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Measurements of wavelength dependent scattering and backscattering coefficients by low-coherence spectroscopy

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Abstract. Quantitative measurements of scattering properties are invaluable for optical techniques in medicine. However, noninvasive, quantitative measurements of scattering properties over a large wavelength range remain challenging. We introduce low-coherence spectroscopy as a noninvasive method to locally and simultaneously measure scattering μs and backscattering μb coefficients from 480 to 700 nm with 8 nm spectral resolution. The method is tested on media with varying scattering properties (μs = 1 to 34 mm−1 and μb = 2.10−6 to 2.10−3 mm−1), containing different sized polystyrene spheres. The results are in excellent agreement with Mie theory. © 2011 Society of Photo-Optical Instrumentation Engineers (SPIE). [DOI: 10.1117/1.3553005]

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Quantitative determination of the optical properties of tissue is invaluable in biomedical optics. The majority of optical diagnostic techniques rely on the spectral absorption and scattering properties of tissue, which provide information on its composition and structure. The same optical properties are of essential importance for the development and optimization of optical therapeutic techniques. However, despite the existence of many spectroscopic methods, it is still a challenge to do noninvasive, quantitative measurements of the absorption and scattering properties in vivo over a large wavelength range.

Recently, we introduced low-coherence spectroscopy (LCS) to do quantitative and localized measurements of absorption coefficients μa over a wavelength range of 480 to 700 nm with a spectral resolution of 8 nm (all wavelength dependent parameters in this paper will be denoted by a boldfaced character). In this study, we use LCS to quantitatively and simultaneously measure scattering μs and backscattering μb coefficients on a wide range of scattering media (μs = 1 to 34 mm−1 and μb = 2.10−6 to 2.10−3 mm−1). Thereby, we demonstrate new opportunities for noninvasive scattering property measurements. In vivo measurements of the quantitative value of μs and μb can assist in differentiating between tissue types and modeling of light-tissue interactions. The spectrally resolved information of μs and μb gives additional valuable information such as the power dependency of μb on wavelength and wavelength dependent oscillations in μb, which have shown to be related to tissue morphology.

Whereas extensive study on tissue (back)scattering has been performed in the areas of light scattering spectroscopy and angle-resolved low-coherence interferometry, these studies lack quantification of μs and μb, since their primary aim has been to retrieve the size of the scattering particles. Quantification of μs and μb has been shown in optical coherence tomography studies, but these studies were limited to the measurement of μs and μb averaged over the bandwidth of the spectrum, i.e., no spectral information was obtained. Moreover, in these studies, quantitative agreement with theory is rarely obtained for highly scattering media, due to multiple scattering contributions to the signal. Other (diffuse) reflectance spectroscopy techniques are able to measure μs and the reduced scattering coefficient μ′s, but this requires additional information on the scattering anisotropy g to obtain μb. Thus, compared to the existing methods for scattering property measurements, LCS offers the unique possibility for a combination of simultaneous, quantitative, and spectrally resolved measurement of μs and μb. Therefore, these measurements will assist in a more complete, and likely more accurate, characterization of the tissue of interest. In addition, like other low coherence interferometry techniques, LCS measures a controlled and confined volume, which is important when measuring local optical properties in an inhomogeneous tissue.

Using LCS, we measured μb and μs of aqueous nonabsorbing suspensions of different sized polystyrene spheres and validated our results with Mie theory. Therefore, we measured backscattered power spectra S(ℓ) at controlled geometrical path lengths ℓ of the light in a sample. Our LCS system, which is described in detail in Ref. 1, consists of a Michelson interferometer and is optimized for 480 to 700 nm. The geometrical round trip path length ℓ (ℓ = 0 to 2 mm, with ℓ = 0 the sample surface) is controlled by translating the reference mirror, in steps of 27 μm. By translating the sample, focus tracking of the 64 μm spot size in the sample is achieved. Around ℓ, the signal is modulated by scanning the piezo-driven reference mirror (23 Hz) resulting in a scanning window of Δℓ ≈ 44 μm. The optical power at the sample is 6 mW.

A multimode fiber (Ø = 62.5 μm) guides the reflected light from both arms to a photodiode. Signal processing after acquisition, which is described in detail in Ref. 1, results in averaged spectra S(ℓ) with 8-nm resolution (≈500 averages per ℓ, to avoid any spectral modulations on S(ℓ) caused by interference between scattering particles). We describe S(ℓ) with a single exponential decay model (Ref. 2) S(ℓ) = S0 · T · Δℓ · μb,NA · exp(−μb · ℓ), where S0 is the source...
power spectrum and $T$ is the system coupling efficiency. When $S(\ell)$ is dominated by a single backscattered light, $\mu_s$ is the attenuation coefficient of the sample and $\mu_b$ equals $\mu_s$ for nonabsorbing samples (this study). The system dependent parameters will be denoted by $S = S_0 \cdot T \cdot \Delta \ell$. The spectra $S(\ell)$ are collected over the detection numerical aperture (NA) of the system, therefore, we define the measured backscattering coefficient $\mu_{b, NA}$ as the product of $\mu_b$ and the phase function $p(\theta)$, integrated over the solid angle of the NA in the medium:

$$\mu_{b, NA} = \mu_b \cdot 2\pi \int_{\theta = \frac{x}{NA}}^{\pi} p(\theta) \cdot \sin(\theta) \cdot d\theta.$$  \hspace{1cm} (1)

We measured the wavelength dependent point spread function in the medium and derived the NA (ranging from 0.035 to 0.045 between 480 to 700 nm) from the resulting Rayleigh length of the system. The terms $\xi$, $\mu_{b, NA}$, and $\mu_s$ are obtained by fitting a two-parameter (amplitude and decay, respectively) exponential function to $S(\ell)$ versus $\ell$. Uncertainties are estimated by the 95% confidence intervals (c.i.) of the fitted parameters. $\xi$ is determined in a separate calibration measurement in which $\mu_{b, NA}$ can be calculated using the fitted amplitude $\xi$ and $\mu_{b, NA}$, if $\xi$ is determined in a separate calibration measurement in which $\mu_{b, NA}$ is exactly known from Mie theory and Eq. (1). To this end, we used National Institute of Standards and Technology (NIST)-certified polystyrene spheres of $\phi = 409 \pm 9$ nm (diameter $\pm$ SD, Thermo Scientific, USA). The obtained $\xi$ was used to determine $\mu_{b, NA}$ in subsequent measurements.

In our Mie calculations, we used wavelength dependent refractive indices of water and polystyrene and integrated over the size distribution of the spheres (2"SD), given by the manufacturer. Brownian motion of the polystyrene spheres causes Doppler broadening of the measured LCS spectra. For adequate comparison, we convolved the Mie spectra with a Lorentzian, with a linewidth of 5 to 13 nm, depending on the size-dependent Doppler frequency distribution of the Brownian motion of the spheres, similar to our analysis in Ref. 1.

Figure 1(a) shows LCS measurements (dots) of $\mu_b$ for four aqueous suspensions of different sized NIST-certified polystyrene spheres: 0.071% with $\phi = 409 \pm 9$ nm, 0.048% with $\phi = 602 \pm 6$ nm, 0.038% with $\phi = 799 \pm 9$ nm, and 0.033% with $\phi = 1004 \pm 10$ nm, which lie within the range of scattered sizes in biological cells. $\xi$ are in agreement with Mie theory (thick solid lines), showing the characteristic size dependent oscillations. $\mu_{b, NA}$ are in agreement with Mie theory (thick solid lines), showing the characteristic size dependent oscillations. $\mu_{b, NA}$ is in agreement with Mie theory [(Fig.2(b)], except for the two highest volume concentrations, where the measurement overestimates $\mu_{b, NA}$ at the shorter wavelengths. The measurements of $\mu_b$ in Figs. 1(a) and 2(a) demonstrate that disagreement with the Mie calculated values for the highest volume concentrations (Fig. 2) is only manifested in $\mu_{b, NA}$ and not in $\mu_b$ (i.e., $\mu_b$ agrees with the Mie calculated $\mu_b$ within the 95% c.i.). For these samples (0.533% and 0.950%), the average surface-to-surface distance between the spheres is comparable to the wavelength: 760 and 556 nm, respectively. Since the effect of multiple scattering would be visible in the measured value of both coefficients, we speculate that another effect may cause this disagreement, i.e., the total scattered field cannot be treated as the superposition of the scattered field by the individual particles (dependent scattering). Our results indicate that for these sphere concentrations, $\mu_{b, NA}$ is altered to favor more backward than forward directed scattering. Further study is needed to assess the influence of the particle phase function and interparticle distance on the measured $\mu_b$ and $\mu_{b, NA}$.1
In conclusion, we present quantitative and wavelength dependent measurements of scattering and backscattering coefficients from polystyrene sphere suspensions. Our method applies for a broad range of sphere sizes and particle densities, and is in excellent agreement with Mie theory up to scattering coefficients as high as 34 \( \text{mm}^{-1} \). LCS measures \( \mu_s \) and \( \mu_b \) simultaneously, over a large wavelength range and with good spectral resolution. The combined wavelength dependent information of \( \mu_s \) and \( \mu_b \) is likely to assist in more accurate tissue characterization in tissue optics.

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References