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Bosschaart, N.; Faber, D.J.; van Leeuwen, T.G.; Aalders, M.C.G.

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Nienke Bosschaart
Dirk J. Faber
Ton G. van Leeuwen
Maurice C. G. Aalders
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Nienke Bosschaart, a, Dirk J. Faber, a,b Ton G. van Leeuwen, a,c and Maurice C. G. Aalders a

a University of Amsterdam, Academic Medical Center, Department of Biomedical Engineering and Physics, P.O. Box 22700, NL-1100 DE Amsterdam, The Netherlands
b University of Amsterdam, Academic Medical Center, Department of Ophthalmology, P.O. Box 22700, NL-1100 DE Amsterdam, The Netherlands
c University of Twente, Biomedical Photonic Imaging Group, P.O. Box 217, NL-7500 AE Enschede, The Netherlands

Abstract. Localized spectroscopic measurements of optical properties are invaluable for diagnostic applications that involve layered tissue structures, but conventional spectroscopic techniques lack exact control over the size and depth of the probed volume. We show that low-coherence spectroscopy (LCS) overcomes these limitations by measuring local attenuation and absorption coefficient spectra in layered phantoms. In addition, we demonstrate the first in vivo LCS measurements of the human epidermis and dermis only, from the measured absorption in two distinct regions of the dermal microcirculation, we determine total hemoglobin concentration (3.0 ± 0.5 g/l and 7.8 ± 1.2 g/l) and oxygen saturation. © 2011 Society of Photo-Optical Instrumentation Engineers (SPIE). [DOI: 10.1117/1.3644497]

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The derivation of physiological parameters from the spectroscopic determination of tissue optical properties can offer a fast and painless alternative to invasive diagnostic procedures such as tissue biopsies and drawing of blood. For instance, the absorption coefficient of the dermal microcirculation is directly related to the tissue hemoglobin concentration, which provides information on oxygen saturation, blood volume, and potentially the hemoglobin concentration in whole blood. A variety of spectroscopic techniques is available for measuring tissue optical properties. However, these techniques have limited ability to confine their probing volume to embedded structures such as the dermal microcirculation (located beneath the epidermis), or require long photon path lengths (several mm to cm) which exceed the adult dermal thickness (± 0.2 – 1.2 mm). Consequently, many of those techniques rely on assumption-based algorithms to account for layered media. Low-coherence interferometry techniques, such as low-coherence spectroscopy (LCS) and spectroscopic optical coherence tomography (sOCT) do not suffer from this limitation, since they control the size and position of the probed volume from which the optical properties are determined (lateral and in depth)—i.e., they reject the detection of photons that originate from outside the volume of interest.

We recently validated LCS on homogeneous phantoms with controlled optical properties, to quantitatively obtain the attenuation \( \mu_a \), absorption \( \mu_a \), scattering \( \mu_s \), and backscattering \( \mu_b \) of an Intralipid-dye phantom \( \mu_a = 4 \text{ to } 6 \text{ mm}^{-1}, \mu_s = 0 \text{ to } 5 \text{ mm}^{-1} \), covered by light attenuating layers with varying optical densities (0.39 to 0.89). Subsequently, we demonstrate the first in vivo LCS measurements of \( \mu_t \) and \( \mu_a \) of the human epidermis and dermal microcirculation, from which we determine total hemoglobin concentrations and oxygen saturation.

To obtain \( \mu_t \) and \( \mu_a \) from a target volume, we measured backscattered power spectra \( S(\ell) \) at controlled geometrical path lengths \( \ell \) of the light in the medium (path length and depth related parameters are corrected for the refractive index of the medium). Our LCS system, which is described in detail in Ref. 5, is optimized for the wavelength range of 480 to 700 nm. We controlled \( \ell \) by translating the reference mirror in steps of 27 \( \mu \text{m} \). By translating the sample in the axial direction, focus tracking of the spot size (\( \sigma = 5 \mu \text{m} \)) in the medium is achieved. Around \( \ell \), the signal is modulated by scanning the piezo-driven reference mirror (23 Hz), resulting in a scanning window of \( \Delta \ell \approx 44 \mu \text{m} \) in the medium. The optical power at the sample is 6 mW.

A multimode fiber (\( \phi = 62.5 \mu \text{m} \)) guides the reflected light from both arms to a photodiode. Fourier transformation of the acquired time signal results in spectra \( S(\ell) \) with spectral resolution \( \Delta \lambda = \lambda^3/(\ell \Delta \ell) \) (4 nm < \( \Delta \lambda < 9 \text{ nm} \)). To minimize the influence of speckle noise on \( S(\ell) \), we spatially average 90 to 250 spectra by translating the sample and measuring \( S(\ell) \) every 5 \( \mu \text{m} \). Fitting the single exponential decay model \( S(\ell) = A \cdot \exp( - \mu_t \cdot \ell ) \) (free running fit parameters \( A \) and \( \mu_t \)) to the background corrected \( S(\ell) \) versus \( \ell \), results in a \( \mu_t \) spectrum. Uncertainties in \( A \) and \( \mu_t \) are estimated by their 95% confidence intervals (c.i.).

When \( S(\ell) \) is dominated by single backscattered light, the attenuation coefficient \( \mu_t = \mu_s + \mu_b \). Since the dependence of \( \mu_s \) on wavelength can be described by \( a \cdot \lambda^{-b} \), least-squares fitting of \( \mu_t = a \cdot \lambda^{-b} + \sum_i (c_i \cdot \mu_a) \) results in the individual contributions of \( \mu_s \) and \( \mu_a \) to the measured \( \mu_t \). The free running fit parameters \( a, b, \) and \( c_i \) are constraint to positive values. The wavelength dependent \( \mu_a \) are the known absorption spectra (unit: \( \text{mm}^{-1} \) per unit concentration) of the contributing chromophores \( i \) with contribution \( c_i \), which are \( \mu_a \) of the dye for the phantom measurements and \( \mu_a \) of deoxygenized hemoglobin (Hb) and oxygenized hemoglobin (HbO2) for the in vivo skin measurements. The \( \mu_a \) of the dye was obtained from a
transmission measurement. Since we are primarily interested in the total hemoglobin concentration ([tHb] = cHbO2 + cHb) and the oxygen saturation (SO2 = cHbO2/[tHb]) with their uncertainty estimates (±95% c.i.) for the in vivo measurements, we directly fit the [tHb] and the SO2 by substituting cHbO2 = SO2 · [tHb] and cHb = (1−SO2) · [tHb] in the fitting algorithm.

The volume from which we obtain μt is controllable in both size and position inside the medium, by choosing the region for in optical densities (OD) of the layers. Fitted scatter powers to the fits (dotted lines) are minimally affected by the covering layers. Fitted scatter powers indicate the presence of blood and can be related to normal dermal blood volume fractions of 2% and 5%, respectively, when assuming a fixed hemoglobin concentration of 150 g/l for whole blood. The fitted SO2 of 81 ± 34% in Region 2 and 100 ± 31% in Region 3 are also within physiological range.

Presumably, Region 3 encloses a flexure line and Region 2 encloses surrounding skin, since relative differences in hemoglobin absorption up to 63% were found between those two palmar skin regions during stretching, which is consistent with our [tHb] results. This also explains the difference in scattering between the two regions, because tissue homogeneity and organization of collagen fiber content, the major contributor to dermal scattering, differ significantly between these skin regions. The value of the μs-contributions to the measured μs fall within the physiological range of 1 to 100 mm−1, but the actual dermal μs may be underestimated due to the contribution of multiple scattering to the LCS signal. Nevertheless, since absorption takes place along the controlled photon path, this contribution does not influence the determination of μs.

These in vivo measurements show that LCS can be used to measure hemoglobin concentration and oxygenation in the microcirculation. Although no gold standard exists to confirm our in vivo [tHb] and SO2 determinations, their values are convincing biologically and the optical properties from which they were derived are within the range of optical properties that were validated in our phantom study. The accuracies at which we determined [tHb] (~15%) and SO2 (~30%) are influenced by the homogeneity of their distribution within the investigated region, and by the accuracy of the determination of μt. The latter is affected by the size of the investigated region (i.e., the number of spatial averages and the length of the t-interval for fitting the exponential decay model) and the OD of the medium covering this region, which limits the maximum probing depth of LCS to ~0.5 to 1 mm in vivo. The limiting accuracy of the in vivo determination of [tHb] can be expected to be 8%, as found for the dye concentration in the phantoms. Since the epidermal OD (~0.8) is comparable to the OD’s of the layers in our phantom study, the extra inaccuracy of the [tHb] and SO2 determination can be ascribed to the skin’s heterogeneity. Although the size of the investigated region improves accuracy, it negatively affects measurement speed. Faster acquisition can be achieved by optimizing this trade-off, and by investigating the possibility for Fourier domain acquisition. In contrast to time domain
We did not observe any of those influences on the measured $\mu_t$ fits on $\mu_t$ and $\mu_r$ contributions to the $\mu_t$ fits are shown in (a) for the epidermis (Region 1) and (b) for the dermis (Regions 2 and 3). The $\mu_r$ contributions to the $\mu_t$ fits are shown in (c) for Region 3 and (d) Region 2 (note the difference in vertical axis scaling). The selected regions are shown in the OCT image in the upper right corner.

In conclusion, we have shown that we can use LCS to locally obtain absorption coefficient spectra within confined volumes of optically inhomogeneous media. This enabled us to perform the first in vivo LCS measurements of hemoglobin concentration and oxygen saturation inside the dermal microcirculation. By confining the measurement volume to specific tissue structures, LCS overcomes the limitations of conventional spectroscopic techniques. LCS therefore offers a potential alternative to invasive drawing of blood for the determination of whole blood hemoglobin concentration and oxygen saturation.

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References