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Abstract. Multifrequency (0 to 0.3 mm−1), multiwavelength (633, 680, 720, 800, and 820 nm) spatial frequency domain imaging (SFDI) of 5-aminolevulinic acid-induced protoporphyrin IX (PpIX) was used to recover absorption, scattering, and fluorescence properties of glioblastoma multiforme spheroids in tissue-simulating phantoms and in vivo in a mouse model. Three-dimensional tomographic reconstructions of the frequency-dependent remitted light localized the depths of the spheroids within 500 μm, and the total amount of PpIX in the reconstructed images was constant to within 30% when spheroid depth was varied. In vivo tumor-to-normal contrast was greater than ∼1.5 in reduced scattering coefficient for all wavelengths and was ∼1.3 for the tissue concentration of deoxyhemoglobin (ctHb). The study demonstrates the feasibility of SFDI for providing enhanced image guidance during surgical resection of brain tumors.

Keywords: fluorescence; Fourier transforms; tomography.

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1 Introduction
Fluorescence guidance in neurosurgery using 5-aminolevulinic acid (ALA)-induced protoporphyrin IX (PpIX) is gaining popularity as a surgical adjunct to improve the extent of tumor resection.1–3 A randomized controlled phase III clinical trial by Stummer and colleagues has shown greater extent of resection and significantly higher 6-month progression-free survival in patients with high-grade glioma (GBM) undergoing fluorescence guidance compared to conventional white-light-guided surgery.4,5 Methods to quantify PpIX concentration in tissue using a fiber-optic “point” probe and a spectrally constrained model of light propagation have been shown to improve tumor detection with enhanced accuracy (87%) compared to the subjective visual fluorescence assessment (66%) used in the Stummer studies.6,7 To better integrate this concept into surgical guidance, this work explores the development of a model-based wide-field fluorescence imaging method integrated into a surgical microscope. Our goal is to develop quantitative, depth-resolved imaging for fluorescence guidance during neurosurgery.

To date, ALA-induced PpIX fluorescence guidance has been based on violet/blue light to excite the Soret-band of the molecule (around λ ~ 405 nm), which limits the penetration of the probing signal to a few hundred microns at best, potentially leaving residual tumor undetected when located below the surgical surface. Fortunately, the excitation spectrum of PpIX also possesses several longer wavelength excitation bands up to around λ ~ 633 nm (Q-bands) and a broad emission spectrum extending up to ~720 nm that could be used to stimulate (and detect) emissions from subsurface tumor. Despite the potential benefits during surgery, efforts to detect fluorescence at depth using model-based techniques are still in an early stage of development with promising preliminary results reported by Kim et al.8 and Leblond et al.9

Spatial frequency domain imaging (SFDI) is a relatively new method that uses red or near-infrared light together with CCD detection to acquire wide-field images of tissue. Sinusoidal patterns of varying spatial frequencies are projected onto tissue, and the attenuation in the amplitude of the spatial patterns at each CCD pixel is fitted to a model of light propagation in order to form images of the tissue optical properties.10,11 SFDI has recently been used to quantify PpIX concentrations in vivo by accounting for tissue optical properties.12 Spatially modulated excitation also encodes the depth from which measured optical signals originate because high spatial frequency patterns penetrate less deeply into tissue than their lower frequency counterparts because of multiple light scattering. This frequency encoding enables depth information to be recovered through three-dimensional (3-D) optical tomography.13

Most SFDI research has focused on generating two-dimensional (2-D) images by fitting the reflectance measure at each pixel of a CCD array to a homogeneous model of light propagation. With the approach, SFDI has been applied to quantitatively image stroke,14 brain injury,15 cortical spreading depression,16 and layered structures in skin.17 SFDI techniques have also been used to localize subsurface inhomogeneities,18 and to determine tissue optical absorption and scattering properties over a...
broad spectral range. Recently, several groups have reconstructed 3-D tomographic images using sinusoidal illumination patterns, but these studies were limited to measuring the optical properties of tissue simulating phantoms and did not include in vivo or fluorescence measurements. While we have performed fluorescence SFDI, these preliminary studies did not involve tomographic image reconstruction and were limited to phantoms and/or skin. Likewise, while fluorescence optical tomography with wide-field illumination has been reported for phantom studies and whole-body mouse imaging, it has not been reported either for spatial frequency domain illumination or brain imaging.

In this study we implement, for the first time to our knowledge, fluorescence optical tomography using spatial frequency domain illumination. We then demonstrate the ability of SFDI to image GBM tumors based on endogenous contrast from hemoglobin and the tissue-scattering parameters as well as exogenous contrast from ALA-induced PpIX fluorescence. We present experiments in tissue simulating phantoms with embedded GBM spheroids, and in an in vivo GBM mouse model. We use 2-D SFDI to image endogenous optical properties and observe that the reduced scattering coefficient ($\mu_s'$) exhibits significant contrast between normal brain and tumor. We then successfully implement 3-D SFDI fluorescence tomography to localize tumors through ALA-induced PpIX fluorescence.

2 Methods

2.1 Sample Preparation

The GBM spheroids used in the phantoms were grown from grade IV GBM cells (ACBT) to tumor spheroids of ~1 to 2 mm diameter according to established methods. Selected spheroids were then positioned in tissue simulating phantoms consisting of 85% water, 10% PBS, 5% of a fat emulsion to simulate tissue scattering (Liposyn® 20%) resulting in a 1% final concentration of fat particles, and gelatine (10 g/100 mL). Porcine erythrocytes were added to simulate tissue scattering (Liposyn® 20%) resulting in a 1% final concentration of fat particles, and gelatine (10 g/100 mL). Porcine erythrocytes were added to simulate tissue scattering (Liposyn® 20%) resulting in a 1% final concentration of fat particles, and gelatine (10 g/100 mL). Porcine erythrocytes were added to simulate tissue scattering (Liposyn® 20%) resulting in a 1% final concentration of fat particles, and gelatine (10 g/100 mL). 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We reconstructed 3-D images of the concentration of PpIX by adapting our previously published method for tomographic imaging of absorption using SFDI. For continuous wave illumination, the measured emission light at position \( \rho_d \) due to illumination wavenumber \( k \) is related to the fluorophore concentration by

\[
I_{\text{tot}}(k, \rho_d) = \int d^3r I_{\text{ex}}(k, r) G_{\text{em}}(r, \rho_d) \varepsilon(\lambda_{\text{ex}}) \eta(\lambda_{\text{ex}}, \lambda_{\text{em}}) C(r),
\]

(3)

where \( I_{\text{ex}}(k, r) \) is the photon density of excitation light in the medium, \( \varepsilon(\lambda_{\text{ex}}) \) is the extinction coefficient of the fluorophore at the excitation wavelength, \( \eta(\lambda_{\text{ex}}, \lambda_{\text{em}}) \) is the fluorescence quantum yield (<1% for PpIX), and \( C(r) \) is the fluorophore concentration. The Green’s function \( G_{\text{em}}(r, \rho_d) \) obeys the diffusion equation at the emission wavelength

\[
[-\nabla \cdot D' \nabla + c \mu_a] G(r, r') = \delta(r - r'),
\]

(4)

subject to the boundary condition

\[
G(r, r') + \hat{n} \cdot \nabla G(r, r') = 0.
\]

(5)

Here \( D = 3/c \ell s \) is the diffusion coefficient, \( \ell \) is the extrapolation length, and \( \hat{n} \) is the outward-pointing unit normal. For the semi-infinite geometry, the Green’s function can be decomposed into plane waves\(^{28}\) according to

\[
G(r, r') = \frac{1}{(2\pi)^2} \int d^2q g(q, z, z') \exp \{i(q \cdot (\rho' - \rho))\},
\]

(6)

where

\[
g(q, z, z') = \frac{\ell}{D} \frac{\exp[-Q(q)|z - z'|]}{Q(q)\ell + 1},
\]

(7)

when either \( z \) or \( z' \) is located on the surface, and

\[
Q(q) = (c \mu_a/D + q^2)^{1/2}.
\]

(8)

Equations (6) and (7) allow us to represent the photon density of the excitation light in the tissue appearing in Eq. (3) as

\[
I_{\text{ex}}(k, r) = g_{\text{ex}}(k, 0, z) \exp[k \cdot \rho].
\]

(9)

Substituting Eqs. (6), (7), and (9) into Eq. (3) and taking a Fourier transform with respect to the detector positions on the surface result in

\[
I_{\text{tot}}(k, q_d) = \varepsilon(\lambda_{\text{ex}}) \eta(\lambda_{\text{ex}}, \lambda_{\text{em}}) \int dz \kappa(k, q_d, z)
\times \int d^2\rho \exp[i(q \cdot \rho)C(r)],
\]

(10)

where

\[
\kappa(k, q_d, z) = \frac{\ell_q}{D_{\text{ex}} D_{\text{em}} [Q_{\text{ex}}(k)\ell_q + 1][Q_{\text{em}}(q_d)\ell_{\text{em}} + 1]} \exp[-Q_{\text{ex}}(k)Q_{\text{em}}(q_d)|z|].
\]

(11)

and

\[
I_{\text{tot}}(k, q_d) = \int d^2\rho \exp[i(q_d \cdot \rho)I_{\text{tot}}(k, \rho_d)].
\]

(12)

The inverse of Eq. (10) is used to calculate the fluorophore distribution. It appears as

\[
C(r) = \frac{1}{\varepsilon(\lambda_{\text{ex}}) \eta(\lambda_{\text{ex}}, \lambda_{\text{em}})} \int d^2q \exp(-i\mathbf{q} \cdot \rho)
\times \sum_{m,n} \kappa^*(m, q_d, z) M_{m,n}^{-1}(q) I_{\text{tot}}(n, q_d),
\]

(13)

where

\[
M_{m,n}(q) = \int_0^\infty dz \kappa^*(m, q_d, z) \kappa^*(n, q_d, z).
\]

(14)

In these formulas, the Fourier variable \( q \) is related to the projection wave number \( k \) and the Fourier variable conjugate to the detector locations \( q_d \) by \( q = k + q_d \). The subscripts ex and em indicate whether the optical properties (\( \mu_a \) and \( \mu'_a \)) for the respective variable refer to the excitation or emission wavelength (~633 nm and 680 nm, respectively). The average optical properties at 633 nm (excitation) and 680 nm (emission) obtained from the endogenous SFDI were used as inputs in Eq. (5) for the tomographic fluorescence reconstructions. The inverses of the \( M \) matrices [see Eq. (14)] are needed in Eq. (13). We take the inverse of each of these matrices by solving its eigenvalue problem and setting all eigenvalues below a threshold to zero. For this study the threshold was kept constant at \( 10^{-4} \). We solved for the fluorophore

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**Fig. 2** Flowchart of the three-dimensional fluorescence optical tomography algorithm.
concentration over a 50 × 50 grid of points in the Fourier domain with $q_{\text{max}} = 2.5\pi \text{ mm}^{-1}$ and $\delta q = 0.1\pi \text{ mm}^{-1}$. Image reconstruction required only $\sim 10$ s on a single 2.80-GHz processor. A flowchart of the tomographic reconstruction algorithm appears in Fig. 2.

### 3 Results

As an initial test of the ability to image relative PpIX concentrations in turbid media, two GBM spheroid phantoms were evaluated. In the first phantom the spheroid was located with its top surface level with the surface of the phantom, whereas...
in the second phantom the spheroid was embedded such that its
top surface was located ∼1.5 mm beneath the phantom. After imaging, the phantom containing the subsurface spheroid was cut in half to confirm its location [Fig. 3(a)]. In the reconstructed images, the maximum PpIX concentration of the superficial spheroid [Fig. 3(b)] was reconstructed at a depth of 1 mm, while the maximum for the subsurface spheroid [Fig. 3(c)] was at a depth of 2 mm. Although both spheroids were approximately the same size (∼1.5 mm), the image of the superficial spheroid had a lateral FWHM of 2.4 mm compared to 4.7 mm for the deeper spheroid. The volume of contrast enhancement in the reconstructed image with concentration greater than 50% of the maximum was also less for the superficial (13.7 mm³) than for the deeper (66.8 mm³) spheroid. However, the reconstructed maximum concentration of the superficial spheroid was 3.6 times that of the deeper spheroid, such that the estimated total amount of PpIX in the two images agreed within 30%.

Table 1 Measured in vivo values in the mouse brain for tumor and normal tissue (Shown are the means and standard deviations of pixel values in the tumor and normal regions of interest).

<table>
<thead>
<tr>
<th></th>
<th>Skull intact</th>
<th></th>
<th>Skull removed</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Tumor</td>
<td>Normal</td>
<td>Tumor</td>
<td>Normal</td>
</tr>
<tr>
<td>(\mu') [(\lambda = 680 \text{ nm})] [mm(^{-1})]</td>
<td>2.2 ± 0.2</td>
<td>1.3 ± 0.1</td>
<td>2.1 ± 0.4</td>
<td>1.2 ± 0.1</td>
</tr>
<tr>
<td>A</td>
<td>1.8 ± 0.08</td>
<td>1.1 ± 0.05</td>
<td>1.8 ± 0.2</td>
<td>1.1 ± 0.05</td>
</tr>
<tr>
<td>b</td>
<td>1.4 ± 0.5</td>
<td>1.2 ± 0.4</td>
<td>1.3 ± 0.6</td>
<td>1.0 ± 0.5</td>
</tr>
<tr>
<td>ctHb ((\mu\text{M}))</td>
<td>43 ± 5</td>
<td>33 ± 3</td>
<td>47 ± 6</td>
<td>36 ± 5</td>
</tr>
<tr>
<td>ctHbO2 ((\mu\text{M}))</td>
<td>52 ± 9</td>
<td>52 ± 6</td>
<td>60 ± 8</td>
<td>60 ± 9</td>
</tr>
</tbody>
</table>

Figure 5(a) shows a picture of the skull illuminated with spatially modulated 633-nm light \(f = 0.25 \text{ mm}^{-1}\). The dotted red line denotes the region of interest used in the fluorescence image reconstruction. In the reconstructed images of PpIX concentration, the tumor appears slightly deeper with an intact skull [Fig. 5(b)] than with the skull removed [Fig. 5(c)], with the maximum intensity occurring 1.5 mm and 1.0 mm below the surface, respectively (actual skull thickness ∼0.5 mm). The presence of the skull decreased the maximum concentration in the image by 8%, increased the FWHM from 1.1 to 2.1 mm, and increased the apparent tumor volume from 4.5 to 5.5 mm³. Similar to the phantom experiments, the effects of increased size and decreased concentration offset each other such that the total estimated amount of PpIX remained constant to within 10%.

4 Discussion

One feature common to all of the reconstructions in this study is that deeper objects appear larger and with less contrast than superficial ones. This is the result of a partial volume effect that occurs because the interrogation volume of each measurement is larger than the features we are trying to reconstruct. As a result, the reconstructed features have values somewhere between those of their actual values and the background. The effect is more pronounced for deeper structures because the sensitivity functions are peaked near the surface and become broader with depth. We have analyzed the partial volume effect for SFDI in a previous publication where we showed that tomographic image reconstruction ameliorated (but did not eliminate) this effect in comparison to our 2-D imaging methods.13
We do not expect this partial volume effect to prevent the use of SFDTI tomography for neurosurgical resection of tumors. Fluorescence-assisted surgical resection is guided by the qualitative presence or absence of fluorescent signal. Thus, the exact peak concentration is of less importance than a measurement of the fluorophore depth or the total amount of fluorophore present. Knowledge of fluorophore depth could improve the efficiency of surgical resection by allowing a surgeon to more confidently dissect down to tumor below the cortical surface; then proceed with more aggressive tumor resection while far from the deep margin; and finally convert to slower, more meticulous removal when the deep interface is near. The increase in size of the reconstructions with depth is expected to interfere with accurate estimates of deep tumor extent. However, as tumor can only be removed from the surface of the operative field, the higher volume estimates at depth are of less concern because they are expected to decrease as tumor is removed and overall depth is reduced.

It may be possible to obtain more accurate reconstructions by using algorithms that allow for inhomogeneous tissue optical properties. Such algorithms could be implemented with finite-element or Monte Carlo methods to model light transport. However, one of the main advantages of our reconstruction method is its speed. This speed is necessary to make surgical applications feasible. The current reconstructions use $10^6$ data points and take only $\sim 10$ s. In addition, they could be made faster by parallel computing, by pre-calculating the eigenvalues/functions of the M matrices, or by reconstructing over a reduced bandwidth. We are able to attain this speed because we use homogeneous tissue optical properties and a planar geometry to model light transport. These assumptions make the Green’s function for the diffusion equation translationally invariant and enable us to write Eq. (3) in block diagonal form [see Eq. (10)] such that it can be solved rapidly. Thus while we plan to explore the potential benefits of modeling light transport through inhomogeneous tissue, we expect that for applications requiring rapid feedback (e.g., surgery), the algorithm proposed here will be better suited.

5 Conclusion

In summary, this proof-of-principle study demonstrates the ability of multiwavelength, multifrequency SFDTI tomography to localize small (~mm) subsurface (~2 to 3 mm deep) tumors in tissue-simulating phantoms and in vivo using ALA-induced PpIX fluorescence. These results suggest that SFDTI may offer a wide-field fluorescence imaging approach that accounts for intrinsic tissue optical properties, characterizes subsurface nests of tumor cells, and potentially provides contrast for improved intraoperative neurosurgical guidance. Future studies will provide more detailed insight into factors that limit SFDTI accuracy and performance, and explore the combination of image guidance with probe-based methods.6,8,9

Acknowledgments

We thank Chung-Ho Sun for preparation of the GBM spheroids, G. Granger for his gift of the ACBT GBM cell line, Yi-hong Zhou for use of the nude mouse glioma model, and Abhishek Chaturbedi for preparation of this model. Support for this work was provided to University of California–Irvine by the National Institutes of Health (NIH) NCCR/NIBIB Laser Microbeam and Medical Program (LAMMP, P41-EB015890), the Military Photomedicine Program (AFOSR Grant # FA9550-08-1-0384), and the Beckman Foundation. Support to Dartmouth College/University of Toronto was provided through NIH grants RO1-NS052274-01A2 and K25-CA138578. Soren D. Konecky was supported by a fellowship from the Hewitt Foundation for Medical Research.

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