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Published in:
Journal of Biomedical Optics

DOI:
10.1117/1.3644497

Citation for published version (APA):
Bosschaart, N., Faber, D. J., van Leeuwen, T. G., & Aalders, M. C. G. (2011). In vivo low-coherence spectroscopic measurements of local hemoglobin absorption spectra in human skin. Journal of Biomedical Optics, 16(10), [100504]. https://doi.org/10.1117/1.3644497

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In vivo low-coherence spectroscopic measurements of local hemoglobin absorption spectra in human skin

Nienke Bosschaart
Dirk J. Faber
Ton G. van Leeuwen
Maurice C. G. Aalders
In vivo low-coherence spectroscopic measurements of local hemoglobin absorption spectra in human skin

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

aUniversity of Amsterdam, Academic Medical Center, Department of Biomedical Engineering and Physics, P.O. Box 22700, NL-1100 DE Amsterdam, The Netherlands
bUniversity of Amsterdam, Academic Medical Center, Department of Ophthalmology, P.O. Box 22700, NL-1100 DE Amsterdam, The Netherlands
cUniversity 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 tissue 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]

Keywords: spectroscopy; low-coherence; absorption; attenuation; hemoglobin; human skin.

Paper 11408LR received Jul. 28, 2011; revised manuscript received Sep. 8, 2011; accepted for publication Sep. 9, 2011; published online Oct. 14, 2011.

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 μa, absorption μa, scattering μs, scattering μs, and backscattering μb coefficients between 480 and 700 nm [bold-faced characters denote wavelength (λ) dependent parameters]. In this study we present, for the first time to our knowledge, quantitative measurements of local μs and μb spectra within selected volumes of inhomogeneous turbid media. We validate our method by measuring the dye concentration from the measured μs of an Intralipid-dye phantom (μs = 4 to 6 mm−1, μa = 0 to 5 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 μs and μb of the human epidermis and dermal microcirculation, from which we determine total hemoglobin concentrations and oxygen saturation.

To obtain μs and μb from a target volume, we measured backscattered power spectra S(ℓ) at controlled geometrical path lengths ℓ of the light in the medium (path length and depth related parameters are corrected for the refractive index n 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 ℓ by translating the reference mirror in steps of 27 μm. By translating the sample in the axial direction, focus tracking is optimized for the wavelength range of 480 to 700 nm. We controlled ℓ by translating the reference mirror in steps of 27 μm. By translating the sample in the axial direction, focus tracking is optimized for the wavelength range of 480 to 700 nm.

A multimode fiber (φ = 62.5 μm) guides the reflected light from both arms to a photodiode. Fourier transformation of the acquired time signal results in spectra S(ℓ) with spectral resolution Δλ = λ3/(nΔℓ) (4 nm < Δλ < 9 nm).5 To minimize the influence of speckle noise on S(ℓ), we spatially average 90 to 250 spectra by translating the sample and measuring S(ℓ) every 5 μm. Fitting the single exponential decay model S(ℓ) = A · exp(−μi · ℓ) (free running fit parameters A and μi) to the background corrected S(ℓ) versus ℓ, results in a μi spectrum.6 Uncertainties in A and μi are estimated by their 95% confidence intervals (c.i.).

When S(ℓ) is dominated by single backscattered light, the attenuation coefficient μi = μs + μb. Since the dependence of μb on wavelength can be described by 3 · λ−b, least-squares fitting of μs = a · λ−b + ∑ci (μi · ℓ) results in the individual contributions of μs and μb to the measured μs. The free running fit parameters a, b, and ci are constraint to positive values. The wavelength dependent μs are the known absorption spectra (unit: mm−1 per unit concentration) of the contributing chromophores i with contribution ci, which are μs of the dye for the phantom measurements and μb of deoxygenized hemoglobin (Hb) and oxygenized hemoglobin (HbO2) for the in vivo skin measurements. The μs of the dye was obtained from a...
Fig. 1 Measured μₜ (dots) of a 1% Intralipid-10% dye phantom behind no layer and three TiO₂-silicon layers (1: D = 155 μm, μₑ = 2.5 mm⁻¹, 2: D = 170 μm, μₑ = 5 mm⁻¹; 3: D = 355 μm, μₑ = 2.5 mm⁻¹). Fits on μₛ and μₛ-contributions to the μₓ-fits are shown for all measurements.

Transmission measurement.⁵ 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 μₛ is controllable in both size and position inside the medium, by choosing the region for lateral averaging and the ℓ-interval for fitting the exponential decay model. When measuring on inhomogeneous media such as skin, spatial information is needed to confine our region of interest to, e.g., the epidermal or dermal layer. Therefore, we support our analysis by reconstructing an OCT image from the individual LCS time signals within every ∆ℓ in the axial and lateral direction, using depth scaling d = ℓ/2. The axial resolution of these “fused” OCT images is given by the coherence length of the light source of ~1.5 μm and is therefore higher than the axial resolution of 22 μm for SLD.

Figure 1 shows the measured μₛ (dots) of a medium consisting of 1% Intralipid (Intralipid® 20%, Fresenius Kabi, Germany) and 10% magenta dye (Ecoline #337, Royal Talens, The Netherlands), uncovered and covered by nonabsorbing silicon-titanium dioxide (TiO₂) layers.¹⁰ The three covering layers varied in thickness D and scattering (layer 1: D = 155 μm, μₑ = 2.5 mm⁻¹; layer 2: D = 170 μm, μₑ = 5 mm⁻¹; layer 3: D = 355 μm, μₑ = 2.5 mm⁻¹) resulting in optical densities (OD = D · μₑ) of 0.39 to 0.89. The measurement volume (1250×484 μm², width × depth) from which we acquired μₛ was confined to the Intralipid-dye medium, directly behind the layer-medium interface. μₛ agrees within ~10% of the measured values, indicating that the measurement of μₛ is unaffected by the optical density of the layer covering the medium. Also, the fits on μₛ (solid lines) and the μₛ-contribution to the fits (dotted lines) are minimally affected by the covering layers. Fitted scatter powers b of the medium are 1.8 ± 0.1, 1.8 ± 0.2, 1.6 ± 0.2, 1.1 ± 0.3 for the medium covered by no layer, layer 1, layer 2, and layer 3, respectively. The dye concentration was fitted with 10.2 ± 0.5%, 9.3 ± 0.7%, 9.2 ± 0.9%, and 9.3 ± 0.6%, resulting in a maximum deviation of 0.8% from the expected dye concentration of 10%.

The upper right corner of Fig. 2 shows the OCT image that was reconstructed from an in vivo measurement on the skin of the palmar side of a stretched finger joint that was stabilized with light pressure against a glass slide. Index-matching gel (Euroband Pedicat, Pollak, France) was applied at the glass-tissue interface to minimize specular reflections. We selected regions in the presumed epidermis (Region 1: 715×88 μm²) and dermis (Region 2: 440×418 μm², Region 3: 1080×418 μm²) for obtaining μₑ. The fit on the measured μₑ of the epidermis [Fig. 2(a)] only shows the contribution of scattering (b = 3.5 ± 0.3) and neglects the absorption of hemoglobin ([tHb] = 0 g/l), which agrees with the expected absence of blood vessels in this skin layer.

Within the dermis, we can distinguish two regions with relatively high (Region 2) and low (Region 3) homogeneity. The measured μₑ differ considerably between the two regions [Fig. 2(b)], which can be ascribed to a difference in both scatter power (b = 0.5 ± 0.1 in Region 2, b = 2.0 ± 0.1 in Region 3) and absorption. The μₑ-spectra of both regions are shown in Figs. 2(c) and 2(d). The fitted [tHb] of 3.0 ± 0.5 g/l in Region 2 and 7.8 ± 1.2 g/l in Region 3 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.⁴ 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,⁷ 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.³ The value of the μₑ-contributions to the measured μₑ fall within the physiological range¹ of 1 to 100 mm⁻¹, but the actual dermal μₑ may be underestimated due to the contribution of multiple scattering to the LCS signal.⁶ Nevertheless, since absorption takes place along the controlled photon path,⁵ this contribution does not influence the determination of μₑ.

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 μₑ. The latter is affected by the size of the investigated region (i.e., the number of spatial averages and the length of the ℓ-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.⁵ 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
acquisition, the latter will require correction for unwanted signal attenuation due to out-of-focus detection and sensitivity roll-off in depth.\textsuperscript{8,12}

Potentially, the dermal [tHb] can be related to the hemoglobin concentration in whole blood, if the blood volume within the investigated region can be assessed. A possible method that deserves further investigation for this purpose is obtaining the blood volume from the OCT image by assessing the vessel density using advanced signal analysis, for instance as described in Ref.\textsuperscript{11}. When measuring other tissue types, the contribution of additional chromophores (e.g., bilirubin, melanin) to the measured \(\mu_t\) may need to be incorporated in our analysis. Also, a correction for Doppler broadening or shifting of the measured \(\mu_s\) spectra may be needed for tissues that exhibit blood flow. We did not observe any of those influences on the measured \(\mu_t\) spectra in Fig. 2(b), which can be explained by a temporary decrease of blood flow due to applied pressure on the skin.

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 \textit{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.

Acknowledgments

This research is funded by personal grants in the Vernieuwingsimpuls program (DJF: AGT07544; MCGA: AGT07547) by the Netherlands Organization of Scientific Research (NWO) and the Technology Foundation STW.

\begin{figure}
\centering
\includegraphics[width=\textwidth]{Fig_2}
\caption{\textit{In vivo} measurement on the skin of the palmar side of a finger joint. The measured \(\mu_t\), fits on \(\mu_t\) and \(\mu_s\)-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 \(\mu_s\)-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.}
\end{figure}

\textbf{References}