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# Adaptive Beam Divergence, and Intensity Decay Compensation in OCT for Enhanced Tissue Imaging

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**ABSTRACT:**

This paper expands on our foundational research in numerical dispersion compensation within Fourier Domain Optical Coherence Tomography (FD-OCT), by introducing a sophisticated method that synchronizes adaptive beam divergence compensation with our established dispersion correction techniques, now further refined to address intensity decay due to tissue layers reflectance. Our prior work set a significant benchmark, achieving substantial reductions in thickness measurement errors. The present study escalates our pursuit to conquer the added complexities within OCT imaging—beam divergence, intensity decay during light propagation, and dispersion effects due to tissue layer refractive index, which collectively impact image fidelity, especially in the intricate milieu of heterogeneous tissue samples. We have applied this comprehensive approach to a real human tooth, obtaining promising results that underscore the efficacy of the method. The improvements in visualizing tissue microstructures and interfaces were substantial, demonstrating the potential of this integrated approach for complex tissue imaging.

**Key words:**

Fourier Domain Optical Coherence Tomography (FD-OCT); Dispersion Compensation; Adaptive Beam Divergence Compensation; Biomedical Imaging

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## 1. Introduction

Optical Coherence Tomography (OCT) has proven indispensable in biomedical imaging, offering non-invasive, high-resolution insights into biological tissues. As OCT technology progresses, overcoming inherent optical challenges that impair image quality is crucial. Dispersion-induced layers thickness measurement errors, beam divergence, and signal attenuation due to tissue layers reflectance are primary obstacles that limit OCT's diagnostic capabilities, particularly within tissue imaging.

OCT has become a mainstay in high-resolution imaging. However, challenges such as dispersion, beam divergence, and intensity decay during light propagation through tissues necessitate advanced compensation techniques. The evolution of OCT from a novel imaging technique to a fundamental diagnostic tool in various medical fields underscores its significance. Initially introduced for retinal imaging, OCT's capabilities have expanded due to technological advancements, notably in photonics, facilitating its application across ophthalmology, gastroenterology, dermatology, cardiology, and oncology [1], [2]. This development is largely attributed to the advent of Fourier domain OCT (FD-OCT), which significantly enhanced sensitivity and imaging speed, enabling detailed three-dimensional imaging [3]. Such advancements have not only improved clinical diagnostics but also fostered a deeper understanding of tissue physiology and pathogenesis, particularly in ophthalmology [3].

In dermatology, OCT has mirrored advancements seen in ultrasonography, incorporating measures such as blood perfusion and polarization of light, thereby enhancing its clinical utility [4], [5]. This advancement has been particularly notable in diagnosing Nonmelanoma Skin Cancer (NMSC), where OCT's enhanced diagnostic accuracy potentially reduces the need for biopsies. Despite these advancements, challenges remain in diagnosing melanoma due to OCT's resolution limits and the optical properties of melanin [4], [5]. OCT's capability to visualize collagen in normal and fibrotic skin further illustrates its versatility, offering a non-invasive alternative for diagnosing and managing fibrotic skin conditions [6].

The integration of Optical Coherence Tomography Angiography OCTA [7] and Photoacoustic Imaging PAI in dermatology exemplifies OCT's expanding utility, improving the imaging of skin diseases by resolving microvasculature and differentiating skin layers [8]. However, OCT's diagnostic accuracy for non-melanoma skin cancer currently falls short of clinical diagnosis, particularly for inexperienced observers [9]. This limitation underscores the need for continued research and development in OCT technology to enhance its diagnostic capabilities. The review of OCT applications in diagnosing onychomycosis highlights OCT's potential in identifying morphological changes associated with the condition, suggesting its utility beyond mere diagnosis to include treatment assessment in clinical practice [10]. However, further research is

 

needed to refine these criteria and explore OCT's diagnostic accuracy and its distinction from other nail diseases like nail psoriasis.

The adaptation of OCT for in vivo endoscopic optical biopsy represents a significant advancement in medical imaging, offering high-speed, micrometer-scale resolution visualization of tissue structures within a living animal, surpassing the capabilities of traditional imaging technologies [11]. This "optical biopsy" method could revolutionize patient management in fields such as gastroenterology, neurology, cardiovascular medicine, and surgical procedures by enhancing clinical diagnostics in areas where traditional biopsy methods are ineffective, risky, or require high precision.

The sensitivity superiority of swept source (SS) [12] and Fourier domain (FD) OCT techniques over traditional time domain (TD) OCT further underscores the technological advancements in OCT imaging [13]. These advancements facilitate ultrahigh imaging speeds and enhanced visualization capabilities, as demonstrated in swept source/Fourier domain OCT at 1050nm wavelength, achieving imaging speeds of 100,000 to 400,000 axial scans per second [14].

The continuous evolution of OCT technology, marked by significant advancements in sensitivity, speed, and imaging capabilities, has expanded its application across various medical fields. The development of adaptive beam divergence, intensity decay, and dispersion compensation techniques within OCT promises to further enhance tissue imaging, potentially revolutionizing the diagnosis and management of a wide range of conditions. As OCT technology continues to evolve, its integration into clinical practice is expected to provide invaluable insights into tissue physiology and pathogenesis, improving patient care and treatment outcomes.

The following factors arise during imaging using OCT technique:

- Dispersion in OCT has long been recognized as a factor causing error in estimating tissue layers thickness [15].
- Beam Divergence in OCT reduces lateral resolution and causes signal loss leading to error in calculating intensity of reflected signals at tissue layers interface [16].
- Intensity Decay in OCT, as light travels through tissue, reflection at tissue interfaces lead to intensity decay, which has been less addressed in traditional OCT imaging approaches.

The variability in tissue refractive indices and complex microstructures exacerbates the effects of dispersion, beam divergence, and intensity decay, demanding sophisticated, adaptive compensation strategies.

Our work aims to bridge these gaps by introducing a comprehensive method that synergizes adaptive beam divergence compensation with enhanced dispersion correction techniques, now including intensity decay compensation—a tri-fold strategy promising to advance OCT imaging quality significantly.

In prior research, we established a robust framework for addressing dispersion effects in Fourier domain OCT, developing numerical techniques that significantly reduced depth measurement errors and enhanced resolution, based on the refractive index's wavelength dependency. This work laid the foundation for further innovation in OCT imaging [15].

Building upon this, our current study introduces a novel, integrated method combining adaptive beam divergence compensation with advanced dispersion correction techniques, and accounting for the additional factor of intensity decay during light propagation. This method adapts to the specific optical properties of tissues, ensuring high-fidelity imaging across different tissue types.

By applying this method to real tissue samples, focusing on a human tooth, we aim to showcase the effectiveness of adaptive beam divergence, intensity decay compensation, and dispersion correction in enhancing OCT imaging. The promising results set a new standard in image clarity, and diagnostic value, heralding broader applications in biomedical research and clinical practice.

 

## 2. Compensation Strategy Analysis

From the theory of optical coherence tomography, the reflected signals comprising the measured interferogram have the amplitudes given by,

$$I_o(\lambda)\{R_o + R_1 + R_2 + \cdots + 2\sqrt{R_0R_1} + 2\sqrt{R_0R_2} + \cdots + \sqrt{R_1R_2} + \cdots\} \quad (1)$$

$I_o(\lambda)$ represents the intensity of the incident light from the fiber probe tip as a function of wavelength. The first line of this equation represents a DC term, while the second row represents reflectance of tissue layers, and the third row represents unwanted autocorrelation noise. The term $R_o$ represents the reference signal where $R_1$, $R_2$, .... represent the mutual multiple reflectivity of different tissue layers which accounts for autocorrelation noise. $R_o$ is given by,

$$R_o = \left(\frac{n_g - 1}{n_g + 1}\right)^2 \quad (2)$$

Where $n_g$ is the refractive index of the material of the fiber core which is 1.47 in our experiment.

The first step in our algorithm is to approximate the DC term by $I_oR_o$ and hence evaluate the intensity of the reference signal. Then according to figure (1), adjusting for the effect of reflectivity of tissue layers is achieved by multiplying the reflectivity of each layer $R_i$ by the following Reflectivity Correction Factor RCF

$$RCF = \frac{R_i}{\prod_{k=1}^{i-1}(1 - R_k)} \; i = 2,3, \ldots \quad (3)$$

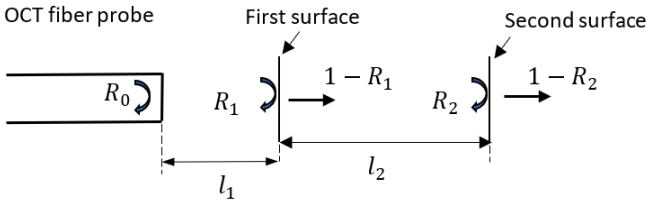


Fig.1. Illustrative representation of an OCT fiber probe interfacing with a two-layer sample. The diagram details the thickness of each layer ($l_1$ and $l_2$) and demonstrates the signal reflectance coefficients (R1, R2) at each interface, as well as the signal transmission coefficients (1 - R1, 1 - R2).

 

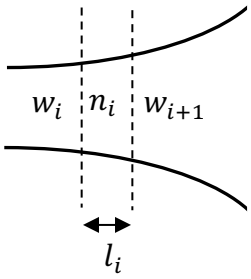


Fig. 2. Depiction of a Gaussian beam profile diverging as it exits the OCT fiber probe and propagates through a sample, with $w_i$ and $w_{i+1}$ representing the beam waist before and after traveling the distance $l_i$, respectively.

The second and third compensation techniques account for beam divergence occurring during its propagation through different sample layers considering tissue material refractive index. Referring to figure (2) and assuming the output from OCT probe as a Gaussian shaped beam, its beam width varies as it traverses a certain layer according to the following relation,

$$w_{i+1} = w_i\sqrt{1 + \left({}^{l_i}/_{z_R}\right)^2} \tag{4}$$

where:

- $w_{i+1}$ is the beam width at the end of layer $i$
- $w_i$ is the beam width at the entrance of layer $i$
- $l_i$ is the thickness of layer $i$
- $z_R$ is the Rayleigh distance given by

$$z_R = \frac{\pi w_i^2 n_i}{\lambda}$$

Hence the correction factor of beam divergence will be

$$LSF = \left({}^{w_{i+1}}/_{w_i}\right)^2 \tag{5}$$

The locations of the tissue layer interfaces are initially determined by conducting a Fourier transform on the interferogram obtained from each A-scan. Subsequently, to account for the beam divergence and the consequent reduction in signal power reflected back to the OCT probe from each layer interface, Equation (5) is employed for compensatory adjustments. The following steps explain the application of the Layer Scaling Factor (LSF) algorithm based on equation (5):

1. Determine Tissue Interfaces:
    - Use Fourier Transform on the interferogram from each A-scan to identify the locations of tissue layer interfaces.
2. Beam Divergence Effect:
    - As light travels through tissue layers, beam divergence causes the beam to spread out, reducing signal power.
3. Calculate LSF:
    - Compute the LSF for each layer using equation (5)

 

4. Apply LSF:
    - Adjust the signal power for each layer based on the calculated LSF to compensate for beam divergence and maintain signal strength.
5. Preserve Image Fidelity:
    - By applying LSF, the OCT system corrects the divergence effect, ensuring accurate imaging across different depths.

This compensation maintains consistent image quality, enhancing visualization of complex tissue structures and allowing detection of subtle features for accurate diagnosis and treatment. These steps streamline the application of the algorithm while ensuring clarity and detail in OCT imaging.

In the subsequent analysis, we employ the developed correction algorithms to enhance the visualization of a treated human tooth that has been coated with a composite layer. This application aims to demonstrate the efficacy of our algorithms in providing clearer and more detailed images of dental structures that have undergone restorative treatment. The focus will be on assessing the composite layer's integration with the tooth's natural structure and identifying any anomalies that may indicate underlying issues or the quality of the treatment applied.

## 3. Results and Discussion

The implementation of our tailored algorithms has markedly improved the interpretative quality of Optical Coherence Tomography (OCT) images. Specifically, the application of these algorithms to a human tooth treated with a composite material illustrates their effectiveness. The adaptive beam divergence compensation algorithm has sharpened the OCT images by addressing the light beam's expansion within the tissue, which often results in a loss of detail at greater depths. This dynamic adjustment preserves the integrity of the OCT image, maintaining high resolution even as the light beam travels deeper into the tooth's structure.

The intensity decay algorithm counters the natural diminishment of the light signal due to absorption by the various tissue layers. Traditionally, this decay would veil the intricate details of the tissue's microarchitecture, but the adaptive compensation ensures these microstructures are captured with clarity, enhancing the image's diagnostic value. Moreover, our refined dispersion compensation technique has successfully mitigated the smearing effects typically caused by the wavelength-dependent refractive index of the composite material within the tooth. By iteratively optimizing the algorithm for the tooth's specific optical properties, we have substantially improved the OCT image's resolution, unveiling details that were previously obscured.

The algorithms' combined application has yielded OCT images with remarkable detail and contrast, enabling precise identification of nonhomogeneous zones within the composite material. These zones, which appear as bubbled or irregular areas, could indicate underlying issues like air entrapment or incomplete curing of the composite, which are critical for dental diagnostics and subsequent treatments.

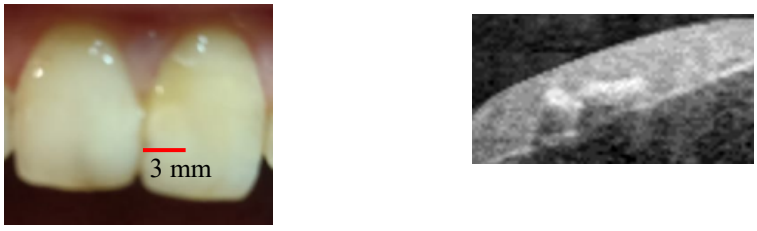


Fig. 3. Optical Coherence Tomography (OCT) B-scan of a human tooth after composite treatment. The scan reveals an area within the composite layer that exhibits a nonuniform texture with bubble-like features.

 

## 4. Conclusions

This research advances our numerical techniques to accommodate the diverse optical profiles of tissues, integrating a dynamic, wavelength-specific correction algorithm that adjusts in real-time to the tissue's optical properties. This novel, adaptive approach results in a tri-fold correction: managing beam divergence, compensating for intensity decay, and mitigating dispersion, thereby yielding significant improvements in image clarity and depth resolution.

The integration of the proposed adaptive algorithms has led to a significant leap forward in OCT imaging capabilities, delivering a higher level of detail and improved visualization of complex tissue structures. These advancements not only facilitate a more accurate assessment of dental treatments but also enhance the overall potential of OCT for broader clinical applications.

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