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روشی جدید برای تعیین ضخامت و ضریب‌شکست لایه‌ها در تصاویر عمق‌نگاری همدوس نوری

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چکیده- عمق‌نگاری همدوسی نوری در حوزه‌ی فوریه طی دو دهه‌ی گذشته به دلیل ماهیت غیرتهاجمی و ضریب تفکیک بالای آن به ابزاری پرکاربرد در چشم پزشکی برای تشخیص و کنترل بیماری‌های مختلف چشمی تبدیل شده است. در این مقاله یک روش جدید براساس تفاوت شدت در پیکسل‌های تصویر در مرز بین دو لایه‌ی برای محاسبه‌ی ضریب شکست و ضخامت لایه‌های مختلف به صورت مجزا پیشنهاد شده و به صورت تئوری مورد بررسی قرار گرفته است. براساس تحلیل‌های انجام شده روی داده‌هایی که به صورت تصادفی تولید شده‌اند میزان خطای میانگین کمتر از ۲ درصد بوده است. همچنین تحلیل‌های مختلفی نیز روی حساسیت به عوامل خطا انجام شده و نتایج قابل قبولی داشته است.

کلید واژه- عمق‌نگاری همدوسی نوری، ضریب شکست، عمق‌نگاری در حوزه‌ی فوریه، تصویربرداری پزشکی، چشم‌پزشکی

A novel method for decoupling diffraction coefficient and thickness in an optical coherence tomography imaging system

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Abstract- Optical coherence tomography in the Fourier domain due to its non-invasive nature and high resolution has become a routine tool in ophthalmology for detection and control of many diseases in the last two decades. In this paper, a novel method based on the image contrast the scans at the interfaces has been proposed and theoretically explained for decoupling refractive index and thickness of the layers. Also, error analysis has been done on this method which results in mean error lower than 2% on the generated reference data. The proposed method also has been checked for sensitivity to the background and foreground refractive indexes. In case of 10% error in background and foreground refractive indexes, mean error in calculation of the refractive index and thickness of the layers are less than 2.5%. Additionally, numerical procedure of this method has been explained and applied to several scan images of the retina and the resulted image has been illustrated.

Keywords: Optical coherence tomography, Refractive index, Fourier domain OCT, Medical imaging, Ophthalmology

1. Introduction

Optical tomographic techniques have significant importance in various fields of medicine, biology, and industry [1]. These techniques can provide non-invasive diagnostic images with high depth resolutions [1]. Among them, modern Fourier domain optical coherence tomography (OCT) systems using low coherency sources has become important especially in ophthalmology [1].

In the OCT imaging method, the thickness of a layer is obtained by calculating the optical path length in that layer which is the layer physical thickness multiplied by its refractive index. Separating the refractive index from the thickness has been discussed in a few pieces of research before [2]. Correction of OCT images based on the exact thickness of layers might have diagnosing applications [3]. In this paper, we have introduced a new method for decoupling the refractive index from the thickness in a z-space image of an N-layer sample.

2. Governing theories

Modern OCT imaging devices are mostly based on Fourier domain analyses (Spectral domain and swept source) which have significant advantages over time-domain devices [1].

Spectral-domain OCT (SD-OCT) imaging is an interferometry technique based on Michelson-Morley interferometer [1]. In this setup resulting signal from the superposition of the reference and sample intensities has fed to the spectrometer and after Fourier transform, the final result will show some peaks corresponding to the interfaces between the layers.

In this paper summation method has been used for the calculations. [4] This method requires two assumptions which are reasonable for multilayer biological samples with nearly the same refractive indexes like retina [4]. Firstly, the inter-layer multiple reflections are neglected and the input field into the sample presumed to reflect once from each layer. Secondly, forward field through the subject is presumed to remain constant. Accordingly, spectral intensity in the spectrometer (interference intensity) can be found and its Fourier transform equals [2]:

$$I(z) = 2s(z) * \left[\begin{array}{l} r_{(R)} r_{(0)} \delta\left(z - 2Z_R + 2Z_S\right) + \\ \sum_{j=1}^{N} r_{(R)} r_{(j)} \delta\left(z - 2\left(\begin{array}{l} Z_R - Z_S + \\ \sum_{m=0}^{j} n_m d_m \end{array} \right) \right) \end{array} \right]$$

$$\rightarrow I(z) = r_{(R)} r_{(0)} s\left(z - 2Z_R + 2Z_S\right) + \sum_{j=1}^{N} r_{(R)} r_{(j)} s\left(z - 2\left(Z_R - Z_S + \sum_{m=0}^{j} n_m d_m \right) \right) \quad (1)$$

Where $r_{(R)}$ and $r_{(0)}$ are the reference mirror and first sample layer reflection coefficients respectively. Z_R and Z_S are reference and sample arm lengths respectively. $r_{(j)}$ is the j'th layer reflection coefficient and n_m and d_m are refractive index and thickness of the m'th layer respectively. s(z) is Fourier transform of source spectrum and $\delta(z)$ is Dirac delta function. Thus, intensity in the z-space includes N+1 peaks (number of interfaces) displaced by optical path of the layers and magnitude of these peaks provide us information about reflection coefficient of each interface. As a result, diffraction index and the layer thickness of each layer can be separately calculated as:

$$\begin{array}{l} j^{th} \text{ interface:} \\ peak\ value = s(0) r_{(R)} r_{(j)} = s(0) r_{(R)} \dfrac{n_j - n_{j+1}}{n_j + n_{j+1}}, \quad j = 0:N \\ displacement = z_{j^{th}} - z_{j-1^{th}} = n_j d_j, \quad j = 1:N \end{array} \quad (2)$$

By assuming known foreground and background refractive indexes (n_0,n_{N+1}) which are often available, (2) can be numerically solved. For instance, in imaging retina layers, the refractive index of foreground layer (vitreous humor) and background layer (choroid) are about 1.337 and 1.45 respectively in the near infrared region [5]. As can be witnessed in (2), ratio of reflection coefficient of each layer on the corresponding peak is constant ($r_{(R)}$s(0)) which provides a tool for recursively calculating every refractive index and thickness. Using known refractive index values (n_i's) and optical path lengths ($n_i d_i$'s) thickness of layers (d_i's) can be calculated.

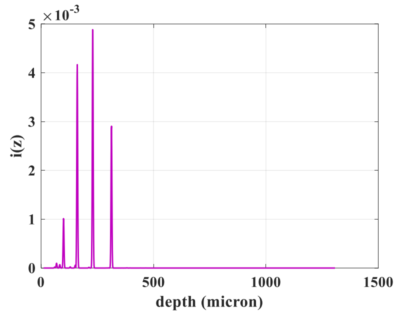
3. Numerical approach

For each A-scan, the first step is to find the peaks related to the interfaces in the z-space signal. This can be done by finding all of the local maximums and omitting the auto-correlation and noise peaks using a wisely guessed threshold. This threshold can be found by sorting the signal maximums and dividing them into two clusters. Then, using the calculated peak positions, peak magnitudes and displacement between them can be derived. The next step is guessing an initial value for the refractive indexes. In our calculations all of the refractive indexes are assumed to be identical to the vitreous medium. This value will be recursively corrected until for all of the interfaces ratio of peak magnitude over the Fresnel reflection coefficient converges to a similar value by assuming a desired error margin between the steps.

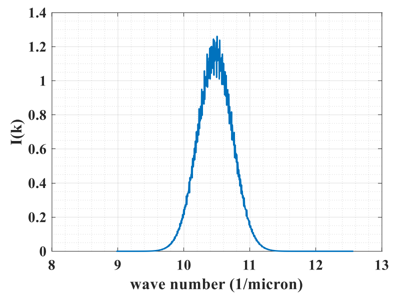
As an example A-scan of a 3 layer system with refractive indexes equal to 1.35, 1.377, 1.41 and corresponding thicknesses equal to 45µm, 50µm, 59µm has been generated using the Transfer matrix(TMM) method and depicted in Fig.1 [6]. Resulting from the proposed algorithm for the generated data, calculated values are 1.3306, 1.3721 and 1.4206 for refractive indexes and 46.0845µm, 50.3608µm, and 58.5668µm for thicknesses. Mean error in calculation of refractive index is 0.85% and thickness error is 1.22%.

The discussed method is appropriate for a single A-scan. However, output of commercial OCT devices are scan images containing both A-scans and B-scans. Also, these images are often very noisy and detection of interfaces in them is quite difficult. In order to adapt our method to the commercial OCT scan outputs, some additional steps should be followed.

The first step is to segment the scan input for finding the interfaces between the layers. In commercially licensed softwares of the famous companies like Heidelberg™ Inc., segmentation algorithms based on artificial intelligence can recognize up to 11 layers in a retina scan. However, in this paper an open source simple graph-cut code [7] has been utilized for the segmentation process. This simple, fast and easy to implement method can provide us 6 retinal layers which are enough for demonstration purpose of this paper. For practical implementation of this method in a commercial device, more precise segmentation algorithms should be used. The next step is to find derivative of the image in each A-scan to have peaks on each interface like the single A-scan case. Then for each A-scan as mentioned in before, refractive indexes of layers can be found. The final step is to correct thickness of the layers in each A-scan according to the calculated refractive index. Eventually the correct scale image can be regenerated and coloured with respect to the refractive indexes. Also, refractive index profile can be plotted in any desired A-scan.



(a)



(b)

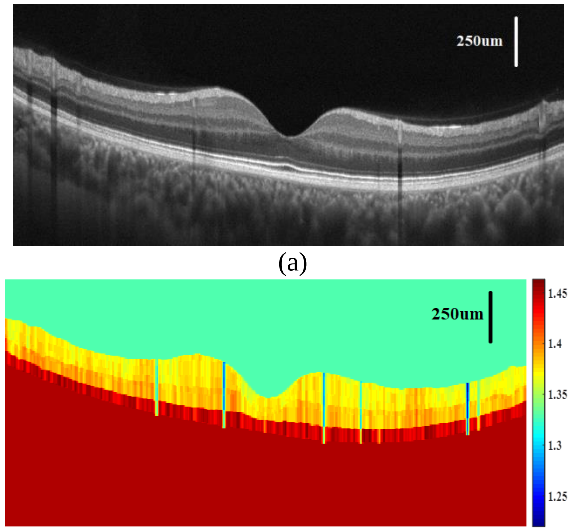
Fig. 1: A sample generated data, (a) z-space output (b) k-space output

4. Results

In order to evaluate the precision of the proposed method, four different analyses have been done. First, 15 A-scan data has been randomly generated using the TMM method to find the errors in various cases [6]. These reference A-scans have from 2 to 6 layers with thicknesses from 30μm to 70μm and generated by assuming an SLD source with centre wavelength of 880nm and FWHM equal to 50nm. Also, sample and reference arm mismatch is assumed to be 100μm. According to our calculations, average error in calculation of refractive index is 1.447% and the mean error in calculation of thicknesses is 1.775%. In addition, for a 6-layer sample the maximum calculated error is less than 8%. Then, sensitivity of the mean error in refractive index and thickness to foreground and background refractive indexes is calculated for 0 to 10% error in n_0 and n_f for the first sample which leads to less than 2.5% error in calculation of refractive indexes and corresponding thicknesses. The last error analysis done in this paper is adding white noise with mean intensity about 10% of the intensity of the peaks to the reference samples. This noise will result in a signal to noise ratio (SNR) about 20dB which in the worst case means lower than 5% error in calculation of the refractive indexes and thicknesses.

Eventually, in Fig.2 (a) a cross section of a healthy retina has been shown in which thickness of each layer is equal to its optical path length (refractive index × physical thickness). Fig.2 (b) is extracted by our algorithm and provides extra information. In this figure refractive indexes are presented according to the color map and thickness of layers are equal to their respective physical thicknesses.

Accordingly, it can be concluded that the presented algorithm extracts additional data from an OCT scan data which is beneficial in medical indications. Moreover, characterizing refractive index of the layers in the object might have other potential medical and industrial applications. Eventually, this method is beneficial when raw spectral data of the scan do not exist or partially unavailable, and the spectral method of [2] can provide more reliable results in case of accessing to the spectral response which is not available for many of the commercially available OCT devices.



(a)

(b)

Fig. 2: (a) OCT image of a healthy retina sample and (b) Result of the presented algorithm on a healthy retina sample

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