Wavelet-domain de-noising of optical coherent tomography data for biomedical applications

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Abstract. Optical coherent tomography (OCT) is a rapidly developing method of fundamental and applied research. Detection and processing of OCT images is a very important problem of applied physics and optical signal processing. In the present paper we are demonstrating the ability for effective wavelet-domain de-noising of OCT images. We are realizing an algorithm for wavelet-domain de-noising of OCT data and implementing it for the purpose of studying test samples and for in vivo nail tomography. High de-noising efficiency with no significant losses of information about the internal sample structure is observed.

1. Introduction and background

Optical coherence tomography (OCT) is an imaging technique allowing non-invasive in vivo and in vitro three-dimensional imaging of both transparent and nontransparent biological tissues on a micrometer scale with no necessity of contact between the probe and the tissue [1–14].

OCT produces an image of optical scattering from internal tissue microstructures by measuring the echo time delay and magnitude of backreflected light in a way that is analogous to ultrasonic pulse-echo imaging [15] or imaging the object with terahertz time-of-flight tomography [16–19]. The result of OCT imaging is a profile of backreflection intensity, containing information about the internal structure of the sample. Because the high speed of light makes a direct measurement of optical echo impossible, OCT uses interferometric techniques with ultrashort light pulses or low coherent light sources to range distances at the level of single micrometers [2-14]. OCT was first developed by Fujimoto’s group [1], and the method has since matured into an important clinical imaging modality. Since OCT is non-invasive technique, it becomes widely applicable in medical disciplines [20] such as ophthalmology [21–24] gastroenterology [25–28] dentistry [29–32] and dermatology [33–36]. An important aim of OCT is to acquire morphological changes within biological tissue. Due to its capability to study the structure of inhomogeneous objects, ability to work in strongly dispersive media, high potential in obtaining cellular level resolution, and non-ionizing nature of radiation presence, OCT appears to be the most promising method of examination and characterization of epithelial tissue in vivo.
This paper presents the algorithm for effective de-noising of OCT data in wavelet domain. We are developing an algorithm for wavelet-domain de-noising of OCT signals and implementing it for studying the test samples, as well as for tomographic reconstruction of nail structure in vivo. The results of algorithm implementation and possible applications of the proposed technique are discussed.

2. Materials and methods

At first, we are briefly discussing the techniques of time-domain and Fourier-domain OCTs [3]. The principle of OCT operation is based on the low-coherence interferometry, and the key component of every OCT system is a two-beam interferometer (typically Michelson’s interferometer) with low-coherence and broad-bandwidth light source (figure 1, 2). The light beam generated by the source is divided into two parts at the beam splitter. The first one is propagating towards the reference mirror, and the second one is directed to the object of interest. After the waves were scattered on the surface of reference mirror and on the surface of the object, the scattered light components travel to the detector. Let us define the optical time-delay of the reference beam with duration \( t_r \), and the sample beam time-delay with \( t_s \). The interference pattern occurs on the surface of the detector as a result of coherent superposition of the wave amplitudes, but interference could appear only for time-delays of optical beams satisfying the condition

\[
|t_s - t_r| \leq \frac{t_c}{2} = \frac{l_c}{2}
\]

where \( t_c \) and \( l_c \) are coherence time and coherence length, respectively, and \( c \approx 3 \times 10^8 \text{ m/s} \) is the speed of light in vacuum. By time-gating the echo response, we are able to create a depth-resolved line profile of the tissue (or A-scan). An OCT image (B-scan) is built up as a series of adjacent axial depth scans (A-scans). In OCT the depth-resolved information about internal structure of the sample can be obtained from either the time-domain measurements (TD-OCT) or the Fourier-domain (FD-OCT) measurements [3].

Figure 1 shows a schematic representation of TD-OCT set-up and describes general principles of measuring the profile of backscattering intensity, \( I(z) \) [3]. The main advantage of TD-OCT is an automatic removal of coherence noises caused by interference of multiple beams reflected from various components of the interferometer optical systems, and this is due to the time-gating principle of TD-OCT operation. The principle of FD-OCT operation is based on the Wiener-Khintchine theorem [45]. It approves that the profile of backscattered light intensity

Figure 1. Schematic representation of TD-OCT measurements: M stands for the reference mirror of Michelson’s interferometer, BS stands for the beamsplitter, and PD stands for the photodetector. In TD-OCT the backscattering intensity profile \( I(z) \) is detected by means of post-processing the series of interference patterns detected at PD while continuous translating M. Since the low-coherence light source is utilized for pumping the sample, each position of M corresponds to the certain optical depth of light backscattering. Coherence length \( l_c \) defines the theoretical limit of TD-OCT depth-resolution as \( \Delta z \geq l_c \).
Figure 2. Schematic representation of FD-OCT measurements: M stands for the reference mirror of Michelson’s interferometer, and BS stands for the beamsplitter. In FD-OCT the spectral dependence of sample complex amplitude reflectivity $\tilde{R}(\nu)$ is detected by means of post-processing the interferometric data. The interference pattern is being detected for a different light frequencies $\nu$ either by tuning the monochromatic wavelength of light source or by simultaneous detection of interference patterns at different wavelengths. Reconstruction of the internal structure of the sample is based on complex reflectivity $\tilde{R}(\nu)$ and the Eq. (2). The bandwidth of sample reflectivity registration $\Delta \nu$ limits the depth resolution of FD-OCT measurements, $\Delta z \geq 1/\Delta \nu$.

$I(z)$ is directly related to the spectral dependence of the sample amplitude reflectivity, and the latter could be reconstructed after interferometric measurements. Let us define the amplitude reflectivity of sample with $\tilde{R}(\nu)$, where $\nu$ is a frequency of electromagnetic wave, then the backscattering intensity profile could be reconstructed in the following way

$$I(z) \propto |R(t)|^2 = \left| \mathcal{F}^{-1}_\nu [\tilde{R}(\nu)] \right|^2$$

(2)

where $R(t)$ corresponds to the sample pulse response, and $\mathcal{F}^{-1}_\nu [...]$ stands for the inverse Fourier transform. Figure 2 shows the principles of FD-OCT operation.

Two types of FD-OCT exist: 1) spectral-domain OCT captures the sample reflectivity simultaneously in the entire spectral range of the system sensitivity by using the spectrometer equipped with a high-speed detector; 2) swept source OCT captures the spectral dependence of the sample amplitude reflectivity with a single detector by means of sweeping the output frequency of tunable laser source. The main advantage of FD-OCT is associated with the ability for fast detection of backscattering intensity profile without any mechanical scanning, and the acquisition speed is only limited with the detector frame rate. Modern OCT systems with enhanced axial resolution of 3 $\mu$m allows obtaining in vivo backscattering intensity profiles. In clinical practice, the ability for visualizing the subtle morphological features of tissue significantly improves the early diagnosis and the assessment of tissue pathologies.

3. Wavelet-domain de-noising approach

Modern OCT systems, based on various operational principles, allow accurate characterization of in vivo biological tissue [1-36], however, the problem of OCT signal post-processing, including solution of the backscattering inverse problem and signal de-noising, still remains very important for further development of OCT applications. We are demonstrating the ability for highly-efficient de-noising of OCT data by means of wavelet-domain prossing [37–39]. The techniques for wavelet-domain de-noising seem to be very effective for processing of optical signals [40–41], including de-noising of spectroscopy data [42–44] due to high locality of wavelet transform kernels both in time-domain and frequency-domain. We are experimentally implementing wavelet-domain de-nosing for processing the OCT measurements.
Let us define $I(x, z)$ to be the results of OCT measurements, where $z$ stands for the sample depth, and $x$ stands for the lateral scanning direction. Thus, the signal $I(x, z)$ is the result of measuring of the backscattering intensity profile $I(z)$ at different points $x$ of sample surface. We are applying the wavelet-domain de-noising separately for various lateral coordinates, thus, let us consider the procedure for single backscattering intensity profile $I(z)$ corresponding to the certain $x = x'$.

The wavelet-domain de-nosing procedure is used in the following way. At first, the wavelet-domain decomposition is applied to the backscattering intensity profile $C(a,b) = W[I(z) − m.e.] = \int_{−\infty}^{+\infty} (I(z) − m.e.) \psi(a,b,z) dz$, \hspace{1cm} (3)

where $m.e. = \frac{1}{Z} \int I(z) dz$ is a mathematical expectation of the intensity profile ($Z$ is an integration interval), $W[...]$ is a direct wavelet transform operator, $C(a,b)$ are wavelet decomposition coefficients, and $\psi(a,b,z)$ are wavelet decomposition kernels. Note, all $\psi(a,b,z)$-kernels are constructed by scaling and shifting of the mother wavelet, $\psi(z)$, as $\psi(a,b,z) = \left|a\right|^{-1/2} \psi(\frac{z-b}{a})$ (a defines the kernel scale and $b$ defines the kernel translation).

De-noising is implemented via the thresholding of the wavelet decomposition coefficients $C_T(a,b) = \begin{cases} C(a,b), & \text{if } C(a,b) \geq T, \\ 0, & \text{if } C(a,b) < T, \end{cases}$ \hspace{1cm} (4)

where $T = \sqrt{2\ln(n)}$ defines the universal threshold, utilized in the present work, and $n$ stands for the number of wavelet-decomposition coefficients. Inverse continuous wavelet transform is applied to de-noised wavelet spectrum and the mathematical expectation is being added $I_T(z) = W^{-1}[C_T(a,b)] + m.e. = C^{-1}_\psi \int_{−\infty}^{+\infty} \int_{−\infty}^{+\infty} C(a,b) \tilde{\psi}(a,b,z) dadb + m.e.$, \hspace{1cm} (5)

where $W^{-1}[...]$ is an inverse wavelet transform operator, $\tilde{\psi}(a,b,z)$ is the dual function of $\psi(a,b,z)$, and

$C_\psi = \int_{−\infty}^{+\infty} \frac{\Psi(\omega) \tilde{\Psi}(\omega)}{|\omega|} d\omega < \infty$ \hspace{1cm} (6)

is the admissible constant. Functions $\Psi(\omega)$ and $\tilde{\Psi}(\omega)$ in (6) correspond to Fourier spectra of $\psi(z)$ and $\tilde{\psi}(z)$, respectively; and the admissible constant (6) restricts the diversity of functions suitable for definition of $\psi(z)$.

We use fast wavelet-transform (FWT) algorithm with ’coif5’ mother wavelet, which was found to be optimal for the purpose of OCT data decomposition [42–44], and FWT helps to implement real-time data processing.

4. Results and discussion
Let us describe the results of the wavelet-domain de-noising implementation for processing the test OCT signals. We utilize TD-OCT set-up with fiber probe for this purpose. The set-up is based on picosecond Er-doped fiber laser and fiber-based Michelson’s interferometer equipped with mechanical scanning system.
Figure 3. Results of wavelet-domain de-noising of time-domain OCT data: (a) corresponds to the test sample made of single-layer polymer film, (b) corresponds to the test sample made of double-layer polymer film, and (c) corresponds to the test sample made of bint. The images of left and right columns show initial noisy tomographic images and tomographic images after de-noising, respectively.

Figure 3 shows the results of studying the test samples of polymer films and complex scattering sample made of bint. One should notice significant noise suppression for all of the test signals. The contrast in images has been raised dramatically with no losses of important information about the sample structure. Note, the top response in all tomographic images corresponds to the light backscattering on the reference plane window, which is utilized in order to possess OCT probe on the surface of the studied object. We could see strong backscattering from all the interfaces of single-layer and two-layer polymer film. Moreover, the backscattering on the internal inhomogeneities of the film is still notable after the noise suppressing. The noises in bint image are also suppressed; and even small scatterers are notable in the tomographic image.

We have studied in vivo samples of human nail as a representative example of biomedical...
algorithm implementation. Tomographic images of nail body and nail lunula have been considered. We have significantly suppressed the noise in nail images, and all the structural features of tissue are notable in the tomogram. Backscattering on small inhomogeneities of nail is present in the image of nail body both before and after data de-noising. Image of nail lunula contains the responses from several interfaces between different tissues in a sample, allowing us to analyze the internal structure of nail and underlying tissue.

5. Conclusions
In the present paper we demonstrated an ability for effective wavelet-domain de-noising of OCT data. We have realized an algorithm for TD-OCT data de-noising in wavelet-domain and implement it for studying the test samples. High de-noising efficiency with no significant losses of information about the object structure have been demonstrated.

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