

Implementation of spectral optical coherence tomography for applications in Odontology

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Abstract

Optical coherence tomography (OCT) is a low depth imaging technique based on a low coherence interferometer. The back scattered light intensity as function of depth can be obtained through different configurations and various methods have been proposed and implemented. Reported here is the implementation of spectral OCT (SOCT). Preliminary results are shown, and an image of tooth enamel is presented.

Introduction

Optical Coherence Tomography [1] has, successfully, been able to produce micrometer scale cross-sectional *in vivo* images of the human retina [2] in near real time refresh rates. Others medical applications were reported such as diagnoses of skin diseases [3], burn depth measurement [4], and red blood cells flow [5]. Our group has been working with dental imaging, being able to identify carious tissues [6] and restorations failures [7].

Till recently, we have been working solely with time domain OCT (TD-OCT), which works by interfering the light coming from the sample with the light coming from a moving mirror [6,7]. With this method, the interference signal is recorded as a function of the mirror displacement, obtaining in this way, the intensity of the back reflected light from the sample as function of depth. To generate a 2D image, the sample is moved transversely and another set is taken, repeating this process until the desired image is formed. The problem with this approach is that the mechanical sweeping limits the scanning rate. Spectral OCT (SOCT), on the other hand, uses the signal spectra to obtain the sample depth information [2]. The availability of spectrometers with CCD arrangements having microseconds integration time, allows the implementation of high speed OCT systems. Also, it was demonstrated [2] that SOCT has superior sensitivity than TD-OCT, making this configuration very attractive.

The aim of this work is to describe the implementation of an SOCT system and exploit its applications to dental imaging, to provide a non invasive real time diagnostic tool.

Theoretical Basis

To obtain a theoretical insight, consider a beam of light, with Gaussian profile whose width is σ and centered on $\omega_0 = \omega + \Delta\omega$, propagating through an interferometer with arm R with length x_r and an arm S with length $x_r + \delta$. The output complex field would be:

$$E_{out} \propto \exp\left(-\frac{\Delta\omega^2}{2\sigma}\right) \exp(n\omega x_r / c - \omega t) + \exp\left(-\frac{\Delta\omega^2}{2\sigma}\right) \exp\left(\frac{n}{c}\omega(x_r + \delta) - \omega t\right), \quad (1)$$

And the calculated intensity would then be,

$$I_{out} = |E_{out}|^2 \propto \exp(\Delta\omega^2 / \sigma) \left[2 + \cos\left(\omega \frac{n\delta}{c}\right) \right], \quad (2)$$

It can be shown that the output power spectrum would have a Gaussian line shape with a cosine modulation, with period inversely proportional to the difference of the arms length δ , as shown in Figure 1.a. To obtain the arms path difference δ one simply have to take the amplitude of the Fourier Transform, depicted in figure 1.b, on of the output intensity,

$$F[I_{out}] \propto \exp(-\sigma\pi^2 x^2) + \exp\left[-\sigma\left(\frac{2\pi x - n\delta/c}{2}\right)^2\right] + \exp\left[-\sigma\left(\frac{2\pi x + n\delta/c}{2}\right)^2\right], \quad (3)$$

The first term on the right hand side is due to the non-interfering parcel of the intensity, the remaining terms are Gaussian curves centered on $\delta n/c$ with σ dependent width. The equation 3 shows the inversely proportional resolution dependency with the light bandwidth. The mirror image occurs because of the Real nature of the light intensity halves the total accessible scan depth, and solutions to avoid this artifact have already been demonstrated [8]. The main idea is to acquire a second set of data with known phase delay, and from that obtain the complex intensity whose Fourier transform is free of the mirror image.

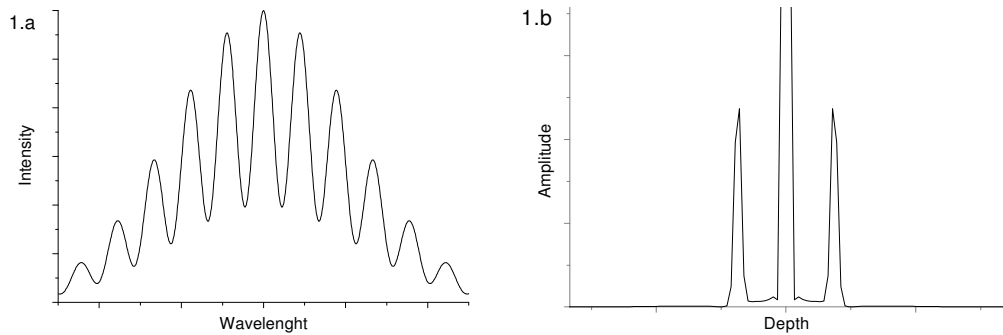


Figure 1: a. Output intensity, I_{out} , spectrum of different length arms interferometer. b. Fourier transform of I_{out} .

In a multilayered sample, the spectrum will consist of the sum of the intensities due to different layers, plus some small parasitic terms due to the interference among the different layers. By taking the Fourier Transform we can obtain the information about sample layers reflectivity.

Experimental Setup

Figure 1 show the basic setup used in our experiment. A femtosecond Ti:Sapphire laser (150 fs), operating at 76 MHz, with $\Delta\lambda = 10$ nm was adjusted to have its central wavelength at $\lambda_0 = 800$ nm. The light was then propagated through a single-mode fiber (for 800 nm) that broadened, through nonlinear effects (self-phase modulation), the bandwidth to $\Delta\lambda = 50$ nm. The tooth was placed in a computer controlled XYZ translation stage and an achromatic lens ($f = 50$ mm) was used to collimate the beam to a <30 μm waist. A piece of glass was placed in the reference arm to match the dispersion that is introduced by using the collimating lens on the sample arm. The output light was collected in a spectrometer (Ocean Optics) that uses a 2048 array CCD which was manufactured to operate from 520 nm to 1186 nm. Since we were only interested on the band of 100 nm centered around 800 nm, we have effectively only used 300 pixels, which limits the fringe contrast. The system was controlled by a program in LabVIEW, which processed the data after acquisition.

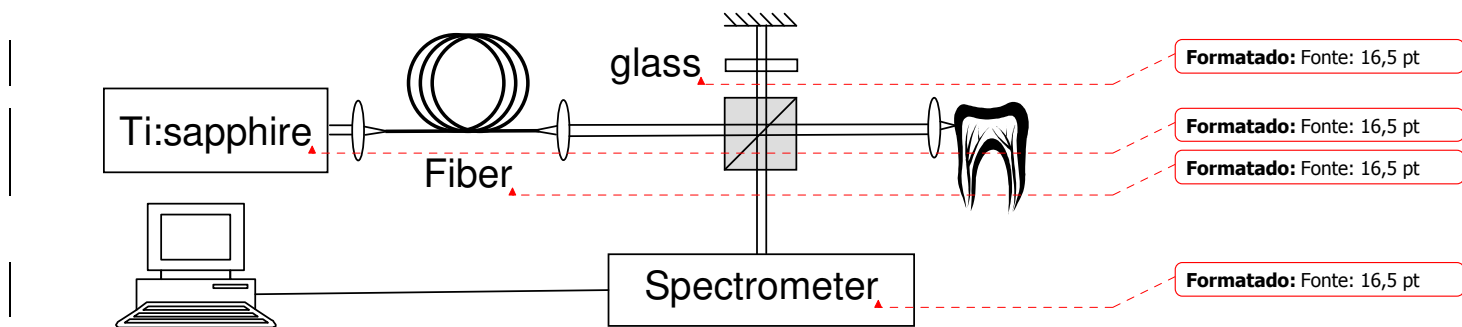


Figure 1: A fiber is used to broaden the light spectrum and a piece of glass was used to match the dispersion.

Results and Discussions

Shown in figure 2 is image generated of an extracted human tooth. It is a mirrored image, and it shows part of the enamel. The surface of the tooth can be identified, and the image corresponds of a 3x1 mm section. A 30 μm step size was used giving us an image of 150x100 points in a 5s scanning time. For comparison, an illustrative drawing of the part of tooth measured is shown in fig. 3, which corresponds to half of figure 2. The black line in the center is also an artifact occurring at the zero delay point.

As expected, in the measurement a mirror image was formed after the Fourier transform of the detected signal. The zero delay point was positioned in the middle of the sample (see arrow in the picture). One way to avoid this is to position the zero delay outside of the sample or to use a much complex detection configuration to obtain the complex information of the signal, therefore being able to separate the Real and the Imaginary parts of the Fourier transform [8]. Some artifacts are also present in the calculation and appear in the picture, but with the correct algorithm they can be properly removed.

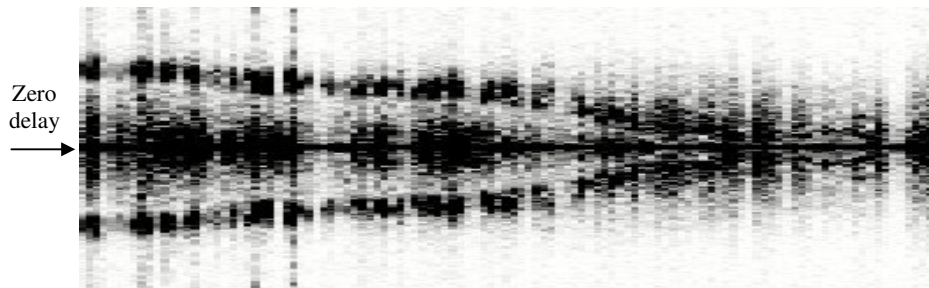
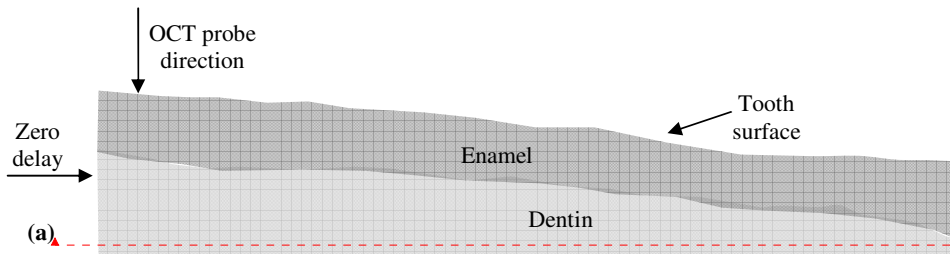
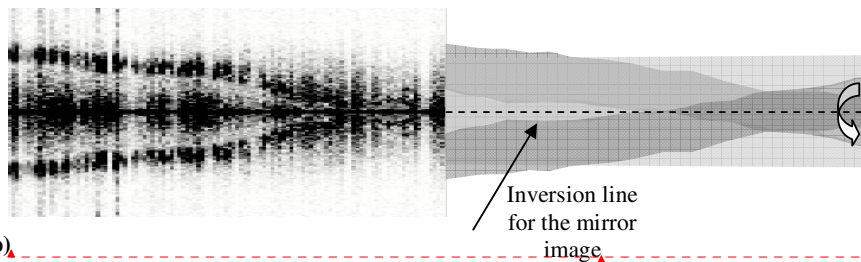


Figure 2: Mirrored image of a 3x1 mm section of a tooth.



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Figure 3: Schematic drawing. (a) Illustrative drawing of the imaged tooth, showing the enamel, the dentin and the tooth surface. We believe we adjusted zero delays point in the dentin, but our results cannot confirm that.

(b) Comparison and illustration of the mirrored images. The inversion line corresponds to the zero delay, the image is duplicated and flipped around this line.

Conclusions

We showed that we were able to make a rough image of a tooth surface, which still contains some artifacts that can be removed with appropriate experimental design. The preliminary results raise good expectations on the possibility to dominate the SOCT technique, develop a compact device, and apply as a diagnostic tool for dentistry.

Acknowledgements

The authors thank the CNPq (BBCK) and CAPES (MTC) for financial support and scholarship provided.

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