of the detector along the \(x\)-axis (located at \((x,0,10)\), \(x\) varying from 0 to 6 mm). We found that the exponential term in Eq. (12) is equal to or less than one and decreases slightly when \(n\) increases, being equal to one for \(x = 0\). Hence, the change in normalized data is almost linear with a change in scattering and is almost unaffected by a change in absorption (Fig. 3).

3.2 Quantification Analysis

3.2.1 Homogeneous slab phantom

Quantification errors increased almost linearly with a change in background scattering and increased slightly with a change in background absorption (Fig. 4). For the homogeneous case, the ratio of these two should be linear, as it can be extrapolated from Eq. (18), where the exponential term and the spatial dependence vanished.
3.2.2 Heterogeneous slab phantom

Figure 5 shows the relative fluorophore concentration values [Eq. (25)] for changes in absorption and scattering on the quantification. Reconstructed images are shown in Fig. 6. Quantification errors induced by assuming a cylindrical heterogeneous region. Reconstructed images are also shown (Fig. 7). It seems that positional accuracy is not affected by cylindrical heterogeneity. In contrast, quantification is highly affected by scattering heterogeneity.

3.2.3 Mouse model phantom

Quantification results of the relative concentration of fluorophore spots, one inside the lung (LU) and one inside the liver (LI). The first row corresponds to simulation and reconstruction using the homogeneous model (first row), the reference homogeneous model (second row), and a homogeneous model with scattering value computed as a weighted average of the scattering values (homogeneous heterogeneous 0.93 1.07 0.87) between the recovered concentrations in LU and LI was used as a reference for the quantification. The second to fourth rows correspond to heterogeneous model simulations reconstructed with the heterogeneous model (second row), the reference homogeneous model (third row), and a homogeneous model with scattering value computed as a weighted average of the scattering values (homogeneous heterogeneous* 0.61 1.14 0.54) (fourth row).

Data model | Rec. model | LU | LI | LU/LI |
---|---|---|---|---|
homogeneous | homogeneous | 1.03 | 0.97 | 1.06 |
heterogeneous | heterogeneous | 0.93 | 1.07 | 0.87 |
heterogeneous | homogeneous | 0.78 | 1.44 | 0.54 |
heterogeneous | homogeneous* | 0.61 | 1.14 | 0.54 |

Table 1 Image quantification of the relative fluorophore concentration [Eq. (25)] in a computer-simulated mouse model with two fluorophore spots, one inside the lung (LU) and one inside the liver (LI). The first row corresponds to simulation and reconstruction using the homogeneous model (homogeneous, $\mu_s' = 1 \text{ mm}^{-1}$), where the average between the recovered concentrations in LU and LI was used as a reference for the quantification. The second to fourth rows correspond to heterogeneous model simulations reconstructed with the heterogeneous model (second row), the reference homogeneous model (third row), and a homogeneous model with scattering value computed as a weighted average of the scattering values (homogeneous heterogeneous* 1.27 mm$^{-1}$) (fourth row).
and liver. As the liver has a smaller scattering coefficient ($\mu_s = 0.65 \text{ mm}^{-1}$) than the background and the lung has a larger one ($\mu_s = 2.12 \text{ mm}^{-1}$), the fluorophore concentration can be both underestimated and overestimated (Fig. 8).

4 Discussion

Reconstruction based on normalized data has previously been found to be robust in the presence of absorption differences. To our knowledge, this is the first report presenting evidence of the effect of scattering heterogeneities and quantification accuracy using normalized data. Our results suggest that scattering heterogeneity has a noticeable negative effect on quantification accuracy.

For a homogeneous infinite space, normalized data present a dominant linear relationship with the scattering coefficient while remaining invariant with the absorption coefficient. Accordingly, for a homogeneous slab phantom, a reconstruction assuming incorrect optical parameters leads to quantification errors that increase almost linearly with the scattering coefficient while increasing slightly with the absorption coefficient.

For heterogeneous slab phantoms with different scattering coefficient in a cylindrical fluorescent region, we found that quantification errors induced by assuming a homogeneous medium ranged from +41 to +94%, taking quantification from a homogeneous phantom as a reference. Additionally, the presence of regions with absorption heterogeneities led to much smaller quantification errors, ranging from +0.1 to −7%, which agrees with the findings of previous reports. Quantification errors decreased significantly (range, −4 to +4%) when using a reconstruction model that accounted for the differences in scattering and absorption coefficients.

For the reconstruction of a computer-simulated mouse model that included different liver and lung scattering coefficients, we obtained quantification errors ranging from −39 to +44% when assuming a homogeneous medium and −7 to +7% when using a reconstruction model that accounted for the differences in scattering. Thus, neglecting the effect of scattering heterogeneities led to a systematic overestimation of the calculated concentration of fluorophore; however, the effect of absorption was remarkably smaller. In more realistic situations, the actual effect may be more complicated.

These results are consistent with those of a previous study that analyzed the effect of modeling errors simultaneously in both scattering and absorption parameters. The model used in this study was a computer-simulated reconstruction of a mouse torso comprising five regions (liver, heart, lung, bone and background tissues), and relative 2-norm errors only increased 0.74% when a heterogeneous reconstruction model was used. The averaged optical property values in this model were taken from the literature. However using a homogeneous model, errors increased 4.8%. These errors were defined as 2-norm errors that accounted for the entire image and were relative to the error associated with the exact model; consequently, they do not correspond exactly to our quantification errors. Our results correspond to a model that accounted for two organs with distinct scattering (reduced scattering in the range 0.65 to 2.34 mm$^{-1}$ at 700 nm). Larger errors should be expected when the biological tissue diversity in real animal experiments is taken into account.

From the profiles drawn on the reconstructed images for the heterogeneous slab phantom and for the mouse model phantom, it seems that positional accuracy is not much affected by optical heterogeneity. In contrast, quantification is highly affected by scattering heterogeneity.

The heterogeneous phantoms used in the present study emulate quantification in two organs with different scattering coefficients. Hence, these results cannot be generalized to other arrangements, such as quantification in the presence of organs with a distinct scattering coefficient. Further studies, especially in vivo validations, would help to establish the limits of use for normalized data in preclinical studies.

In conclusion, the present study makes use of theoretical results for homogeneous media and computer simulations for validation.
heterogeneous models to demonstrate a) that inaccuracies in modeling the scattering parameter may lead to noticeable quantification errors when using a normalized data reconstruction approach and b) that more sophisticated models of the forward problem yield much better results.

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