Automated Estimation of OCT Confocal Function Parameters from two B-Scans

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Abstract: We present a method for automatically determining the confocal function parameters of an OCT system for quantification of the attenuation coefficient in B-scans.

OCIS codes: (110.4500) Optical Coherence Tomography, (100.0100) Image Processing.

1. Introduction

In optical coherence tomography (OCT), the attenuation of near-infrared light is governed by the Beer-Lambert Law, and the magnitude of this attenuation is characterized by the attenuation coefficient. Several studies have shown the utility of quantifying the attenuation coefficient for medical purposes (e.g., glaucoma diagnosis [1], bladder cancer staging [2]). Accurate quantification of the attenuation coefficient requires knowledge of the confocal function parameters of the OCT system employed (i.e., the focal plane depth and apparent Rayleigh range). Identifying these parameters, however, usually requires a time-intensive calibration process. In this work, we present a simple, fast, and automatic technique for estimating the confocal function parameters of an OCT system from the imagery itself. Due to its simplicity and speed, this measurement can easily be performed in the clinic by medical personnel using clinical instruments. When combined with a technique for estimating the attenuation coefficient, this algorithm enables automated estimation of the attenuation coefficient from clinical systems.

2. Theory

The irradiance at each depth z of an A-scan may be modeled as $I(z) = h(z)\,\beta\,L_0\,\alpha\,\mu(z)\,e^{-2\int_0^z \mu(\theta)d\theta}$, where h is the confocal function, β is the quantum efficiency of the detector, L_0 is the intensity of the light source, α is the fraction of attenuated light that is backscattered, $\mu(z)$ is the attenuation coefficient at depth z, and θ is a variable of integration [3]. The confocal function h describes the intensity profile emitted from and coupled back into the fiber as a function of depth. It may be modeled (for irradiance) as $h(z) = \left(((z-z_0)/z_R)^2+1\right)^{-1}$, where z_0 is the depth of the focal plane and z_R is the apparent Rayleigh range [4]. Note that this model ignores the effect of fall-off.

Our goal is to determine z_0 and z_R. Here, we propose to leverage the common information that may be obtained from two images of the same structure to solve for these variables. Let I_1 and I_2 be two OCT images such that the structure imaged in I_2 represents a vertical translation of the image content in I_1. That is, there exists $\Delta z > 0$ such that $\mu_2(z+\Delta z,:) = \mu_1(z,:)$ for all z. In this case, the structure in I_2 will appear further from the location of zero pathlength delay. Additionally, we assume that the structure in I_2 fits within the Nyquist-limited ranging depth of the system.

Let us consider the values of I to be in units of decibels (dB), the standard unit for presenting A-scan data. The conversion to dB is $I_{\text{dB}} = 10\log_{10}(I)$. Working in dB enhances the contrast across the image, which makes the confocal function parameters easier to determine.

The first step of the algorithm is to determine Δz, which we do with an exhaustive search. That is, we vary Δz to minimize $\|I_1(z,j)_{dB} - I_2(z+\Delta z,j)_{dB}\|_2 \,/\, \|I_1(z,j)_{dB}\|_2$, where j is the index of each A-scan in the B-scan. For each shift, the cost is calculated using all depths for which there are data in both images.

Once Δz is determined, we can eliminate the unknown quantities and find an expression with z_0 and z_R as the only unknowns: $I_1(z,j)_{dB} - I_2(z+\Delta z,j)_{dB}$. The values of z_0 and z_R are found by using an exhaustive search to minimize expression (1), where the L_1 norm is used to improve robustness against outliers.

$$\left\| [I_1(z,j)_{dB} - I_2(z+\Delta z,j)_{dB}] - \left[10\log_{10}\left(\left(\frac{z+\Delta z - z_0}{z_R}\right)^2 + 1 \right) - 10\log_{10}\left(\left(\frac{z-z_0}{z_R}\right)^2 + 1 \right) \right] \right\|_1 \tag{1}$$

Once the focal plane and Rayleigh range are known, the attenuation coefficient for each pixel in the B-scan can be determined using the Depth Resolved Confocal (DRC) algorithm [5], as we have previously shown.

3. Results and Conclusion

OCT measurements were collected with a commercial spectral domain OCT system (λ_0 = 1325 nm, TELESTO, Thor-Labs). Figure 1 shows the result on a five-layered phantom; the attenuation coefficients of the layers were controlled by dispersing TiO_2 in PDMS. Layers 1, 3, and 5 were made with a higher concentration of TiO_2 than layers 2 and 4. Figures 1 (a) and (b) show the input intensity images representing the imaged structure, which are related by a z-translation. The horizontal black arrow indicates the focal plane location is at 1.07 mm, as determined visually by manually translating the sample in and out of the focal plane and identifying the plane of maximum brightness; the bars on the black arrow represent $z_R = 0.28$ mm. The white arrow and bars indicate the values of the focal plane and Rayleigh range (1.11 mm and 0.35 mm, respectively) determined by the algorithm; note, the differences in these values are within the tolerance for accurate quantification of the attenuation coefficient required by DRC (i.e., the estimated focal plane is within a Rayleigh range of the actual focal plane). The discrepancy between the values extracted manually and automatically may be due to the specular nature of the surface of the sample, which is not accounted for in the model of image intensity. Figure 1 (c) shows the attenuation coefficient image calculated by DRC; the alternating structure of attenuation in the layers is clearly seen.

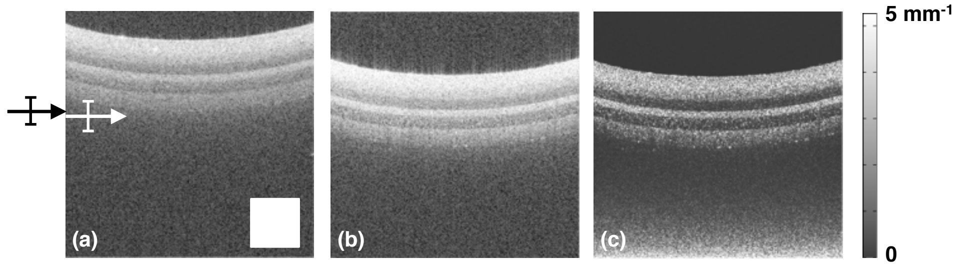

Fig. 1. Results shown on a five-layered phantom. (a) and (b) are the input B-scans. The white square in (a) represents 0.5 mm × 0.5 mm. The black arrow indicates the true z_0; the bars indicate $\pm z_R$. The white arrow and bars indicate the estimated z_0 and z_R. (c) is the resulting attenuation coefficient image. The alternating structure of attenuation in the layers is clearly seen.

In conclusion, we have described a method to automatically determine the confocal function parameters of an OCT system from two B-scans. These parameters are then successfully used to calculate the attenuation coefficient via the DRC algorithm. The ability of this algorithm to work on B-scan images directly (as opposed to raw interferometric data) makes it device agnostic and will allow clinicians to perform automated attenuation coefficient measurements on clinical systems without the need for a laborious calibration process or access to specialized samples.

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