Structured illumination enhances resolution and contrast in thick tissue fluorescence imaging

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Abstract. We introduce a noncontact imaging method utilizing multifrequency structured illumination for improving lateral and axial resolution and contrast of fluorescent molecular probes in thick, multiple-scattering tissue phantoms. The method can be implemented rapidly using a spatial light modulator and a simple image demodulation scheme similar to structured light microscopy in the defraction regime. However, imaging is performed in the multiple-scattering regime utilizing spatially modulated scalar photon density waves. We demonstrate that by increasing the structured light spatial frequency, fluorescence from deeper structures is suppressed and signals from more superficial objects enhanced. By measuring the spatial frequency dependence of fluorescence, background can be reduced by localizing the signal to a buried fluorescent object. Overall, signal-to-background ratio (SBR) and resolution improvements are dependent on spatial frequency and object depth/dimension with as much as sevenfold improvement in SBR and 33% improvement in resolution for ~1-mm objects buried 3 mm below the surface in tissue-like media with fluorescent background. © 2010 Society of Photo-Optical Instrumentation Engineers. [DOI: 10.1117/1.3399321]

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NIR excitation and subsequent emission can significantly improve imaging depth compared to visible fluorescent probes. Despite this advantage, accurate mapping of subsurface (i.e., mm to cm) fluorescence with high spatial resolution remains a critical challenge. This is due to the fact that lateral and axial localization of fluorescence is “blurred” by multiple light scattering. In order to address this limitation, imaging systems that utilize multiple detectors, source scanning, and time/frequency domain techniques have been successfully implemented based on diffuse optical tomography principles.

In this work, we demonstrate a simple alternative nonscanning method utilizing multifrequency structured illumination patterns to improve axial and transverse resolution of buried fluorescent inhomogeneities. Our approach utilizes similar principles to technologies employing structured illumination developed for the diffraction regime. However, in this case, lateral and axial resolution improvements are obtained from thick, turbid materials using principles of scalar photon density waves propagating in the diffusion regime.

A spatial light modulator is introduced at the source, and a number of sinusoidal illumination patterns with different spatial frequencies, \( k \), are projected over a tissue sample at three offset phases (typically 0 deg, 120 deg, and 240 deg). During data analysis, the images at each phase, \( I_{1,2,3} \), are demodulated using Eq. (1) in order to analyze the AC component of the remitted light. This concept has also been applied by Neil et al. for the purposes of optical sectioning microscopy.

\[
M_{AC}(x_1)_{I_{\mu}} = \frac{2}{3} \left[ (I_1 - I_2)^2 + (I_2 - I_3)^2 + (I_3 - I_1)^2 \right]^{1/2}. 
\]

In the spatial frequency domain, the reflected AC component of the spatially modulated waves can be used to characterize the modulation transfer function (MTF) of the sample. We now extend this concept to fluorescence imaging where structured light illumination specifically impacts the excitation fluorescence rate. A forward fluorescence model for spatially modulated light—specifically, sinusoidal waves—can be adapted from a 1-D second-order Helmholtz equation in order to calculate fluorescence rate (\( \varphi \)) as a function of depth, \( z \):

\[
\frac{d^2}{dz^2} \varphi_0(z) - \mu_\text{eff}^2 \varphi_0(z) = -3 \mu_a \varphi_0(z).
\]

In these equations, \( q \) is the source, \( \mu_a = (\mu_s + \mu_r) \) is the transport coefficient, \( \mu_s = \mu_s(1 - g) \) is the absorption coefficient, \( \mu_r = \mu_r/k^2 \) is the reduced scattering coefficient, \( q \) is the mean cosine of the scattering angle (\( \cos \theta \)), and \( k \) is the spatial frequency of the illumination pattern, \( \mu_\text{eff} = [3\mu_a + \mu_r]^{1/2} \), and \( \mu_\text{eff} = [\mu_r(1 - g)]^{1/2} \). The solution for this equation is:

\[
\varphi_0(z) = \left( \frac{3P_{out}}{\mu_\text{eff}^2 \mu_a} \right) \exp(-\mu_a z) + C \exp(-\mu_\text{eff} z),
\]

where \( a' \) is the reduced albedo (\( \mu_s'/\mu_a' \)), is the incident optical power, and \( C \) is a constant determined by the choice of a boundary condition. A normalized fluence rate profile for

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three frequencies (Fig. 1) is plotted as a function of depth for a given set of optical properties ($\mu_a=0.005 \text{ mm}^{-1}$, $\mu'_a=0.8 \text{ mm}^{-1}$). Tissue behaves as a low-pass spatial filter as the normalized excitation fluence rate decays more rapidly at higher illumination frequencies, allowing depth-sensitive excitation of buried fluorescence inclusions.

For experiments, structured NIR light is projected onto the sample using a modified digital light projector (NEC HT1000). The micromirrors allow generation of desired patterns as shown in other applications. For fluorescence imaging, the color’s filter wheel was removed and an interference bandpass filter ($\lambda=660 \text{ nm}$, $\delta=10 \text{ nm}$ FWHM) was placed at the source, and a second bandpass filter ($\lambda=720 \text{ nm}$, $\delta=10 \text{ nm}$ FWHM) was placed at the CCD camera (Roper QuantEM: 512SC). Cross-linear polarizers were also placed at the source and detector to eliminate specular signals in the case of reflectance imaging, although they are not necessary for fluorescence.

In order to explore axial and lateral resolution, respectively, imaging was performed on two polydimethylsiloxane (PDMS) phantoms with embedded channels and tissue-like optical properties ($\mu_a=0.005 \text{ mm}^{-1}$, $\mu'_a=0.8 \text{ mm}^{-1}$). Phantom 1 contained four 1-mm-diam tube inclusions, each separated laterally by 1 cm at depths of 1 mm, 2 mm, 3 mm, and 4 mm, respectively [Video 1(a)]. The fluorophore injected into the phantom tubes consisted of a mixture of 250 nM IRDye 700DX dye (LyCOR, Omaha, Nebraska) dissolved in a solution of 0.8% Intralipid diluted with deionized water in order to match background scattering. Imaging was performed on each phantom using 16 evenly spaced illumination frequencies ranging from 0 $\text{ mm}^{-1}$ to 0.3 $\text{ mm}^{-1}$ [Video 1(b)]. Sinusoidal illumination patterns were projected with frequency, $k$, oriented in the horizontal ($x$) direction. Three phases (0 deg, 120 deg, 240 deg) were acquired for each frequency at constant exposure times (typically 500 ms), and images were demodulated using Eq. (1). Demodulated images were calibrated by dividing by a reference unstructured planar (i.e., 0 $\text{ mm}^{-1}$) illumination image obtained from a gelatin phantom with evenly distributed fluorophore [Video 1(c)]. Video 1 illustrates the entire image acquisition and demodulation process and as predicted by the forward fluence model, the fluorescence contribution from deeper tubes diminishes as the illumination frequency increases due to attenuation of longer path length excitation photons. Horizontal line profiles of these images highlight the frequency-dependent depth sensitivity, revealing only the most superficial tubes at the higher spatial frequency [Video 1(d)].

Imaging was performed in the same manner on Phantom 2, consisting of two 2-mm-diam tubes separated by 5 mm and positioned 4 mm below the surface [Fig. 2(a)]. Calibrated and demodulated fluorescence images [Fig. 2(b)] illustrate that planar unstructured illumination (0 $\text{ mm}^{-1}$) yields an image of a single tube at the surface, while an increase in spatial frequency (0.1 $\text{ mm}^{-1}$) produces two optimally resolved tubes [Fig. 2(c)]. The resolution improvement seen at 0.1 $\text{ mm}^{-1}$ is lost at 0.18 $\text{ mm}^{-1}$ due to suppression of long path length excitation photons resulting in an interrogation region predominantly above the buried tubes. Thus, although increasing spa-
using a simple demodulation scheme that effectively provides spatial frequency “gating” due to the frequency-dependent penetration depth of the excitation fluence. This method improves localization and SBR compared to planar imaging techniques and is amenable to the development of further tomographic algorithms. Potential applications of this approach include imaging of small animal models as well as imaging subsurface structures in clinical settings.

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References