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RESEARCH ARTICLE

Medium chromatic dispersion calculation and correction in spectral-domain optical coherence tomography

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Abstract A method for determining and correcting distortions in spectral-domain optical coherence tomography images caused by medium dispersion was developed. The method is based on analysis of the phase distribution of the interference signal recorded by an optical coherence tomography device using an iterative approach to find and compensate for the effect of a medium's chromatic dispersion on point-spread function broadening in optical coherence tomography. This enables compensation of the impact of medium dispersion to an accuracy of a fraction of a radian (units of percent) while avoiding additional measurements and solution of the optimization problem. The robustness of the method was demonstrated experimentally using model and biological objects.

Keywords optical coherence tomography (OCT), dispersion, image resolution restoration

1 Introduction

Optical coherence tomography (OCT) is a technique for noninvasive visualization of the internal structure of optically turbid media, primarily biological media. OCT provides imaging at a resolution of a few micrometers, where the fundamental limitation is specified by the bandwidth of the probe radiation. Other factors influencing the axial resolution of OCT images are the dispersion of the group velocity of light determined by its optical path in the medium, the shape of the probe light spectrum, the chromatic aberration and polarization anisotropy of optical elements, etc. [1]. OCT technology is applied mainly in ophthalmology [2], where the method has become a gold standard. The technology enables a dramatic increase in both the diagnostic accuracy and the accessibility of human eye therapy in comparison with other methods [3]. In addition to conventional OCT imaging, many methods to obtain concomitant information have been developed. In particular, OCT-guided technology provides an opportunity to reconstruct vascular bed images, map the elastic and polarization properties of tissue [4], retrieve information about the thickness of individual layers of the retina [5,6], etc. One of the main factors significantly reducing the axial resolution of OCT images (manifested in broadening of the point-spread function (PSF) in the direction of the probe beam axis) is chromatic dispersion of the refractive index of the optical path [7]. Despite the relevance of the problem and significant efforts of researchers toward its solution [8 – 10], no unified approach has been formulated yet that would find and compensate for image distortions caused by medium dispersion. In most cases, dispersion distortions are compensated by using hardware and software approaches sequentially. The former approach provides chromatic dispersion correction directly in the interferometer by using materials possessing dispersion properties similar to those of the media under study. This approach enables correction of the effect of primarily the quadratic terms in the polynomial expansion of the dispersion spectrum. Although it is quite sufficient for many tasks [10], it is worth noting, following [11], that the uncompensated effect of medium dispersion on the third decimal place of the refractive indices may result in an interference pulse delay up to 10 times longer than the pulse duration. This necessitates accurate selection of both the compensator material and its optical thickness.

The software approach makes it possible to achieve a higher level of correction (up to full elimination) of the effect of medium dispersion on the image resolution, especially if it is used in combination with the hardware approach. In general, this is done in two stages: measuring the phase distribution of the filling of the interference-induced frequency comb during calibration signal recording (for example, reflection from a thin border) and

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subsequent subtraction of the distribution from the real signal phase component [12].

Unfortunately, both the above approaches require prior information about the optical path dispersion. In some cases, this information is unavailable — in particular, if the dispersive medium is part of the object under study. One such object is the human eye, where the probe light should pass through several centimeters of vitreous humor. In these cases, correction of the effect of medium dispersion is usually limited to a few minor terms of the dispersion curve expansion [13] (although one can use a wavelet transform to obtain a correction [14]). The expansion coefficients are selected algorithmically and can be corrected for an entire image or for its separate layers [8,15,16]. This makes it possible to obtain a resolution close to the spectrally limited one, but it does not allow exact reconstruction of the original dispersion curve, which may have singularities caused by optical absorption in the probe light's spectral band.

An alternative method of acquiring OCT images free of any dispersion-caused distortions is worth mentioning. This method, which is based on the use of two-photon interference [17,18], is a handy tool for optical science rather than a technique for medical applications.

However, some numerical methods do not require prior knowledge of the dispersion values in the object under study. One of them is based on speckle analysis [19,20], but the main propose of the method is the absence of PSF widening of the image of the front surface of the object. Another method reported in 2004 [21] and then published in 2007 [22] is based on determination of the autoconvolution function and provides automatic improvement of the image resolution up to the spectral width produced.

The present work develops a method for determining and correcting the effect of medium dispersion in OCT images without using additional calibration measurements or choosing dispersion curve expansion coefficients.

2 Principle of operation

The principle underlying the proposed method is similar to that of the method for reconstructing a 2D surface of optical aberrations in the digital holography setup described in Ref. [23]. It consists of quantitative determination of the phase distortions of a complex OCT signal induced by dispersion using image analysis. After eliminating coherent noise and autocorrelation terms [24,25], one can obtain an optical frequency comb, the scale of which is determined by a set of optical delays between the interfering waves. This allows reconstruction of the OCT image by Fourier transformation of the registered optical spectrum [26]. The presence of dispersive media in the optical path may be taken into account by introducing a phase term into a reference or probe wave in a complex exponential representation,

$$E(k) = E_X(k)\cdot \mathrm{e}^{\mathrm{i}\psi(k)},$$

where $k = 2\pi/\lambda$ is the wavenumber of a single spectral component registered by a separate spectral channel, $E_X(k)$ is the spectral component amplitude, and $\psi(k)$ is a phase term determined by the chromatic dispersion of the optical path length. Here we assume that $\psi(k)$ has no linear component in its polynomial expansion, as this component causes only an insignificant shift of the scatterer depth and does not affect the PSF broadening [27].

In conventional spectral-domain (SD)-OCT, the optical frequency comb is insensitive to the sign of the mutual delay between the reference and probe waves. Many approaches have been developed for determining the complex value of the optical field in interferometric registration (i.e., complex OCT) [24]. In such systems, the registered values of the optical frequency comb function $F(k)$ can be represented by the expression

$$F(k) = |F(k)|\cdot \exp\Big(\mathrm{i}\phi(k) + \mathrm{i}\psi(k)\Big), \tag{1}$$

where $\phi(k)$ denotes the inherent argument of the complex optical frequency comb in the absence of dispersion. The $\phi(k)$ term in Eq. (1) is determined by the total phase of all the interfering waves registered by the interferometer.

Most biological structures consist of a high number of scatterers that are smaller than both the probe beam diameter and its coherence length. Thus, the distribution of $\phi(k,x)$ during probe beam scanning of the sample in the x direction can be considered to be delta-correlated. This makes it possible to find $\psi(k)$ from Eq. (1) by averaging the phase term over a large number of scatterer images.

The proposed method operates with a number of separate fragments of an OCT image, assuming that the impact of medium dispersion is similar for a large number of scatterers in a highlighted area of the image.

The method is realized iteratively, and each iteration consists of two stages.

In the first stage, a number of depth-localized complex-valued optical frequency combs are generated. In the first iteration, an inverse Fourier transformation is performed for each complex-valued optical frequency comb $F(k,x)$ from the set corresponding to an OCT B-scan yielding a complex-valued image $Q(z,x)$. The next step is to automatically select a number of A-scans containing strong local maxima in a certain range of depth z.

Each selected A-scan is multiplied by a window function centered at the A-scan maximum. The window function width is determined on the basis of the preliminary estimation of possible PSF broadening. The truncated A-scans, denoted as $q(z,J)$, where J is the selected A-scan number, are used to reconstruct the in-depth localized spectra:

$$f_J(k) = FT_z\Big(q(z,J)\Big).$$

Because the selected A-scan is windowed in the area containing a strong maximum, the filtered spectra $f_J(k)$ may be written in quasi-harmonic representation:

$$f_J(k) = |f_J(k)| \cdot \exp\left(\mathrm{i}\left(\phi_J(k) + \tilde{\phi}_J(k) + \psi(k)\right)\right), \quad (2)$$

where the phase component contains the dispersion-induced term $\psi(k)$, which is the same for every A-scan number J; the regular phase $\phi_J(k)$ is determined by the depth of the selected maximum, and the irregular phase component $\tilde{\phi}_J(k)$ represents the contribution of neighboring scatterers. The last component is assumed to have no linear component in its polynomial expansion.

In the second stage, the dispersion-induced image distortions are found and eliminated. As the first step, the gradients of the phase component of Eq. (2) are calculated:

$$\delta\theta_J(k) = \frac{\partial}{\partial k}\arg\left(f_J(k)\right).$$

Because $\phi_J(k)$ depends linearly on k, this term may be excluded by subtracting the mean of $\delta\theta_J(k)$:

$$\delta\tilde{\theta}_J(k) = \delta\theta_J(k) - \langle\delta\theta_J(k)\rangle_k = \frac{\partial}{\partial k}\tilde{\phi}_J(k) + \frac{\partial}{\partial k}\psi(k).$$

Different A-scans were assumed to be delta-correlated in x space, so by averaging these gradients one can eliminate the $\tilde{\phi}_J(k)$ term:

$$\langle\delta\tilde{\theta}_J(k)\rangle_J = \left\langle\frac{\partial}{\partial k}\phi_J(k)\right\rangle_J + \left\langle\frac{\partial}{\partial k}\psi(k)\right\rangle_J = \frac{\partial}{\partial k}\psi(k).$$

After this equation is integrated, the approximated $\hat{\psi}(k)$ is obtained and then used to reconstruct the improved complex-valued image $Q'(z,x)$:

$$Q'(z,x) = FT^{-1}\left(F(k,x) \cdot \exp(-\mathrm{i}\hat{\psi}(k))\right).$$

The improved image is used as the starting image for the next iteration.

3 Experiment

The proposed method was verified using a common-path SD-OCT setup based on a light source with a central wavelength of 1055 nm and a bandwidth of 71 nm (Superlum Diodes Ltd.).

Chromatic dispersion distortions were introduced by placing a piece of optical glass (*N-SF*4, 9 mm thick) in the reference arm of the main interferometer.

The first investigation examined the ability of the method to reconstruct the nonlinear profile of the medium dispersion spectrum. Toward this end, images of a human finger (Fig. 1(a)) were recorded with and without the additional dispersive element. Consequently, the effect of the OCT setup's medium dispersion could be excluded. The proposed method provides the possibility of reconstructing the nonlinear term of the dispersion profile, which seems to be unattainable using the method described in Ref. [22]. The reconstructed profile of the introduced dispersion was obtained by selecting 32 reference A-scans (the 32 brightest elements of the image in different A-scans). Owing to the relatively small dispersion-caused distortions, the first iteration provides a quite acceptable result. The profile is represented by the solid curve in Fig. 1 (b). A comparison with the known dispersive parameters of the element (*N-SF*4, 9 mm thick, calculated by Sellmeier's equation [28] and shown as a dashed curve in Fig. 1(b)) shows that the proposed method reconstructs the nonlinear profile of the medium dispersion to an accuracy not worse than 7% in the entire range of the dependence in the central spectral region. The larger differences on the left end of the spectrum (starting at a wavelength of 1.075 μm) are caused by the low values of the spectral intensity resulting from asymmetry induced during spectrometer tuning.

The PSF broadening in the image in Fig. 1(a) is barely noticeable owing to the relatively small width of the probe light spectrum. A modern trend in OCT is to increase the axial resolution by using broader spectra (up to and even exceeding 10% of the central wavelength [29]). As the effect of dispersion on the increase in PSF broadening depends quadratically on the bandwidth, the proposed method is appropriate for such systems.

The ability of the method to restore the resolution after strong dispersion was demonstrated in a retina-imaging experiment. To do so, an additional dispersive element (two pieces of *N-SF*4, 9 mm thick) was placed in the reference arm of the main interferometer of the common-path SD-OCT system mentioned above. The probe end of the setup was connected to a retina scanning probe (the entire system was not corrected for dispersion distortions). The recorded OCT image of a human volunteer's retina is shown in Fig. 2(a). For clarity, the magnified central area of the image is presented in Fig. 2(c). The dispersion-induced distortion results in dramatic broadening of the PSF, and the inner/outer segment junction layer is not visible in the image. The use of the proposed method (32 reference A-scans, two iterations) results in an appreciable decrease in the PSF width (Figs. 2(b) and 2(d)) and the appearance of layers that were indistinguishable before. The observed width of the layer is about one resolution element of the image, which corresponds to the spectrally caused resolution of the OCT setup.

Estimates have shown the low computational capacity of the proposed method in comparison to the method presented in Ref. [22], which requires calculation of the autoconvolution for a relatively large numerical dataset. Moreover, the proposed method's computational capacity seems to be much smaller than that of the reconstruction procedure for the entire SD-OCT image. Thus, it may be successfully used in software for OCT complexes based on laptop platforms, in particular for passively cooled CPUs.

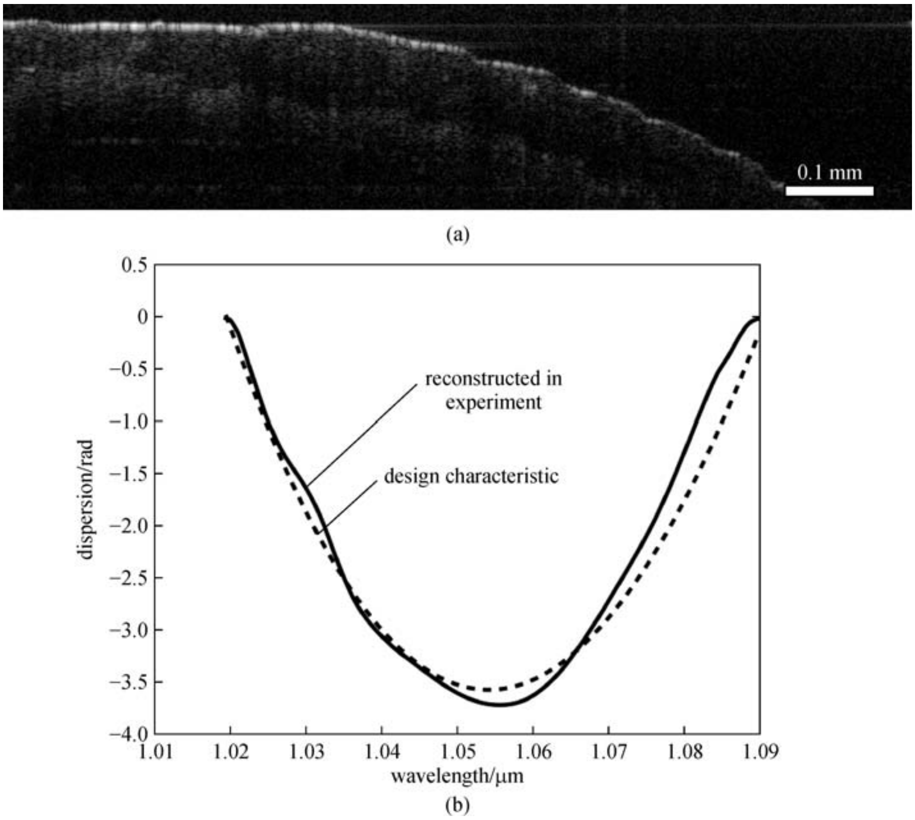

Fig. 1 Experimental verification of the possibility of dispersive profile reconstruction. (a) Initial OCT image of human finger skin; (b) reconstructed nonlinear profile of medium dispersion spectrum (solid curve) and its numerical model based on Sellmeier's equation [28] (dashed curve)

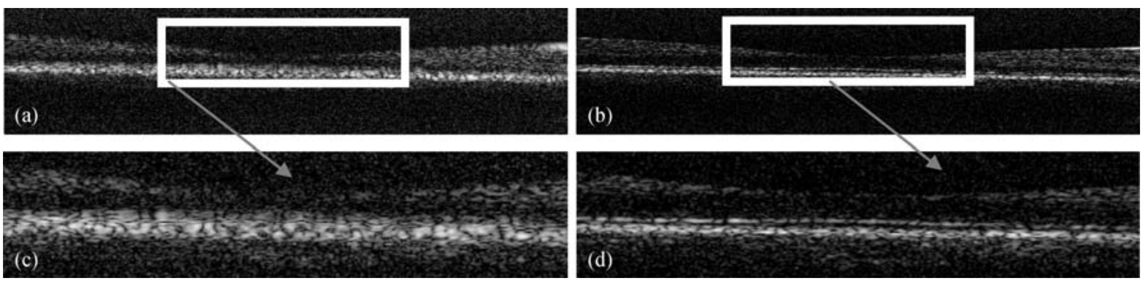

Fig. 2 Human volunteer retina OCT image. (a) Initial image in the case of dispersion-induced distortion; (b) restored image; (c) and (d) magnified fragments of (a) and (b) images, respectively

This is important for the development of OCT devices intended for sterile rooms. The method may also be useful for OCT complexes based on ordinary computers in combination with high-level modalities (3D visualization, OCT angiography and electrography, etc. [30]) requiring high computing power without degradation.

4 Conclusion

The proposed method of determining the effect of medium dispersion on OCT images enables automatic reconstruction of the nonlinear profile of the medium dispersion spectrum. Application of the method to OCT images with large dispersion-induced distortions results in a significant increase in the axial resolution up to the spectrally determined level. The method was verified experimentally by imaging model media and biological tissues.

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