

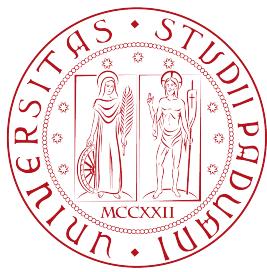
UNIVERSITÀ DEGLI STUDI DI PADOVA
DEPARTMENT OF INFORMATION ENGINEERING
Master Thesis in Telecommunication Engineering

DESIGN OF REAL-TIME OPTICAL COHERENCE
TOMOGRAPHY

Master Candidate
GIANLUCA MARCON

Supervisor
LUCA PALMIERI

April 18, 2018
Academic Year 2017/2018



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*If one feels the need of something
grand, something infinite,
something that makes one feel
aware of God, one need not go
far to find it. I think that I see
something deeper, more infinite,
more eternal than the ocean in
the expression of the eyes of a
little baby when it wakes in the
morning and coos or laughs
because it sees the sun shining
on its cradle.*

Vincent van Gogh

Dedicated to my niece, Teresa.

April 2018

ABSTRACT

Optical Coherence Tomography ([OCT](#)) is a non-invasive imaging technique that exploits the coherence property of light to generate 2-D (cross-sectional) and 3-D (volumetric) images of a live sample from the backscattered electromagnetic field. OCT imaging has found widespread application in medicine, mainly in the areas of *Ophthalmology* and *Angiography*, but also in industrial processes where non-destructive or contactless testing is necessary.

There are two main categories of OCT systems: Time-Domain OCT ([TD-OCT](#)) and Fourier-Domain OCT ([FD-OCT](#)). As the names suggest, the former technique makes use of time domain measurements, while the latter takes advantage of the frequency contents of the reflected signals. FD-OCT offers significant advantages over TD-OCT, such as faster scanning rates, better imaging resolution, and enhanced sensitivity, while at the same time requiring no mechanical movements of critical components such as lenses or collimators.

In this thesis I focus on a particular FD-OCT technique called Swept-Source OCT ([SS-OCT](#)), which uses a rapidly tunable narrow band laser as a light source. A working SS-OCT system capable of real-time imaging is fully developed, along with the data-acquisition and signal-processing modules needed for a complete tomographic imaging device.

Future development includes the migration of the signal processing stack on a Graphics Processing Unit ([GPU](#)) in order to enhance the performance of the system making use of the General-Purpose computing on Graphics Processing Units ([GPGPU](#)) paradigm. This approach opens up the possibility to implement more advanced and refined OCT schemes, such as Polarization-Sensitive OCT ([PS-OCT](#)) and Speckle Variance OCT ([svOCT](#)).

SOMMARIO

La Tomografia a Coerenza Ottica ([OCT](#)) è una tecnica di imaging non invasiva che sfrutta la proprietà di coerenza della luce per generare immagini 2-D (a sezioni) e 3-D (volumetriche) di un campione in vivo a partire dalla luce retrodiffusa dallo stesso. La tecnica OCT ha trovato ampio utilizzo nel campo della medicina, in particolare nelle aree dell'*Oftalmologia* e dell'*Angiografia*, ma anche in processi industriali in

cui sono necessarie misure non distruttive e senza contatto.

Vi sono due principali categorie di sistemi OCT: TD-OCT (OCT nel dominio del tempo) e FD-OCT (OCT nel dominio della frequenza). Come suggerisce il nome, la prima di queste due tecniche sfrutta delle misure nel dominio del tempo, mentre la seconda utilizza il contenuto spettrale dei segnali riflessi per ricostruire l'immagine del campione in esame. FD-OCT offre vantaggi significativi rispetto a TD-OCT, come velocità di scansione più elevate, risoluzione più fine e migliore sensitività. Tutto ciò avviene senza che vi siano movimenti meccanici di componenti critici come lenti e collimatori.

In questa tesi lavorerò su un particolare sistema FD-OCT chiamato Swept-Source OCT ([ss-OCT](#)) che sfrutta un laser a banda molto stretta e con alta velocità di sintonizzazione. Verrà quindi sviluppato un sistema SS-OCT funzionante, capace di eseguire misure continue e in tempo reale. Il lavoro verterà sulla parte di progettazione e ottimizzazione dello schema ottico e sullo sviluppo di algoritmi per l'acquisizione ed elaborazione dei dati.

Sviluppi futuri verteranno sulla migrazione dell'intero sistema di elaborazione dati su GPU (Graphical Processing Unit) facendo uso del paradigma di General-Purpose computing on Graphics Processing Units ([GPGPU](#)), che renderà più efficiente il dispositivo e permetterà la progettazione di tecniche più avanzate come OCT sensibile alla polarizzazione ([PS-OCT](#)), per ottenere misure di birifrangenza del campione, o Speckle Variance OCT ([svOCT](#)).

In truth, O judges, while I wish to be adorned with every virtue, yet there is nothing which I can esteem more highly than the being and appearing grateful. For this one virtue is not only the greatest, but is also the parent of all the other virtues.

— Marcus Tullius Cicero

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ACRONYMS

OCT	Optical Coherence Tomography
SS-OCT	Swept-Source OCT
TD-OCT	Time-Domain OCT
FD-OCT	Fourier-Domain OCT
SD-OCT	Spectral-Domain OCT
PS-OCT	Polarization-Sensitive OCT
svOCT	Speckle Variance OCT
GPGPU	General-Purpose computing on Graphics Processing Units
GPU	Graphics Processing Unit
MZI	Mach-Zender interferometer
OPD	Optical Path Difference
BPD	Balanced Photodiode
OSA	Optical Spectrum Analyzer
SMF	Single Mode Fiber
EOCT	Endoscopic Optical Coherence Tomography
IVUS	Intravascular Ultrasound

SEM	Scanning Electron Microscope
MRI	Magnetic Resonance Imaging
DOCT	Doppler OCT
PEG	Photonic and Electromagnetic Group
DEI	Department of Information Engineering
DAQ	data acquisition
PZT	piezoelectric transducer
ADC	Analog-to-Digital Converter
SLD	Superluminescent Diode
CCD	charged coupled device
CMOS	complementary metal-oxide semiconductor
SNR	signal-to-noise ratio
FWHM	full-width half-maximum
PSD	power spectral density
OFDI	optical frequency domain interferometry
FFT	Fast Fourier Transform
MEMS	micro electro-mechanical systems
PZT	piezoelectric transducer
EOM	electro-optic modulator
SOA	semiconductor optical amplifier
FDML	Fourier domain mode locking
VCSEL	vertical cavity surface-emitting laser
FPGA	Field Programmable Gate Array
SDK	Software Development Kit
FIFO	First-In First-Out
DMA	Direct Memory Access
LVTTL	Low-Voltage Transistor-Transistor-Logic
OFDR	Optical Frequency-Domain Reflectometer

INTRODUCTION

1.1 SUMMARY ON OPTICAL COHERENCE TOMOGRAPHY

The technique of Optical Coherence Tomography ([OCT](#)), firstly introduced in the late 80's by Fujimoto et al. [19] and later improved by Huang et al. [23], was developed for the noninvasive axial and cross-sectional imaging of biological tissues. In a way similar to ultrasonic imaging, 2D images are generated by combining the incident electromagnetic radiation with its delayed version reflected by the sample under test.

The advantage over conventional ultrasound techniques is that it permits imaging resolutions that range from 1 to 20 μm , which is up to two orders of magnitude smaller than for ultrasound. This enables the diagnosis of pathologies which were previously only detectable by histological techniques, which have the advantage of higher resolution at the expense of the capability of non-destructive measurement. In fact, histology requires the following steps to be performed on the sample:

- Excision
- Fixation
- Embedding
- Microtoming
- Staining

These operations prevent the use of histology in certain areas where the sample to be analyzed cannot be damaged, such as *Ophthalmology*, a field of medicine which studies the pathologies of the human eyeball and orbit. These characteristics, coupled with its real-time imaging capability, make [OCT](#) one of the strongest candidates for this particular branch of medicine.

In [Figure 1.1](#) we can compare the images of a human retina obtained with the first type of OCT technique invented, Time-Domain OCT ([TD-OCT](#)) ([Figure 1.1a](#)) and histology ([Figure 1.1b](#)) [23]. Image quality, while arguably low, is enough to identify and measure the thickness of the main structures of the eye. In [Figure 1.1c](#) we can instead see a high quality image of a live sample obtained with the more modern Swept-Source OCT ([SS-OCT](#)) technique [14], which allows higher quality measurements and fast image acquisitions.

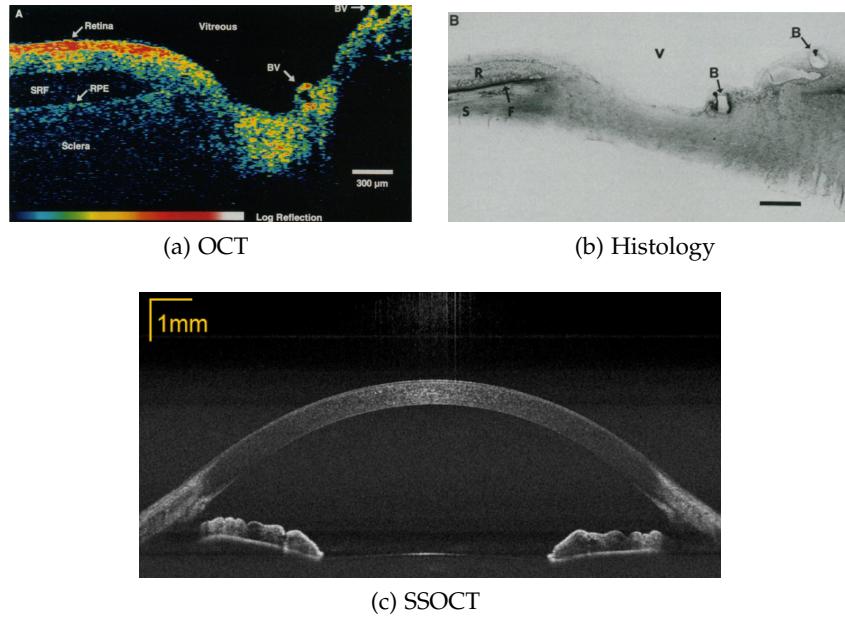


Figure 1.1: Optical coherence tomograph of human retina and optic disk in vitro (top left) and histologic section of the same sample (top right) [23]. On the bottom, a SS-OCT image of a live human eye sample [14].

The first drawback of OCT as a medical imaging technique comes from its relatively low imaging penetration, which is usually between less than a millimeter up to a couple of centimeter, depending on the specific technology and the absorption coefficient of the sample to analyse. In Figure 1.2, a comparison between OCT, Ultrasound, and Confocal Microscopy is available [15]. The trend is that for an increasingly better imaging resolution, imaging depth has to be sacrificed. In this aspect, OCT sits right between the other two techniques, offering micrometer-level resolution for a moderate image penetration.

Whenever cross-sectional images are not sufficient for a correct diagnosis, volumetric data can be exploited for a more in-depth analy-

SOURCE	YEAR	VOXELS	VOL. RATE	SPEED
			Volumes/s	GVoxels/s
Zhang and Kang[47]	2010	6400000	10	0.06
Choi et al. [10]	2012	4194304	41	0.17
Wieser et al. [44]	2014	40960000	26	1.07
Darbrazi [12]	2016	1408800000	2.05	2.89

Table 1.1: Performance of different implementations of volumetric OCT using GPGPU (adapted from [12]).

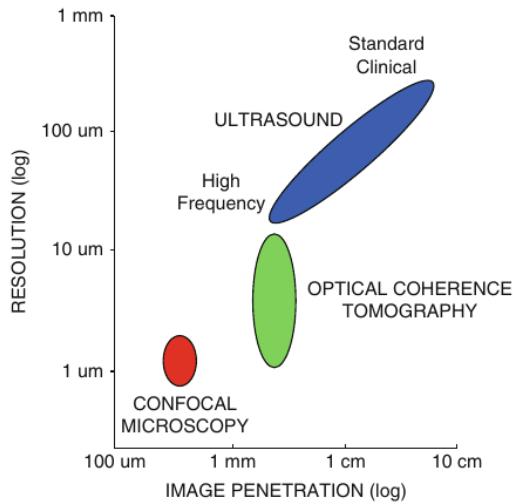


Figure 1.2: Imaging resolution and image penetration compared for different imaging techniques.

sis. Volumetric images are composed by multiple 2D images captured in succession along a certain scanning direction, creating a 3D grid of datapoints called Voxels. . Two examples of 3D-OCT data obtained with SS-OCT (left) [14] and SD-OCT (right) [10] are depicted in Figure 1.3.

With the near-exponential growth in computing power over the last decades, and the advent of GPU computing by means of the GPGPU paradigm, advanced volume rendering and signal-processing algorithms can be applied on large OCT data sets in real-time. In Table 1.1 a few results from the literature are summarized, highlighting the advancement in processing power using GPU solutions (adapted from [12]).

Starting from 3D-OCT data it is also possible to generate *en-face* projections of the sample at different depths, providing an invaluable tool for further analysis. A result of this technique is available in Figure 1.4, which shows a frontal view of the macula, the central part of the retina, affected by an edema.

1.2 APPLICATIONS

1.2.1 Medicine

Ophtalmic applications have already been briefly mentioned in Section 1.1, but other areas of medicine such as Cardiology [7, 24] and Angiography [25, 39] have benefitted from the diagnostic capabilities of OCT. A comparison between OCT and Ultrasound images of a coronary plaque is available in Figure 1.5: the higher resolution of the optical system is substantial.

A voxel is the primary unit of three-dimensional datasets, like the pixel is for 2D data. The word voxel originated by analogy with the word "pixel", with *vo* representing "volume" and *el* representing "element"

Angiography or arteriography is a medical imaging technique used to visualize the inside, or lumen, of blood vessels and organs of the body

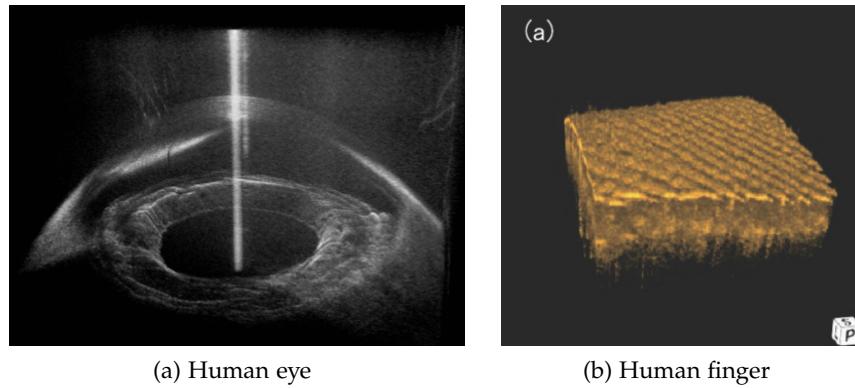


Figure 1.3: Example of 3D OCT data: human eye obtained with SS-OCT [14] (left) and human finger obtained with SD-OCT [10] (right).

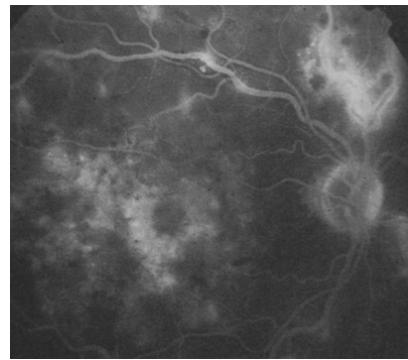


Figure 1.4: En-face view for the assessment of macular edema [32].

These applications are made possible by the small footprint of optical fibers (in the order of $200\text{ }\mu\text{m}$ of diameter) and the use of micro electro-mechanical systems (**MEMS**) mirrors, which can be easily embedded in endoscopes, catheters or other special probes [29, 42]. An example of a focus-adjustable probe is depicted in Figure 1.6.

Endoscopic Optical Coherence Tomography (**EOCT**) became an important tool for the detection of cancers affecting different parts of the human body, including bladder [46], cervix [17] and colon [22]. Other applications in the field of medicine include dermatology and skin damage assessment [21, 26] and dentistry [2, 30]. Low-delay, real-time OCT systems are also employed as a guidance tool for surgeries or tissue removal [4, 5], allowing micrometer-scale resolution and providing depth-resolved images which are unobtainable with other classical methods.

1.2.2 Industrial

OCT has also found widespread application in a variety of non-medical fields, especially where non-contact, high-precision measurements are needed. For example, real-time monitoring and thickness measure-

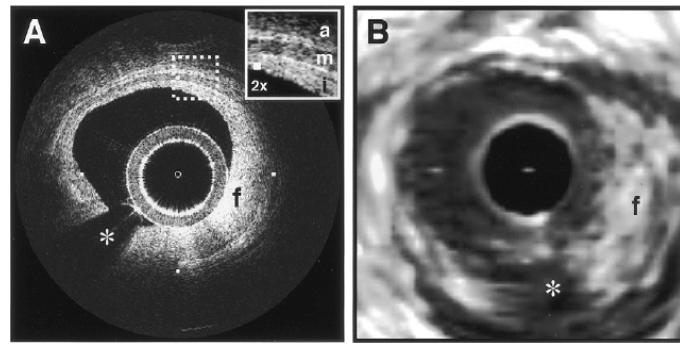


Figure 1.5: Coronary plaque imaged by OCT (left) and Intravascular Ultrasound (IVUS) (right) [24].

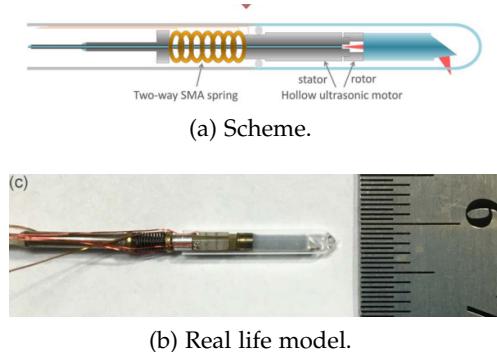


Figure 1.6: Probe for endoscopic OCT [29].

ment of multi-layer structures are important tools in the manufacturing of microelectronics and optical devices. Industrial uses of OCT range from defect detection in ceramic and polymeric materials [40, 43] to quality evaluation of paper products [1, 34]. An interesting usage is found in [28], where the non-invasive examination of museum paintings was demonstrated, paving the way for OCT to the field of art conservation.

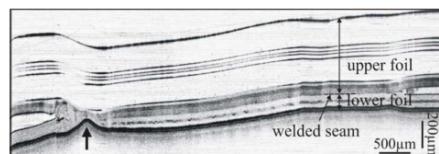


Figure 1.7: Cross-sectional OCT image of a multi-layered plastic foil used in the food packaging industry [43].

1.2.3 *In vivo monitoring of biological specimen*

We've seen that non-destructive measurements are essential in human medical diagnostic, but *in vivo* cross-sectional analysis is suited for a wide range of other biological samples. OCT has been used

for growth monitoring of seeds [35], virus detection in plants [11], and even quality assessment of egg quality in the poultry industry [36]. Alternative methods such as histology, Scanning Electron Microscope (SEM) imaging, Magnetic Resonance Imaging (MRI) and X-Ray radiography are either destructive or can not guarantee the possibility of continuous monitoring with a comparable image resolution.

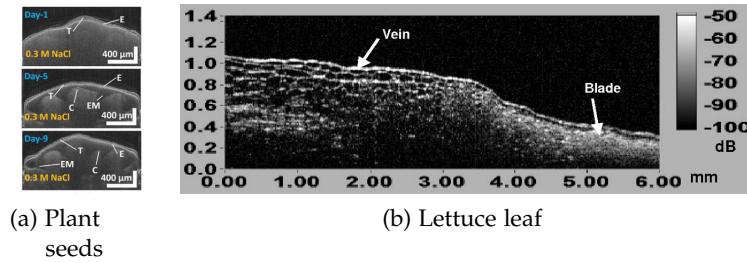


Figure 1.8: Growth monitoring of *Capsicum annuum* seeds [35] (left) and lettuce leaf [20] (right).

1.2.4 Functional imaging

Apart from the structural imaging of biological tissue, OCT can also be utilized to perform *functional* imaging, giving the user insights on the different properties of the material under analysis. PS-OCT schemes can measure properties such as birefringence, dichroism and optic axis orientation of the sample [3], while Doppler OCT (DOCT) and svOCT are able to estimate the direction and velocity of the blood flow in vessels [31]. Birefringence measurement has found use in dentistry, specifically in the monitoring of caries lesions and their progression, enabling early detection and preventing the need for surgical intervention [18].

1.3 OBJECTIVES

The study conducted in this thesis is based on a previous thesis [9] developed in the Photonic and Electromagnetic Group (PEG) Laboratory at the Department of Information Engineering (DEI) of the University of Padova. It consisted in the preliminary characterization of the main components of a high speed Swept-Source OCT (SS-OCT) system working in the 1300 nm range and the experimental determination of different parameters of the device, such as the source coherence length, the achievable scanning speed and the transversal resolution permitted by the focusing optics.

The primary objectives of this thesis are the following:

1. Designing and testing the optical circuitry needed for a stable SS-OCT system.
2. Developing a data acquisition ([DAQ](#)) application capable of continuous and low-delay video stream.
3. Control and synchronization of the Galvanometric System with the optical source and the [DAQ](#) board.
4. Acquiring cross-sectional and volumetric data of different samples.

1.4 THESIS STRUCTURE

The structure of this document is organized in the following manner:

- [Chapter 2](#) consists of the theoretical background on the electromagnetic phenomena of interference and spatio-temporal coherence. The basic operating principle of different OCT schemes will be thoroughly detailed.
- In [Chapter 3](#) I will give a description of the optical and electrical devices that were used. I will also comprehensively describe the data flow of the [DAQ](#) application that was developed.
- [Chapter 4](#) is dedicated to explaining the methods that were used in designing the system and assessing its performance. It will also showcase a series of cross-sectional images and basic volume renderings obtained from a variety of different samples.
- [Chapter 5](#) concludes the thesis, illustrating possible future developments and further optimization of the presented work.

2

BASIC THEORY OF OPTICAL COHERENCE TOMOGRAPHY

In this chapter I will report the basic theoretical background needed to comprehend the working mechanisms behind the OCT technique, introduce the different schemes and compare them in terms of performance. The content of this chapter is partially adapted from [6, 33, 38].

2.1 PRINCIPLES OF COHERENCE AND INTERFERENCE

A solution to a generic electromagnetic problem is completely determined by the vector couple $\mathcal{S} = \{\mathbf{E}(\mathbf{r}, t), \mathbf{H}(\mathbf{r}, t)\}$, whose components represent the time-varying electric and magnetic field, respectively (\mathbf{r} is the three-dimensional vector determining the position in space at which the field is evaluated). The electromagnetic field determined by \mathcal{S} is considered a valid solution if it satisfies both Maxwell's Equations and the boundary conditions specific to the problem.

For monochromatic waves, i.e. fields oscillating at a single frequency ω , we can use Steinmetz notation and write

$$\mathbf{E}(\mathbf{r}, t) = \Re[\tilde{\mathbf{E}}(\mathbf{r})e^{j\omega t}] \quad \mathbf{H}(\mathbf{r}, t) = \Re[\tilde{\mathbf{H}}(\mathbf{r})e^{j\omega t}], \quad (2.1)$$

where $\tilde{\mathbf{E}}$ and $\tilde{\mathbf{H}}$ are complex 3D vectors and $\Re[\cdot]$ is the real value operator. Maxwell Equations can then be rewritten in the frequency domain. In a linear, homogeneous, isotropic, and current-free medium, Maxwell's equations can be written in the following way

$$\nabla \times \tilde{\mathbf{E}} = -j\omega\mu\tilde{\mathbf{H}} \quad (2.2a)$$

$$\nabla \times \tilde{\mathbf{H}} = j\omega\epsilon_c\tilde{\mathbf{E}} \quad (2.2b)$$

where μ is the magnetic permeability of the medium, $\epsilon_c = \epsilon - \sigma/\omega$ is the complex dielectric permittivity. The simplest solution of [Equation 2.2](#) is given by the *homogeneous plane wave*. Using a cartesian coordinate system, the electric field can be expressed as

$$\tilde{\mathbf{E}}(\mathbf{r}) = \sum_{i=1}^3 A_i \cdot \hat{\mathbf{x}}_i = \sum_{i=1}^3 a_i \exp[jg_i(\mathbf{r})] \cdot \hat{\mathbf{x}}_i \quad (2.3)$$

where the amplitudes a_i are constant and $g_i(\mathbf{r}) = \mathbf{k} \cdot \mathbf{r} - \delta_i$ represent the phase functions of the electric field components. The propagation vector \mathbf{k} is given as a function of the wavelength λ as follows

$$|\mathbf{k}| = \frac{2\pi}{\lambda}, \quad (2.4)$$

while the δ_i 's are the phase differences which determine the state of polarization of the electromagnetic field. The magnetic field is instead obtained by applying [Equation 2.2a](#) to [Equation 2.3](#).

The intensity of an electromagnetic field is given by the time average of the amount of energy which crosses in a unit time a unit area perpendicular to the direction of the energy flow. In the case of homogenous plane waves, it is expressed as

$$I \propto \langle \mathbf{E}^2 \rangle \quad (2.5)$$

Using [Equation 2.1](#), we can express the electric field as

$$\mathbf{E}(\mathbf{r}, t) = \frac{1}{2} \left[\tilde{\mathbf{E}}(\mathbf{r}) e^{-j\omega t} + \tilde{\mathbf{E}}^*(\mathbf{r}) e^{j\omega t} \right], \quad (2.6)$$

so that [Equation 2.5](#) can be written as

$$\langle \mathbf{E}^2 \rangle = \frac{1}{4} \left\langle \tilde{\mathbf{E}}^2 e^{-2j\omega t} + \tilde{\mathbf{E}}^{*2} e^{2j\omega t} + 2\tilde{\mathbf{E}} \cdot \tilde{\mathbf{E}}^* \right\rangle. \quad (2.7)$$

Averaging over a time interval sufficiently larger than the period $T = 2\pi/\omega$, we obtain

$$\langle \mathbf{E}^2 \rangle = \frac{1}{2} \tilde{\mathbf{E}} \cdot \tilde{\mathbf{E}}^* = \frac{1}{2} (a_1^2 + a_2^2 + a_3^2) = \text{constant}, \quad (2.8)$$

as the high frequency terms at 2ω are canceled by the detector.

Suppose now that the field \mathbf{E} is split in two components \mathbf{E}_1 and \mathbf{E}_2 which are then superimposed in a certain point \mathbf{r}' in space, then

$$\mathbf{E}(\mathbf{r}') = \mathbf{E}_1(\mathbf{r}') + \mathbf{E}_2(\mathbf{r}'), \quad (2.9)$$

which implies that

$$\mathbf{E}^2(\mathbf{r}') = \mathbf{E}_1^2(\mathbf{r}') + \mathbf{E}_2^2(\mathbf{r}') + 2\mathbf{E}_1(\mathbf{r}') \cdot \mathbf{E}_2(\mathbf{r}'). \quad (2.10)$$

The field intensity is thus

$$I = I_1 + I_2 + J_{12} = \langle \mathbf{E}_1^2 \rangle + \langle \mathbf{E}_2^2 \rangle + 2\langle \mathbf{E}_1 \cdot \mathbf{E}_2 \rangle. \quad (2.11)$$

The last term in [Equation 2.11](#) is called *interference*. Depending on the phase difference between the two fields, the total intensity can assume different values. If the difference between the optical paths traveled by the two components is Δz , the phase difference will be $\delta = \Delta z \cdot 2\pi/\lambda$, and the interference term will be equal to

$$J_{12} = 2\langle \mathbf{E}_1 \cdot \mathbf{E}_2 \rangle = (a_1^2 + a_2^2 + a_3^2) \cos \delta. \quad (2.12)$$

In the simple case in which the two fields are linearly polarized along the x_1 direction, we have that $a_2 = a_3 = 0$, and $I_1 = I_2 = 1/2 a_1^2$, while the interference is given by $J_{12} = a_1^2 \cos \delta = 2\sqrt{I_1 I_2} \cos \delta$. The total intensity is then

$$I = I_1 + I_2 + 2\sqrt{I_1 I_2} \cos \delta = 2I_1 + 2I_1 \cos \delta \in [0, 4I_1]. \quad (2.13)$$

The intensity profile is displayed in [Figure 2.1](#). In this first approximation of perfectly monochromatic waves there are no constraint on the phase offset δ or on the optical path difference Δz for the presence of interference.

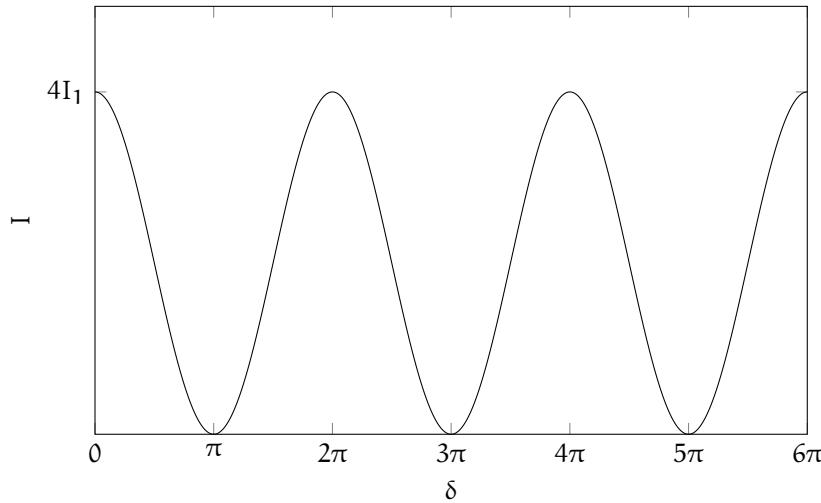


Figure 2.1: Interference fringes created by two beams of equal intensity.

2.1.1 *Michelson interferometer*

A classical example of the application of interference is given by the scheme called *Michelson interferometer*. In [Figure 2.2](#) the schematic diagram of the interferometer is given. A light source emits an electric field \mathbf{E}_0 which is then split in two by a semitransparent mirror. The two replicas, \mathbf{E}_1 and \mathbf{E}_2 , travel along the two *arms* of the interferometer, of length l_1 and l_2 , are reflected by two mirrors and finally recombined. The intensity of the superposition of two fields is then measured by a photodetector. The two replicas arrive at the detector with a time difference given by

$$\tau = 2 \frac{l_2 - l_1}{c}, \quad (2.14)$$

where c is the speed of light in the considered medium. From [Equation 2.11](#) we then obtain

$$I = I_1 + I_2 + 2\langle \mathbf{E}_1(t) \cdot \mathbf{E}_2(t) \rangle = 2[I_1 + \langle \mathbf{E}_1(t)\mathbf{E}_1(t-\tau) \rangle] \quad (2.15)$$

$$= 2I_1 \left\{ 1 + \cos \left[\frac{2\pi}{\lambda} 2(l_2 - l_1) \right] \right\}. \quad (2.16)$$

This type of interferometer can be used to perform high-precision measurements of distances by making one of the arms mobile and maintaining the other fixed as a reference. Counting the number of "peaks" and "valleys" of the intensity profile as the mobile arm is translated, an estimate of path length difference is given with a precision of $\lambda/2$.

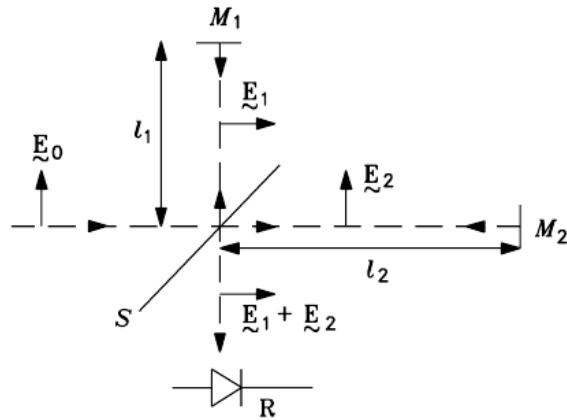


Figure 2.2: Diagram of a Michelson interferometer [38].

2.1.2 Fringe visibility

We can define a useful parameter called *fringe visibility* as follows

$$\nu = \frac{I_{\max} - I_{\min}}{I_{\max} + I_{\min}}. \quad (2.17)$$

For a perfectly monochromatic source, ν will always be constant. In particular, in the case when $I_1 = I_2$ ν is equal to 1, as $I_{\min} = 0$. Real-life optical sources however will always have a bandwidth Δf greater than 0. In these cases ν is a monotonically decreasing function of the time delay τ . The *coherence time* of a source is defined as the value τ_c such that $\nu(\tau_c) = 1/e$ (Figure 2.3).

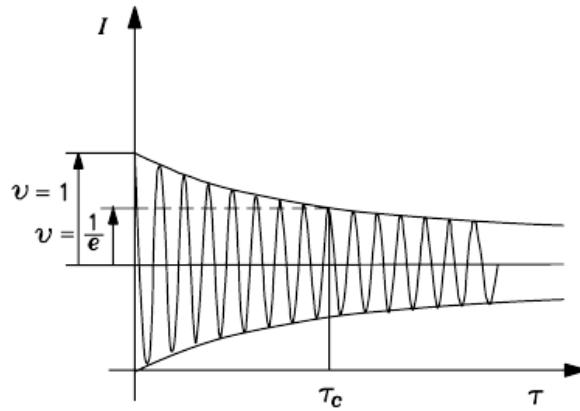


Figure 2.3: Effect of the coherence time of a source on the interference fringes at the output of a Michelson interferometer [38].

Intuitively, we may think of the coherence time as the time slot after which the optical source loses memory of what it was at the beginning. In fact, after this time slot has passed, some properties of the source randomly change due to the stochastic nature of photon

emission. A perfectly coherent source emits a sinusoidal field with a well defined and constant phase relation. Similarly, two electromagnetic fields are perfectly coherent if the phase difference between the two is maintained constant for an infinite amount of time. This property is however impossible to obtain in real life, as every source has a finite coherence time. This concept is explained by the uncertainty-principle, which states that

$$\tau_c \geq \tau_{\min} = \frac{1}{B}, \quad (2.18)$$

where B is the bandwidth of the source. This equation results in the fact that there are no sources which are completely incoherent in time, as they would require an infinite bandwidth.

2.1.3 The coherence function

Another way to describe the coherence property of light other than the fringe visibility parameter, is through the so-called *mutual coherence function*. It is defined for polychromatic, i.e., non-monochromatic fields as follows [6]:

$$\Gamma_{12}(\tau) = \langle E_1(t + \tau) E_2^*(t) \rangle, \quad (2.19)$$

If $E_1 = E_2$, it is called *self-coherence function*, and it's written as $\Gamma_{11}(\tau)$. Notice that when $\tau = 0$ it reduces to the field intensity:

$$\Gamma_{ii}(0) = I_i. \quad (2.20)$$

It is then useful to apply the following normalization

$$\gamma_{ij}(\tau) = \frac{\Gamma_{ij}(\tau)}{\sqrt{\Gamma_{ii}(0)} \sqrt{\Gamma_{jj}(0)}}, \quad (2.21)$$

so that it assumes values in the $[0, 1]$ interval. This normalized function is called *complex degree of coherence*, and it allows the following definitions:

1. Completely incoherent fields: $|\gamma_{ij}| = 0$, no interference fringes are visible.
2. Completely coherent fields: $|\gamma_{ij}| = 1$, the total intensity includes the interference term.
3. Partially coherent fields: $0 < |\gamma_{ij}| < 1$, interference fringes are present, and assume a visibility parameter equal to

$$v = \frac{2\sqrt{I_i I_j}}{I_i + I_j} |\gamma_{ij}|. \quad (2.22)$$

When the two fields have equal intensity, than the two parameters coincide.

Using this notation, the equation for the intensity of two partially-coherent beams is then

$$I = I_1 + I_2 + 2\sqrt{I_1 I_2} |\gamma_{12}(\tau)| \cos(\delta) \quad (2.23)$$

where we can clearly see the effect of coherence on the shaping of the interference fringes. Using the coherence function instead of the fringe visibility as a measure of coherence, the coherence time of a source, τ_c , is defined as the full-width half-maximum (**FWHM**) parameter of its self-coherence function, that is, the value of τ such that

$$|\gamma(\tau)| = \frac{|\gamma(0)|}{2}. \quad (2.24)$$

The shape of the coherence function of an optical source is completely determined by its spectrum $I(k)$ through the Wiener-Khintchine theorem, which states that the autocorrelation function and the power spectral density (**PSD**) of a random process are connected through their Fourier Transform.

2.1.4 Coherence length

Starting from the coherence time of the source, τ_c , we can also define its coherence length as follows

$$l_c = c_0 \tau_c, \quad (2.25)$$

where $c_0 \simeq 3 \cdot 10^8$ m/s is the speed of light in vacuum. Using a light source with a coherence length l_c in the Michelson interferometer previously presented, we would observe an interference pattern at the photoreceiver only if the difference in length of the two arms is such that

$$\Delta l = |l_2 - l_1| \leq l_c/2, \quad (2.26)$$

or, equivalently, if the total Optical Path Difference (**OPD**) is matched to within the coherence length of the source.

While we just defined the coherence *length* of a source, it's important to note that it is a parameter that is directly related to the time coherence and does not describe the phenomenon of spatial coherence.

2.2 OCT TERMINOLOGY

Before introducing the different OCT techniques it is useful to introduce some terminology that will be used throughout this thesis. The most basic OCT measure is called A-scan (Axial-scan), which is a signal that represents the reflectivity of a sample as a function of depth.

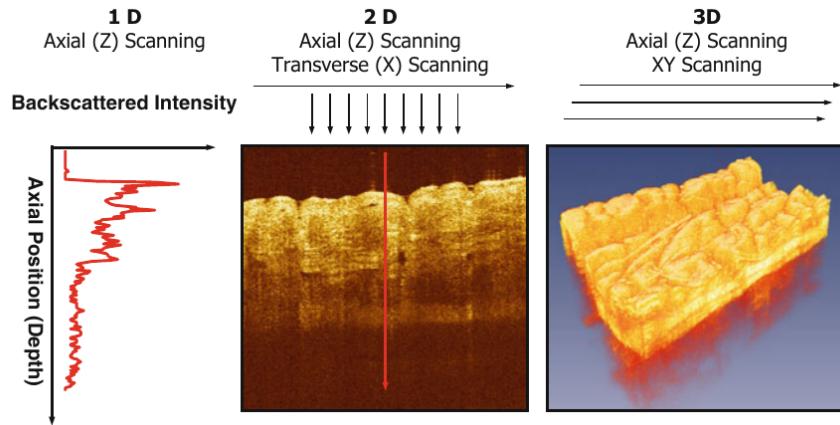


Figure 2.4: Different types of OCT measurements. [15]

The amount of information carried by these measurements is limited, but can be used to perform thickness measurements if the structure of the sample is known a priori.

If multiple consecutive A-scans are acquired along a transverse direction on the sample, a cross-sectional image, called B-scan, is generated. They are the most direct way to determine the anatomy of an unknown sample and are often sufficient for diagnostic purposes, especially in Ophthalmology. Their visualization does not require advanced techniques, but if a real-time high-resolution video stream is desired then careful design choices have to be made.

Finally, 3D volumetric data can be generated by acquiring consecutive B-scans along a second direction on the transverse plane. Intuitively, they are called C-scans. Contrary to B-scans, volume rendering requires sophisticated and efficient algorithms for a correct data interpretation and visualization [8, 16, 27]. Starting from 3D data it's also possible to obtain frontal projections of the sample, called *en-face* views (Figure 2.5), or to generate cross-sections along an arbitrary plane.

2.3 TIME DOMAIN OCT

Time-Domain OCT (TD-OCT) was the first OCT technique that was demonstrated in the literature [23]. The basic setup is that of a Michelson Interferometer in which one of the two mirrors is translatable and the other is replaced by the sample that we wish to analyze. The two arms are called respectively *Reference arm* and *Sample arm*. As explained in Section 2.1.1, the two beams arriving at the photodetector generate interference fringes if the OPD is less than the coherence length l_c of the source. A diagram of this setup is available in Figure 2.6.

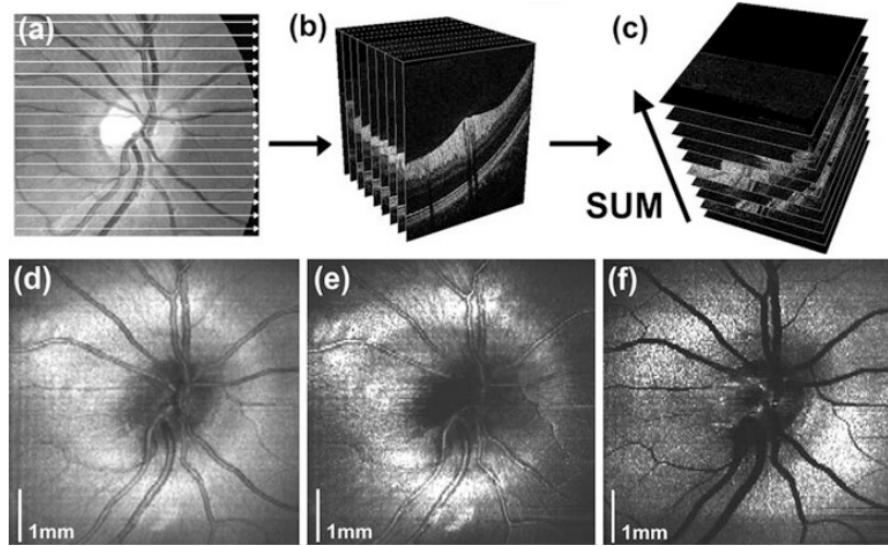


Figure 2.5: [15]

The light beam reflected by the reference mirror while in position z_1 will interfere with all the reflections occurring in the sample at depths $z \in [z_1 - l_c/2, z_1 + l_c/2]$. By measuring the intensity of the fringes it is then possible to obtain an estimate of the sample reflectivity at those depths. The whole axial measurement of the sample can be acquired by moving the translatable mirror along an adequate interval of positions. Since a single mirror position is mapped to an interval of length l_c of axial positions in the sample, the coherence length of the source can be considered to be the axial resolution of the system. As a consequence short coherence lengths are preferred, meaning that broadband sources are more widely employed.

If the light source was perfectly monochromatic, the reference signal would interfere with an infinite number of reflected replicas generated at every depth in the sample, as there would be no constraint on the OPD.

To better understand this concept, suppose that the sample is an ideal reflector with a reflection coefficient ρ_s is positioned in such a way that the OPD is 0. The intensity measured by the detector when a OPD equal to Δz is introduced is then

$$I(\Delta z) \propto |\rho_s|^2 |E_i|^2 + |\rho_r|^2 |E_i|^2 + |\rho_s \rho_r|^2 |E_i|^2 |\gamma(\Delta z)| \cos\left(\frac{2\pi}{\lambda_0} \Delta z\right) \quad (2.27)$$

Apart from the DC offset given by the intensity of the two signals there is an oscillating term with period equal to λ_0 , which is the central wavelength of the source. Figure 2.7 illustrates the oscillating term and its envelope, which is dictated by the coherence function γ and the sample reflectivity ρ_s (set equal to 1). With a perfectly

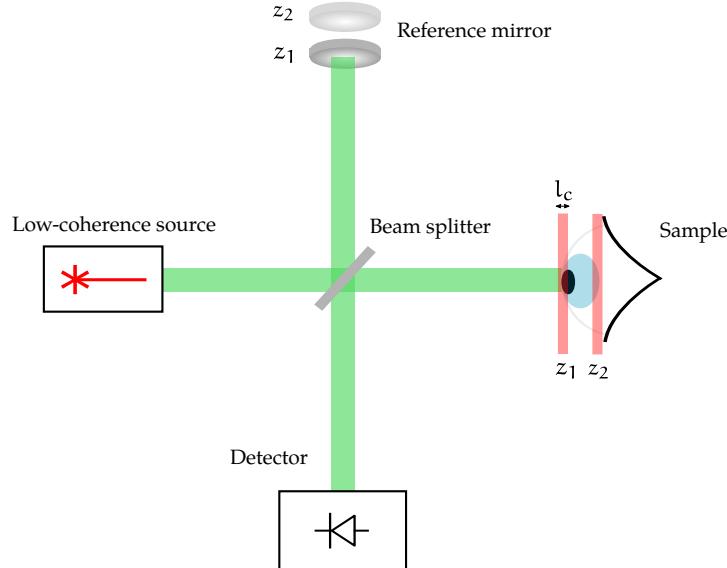


Figure 2.6: Diagram of the basic TD-OCT setup using a Michelson interferometer.

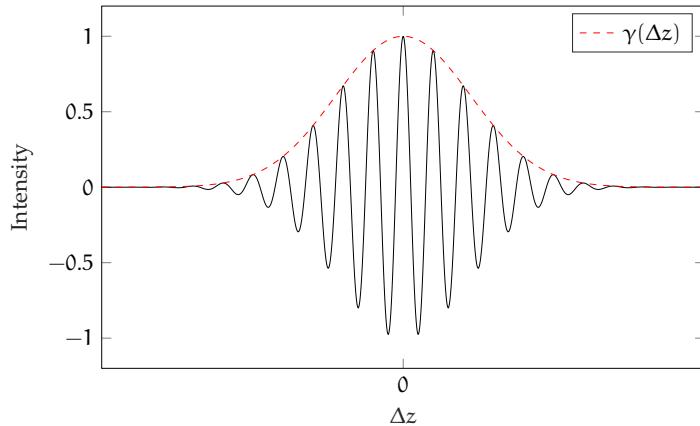


Figure 2.7: Diagram of the basic TD-OCT setup using a Michelson interferometer used in [23].

coherent source γ would be equal to 1 for all values of Δz and we would not be able to identify the reflector. On the other hand, with an increasingly sharper γ , the ideal reflector would be more and more defined.

Since the difference between the arm lengths is $\Delta l = \Delta z/2$, when scanning the reference arm, interference fringes will be generated as a function of time with periodicity equal to $(\lambda/2)/\sigma$ where σ is the scanning speed.

A setup similar to Figure 2.6 can be created using fiber optics components and lenses to focus the optical beam on the sample and collect the reflections, as in Figure 2.8 [23]. The role of splitting and recombining the signals is assigned to a fiber coupler, while a piezoelectric transducer (PZT) is used to frequency-modulate the interference sig-

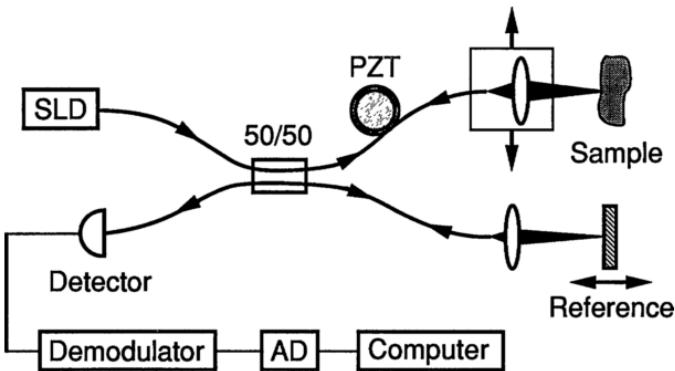


Figure 2.8: Diagram of the basic TD-OCT setup using a Michelson interferometer used in [23].

nal and shift it inside the photodetector's bandwidth. The generated electrical signal is then filtered, demodulated and acquired by an Analog-to-Digital Converter (ADC).

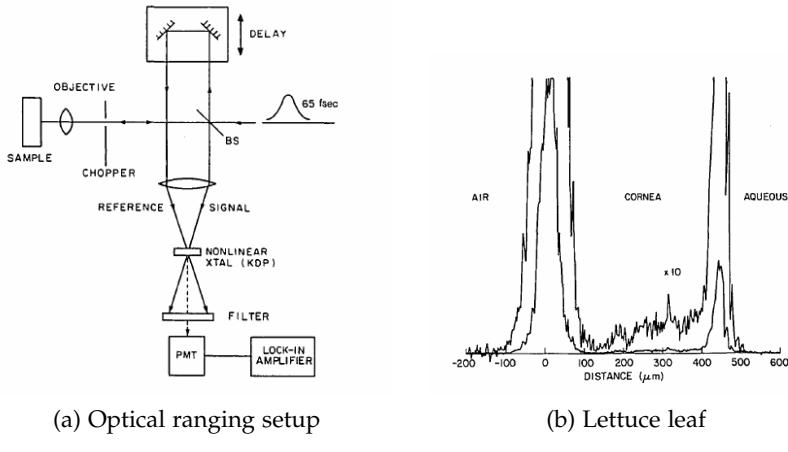


Figure 2.9: Optical ranging setup (left) and measurement (right) of the cornea of the rabbit eye [19].

This imaging technique is also called *Optical Ranging*, and was demonstrated by Fujimoto et al.[19] in 1986. An A-scan of the cornea of a rabbit's eye was performed *in vivo*, and is available in Figure 2.9b. The peaks in signal intensity located at $l = 0$ and $l \sim 450 \mu\text{m}$ are due to the strong reflection occurring at the interface between two different media. Inbetween these two peaks it's also possible to notice the effect of light scattered by the cornea, which is not present in the air and aqueous regions.

To acquire B-scans and C-scans, axial measurements can be repeatedly performed while moving the sample on the orthogonal plane or by deflecting the impinging optical beam by means of galvanometric mirrors.

The main disadvantage of this scheme is the slow acquisition rate, as it requires the mechanical movement of the reference arm. Con-

sequently, *in vivo* imaging is difficult to achieve since the movement of the sample could introduce heavy distortions on both B-scans and C-scans.

2.4 FOURIER DOMAIN OCT

Fourier-Domain OCT (**FD-OCT**) is a group of OCT techniques that encode the depth information of a sample in the frequency content of the interference signal and decode it through a Fourier transform operation. As opposed to **TD-OCT**, an axial scan is obtained without the need of a scanning mirror in the reference arm. This results in a much faster acquisition speed and higher quality measurements, since the distortions caused by the mechanical vibrations and uncertainty on the position of the mirror are removed. On the other hand, **FD-OCT** systems require more advanced laser sources and detection schemes, other than some numerical compensation and high speed **ADCs**.

The two main **FD-OCT** schemes are called Spectral-Domain OCT (**SD-OCT**) and Swept-Source OCT (**SS-OCT**), and will be presented in the next sections.

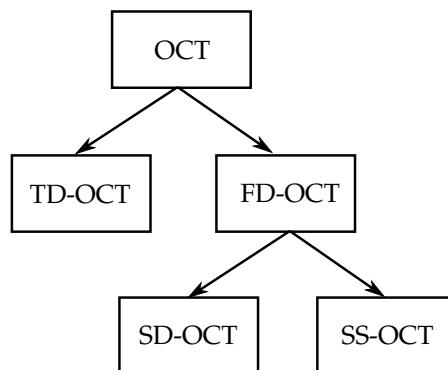


Figure 2.10: Diagram illustrating the different types of OCT modalities.

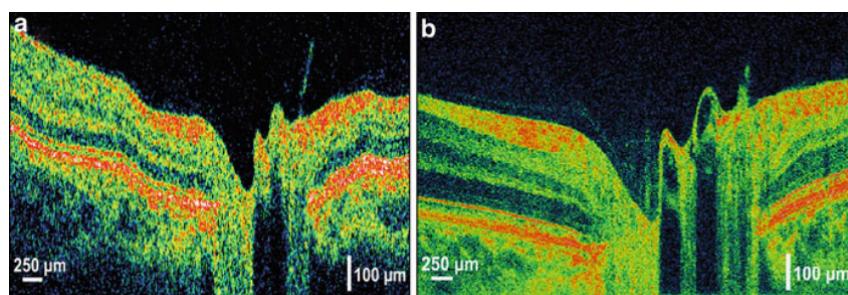


Figure 2.11: Comparison between images of the human retina obtained with standard **TD-OCT** (left) and high-speed, high-resolution **FD-OCT** (right).

2.4.1 Spectral Domain OCT

SD-OCT was the first type of **FD-OCT** to be implemented, and was proposed by Wojtkowski et al. in 2002 [45]. This technique uses a Superluminescent Diode (**SLD**) as a broadband optical source, a Michelson interferometer similar to that in [Figure 2.6](#) and a detector consisting of a spectrometer. A diagram of the setup is available in [Figure 2.12](#). As already mentioned in [Section 2.4](#), **SD-OCT** can acquire an A-scan with a single optical pulse, without the need to select the imaging depth through a scanning reference mirror. This comes at the price of a more complex detection scheme and the need for computationally intensive post-processing solutions. In order to achieve real-time performances, **SD-OCT** scheme typically require the use of fast **GPUs**.

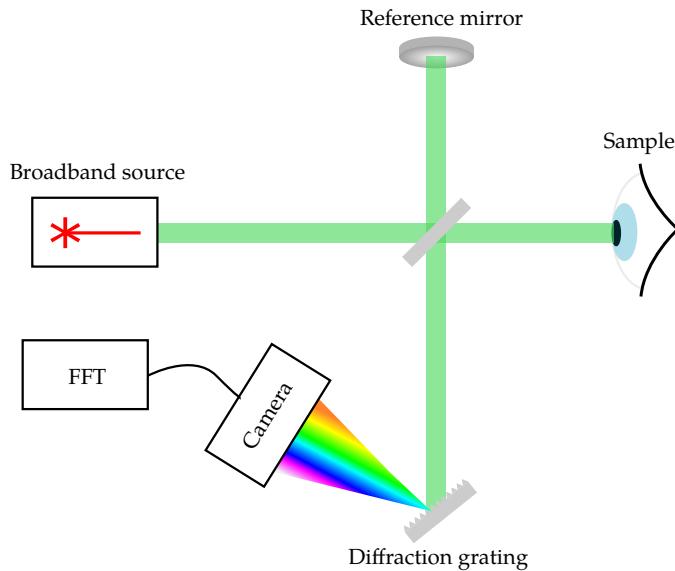


Figure 2.12

To gain insight on the working principle of **SD-OCT**, we can rewrite [Equation 2.27](#) as a function of the wavenumber $k = 2\pi/\lambda$ and fixing the **OPD**:

$$I(k) \propto I_{\text{source}}(k) \cos(k\Delta z) \quad (2.28)$$

This means that a reflector placed at a depth $d = \Delta z/2$ will frequency modulate the source spectrum with a frequency that is linearly dependant on d . This effect is illustrated in [Figure 2.13](#) for a broadband source centered at 1310 nanometers, with ideal reflectors positioned at $d_1 = \Delta z/2 = 12.5 \mu\text{m}$ and $d_2 = \Delta z/2 = 25 \mu\text{m}$.

Using a diffraction grating or a prism, the spectrum of the interference signal is spatially separated and acquired through a linear photodetector, typically a charged coupled device (**CCD**) or complementary metal-oxide semiconductor (**CMOS**) camera. The corresponding A-scan

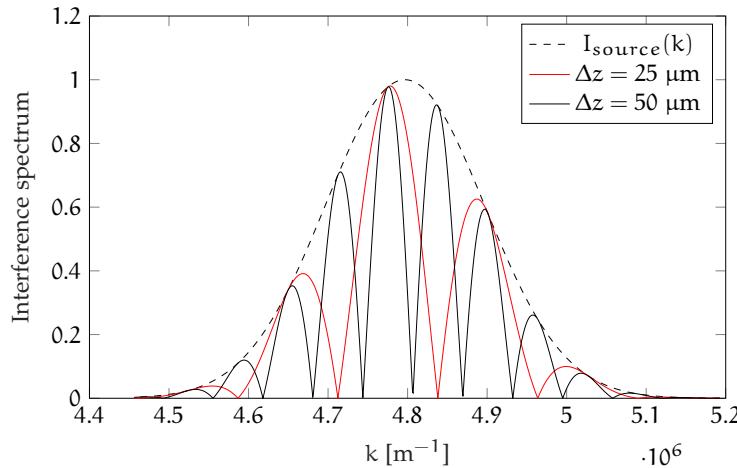


Figure 2.13

is then computed using an inverse Fourier Transform operation, mapping modulation frequency in axial position. The depth of the various layers of the sample are encoded in the modulation frequency of the spectrum, while their reflectivity is encoded in the fringe visibility.

A key observation has to be made regarding the role of the coherence function γ . In fact, the spectrum modulation induced by a reflection at depth $d = \Delta z/2$ can only be detected if $\gamma(d/2)$ is non-zero, that is, only the portion of the sample that is within the coherence length of the source can be imaged. This is the main disadvantage compared to [TD-OCT](#) schemes, in which the depth of focus in the sample was selected with the mechanical movement of the reference mirror.

2.4.1.1 Drawbacks

SAMPLING DISTORTION One of the major problems concerning this detection technique is that the interference spectrum is usually detected linearly in the wavelength. For example, a diffraction grating spatially separates the different wavelengths at an angle β_m with respect to the axis normal to the grating such that

$$\sin \beta_m = m \frac{\lambda}{\Lambda} + \sin \alpha, \quad (2.29)$$

where Λ is the spacing between each line of the grating, α is the angle of the inpinging wave and m is the order of diffraction. This behaviour requires a re-sampling of the acquired spectrum before computing the Fourier Transform in order to obtain a signal which is linearly spaced in frequency instead of wavelength. If this post-processing step is ignored, an oscillating term with a fixed frequency $\propto \Delta z$ in the interference spectrum will result in a broadened peak in the Fourier domain, as can be observed in [Figure 2.14](#). The linear sampling in the λ domain will induce a chirp on the spectrum (red).

Such behaviour has a detrimental effect on the axial resolution of the system.

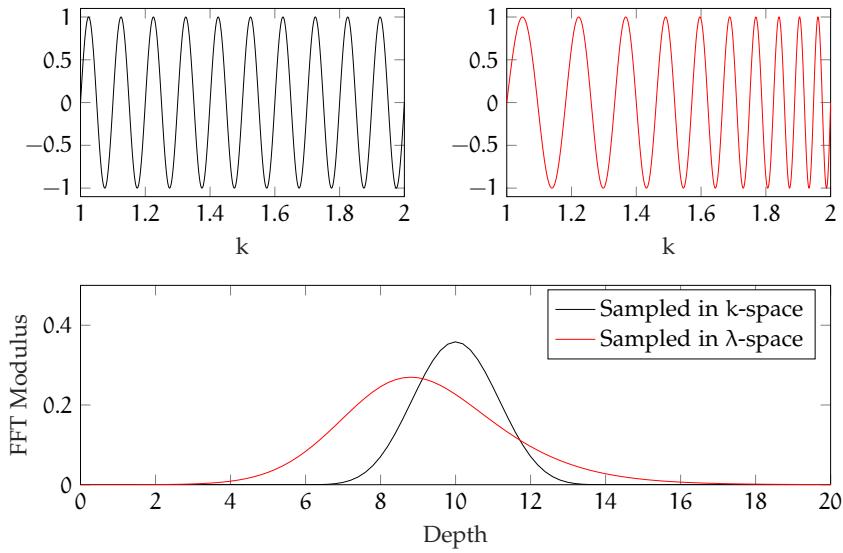


Figure 2.14: Interference distortion induced by linear sampling in the λ space.

SENSITIVITY FALLOFF A second harmful effect on the performance of these types of FD-OCT devices is the so-called *Sensitivity falloff*. Sensitivity is defined as the signal-to-noise ratio ([SNR](#)) when the sample is an ideal reflector. It's been experimentally demonstrated that for increasing imaging depths, the sensitivity value drops. Such effect is illustrated in [Figure 2.15](#). This behaviour is due to the fact that when increasing the [OPD](#) between reference and sample arm, the coherence function $\gamma(\Delta z)$ decreases as the two reflected signals only partially overlap, resulting in a lower signal intensity.

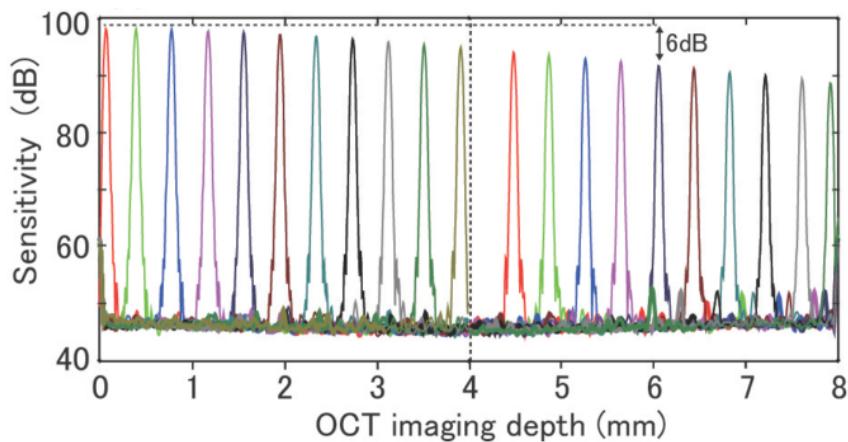


Figure 2.15: Sensitivity falloff in FD-OCT systems [10].

An additional factor that contributes to this detrimental effect is the resolution of the photodetector arrays used to acquire the interference signal. As previously explained, higher imaging depths correspond to higher spectrum modulation frequencies. If the number of pixel N of, e.g., a **CCD** camera is not sufficiently high, the higher frequency components will be undersampled, reducing the range of visibility. Beam diameter and camera's pixel size also influence the loss of sensitivity.

IMAGING ARTIFACTS The massive gain in acquisition rate that comes from detecting the whole imaging range in a single shot is offset by the introduction of different artifacts in the reconstructed A-scans. The interference pattern acquired by the photodetectors array is comprised of three terms:

1. *DC component*. All the reflected replicas of the impinging signal coming from the sample contribute to a constant DC offset in the interference signal. These are equivalent to the non-interfering terms in the Michelson interferometer formula. When Fourier-transformed, they will appear as a reflector positioned at the very start of the imaging range, where the **OPD** is close to zero. A simple solution to this problem is to compute and subtract the average value of the spectrum before applying the Fourier Transform.
2. *Cross-Correlation terms*. These are the interference terms that arise from the correlation between the reference signal and the reflections coming from the sample arm. As previously discussed, they contain the information required to reconstruct the reflectivity profile of the sample.
3. *Auto-Correlation terms*. They result from the interference between reflectors in the sample that are less than a coherence length apart.

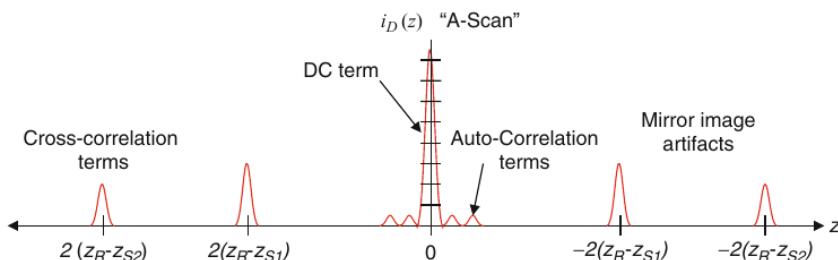


Figure 2.16: [15]

These components are visible in [Figure 2.16](#), who depicts the Fourier Transform of the interference signal generated by two reflectors. A further issue that is highlighted by this plot is the presence of the

so called *complex conjugate artifacts*, or mirror image artifacts. This phenomenon arises from the Fourier Transformation of a real signal, which will have the property of Hermitian symmetry (even modulus and odd phase). This means that **FD-OCT** techniques cannot differentiate between positive and negative frequencies, or equivalently, between a sample placed before or after the zero **OPD** position.

The easiest way to deal with this problem is to simply drop one half of the acquired A-scan, effectively halving the maximum imaging depth of the system. Mirror artifacts will still appear in the image if the sample is too close to the scanning lens and crosses the zero **OPD** position, which is fixed by the reference mirror. The part of the sample which is in the "negative" **OPD** range will appear flipped and superimposed to the other part of the sample. This effect is illustrated in [Figure 2.17](#), in which the scanning lens is progressively moved forward, closer to the patient's retina.

Different methods to solve this issue and restore the full imaging range have been proposed, including complex-signal detection schemes using 3×3 fiber couplers [37], heterodyne detection [13] and phase-shifting techniques [41] using either **PZTs** or electro-optic modulators (**EOMs**).

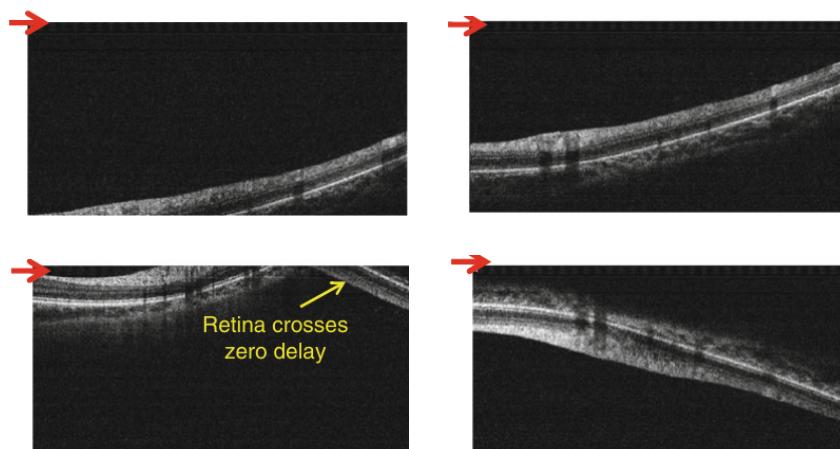


Figure 2.17: Complex conjugate artifacts in **SD-OCT**.

2.4.2 Swept-Source OCT

Swept-Source OCT (**SS-OCT**), also known as optical frequency domain interferometry (**OFDI**), is a **FD-OCT** technique that employs a narrow-bandwidth frequency-swept laser as an optical source. The introduction of tunable sources simplifies the detection scheme, removing the need for diffraction gratings and photodetector arrays. Ideally, a swept-source laser for **SS-OCT** should present a frequency sweep that

is linear, i.e, the instantaneous optical frequency should be expressed as

$$f(t) = f_0 + \sigma_f \cdot t, \quad (2.30)$$

where σ_f , typically measured in Terahertz per microsecond, is the sweep rate of the source. If this requirement is satisfied, the intensity at the output of the interferometer when the sample is a single reflector positioned at the depth $d = \Delta z/2$, can be expressed as

$$I(t) = I_1 + I_2 + 2\sqrt{I_1 I_2} \cos(2\pi\Delta f t) \quad (2.31)$$

where

$$\Delta f = \sigma_f \cdot \Delta t = \sigma_f \cdot \frac{\Delta z}{c} = \sigma_f \cdot \frac{d}{2c}, \quad (2.32)$$

called the *beat frequency*, is linearly dependant on the depth of the reflector. When the reflector is replaced by a real sample, multiple beat frequencies arise in the photodetector's current and the reflectivity profile can be reconstructed with a Fourier Transform. This technique is equivalent to [SD-OCT](#), with the key difference that the frequency sweep enables the detection as a function of time through a simple photodetector, while the diffraction grating spatially separates the different wavelengths and directs them to a detector array.

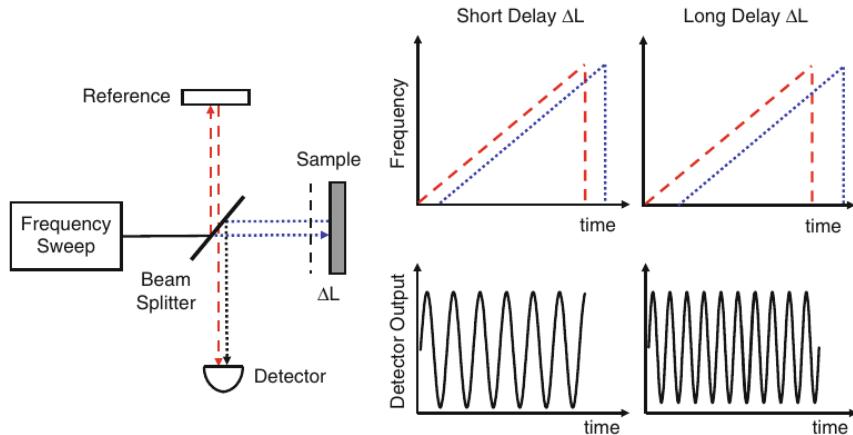


Figure 2.18: [15]

2.4.2.1 Sweep nonlinearity

A tunable laser generally consists of a gain medium, typically a semiconductor optical amplifier ([SOA](#)), a tunable wavelength filter and a laser cavity that supports a large bandwidth. The frequency-sweep is achieved by controlling the tunable filter through electrical signals to select the desired wavelength. [Figure 2.19](#) shows the diagram of an Axsun swept source in which frequency tuning is achieved by tilting a [MEMS](#) mirror at the end of a Fabry-Perot filter. Rapidly changing the selected

wavelength can affect the linearity frequency sweep, contributing to a nonlinear term in [Equation 2.30](#), which then becomes

$$f(t) = f_0 + \sigma_f \cdot t + \eta(t), \quad (2.33)$$

where $\eta(t)$ contains the aforementioned nonlinearities. If the detector's output is digitized with a uniform clock rate, a similar effect to that described for [SD-OCT](#) systems can arise ([Figure 2.14](#)), resulting in a non-linear sampling in the k-space. Consequently, distortions in the acquired A-scans can appear, hindering the image quality. A resampling of the acquired signal is then required in order to linearize the frequency sweep, adding to the total computational cost and possibly reducing the acquisition rate of the system.

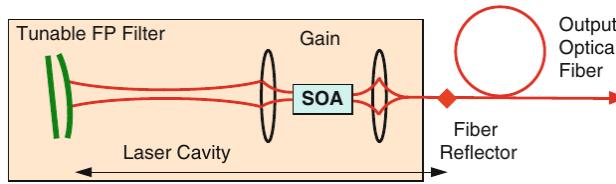


Figure 2.19: Frequency-swept optical source by Axsun [[15](#)].

2.4.2.2 *k-clocking*

While in [SD-OCT](#) the resampling step is the only possible solution for this issue, [SS-OCT](#) enables a hardware approach through a non-uniform clocking of the [DAQ](#) device. This is accomplished by extracting a clock signal, called *k-clock*, by means of an unbalanced Mach-Zender interferometer ([MZI](#)) which detects the frequency-sweep and generates a sinusoidal signal with a variable frequency. This approach removes the need for the computationally-intensive resampling step, but requires special [DAQ](#) devices that can handle external clocks with a wide range of frequencies and duty cycles, which are generally more expensive.

2.4.2.3 *Acquisition rate*

The acquisition speed of [SS-OCT](#) systems is dictated by the source sweep repetition rate, that is, the frequency at which the source is able to sweep the entire bandwidth. This parameter is determined by the laser cavity length. In fact, for a given frequency, the spontaneous emission has to build up in order to reach a saturation limit to be correctly amplified by the active region. Shorter cavities with a smaller round-trip generally guarantee higher repetition rates than longer cavities. Novel swept-source designs enable acquisition speeds in the order of 100,000 A-scans per second, with Fourier domain mode locking ([FDML](#)) and vertical cavity surface-emitting lasers ([VCSELs](#)) reaching repetition rates above 500 kHz. These high rates make [SS-OCT](#) the

primary candidate for real-time 3D visualization of *in vivo* samples, where motion artifacts could hinder image quality. This is possible because the detection schemes is no longer the bottleneck like in **SD-OCT**, where the slow camera response times limited the overall rate of the system.

2.4.2.4 Axial resolution

Just like in the case of **SD-OCT**, the axial resolution depends on the tuning range of the source, $\Delta\lambda$, which is defined as the **FWHM** of the spectrum. In the case of a Gaussian spectrum, the axial resolution, defined as the **FWHM** of the reflection peak generated by a perfect reflector, is given by

$$\delta z \simeq 0.75 \cdot \frac{\lambda_0^2}{\Delta\lambda}, \quad (2.34)$$

where λ_0 is the central wavelength of operation, which is chosen based on the application (1060 nm for Ophthalmology, 1300 nm for tissue imaging).

2.4.2.5 Imaging range

Imaging range of **SS-OCT** systems is, like other **FD-OCT** schemes, limited by the coherence length of the source, which we have previously defined as the **OPD** at which the visibility of the interference fringes drops to 0.5. This parameter is controlled by the instantaneous linewidth of the source $\delta\lambda$ by the following equation:

$$l_c \approx 0.44 \frac{\lambda_0^2}{\delta\lambda}. \quad (2.35)$$

As with **SD-OCT**, complex conjugate artifacts typically limit the imaging range to be half of the coherence length.

When using a k-clock to acquire A-scans, the imaging range is also dependant on the path difference of the **MZI**'s arms, which has to be 4 times the maximum imaging depth, d_{max} . A factor of 2 is needed because a reflector positioned at d_{max} will have **OPD** equal to $\Delta z = 2d_{max}$ with respect to the reference mirror. The other factor of 2 is needed to satisfy Nyquist's criterion, so that the k-clock will have a maximum frequency that is 2 times the beat frequency generated by the reflector at d_{max} .

2.4.2.6 Balanced detection

The use of simple photodetectors instead of more complex setups involving diffraction gratings and prisms allows for a dual balanced detection scheme. By collecting the reference and sample signals through a circulator and feeding them to a 3 dB fiber coupler, two interference signals can be obtained. A balanced detector can then subtract the

common DC component and the excess noise while adding the interference term. This detection scheme has been proved to be more efficient than a simple Michelson interferometer. A complete **SS-OCT** scheme with dual balanced detection and external k-clock is illustrated in [Figure 2.20](#).

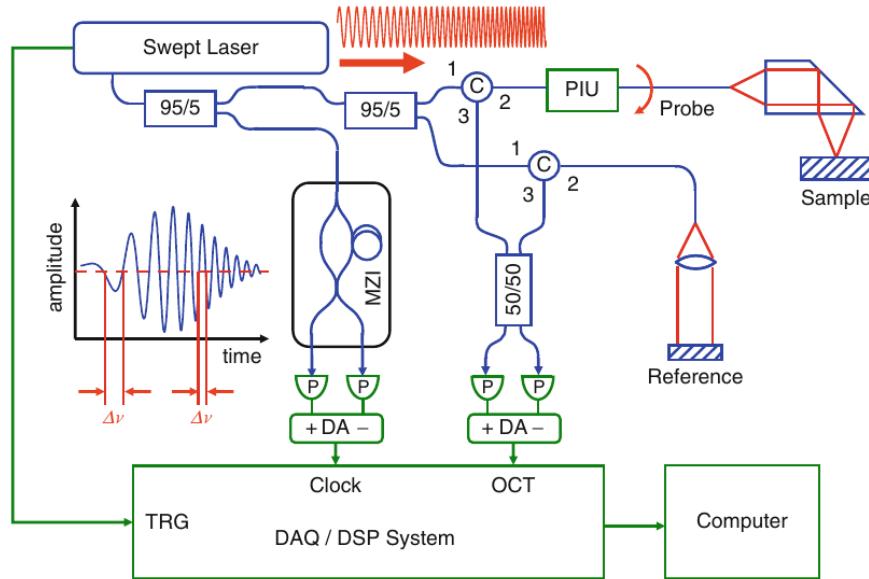


Figure 2.20: Frequency-swept optical source by Axsun [15].

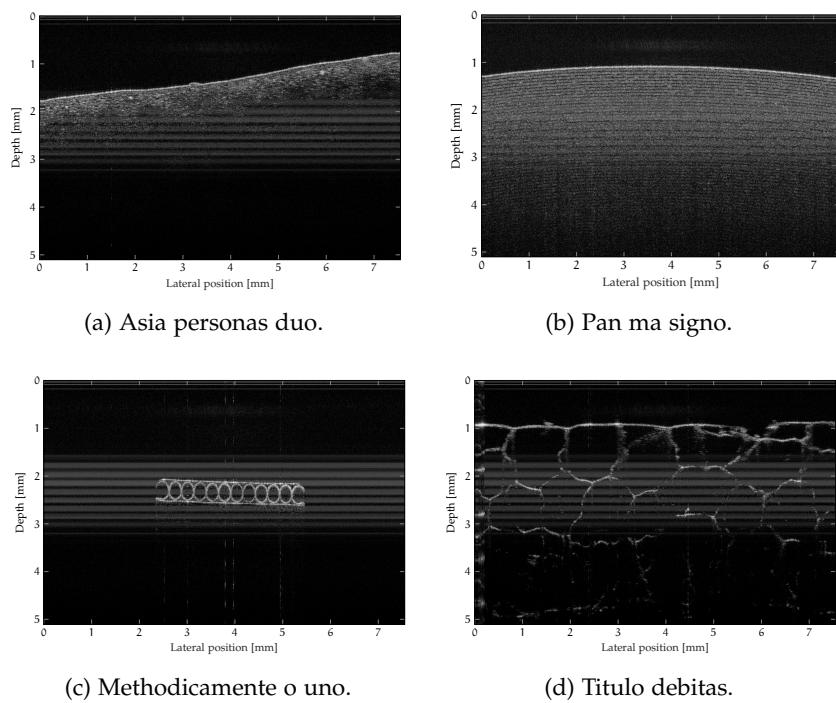


Figure 2.21: Tu duo titulo debitas latente.

3

DESCRIPTION OF THE SETUP

This chapter details the setup that was used in the design of a Swept Source OCT system developed for this thesis work, describing the optical and electrical components that were employed.

Particular care will be given to the characterization of the DAQ board and the design of the acquisition software.

3.1 OPTICAL SOURCE

As already discussed in [Chapter 2](#), SS-OCT systems employ a narrow-bandwidth tunable laser to enable a simple detection scheme and is one of the most critical components of the entire system. The source that was used in this thesis is a SSOCT-1310 by Axsun Technologies, which is a class 3 laser that uses a [MEMS](#) tunable filter to sweep wavelengths in the 1300 nm range.

The two most important features of this laser are the fast sweep rate, which enables high speed imaging, and the presence of the k-clock signal used for equalizing the nonlinear frequency sweep.



Figure 3.1: The acquisition board, AlazarTech ATS9350.

An photo of the laser is available in [Figure 3.1](#). The emitted light is collected by connecting a fiber optic patch cord to the FC/APC connector on the front panel, which also includes two SMA connectors for the sweep-start trigger and the k-clock signal.

PARAMETER	UNITS	VALUE
Sweep Rate	kHz	100.2
Center Wavelength	nm	1305
Wavelength Tuning Range	nm	140.38
Average Power	mW	25.7
Duty Cycle	%	77.3
Sampled Duty Cycle	%	50.5
External Clock Min Frequency	MHz	183.1
External Clock Average Frequency	MHz	307.0
External Clock Max Frequency	MHz	332.1
Sampling Clocks	-	1536

Table 3.1: Axsun laser datasheet.

3.1.1 Optical Spectrum

A critical parameter in swept sources for OCT applications is the wavelength interval over which the laser is able to tune. From Table 3.1, we can see that the Axsun SSOCT-1310 is centered at $\lambda_0 = 1305$ nm with a bandwidth of $\Delta\lambda = 140.38$ nm. These values were verified by connecting the laser output directly to an Optical Spectrum Analyzer (OSA), measuring its spectrum.

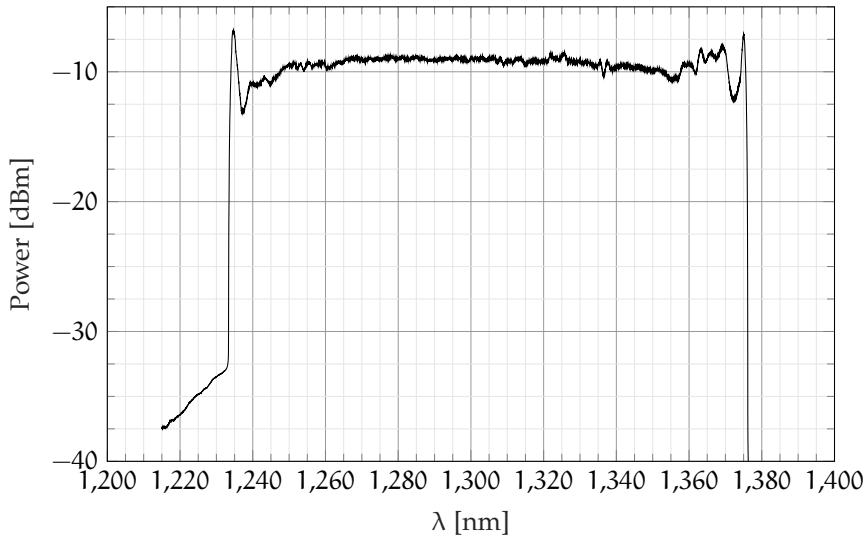


Figure 3.2: Spectrum of the Axsun SSOCT-1310 laser, obtained with a 1 nm resolution.

The measurement was executed with a resolution of 1 nanometer, resulting in the spectrum visible in Figure 3.2. The edge wavelengths

turned out to be $\lambda_1 \approx 1237$ nm and $\lambda_2 \approx 1376$ nm, which result in a bandwidth of $\Delta\lambda \approx 139$ nm and central wavelength of $\lambda_0 \approx 1307$ nm. These values slightly differ from those reported in the datasheet, but not enough to compromise the performance of the laser.

The bandwidth is swept at a frequency $f_a = 100.2$ kHz and with a duty cycle $d_c = 0.505$, meaning that the average sweep speed is

$$\sigma_\lambda = \frac{\Delta\lambda f_a}{d_c} \approx 27.8 \text{ nm}/\mu\text{s}. \quad (3.1)$$

Since the goal of SS-OCT sources is to perform a linear frequency sweep, it is more useful to define this parameter in the following manner

$$\sigma_f = c_0 \left(\frac{1}{\lambda_0 - \Delta\lambda/2} - \frac{1}{\lambda_0 + \Delta\lambda/2} \right) \frac{f_a}{d_c} \approx 4.9 \text{ THz}/\mu\text{s} \quad (3.2)$$

The instantaneous frequency, after linearization, is thus expressed by

$$f(t) = f_0 + \sigma_f t \quad (3.3)$$

where the starting frequency is determined by

$$f_0 = \frac{c_0}{\lambda_0 + \Delta\lambda/2} \approx 218.2 \text{ THz}. \quad (3.4)$$

3.1.2 Axial Resolution

Using [Equation 2.34](#) and the data provided by [Table 3.1](#) it is possible to obtain an estimate of the axial resolution provided by the employed optical source. The axial resolution is found to be

$$\delta z \simeq 0.75 \frac{\lambda_0^2}{\Delta\lambda} \approx 9.2 \text{ }\mu\text{m}. \quad (3.5)$$

This approximation is valid for sources with a Gaussian spectrum, and can lead to an overestimation of the real axial resolution of the system in case this condition is not met.

3.1.3 Sweep trigger

The Axsun laser provides a square-wave signal that determines the start of the frequency sweep and that is used to trigger the acquisition of A-scans. This signal is acquired with a high-speed oscilloscope and illustrated in [Figure 3.3](#). The voltage range was measured as $V_{\text{range}} \approx [0, 1.48]$ V, while the duty cycle is found to be

$$d_c = \frac{t_{\text{high}}}{t_{\text{high}} + t_{\text{low}}} \approx 0.97. \quad (3.6)$$

For a correct acquisition of A-scans, the datasheet recommends to trigger the acquisition device when the signal level reaches the value of $V_{\text{trig}} = 0.71$ V, pictured in [Figure 3.3](#) with a red line.

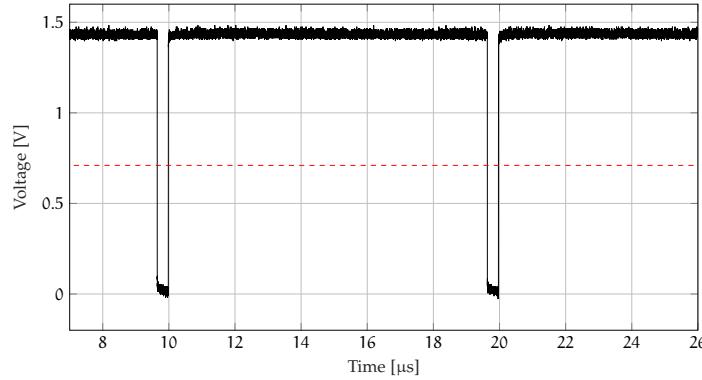


Figure 3.3: Sweep trigger of the SSOCT-1310 laser.

3.1.4 Power profile

Using a photodiode it's possible to measure the instantaneous power profile of the laser emission. The photodiode generates an electrical current which is proportional to the power of the detected electromagnetic field:

$$I_{ph} = \mathcal{R} \cdot P \quad (3.7)$$

The constant \mathcal{R} is called *responsivity* [A/W] and is specific to the detector used. The current can then be measured with an oscilloscope. In order to avoid the saturation of the receiver, a 3 dB coupler was inserted between the source and the photodiode. The obtained power profile is depicted in Figure 3.4 along with the sweep trigger. We can observe that in the first $\sim 5 \mu s$ after the positive edge of the trigger, the instantaneous power behaves like a typical laser pulse while outside of this interval it assumes an irregular profile.

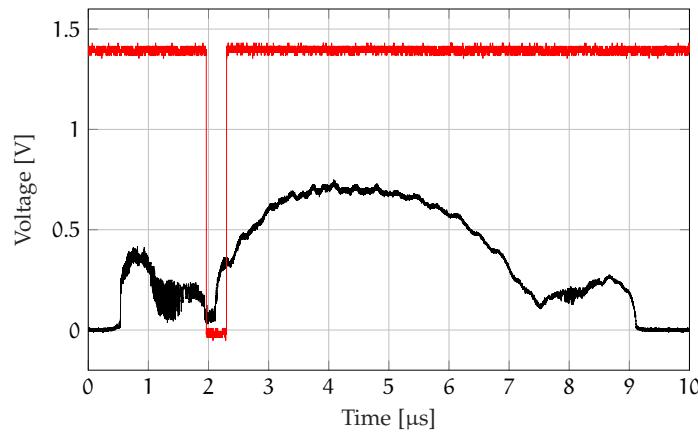


Figure 3.4: Sweep trigger of the SSOCT-1310 laser.

3.1.5 *k-clock*

As already stated, this particular SS-OCT laser is equipped with an internal [MZI](#) which generates the variable-frequency clock, called *k-clock*, used to sample the A-scan signal at a variable rate and equalize the nonlinear terms in the frequency sweep. As we can see in [Figure 3.5](#), this signal assumes values in the [0, 900] mV range with an average value of about 460 mV. This sinusoidal signal will be used to drive the acquisition board for the first 5 μ seconds of the sweep and from this point forward will be referred to as *useful clock*. As per [Table 3.1](#), the total number of samples that can be acquired using this clock is $N_s = 1536$. In the last 5 μ seconds the clock is called "dummy" because of its undefined behaviour that could lead to issues in the clocking of the DAQ devices.

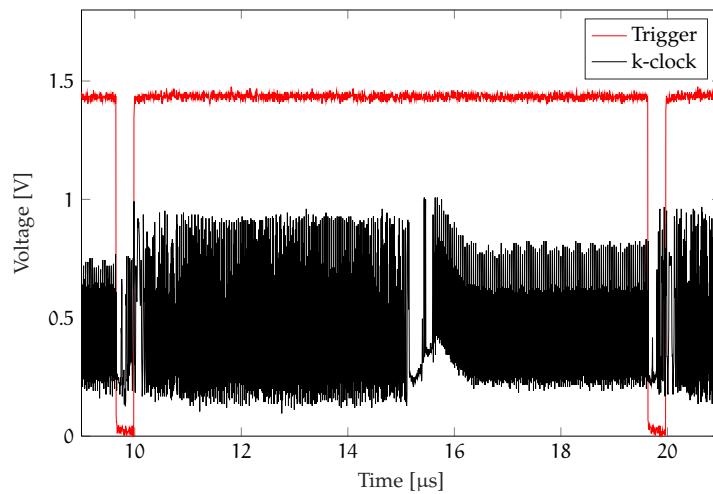


Figure 3.5: Estimate of the instantaneous frequency of the *k-clock*.

In [Figure 3.6](#) are depicted two 50 nanoseconds windows at the start (left) and in the middle (right) of the useful clock, where the variable frequency of signal is clearly visible. Also worth noting is that at the start of the sweep the signal appears much more distorted than in the middle of the sweep. Fortunately, this behaviour did not compromise the performance of the system.

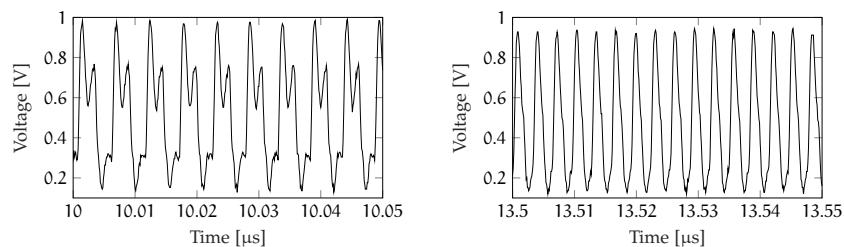


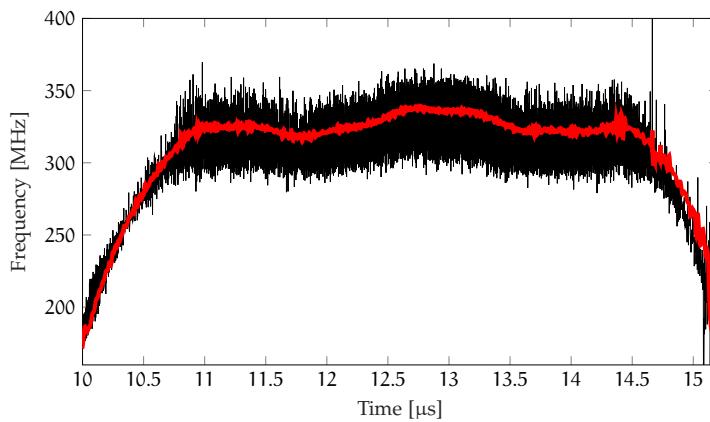
Figure 3.6: Behaviour of the useful *k-clock* at different time instants.

To gain further insights on the nature of the frequency sweep, it is interesting to estimate the instantaneous frequency of the k-clock. A method to perform this estimate is the following

1. Subtract the average value from the signal
2. Interpolate the signal (t, y) to obtain (\hat{t}, \hat{y})
3. Determine the time instants \hat{t}_i at which the signal \hat{y} crosses 0, from which a vector $\mathcal{I} = [\hat{t}_{i_1}, \hat{t}_{i_2}, \dots, \hat{t}_{i_N}]$ is built.
4. Compute the difference between the adjacent elements of \mathcal{I} and obtain $\mathcal{T} = [\hat{\tau}_{i_1}, \hat{\tau}_{i_2}, \dots, \hat{\tau}_{i_{N-1}}]$,
5. The instantaneous frequency of the signal is then estimated as $\hat{f}_i = 1/2\hat{\tau}_i$.

The result of this method is illustrated in [Figure 3.7](#). Since the estimate is rather noisy, a low pass filter was applied to the \hat{f}_i signal, resulting in a much cleaner estimation (depicted in red). As we can observe, the instantaneous frequency rapidly changes at the start and at the end of the sweep, while in the middle it stabilizes at about 320 MHz. Since a constant k-clock frequency implies a linear frequency sweep, we can infer that the sweep nonlinearities are much more prevalent at the edge of the sampling interval.

The maximum estimated frequency is $\hat{f}_{clk}^{\max} = 339$ MHz, as opposed to the value that was reported in [Table 3.1](#) of $f_{clk}^{\max} = 332.2$ MHz. This slight overestimation is also visible in [Figure 3.7](#), where we can see that the low-pass filtered signal is slightly higher than the average value of the non filtered one.



[Figure 3.7](#): Estimate of the instantaneous frequency of the k-clock.

From the maximum frequency of the k-clock signal we can measure the maximum imaging depth obtainable by the system.

Two replicas of the pulse overlapping with a relative time delay τ will generate a beat frequency f_b given by

$$f_b = \sigma_f \tau \quad (3.8)$$

where σ_f , the sweep speed, was determined in [Equation 3.2](#). In a OCT setup, the time delay τ is related to the path mismatch of reference and sample arm, which in turn depends on the depth of the layer of the sample that generates the reflection. In this way

$$\tau = \frac{2dn}{c_0}, \quad (3.9)$$

where c_0 is the speed of light in vacuum, n is the refractive index of the sample and d is the depth. By the Nyquist theorem, the maximum beat frequency that can be digitized using the k-clock is

$$f_b^{\max} = \frac{f_{\text{clk}}^{\max}}{2}, \quad (3.10)$$

which results in a maximum depth in air ($n = 1$) of

$$d_{\max} = \frac{c_0 f_b^{\max}}{2\sigma_f} = \frac{c_0 f_{\text{clk}}^{\max}}{4\sigma_f} \simeq 5.1 \text{ mm}. \quad (3.11)$$

3.1.6 Coherence length

In [Chapter 2](#) we have seen that the coherence length of a source is one of the most important parameters of optical sources, independently of the type of OCT system. In [TD-OCT](#), this parameter determines the axial resolution of the system, while [FD-OCT](#) schemes it also affects the imaging range, as the reference arm is fixed. In particular, complex conjugate artifacts arising from the Fourier transform of the detected signals halve the maximum imaging range. For this reason, the coherence length of a swept source satisfies the following inequality

$$l_c \geq 2 \cdot d_{\max}, \quad (3.12)$$

which implies that the coherence length of the SSOCT-1310 source is at least

$$l_c \geq 2 \cdot d_{\max} \approx 10.2 \text{ mm}. \quad (3.13)$$

The exact value was experimentally determined in [\[9\]](#), measuring the normalized coherence function $|\gamma(\Delta z)|$ by placing a mirror in the sample arm and acquiring the interference signal for different values of [OPD](#). The path length difference was changed by translating the reference mirror. The intensity of the interference signal, normalized by its maximum value, is plotted in [Figure 3.8](#) as a function of the

OPD. When the OPD increases above a certain value, the coherence function decays exponentially as

$$|\gamma(\Delta z)| \propto e^{-\alpha|\Delta z|}, \quad (3.14)$$

which corresponds to a linear decay in the logarithmic domain. The parameter α was estimated by fitting the data points to a line using a Least-Squares fit. The coherence length is finally determined as the Δz value such that $|\gamma(\Delta z)| = 1/2$, resulting in

$$l_c = 2 \cdot \frac{\ln 2}{\alpha} \approx 12.3 \text{ mm}. \quad (3.15)$$

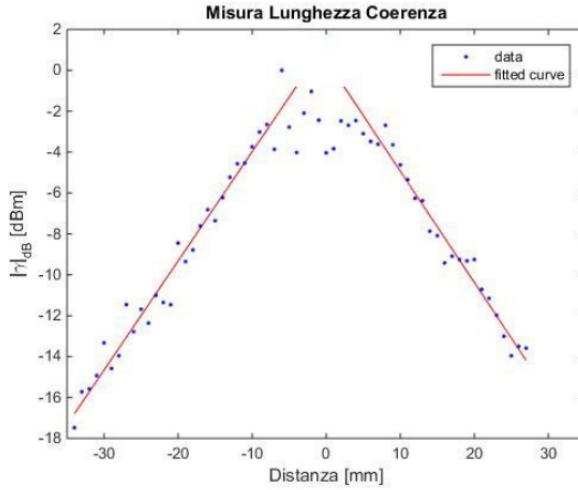


Figure 3.8: Experimental estimation of the normalized coherence function.

3.2 SCANNING SYSTEM

One of the most critical components of an OCT is the scanning system. Its role is to receive the light emitted by the source, focus it on the sample under test, and collect the backreflections. A scanning system usually comprises of three different devices

1. Fiber collimator
2. Galvanometric Mirrors
3. Focusing lens

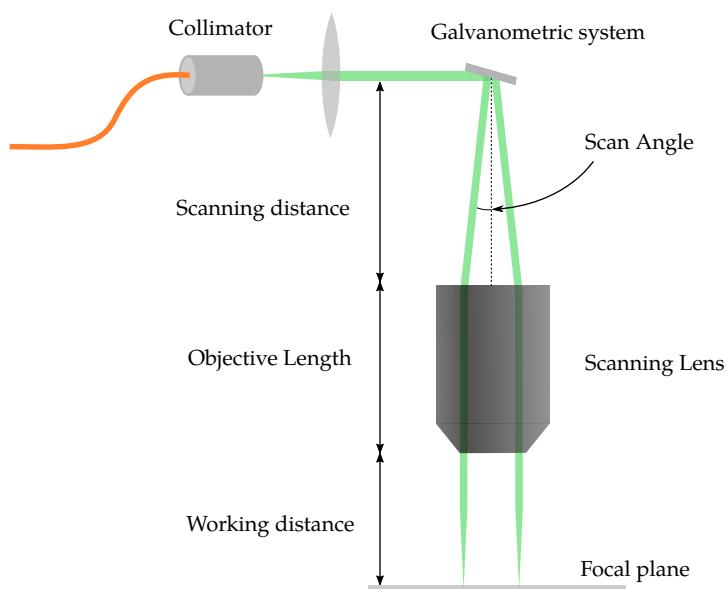


Figure 3.9: Diagram of a OCT scanning system.

A schematic of this system is depicted in [Figure 3.9](#). Light coming from a single mode optical fiber is deflected by a system of galvanometric mirrors towards a lens which in turns focuses the radiation on the sample. By rotating the mirrors the light beam is focused on a different position on the sample. The reflected beam travels the same path in the opposite direction and is collected by the optical fiber. The correct design of these components is paramount in order to guarantee small transversal and axial resolutions of the OCT system.

3.2.1 Collimator

The role of the fiber collimator is to collect light coming from a single mode fiber and collimate the beam on the galvanometric mirrors. The divergence angle at the output of the collimator is approximated with

the following equation when the beams have a Gaussian intensity profile:

$$\theta \approx \frac{D}{f} \frac{180}{\pi}, \quad (3.16)$$

where D is the mode diameter and f is the focal length of the collimator. This approximation works well for single mode fibers, but will underestimate the real divergence angle in the case of multimode fibers, as the intensity profile has a non-Gaussian shape.

The collimator used in this setup is the Thorlabs F280APC-C¹, which is designed to work in the 1310 nm range. Its main specifications are summarized in [Table 3.2](#).

PARAMETER	UNITS	VALUE
Central wavelength	nm	1310
Beam diameter	mm	3.4
Divergence angle	degrees	0.028
Lens numerical aperture (NA)	-	0.15
Focal length	mm	18.67

Table 3.2: Thorlabs F280APC-C datasheet.

3.2.2 Scanning lens

The light beam deviated by the galvanometric mirrors has to be focused on the sample under test in order to guarantee high resolution images. This is accomplished by a telecentric focusing lens. This type of lenses is characterized by flat image planes which are ideal for OCT applications.

The main parameters that characterize this device are the following:

- Entrance Pupil Size (EP): it specifies the diameter of the collimated laser beam that will maximize the resolution of the imaging system. When using a single galvanometric mirror, the EP is located at the pivot point of the mirror. When two mirrors are used, the EP is located between the two mirrors.
- Scanning Distance (SD): the distance between the EP and the base of the lens.
- Scan Angle (SA): the angle between the incoming light beam and the optical axis of the lens.

¹ <https://www.thorlabs.com/thorproduct.cfm?partnumber=F280APC-C>

- Working Distance (WD): the distance between the tip of the lens and the focal plane.
- Parfocal Distance (PD): the distance between the base of the lens and the focal plane. It is equal to the WD plus the objective length.
- Field of View (FOV): the area on the focal plane that can be imaged with a resolution equal or better than the one guaranteed by the lens.
- Depth of View (DOV): it corresponds to the distance between parallel planes on either side of the focal plane, where the beam spot diameter is $\sqrt{2}$ greater than it is at the focal plane. Using lenses with a high DOV value yields images with high lateral resolution in a larger interval of depths.
- Spot size: the diameter of the beam on the focal plane.

The lens used for the SS-OCT system developed in this thesis is the Thorlabs LSM04². Its main specifications are listed in [Table 3.3](#). A simulation of the beam spot size on the focal plane is available in [Figure 3.10](#), where we can see that it increases for bigger scan angles. This behaviour has the effect of reducing the lateral resolution at the edge of the FOV.

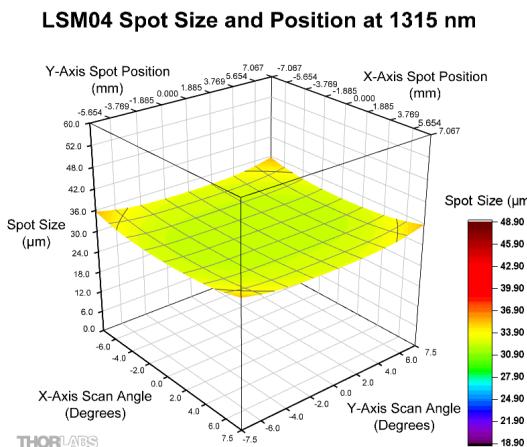


Figure 3.10: The Thorlabs GVSo02 galvo system.

3.2.2.1 Lateral resolution

Using the parameters listed in [Table 3.3](#) it is also possible to obtain an estimate of the lateral resolution of the system, which is dictated

² <https://www.thorlabs.com/thorproduct.cfm?partnumber=LSM04>

PARAMETER	UNITS	VALUE
Wavelength range	nm	1250-1380
Effective Focal Length (EFL)	mm	54
Entrance Pupil size	mm	4
Working Distance	mm	42.3
Parfocal Distance	mm	80.8
Scan Distance	mm	18.9
Maximum Scan Angle	degrees	$\pm 7.5 \times \pm 7.5$
Field of View	mm ²	14.1 × 14.1
Depth of View	mm	0.61

Table 3.3: Thorlabs LSMo4 datasheet.

by the diffraction limited spot size of the focused optical beam. For Gaussian-shaped beams, it is approximated as [15]

$$\delta x \approx \frac{4\lambda_0}{\pi} \frac{f}{d} \quad (3.17)$$

where λ_0 is the central wavelength of operation, f is the focal length of the lens and d is the diameter of the beam at the entrance of the lens. The lateral resolution guaranteed by the LSMo4 is thus

$$\delta x \approx \frac{4\lambda_0}{\pi} \frac{\text{EFL}}{\text{EP}} \approx 22.5 \text{ } \mu\text{m}. \quad (3.18)$$

Since the diameter of the beam exiting the fiber collimator is slightly smaller ($d = 3.4 \text{ mm}$) than the EP size, the resulting lateral resolution becomes

$$\delta x \approx 26.5 \text{ } \mu\text{m}. \quad (3.19)$$

The depth of view, also called confocal parameter, limits the imaging depth of the system. Due to diffraction, this parameter is also governed by the lateral resolution of the system through the following equation

$$b = \frac{2\delta x^2}{\lambda_0}. \quad (3.20)$$

This effect is illustrated in Figure 3.11, where we can see that using lenses with a high numerical aperture (NA), or small spot size, the depth of view is limited. In OCT systems it is preferred to use low NA lenses to sacrifice lateral resolution in favor of a depth of view comparable to the coherence length of the source. Following this criterion the coherence length is fully exploited.

Using the EP size when calculating δx , the confocal parameter is $b \approx 0.61 \text{ mm}$, equal to the value reported in Table 3.3, while using the beam size diameter $d = 3.4 \text{ mm}$ it becomes $b \approx 0.84 \text{ mm}$.

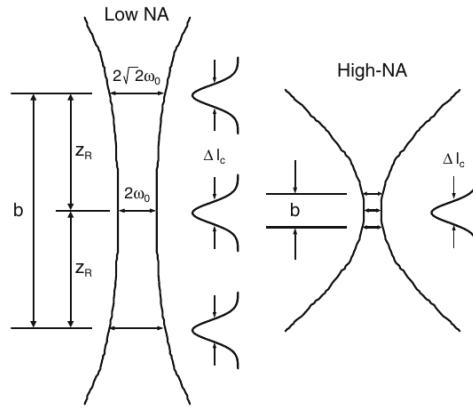


Figure 3.11: Basic diagram of a SS-OCT setup

3.2.3 Galvanometric Mirrors

In order to acquire B-scans and C-scans it is necessary to direct the laser beam emitted from the swept laser over a specified area of the sample. This is accomplished by using a system of *galvanometric mirrors*, also called *galvo mirrors*. A cross-sectional image can be generated with a single 1D mirror, that is, a mirror which can rotate on a single axis, while volumetric data require the use of either a 2D mirror or a couple of 1D mirrors. In the second case, each mirror is responsible for the movement of the beam over a single direction on the focal plane, typically called X and Y.



Figure 3.12: The Thorlabs GVS002 galvo system.

The galvo system employed in this thesis work is the Thorlabs GVS002³, which consists in two 1D silver plated mirrors, the motors that control them, a detector to measure the mirrors' position and the servo driver boards that interpret the error signals coming from the detector to correctly drive the motors to the demanded position. A photo of the dual axis motor/mirror assembly is available in Figure 3.12. A thorough analysis of the electrical circuitry of the system and its performance characterization is found in [9].

³ <https://www.thorlabs.com/thorproduct.cfm?partnumber=GVS002>

3.2.3.1 Controlling the mirrors

The position of each mirror is controlled by sending a voltage signal to the respective driver board which is proportional to the mechanical angle of the respective motor. The maximum mechanical scan angle that the GVSoo2 system can handle is $\pm 12.5^\circ$, but it can vary depending on the laser beam diameter and the voltage scaling factor. This particular galvo system offers three different scaling factors which also govern the maximum input voltage applied to the servo boards:

SCALING FACTOR	MAX. VOLTAGE	MAX. SCAN ANGLE
0.5 V/ $^\circ$	± 6.25 V	$\pm 12.5^\circ$
0.8 V/ $^\circ$	± 10 V	$\pm 12.5^\circ$
1 V/ $^\circ$	± 10 V	$\pm 10^\circ$

Through simple trigonometric, it can be shown that the scan angle seen by the focusing lens, called *optical scan angle*, θ_o , is twice the mechanical scan angle applied to the mirror, θ_m . Setting the scaling factor $\alpha = 1V/^\circ$, the maximum optical scan angle then becomes $\pm 20^\circ$, which allows us fully exploit the maximum angle supported by the lens. In this case, the full FOV is scanned if the applied voltage is $v_{max} = \pm 7.5^\circ/(2\alpha) = \pm 3.75$ V.

The beam spot position on the focal plane is related to the mechanical angle applied to the mirrors and the focal length of the lens through the following relation

$$x = EFL \tan(2\theta_{m,x}) \quad (3.21)$$

$$y = EFL \tan(2\theta_{m,y}) \quad (3.22)$$

With a scaling factor equal to α , the voltage signal $v_{x,y}(t)$ will therefore focus the beam at the following position

$$x(t) = EFL \tan\left[2\alpha v_x(t) \frac{\pi}{180}\right] \quad (3.23)$$

$$y(t) = EFL \tan\left[2\alpha v_y(t) \frac{\pi}{180}\right] \quad (3.24)$$

The two signals v_x and v_y are delivered to the driver boards as two pairs of positive voltage signals, (v_x^+, v_x^-) and (v_y^+, v_y^-) , where

$$v_i^+(t) = \begin{cases} v_i(t) & v_i(t) \geq 0 \\ 0 & v_i(t) < 0 \end{cases}, \quad v_i^-(t) = \begin{cases} -v_i(t) & v_i(t) < 0 \\ 0 & v_i(t) \geq 0 \end{cases} \quad (3.25)$$

3.2.3.2 Image distortion

When using a two-mirror system three different effects can appear

1. The arrangement of the mirrors leads to a distortion of the image field, which will be "pillow-shaped" instead of rectangular (fig:galvo-distortion). This is due to the fact that the distance between the two mirrors depends on the mechanical angle applied to the mirrors.

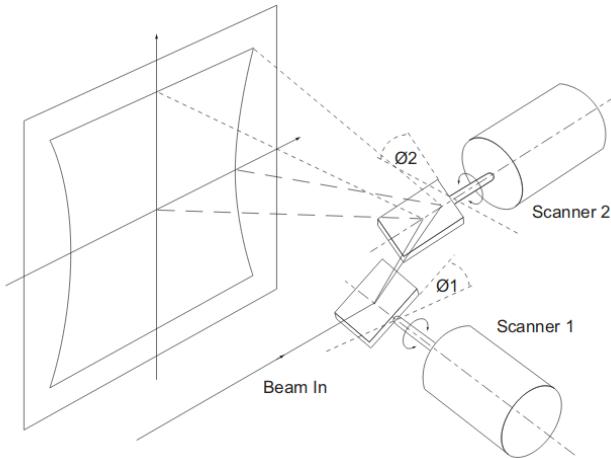


Figure 3.13: Field distortion using two mirrors.

2. From [Equation 3.21](#) and [Equation 3.22](#) we can notice that the distance on the image field is not directly proportional to the applied angle but to its tangent. This can result in an additional distortion effect.
3. If an ordinary lens is used for focusing the laser beam, the focus lies on a sphere. In a flat image field, a varying spot size results.

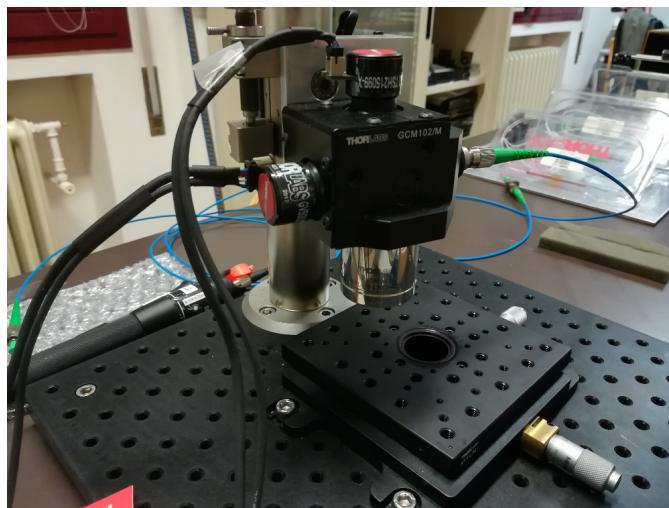


Figure 3.14: The final scanning system mounted on a movable vertical support.

3.3 ACQUISITION BOARD

In order to digitize the interference signals generated by the OCT system and reconstruct images of the analyzed sample in real time, a specialized high speed data acquisition (**DAQ**) device is required. A desired feature in a **DAQ** board for SS-OCT applications is the ability to sample the signals at a variable rate using a supplied k-clock, avoiding the computationally heavy task of data resampling and allowing higher imaging speeds. The **DAQ** board used for this thesis is the AlazarTech ATS9350⁴, which is a specialized digitizer ideal for OCT, ultrasonic and radar applications.



Figure 3.15: The acquisition board, AlazarTech ATS9350.

This device is an 8-lane PCI Express 2.0 (PCIe x8) card with the following characteristics:

- 12 bits sample resolution
- 250 MHz input bandwidth
- 2 input channels with a dedicated **ADC** chip and a user selectable sampling rate which ranges from 2 MHz up to 500 MHz. The raw throughput to the on-board memory is thus $T = 2 \text{ bytes per sample} \times 2 \text{ channels} \times 500\text{MHz} = 2 \text{ GB/s}$.
- User selectable input voltage range, from $\pm 40 \text{ mV}$ to $\pm 4 \text{ V}$ for a better utilization of the 12 bit resolution.
- External trigger support
- Variable frequency external clocking
- AUX (Auxiliary) connector that can be configured either as a trigger output or a "acquisition start" trigger.

⁴ <http://www.alazartech.com/product/ats9350>

- dual-port 128 Megasamples on-board memory
- On board Field Programmable Gate Array ([FPGA](#)) module for the computation of the Fast Fourier Transform ([FFT](#)) on one of the input channels.

The board also comes with the proprietary AlazarDSO software to configure the device, acquire and display signals. For custom applications, the board can be integrated using a supplied Software Development Kit ([SDK](#)) which includes libraries and headers for a variety of different programming languages, including C/C++, Python and MATLAB. The device was installed in a Dell Precision T5810⁵ workstation with the following specifications:

- Intel Xeon E5-1650 v4, 3.6 GHz, 6 cores, 12 threads
- 16 GB of DDR4 ECC RAM at 2400 MHz
- 256GB SATA SSD
- NVIDIA Quadro M4000 with 8 GB of DDR5 memory

3.3.1 Data Acquisition

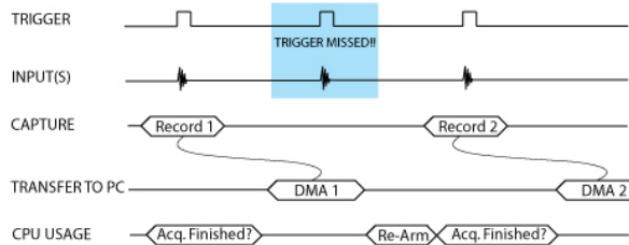
Data acquisition on the ATS9350 is organized in *records* and *buffers*. A record consists in a user defined number of samples acquired by the board for each trigger event. In OCT applications, a record typically corresponds to a single A-scan. A buffer instead is a collection of records and is the format in which data is returned to the user application. For our application, a buffer corresponds to a B-scan.

This [DAQ](#) board can be configured to acquire data in one of the following modes

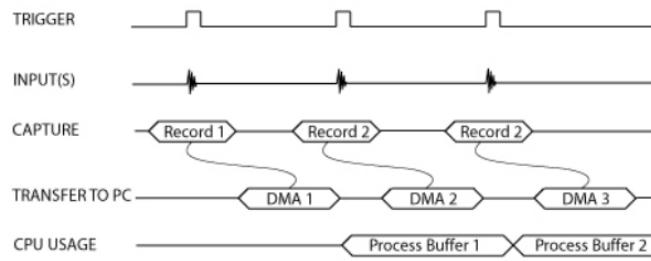
1. Single-Port mode acquires data to the on-board memory and then, after the acquisition is complete, transfers data to the host PC memory. This mode can be used if the application can tolerate to miss trigger events that occur while data is being transferred.
2. Dual-Port mode, also called AutoDMA, acquires to on-board memory while, at the same time, transferring data from on-board memory to the host PC memory. This is the preferred acquisition method, as no trigger events are missed by the board while managing the data transfer, meaning that no B-scans are lost by the system. The on-board memory acts as a very long First-In First-Out ([FIFO](#)) structure that is able to stream data to the host memory at the rate supported by the motherboard.

⁵ <https://www.dell.com/us/business/p/precision-t5810-workstation/fs>

The benchmarking tool integrated in the AlazarDSO software measured an average PCIe throughput of 1.735 GB/s, which is even higher than that advertised by AlazarTech.



(a) Acquisition using Single-Port Memory.



(b) Acquisition using Dual-Port Memory.

Figure 3.16: Diagram highlighting the differences between Single-Port acquisitions and Dual-Port acquisitions

In Figure 3.16 a comparison between the two acquisition techniques is available. For single-port acquisitions, in Figure 3.16a we can see that trigger events happening while the Direct Memory Access (DMA) transfer is ongoing are missed completely, and virtually all CPU cycles are used in managing the data acquisition. This leaves no room for data processing on the CPU. In Dual-port AutoDMA acquisition instead (Figure 3.16b) no trigger events are missed and over 95% of CPU time is available for data processing.

This is possible thanks to proprietary circuitry that allows for data transfers to be initiated by the hardware itself, without the need to waste CPU cycles. On the software side, an asynchronous driver is able to parallelize the tasks of setting up a DMA channel while handling data transmission on another channel.

The board supports different types of AutoDMA acquisitions which will be here shortly explained

3.3.1.1 Traditional AutoDMA

This method allows for the acquisition of both pre-trigger and post-trigger data, which will be returned to the user in form of buffers that contain from 1 to 8192 records, one for each trigger event. Each record can come with its own header containing a 40-bit timestamp. A DMA

transfer is initiated after the acquisition of each record, copying the data from on-board to user supplied memory.

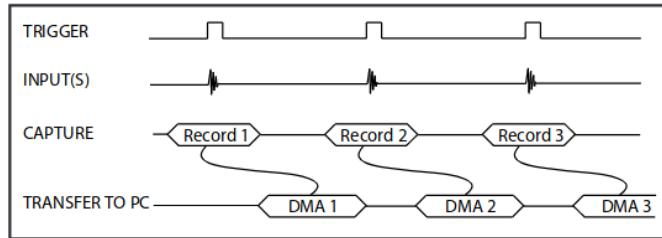


Figure 3.17: Time diagram of a Traditional AutoDMA acquisition.

Fast trigger repeat rates are not supported and can cause the board to overflow even if data transfer is happening at the maximum sustained rate and the on-board memory has not been completely used up.

3.3.1.2 NPT AutoDMA

No Pre-Trigger (NPT) AutoDMA is designed for applications that do not require the acquisition of data before a trigger event. By only storing post-trigger data, the memory bandwidth is optimized. Record headers are also not included, resulting in a more efficient memory utilization. For this reason NPT AutoDMA supports high trigger repeat rates resulting in an acquisition speed which is only limited by the PCIe transfer speed and the rate at which the user application consumes data.

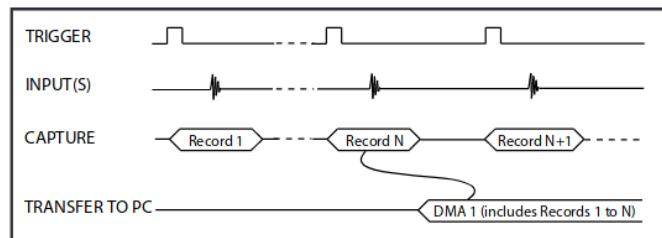


Figure 3.18: Time diagram of a NPT AutoDMA acquisition.

Unlike Traditional AutoDMA, a DMA transfer is initiated only after an entire buffer consisting of multiple records is acquired. This data acquisition method has been specifically designed for ultrasonic scanning, radar and medical imaging such as OCT.

3.3.2 FFT module

The ATS9350 is equipped with a [FPGA](#) module that computes the FFT on data records acquired on the first of the two input channels. For OCT applications this means that a powerful GPU dedicated to data processing is not needed, simplifying the user application both on

the design and efficiency standpoint. In Figure 3.19 the flow diagram of the FFT module is available: a data record is zero-padded to a power-of-2 length (and a maximum of 2048), multiplied by a complex windowing function and fed to the FFT module, whose output is then scaled and optionally converted in a logarithmic scale. At the end of the chain it is possible to obtain either FFT data, time-domain data or both.

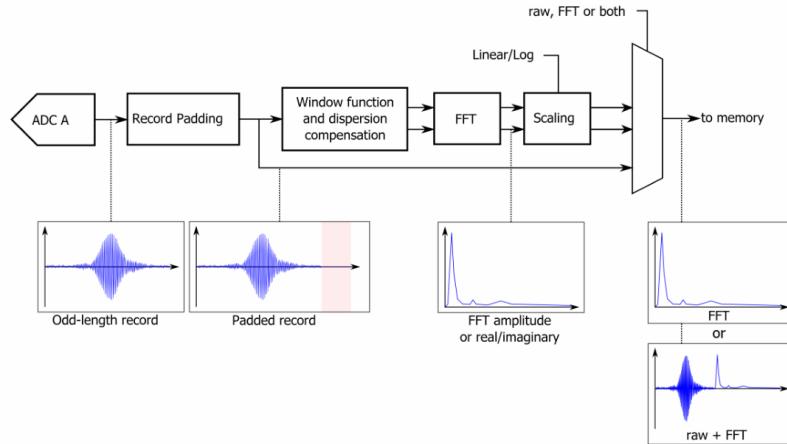


Figure 3.19: Time diagram of a NPT AutoDMA acquisition.

The format of frequency-domain data can be chosen between amplitude, real or imaginary part of the signal and can be converted in either single-precision floating point or unsigned integer.

3.3.3 External clocking

As previously stated, the ATS9350 provides an SMA input for external clocking. This connector supports both sinusoidal and Low-Voltage Transistor-Transistor-Logic (**LVTTL**) signals, sampling on each rising edge of the clock. The input impedance of the port is fixed at $50\ \Omega$ and AC coupling is used.

There are three types of supported external clock operations:

1. 10 MHz Reference Clock
2. Slow External Clock, 0 – 60 MHz
3. Fast External Clock, 2 – 500 MHz

Only the third mode is relevant for our application, as the k-clock provided by the Axsun SSOCT-1310 ranges from 180 to 340 MHz. A useful feature of the ATS9350 digitizer is the *dummy clock switchover*: inbetween frequency sweeps some SS-OCT sources generate unstable clocks that do not respect the specification of the board; such behaviour could lead to missed trigger events. In order to avoid this

issue, the board can switch to an internally generated clock for a specified amount of time after each record, and switch back to the external source once a trigger signal is detected.

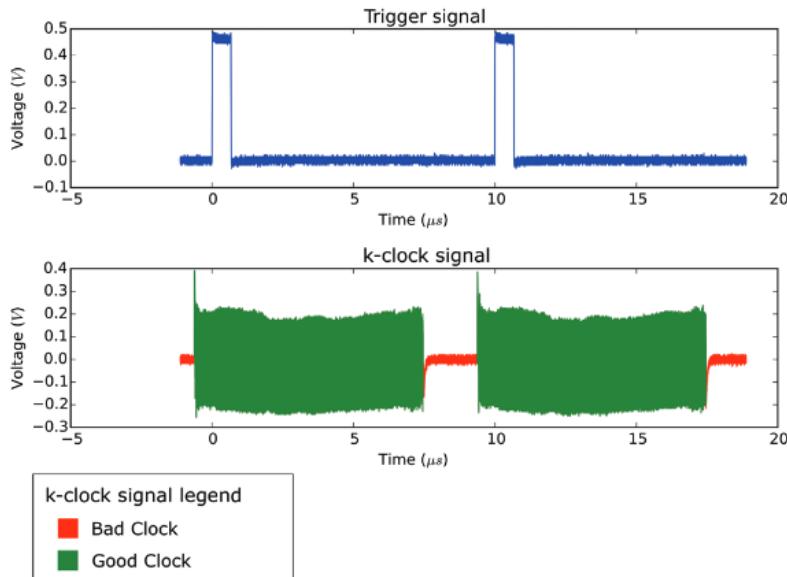


Figure 3.20: Example of bad k-clock that requires the use of the dummy clock switchover.

The use of this feature was not necessary in our case, as the installation of a k-clock deglitching firmware was enough to solve any problem related to bad clocks.

3.3.4 Triggering

The acquisition of A and B-scans requires the ATS9350 to be configured in the following way:

1. The acquisition mode must be set to NPT AutoDMA.
2. The laser sweep trigger, whose rising edge corresponds to the start of an A-scan, is connected to the External Trigger port of the board; as stated above, each record corresponds to an A-scan. The number of samples per A-scan should respect the following rules:
 - a) it should be a minimum of 256
 - b) it should be a multiple of 32
 - c) it should be less or equal to the number of sampling clocks of the k-clock, as specified by the manufacturer of the laser

In our case the number of useful clocks per A-scan is 1536 (Table 3.1), which also satisfies the first two rules. A conservative

The first two rules are needed for a correct alignment of the samples in memory.

approach would require to set this number to $1536 - 32 = 1504$, as a minor instability in the laser-generated clock could lead to a number of sampling clocks which is slightly lower, requiring the board to wait for the following A-scan to fill the record and consequently miss a trigger event.

The number of records per buffer instead determines the width of the B-scan.

3. To start the acquisition of a B-scan, the Auxiliary port of the board must be configured as "Trigger Enable In" and a TTL pulse must be supplied to the connector. Once the board detects a rising edge on the AUX port, it will wait for a sufficient number of A-scan triggers to fill a buffer. Once the buffer has been filled a DMA transfer will be started and the image will be delivered to the user application. Afterwards, the board waits for another pulse on the AUX port and start the acquisition of a new B-scan.

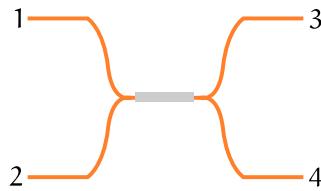
A more in-depth explanation of the programming of the board will be given in [Chapter 4](#).

3.4 OPTICAL CIRCUIT

In this section I will detail the process of designing and balancing the fiber optic interferometer that was used to realize the OCT device. Additionally, some performance parameters will be assessed.

3.4.1 Fiber Couplers

In order to build a fiber optic interferometer, a device called *fiber coupler* must be used. A coupler is a 4-port passive device which is typically used as a signal splitter/adder and that looks like the following:



The two couples of ports (1,2) and (3,4) are isolated, meaning that a signal that enters the device from port 1 (2) will not exit from port 2 (1) (the same principle holds for ports 3 and 4). The "direct" paths ($1 \leftrightarrow 3, 2 \leftrightarrow 4$) only let through a fraction of optical power equal to $|\rho|^2$ with no phase shift while the "crossed" paths ($1 \leftrightarrow 4, 2 \leftrightarrow 3$) let through a fraction $1 - |\rho|^2$ with a $\pi/2$ phase shift. The scattering matrix of this device is thus easily computed:

$$S = \begin{bmatrix} 0 & 0 & \rho & j\sqrt{1-\rho^2} \\ 0 & 0 & j\sqrt{1-\rho^2} & \rho \\ \rho & j\sqrt{1-\rho^2} & 0 & 0 \\ j\sqrt{1-\rho^2} & \rho & 0 & 0 \end{bmatrix} \quad (3.26)$$

For a 50:50 fiber coupler, $\rho = 1/\sqrt{2}$ so that 50% of the optical power entering from e.g. port 1, will exit from port 3 and the other 50% from port 4.

For this thesis, three fiber couplers were used:

- 2 Thorlabs TW1300R5A2⁶ 50:50 fiber couplers
- 1 Thorlabs TW1300R2A2⁷ 90:10 fiber coupler

The arms of the couplers are color coded to easily distinguish the direct paths, which are White↔White and Blue↔Red.

⁶ <https://www.thorlabs.com/thorproduct.cfm?partnumber=TW1300R5A2>

⁷ <https://www.thorlabs.com/thorproduct.cfm?partnumber=TW1300R2A2>

PATH	Path length [mm]		
	50-50A	50-50B	90-10
White \leftrightarrow White	2101.9939	2098.610325	2112.5832
White \leftrightarrow Red	2108.25695	2119.471656	2113.8573
White \leftrightarrow Blue	2101.149125	2095.978125	2129.024925
Blue \leftrightarrow Red	2107.402875	2116.86915	2130.299

Table 3.4: Length of the paths of the fiber couplers.

ARMS COUPLE	Path length difference [mm]		
	50-50A	50-50B	90-10
White-Blue	0.84	2.62	16.44
White-Red	6.25	20.85	1.28

Table 3.5: Path difference of each pair of adjacent ports of the couplers.

3.4.2 Measuring the couplers' length

For the design of the interferometer it is useful to determine the length of the coupler's arms with high precision. This was accomplished using an Optical Frequency-Domain Reflectometer ([OFDR](#)) and the setup in [Figure 3.21](#).

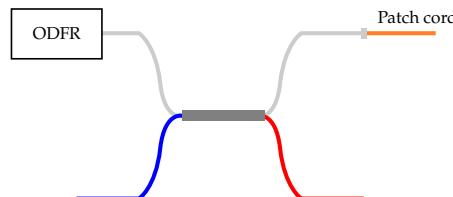


Figure 3.21: Diagram of setup for the measurement of fiber couplers' length.

For each input port, the [OFDR](#) trace will contain two peaks corresponding to the reflections generated by the fiber/air interface of the two non-isolated ports, which can be used to determine the length of the optical paths. In order to associate a specific port to its peak, a fiber patch cord was connected to the port, resulting in the reduction of the corresponding peak in the trace. For the same input port a second trace is acquired, switching the position of the patch cord. This procedure is repeated for each of the 4 input ports, for a total of 8 traces per coupler.

The resulting traces are available in [Appendix A](#). The total length of each path and the difference in length between adjacent arms are summarized in [Table 3.4](#) and [Table 3.5](#) respectively.

3.4.3 Mach-Zender Interferometer

In order to test the DAQ device and to determine the effect of sampling an interference signal with laser-supplied k-clock, a simple unbalanced MZI was set up using two 50-50 fiber couplers. The diagram of the setup for this test is reported in Figure 3.22. In order to acquire a beat frequency with the ATS9350, two replicas of the laser pulse have to overlap with a relative delay τ that is less than the source coherence time. Additionally, the beat frequency has to fall in the electronic bandwidth of 1) the photodiodes and 2) the acquisition device. The delay element in the unbalanced interferometer arises from the difference in length of the couplers' arms.

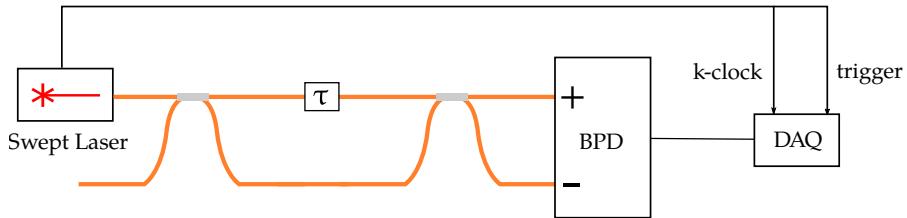


Figure 3.22: Diagram of an unbalanced Mach-Zender Interferometer.

The swept laser emits a signal with a time-varying frequency that can be described as follows

$$E_{ss} = E_{in} e^{j2\pi f(t)t} = E_{in} e^{j\phi(t)}, \quad (3.27)$$

hence the signals at the input ports of the second coupler will be respectively

$$E_1 = \frac{1}{\sqrt{2}} E_{in} e^{j\phi(t-\tau)} \quad E_2 = j \frac{1}{\sqrt{2}} E_{in} e^{j\phi(t)} \quad (3.28)$$

The + and - ports of the BPD will instead see two signals expressed by

$$E_+ = \frac{1}{2} E_{in} e^{j\phi(t-\tau)} - \frac{1}{2} E_{in} e^{j\phi(t)} \quad (3.29)$$

$$E_- = j \frac{1}{2} E_{in} e^{j\phi(t)} + j \frac{1}{2} E_{in} e^{j\phi(t-\tau)} \quad (3.30)$$

which will be converted by the photodiode in the following currents

$$I_+ \propto \langle E_+ E_+^* \rangle = \frac{1}{2} |E_{in}|^2 - \frac{1}{2} \cos [\phi(t) - \phi(t-\tau)] \quad (3.31)$$

$$I_- \propto \langle E_- E_-^* \rangle = \frac{1}{2} |E_{in}|^2 + \frac{1}{2} \cos [\phi(t) - \phi(t-\tau)] \quad (3.32)$$

The output of the BPD will then be the sum the two currents

$$I_{tot} = I_+ - I_- \propto \cos [\phi(t) - \phi(t-\tau)]. \quad (3.33)$$

We can observe that the DC term has been canceled out, leaving only an oscillating term that depends on the phase of the light source.

If the frequency sweep of the laser is assumed linear, the BPD generates a sinusoidal current with frequency that is proportional to the OPD between the arms of the interferometer:

$$I_{\text{tot}} \propto \cos [2\pi\sigma_f \tau t + \phi_0]. \quad (3.34)$$

The beat frequency is determined by the OPD Δz through the following relation

$$f = \sigma_f \tau = \sigma_f \frac{n \Delta z}{c_0}, \quad (3.35)$$

where $n = 1.4674$ is the refractive index of the optical fiber used for the manufacturing of the couplers⁸ and c_0 is the speed of light in vacuum.

Connecting the White and Red ports of coupler A to the White and Blue ports of coupler B we obtain $\Delta z = 6.25 - 2.62 \approx 3.63$ mm (Table 3.5), which corresponds to a beat frequency $f \approx 87$ MHz. This frequency is sufficiently low to satisfy Nyquist's criterion and be sampled using the k-clock, as its maximum frequency is ≈ 330 MHz.

The total BPD current was acquired with the ATS9350 digitizer using the provided AlazarDSO software. Two acquisitions were made, first using the internal 500 MHz clock and then clocking the board with the k-clock. In order to synchronize the acquisition with the laser sweep, the board was externally triggered by the sweep-start signal supplied by the laser.

When acquiring with the internal clock, the number of useful samples to acquire was calculated as follows

$$N = F_s \frac{d_c}{f} = 2525, \quad (3.36)$$

where $F_s = 500$ MHz is the sampling frequency, $d_c = 0.505$ is the laser duty cycle and $f = 100$ kHz is the sweep repetition rate. In order to satisfy the bit alignment requirement of the board, the actual number of samples was set to the nearest multiple of 32, $\hat{N} = 2528$. Instead, when using the external clock option of the board, this number is determined by the number of useful sampling instants of the k-clock, 1536 (Table 3.1).

The acquired signals were multiplied by a Hamming window and then Fourier-transformed. The result of this procedure is illustrated in Figure 3.23, where the x-axis is mapped from frequency to OPD by inverting Equation 3.35. We can notice that when using the internal clock of the board, the frequency content of the signal is spread over a relatively wide band (left). In fact, since the frequency sweep is not linear, uniformly sampling the interference signal results in a distorted

⁸ https://www.corning.com/media/worldwide/coc/documents/PI1463_07-14_English.pdf

sinusoidal term. Thus, a precise estimation of the OPD between the MZI arms is not possible.

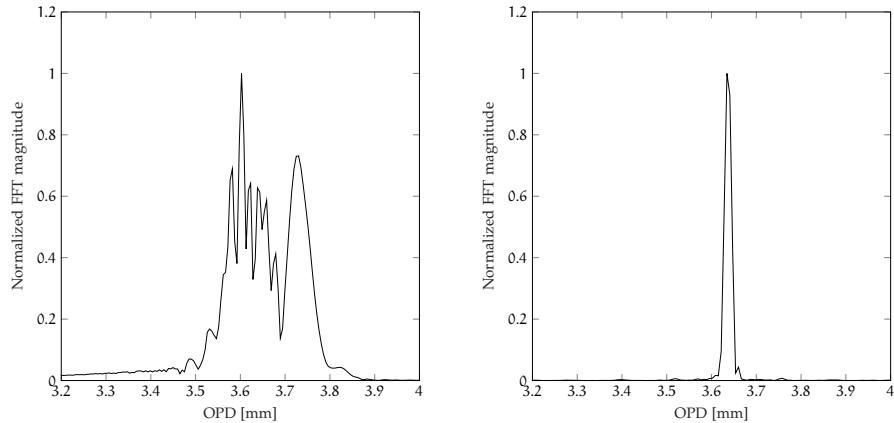


Figure 3.23: Comparison between the beat frequency acquired with the uniform clock (left) and the k-clock (right).

On the other hand, sampling with the provided k-clock produces a clearly defined peak in the frequency domain (right) and permits a precise evaluation of the OPD.

To better appreciate the effect of external clocking on the imaging quality of an OCT system, a number of $W = 1000$ consecutive A-scans were captured, from which an artificial B-scan was generated (Figure 3.24).

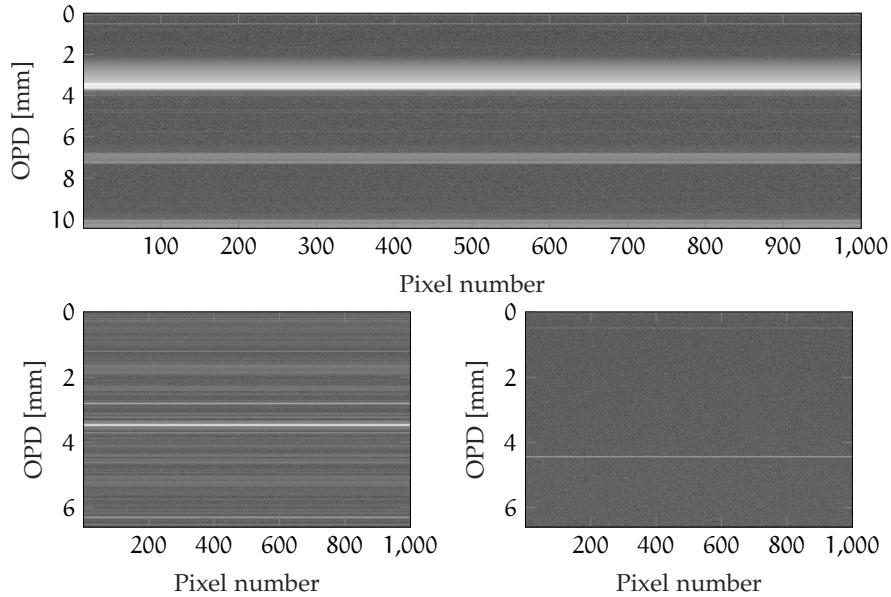


Figure 3.24: Simulated B-scans using the internal 500 MHz clock (top) and the k-clock (bottom).

Using the internal clock and a logarithmic color scale (top), the width of the peak is significantly more pronounced: in this case a perfect reflector would take up a considerable portion of the imaging

range. Moreover, the axial resolution of the OCT would not be acceptable for a functioning system. In this image it is also possible to notice two faint peaks at ~ 7 mm and ~ 10 mm. These are due to reflections accumulating a delay which is a multiple of the OPD between the arms of the interferometer.

As expected, the improvement when sampling with the k-clock is remarkable (bottom left). We can also notice that since the frequency of the k-clock is smaller than 500 MHz, the second of the two peaks generated by reflections is undersampled and appears at ~ 2.5 mm. The third image (bottom right) was captured after distancing the APC connectors on one of the arms of the MZI, introducing a ~ 1 mm layer of air ($n = 1$) which corresponds to an additional ~ 0.68 mm of OPD in the fiber. The reflection peaks in this case do not appear as the received power is too low.

3.4.4 Fiber Michelson interferometer

The first OCT circuit that was tested was a simple Michelson interferometer realized with a 50-50 fiber coupler. The schematic diagram of this setup is illustrated in Figure 3.25. The reference arm consists in a cage system with a silver mirror mounted on a translatable support and an adjustable FiberPort collimator (Figure 3.26). This collimator permits a fine alignment with five degrees of freedom to easily maximize the amount of power that gets coupled back in the optical fiber after being reflected by the mirror. Through a careful alignment procedure, the attenuation of the reference arm is minimized, reaching a value of ~ 4 dB.

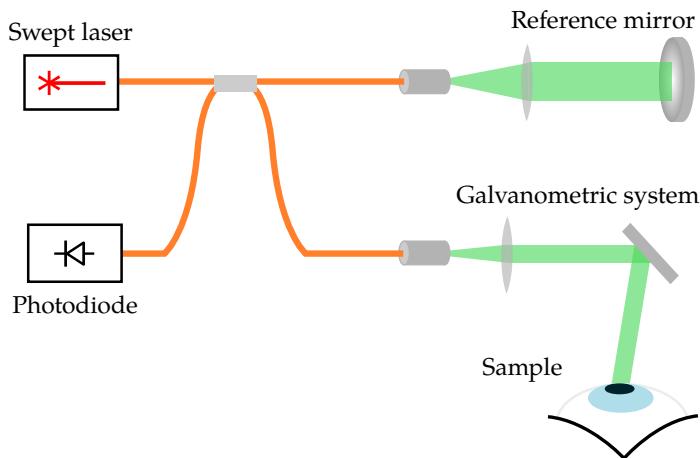


Figure 3.25: SS-OCT setup using a fiber Michelson interferometer.

This setup, while fairly simple, gave rise to two different issues:

1. a balanced detection scheme could not be implemented.

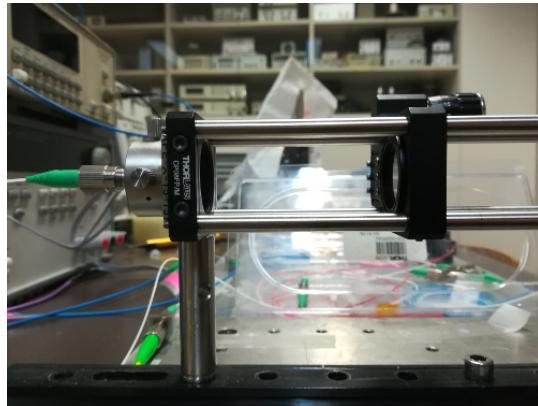


Figure 3.26: SS-OCT setup using a fiber Michelson interferometer.

2. the output of the optical source experienced instability, as half of the power reflected by the two arms is coupled back in the laser output port. This behaviour was tested by replacing the reference arm with a photodiode and monitoring its current. In [Figure 3.27](#) we can see that the power profile of the source is substantially distorted when a high-reflectivity object such as a mirror is placed under the focusing lens of the sample arm. With the additional power reflected by the reference mirror this behaviour can only worsen.

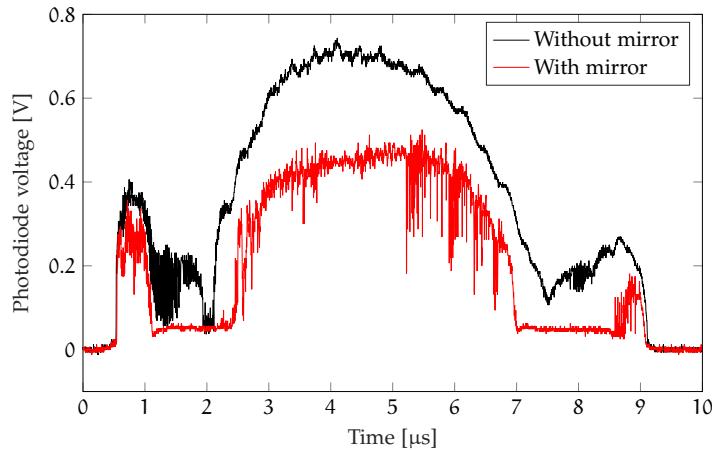


Figure 3.27: Instability of the laser output power.

3.4.5 Second scheme

In order to limit the amount of power returning to the laser and avoid permanent damage, the setup in [Figure 3.28](#) has been designed. Connecting the source to the 90-10 coupler, only 10% of the total output power is directed to the sample and only 10% of the reflected power is directed back to the laser, reducing the reflected power by a factor

of 25. The power profile of the source was measured by connecting a photodiode to connector D and monitoring its current. Unlike before, placing a mirror in the sample arm did not alter the shape of the laser pulse.

With respect to the previous scheme, the power returning to the laser from the reference arm is attenuated by ~ 2 dB thanks to the 90-10 coupler. The effect of the reference arm on the stability of the laser output was tested in a similar way, connecting the photodiode to connector I. With the silver mirror stilll on the sample arm, the FiberPort collimator was slightly mis-aligned until the measured power profile did not exhibit any distortion.

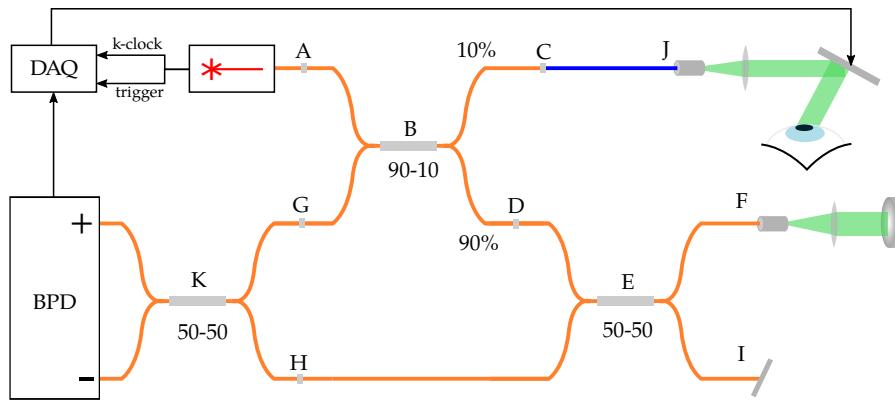


Figure 3.28: Final SS-OCT setup.

3.4.5.1 Balancing the interferometer

As already explained, interference fringes are only visible if the mismatch between the length of sample and reference arm is less than the coherence length of the source.

With reference to Figure 3.28, the length of the sample arm, defined by the letters ABCJCBGK is

$$L_{\text{sample}} = \overline{ABCJCBGK} = 5L_{\text{arm}} + 2L_{\text{patch}}, \quad (3.37)$$

where L_{arm} is the length of the couplers' arm and L_{patch} is the length of the fiber patch cord (illustrated in blue). The reference arm is instead defined by the path ABDEFEHK, whose length is

$$L_{\text{ref}} = \overline{ABDEFEHK} = 7L_{\text{arm}}, \quad (3.38)$$

meaning that, ideally

$$|L_{\text{ref}} - L_{\text{sample}}| = 0 \implies L_{\text{patch}} = L_{\text{arm}}, \quad (3.39)$$

Since $L_{\text{arm}} \sim 1$ m, a patch cord of the same length has to be used. Potentially, light travelling the path ABDEFEDBGK could cause unwanted interference, but since

$$|L_{\text{ref}} - \overline{ABDEFEDBGK}| = 2L_{\text{arm}} \approx 2 \text{ m}, \quad (3.40)$$

the mismatch is far greater than the coherence length of the source, $l_c \approx 12$ mm meaning that no interference will be generated.

This setup is therefore able to protect the laser source from damage caused by strong reflections, removes unwanted coherent noise and enables balanced detection. In order to minimize distortions due to the detection scheme, the BPD is connected to the fiber coupler with the smallest mismatch between adjacent ports: looking at [Table 3.5](#), the Blue and White arms of coupler 50-50 A are the optimal choice.

3.4.5.2 OFDR measurement

For a more precise analysis on the balancing of sample and reference arm, the difference between the two optical paths is measured with an OFDR using the setup in [Figure 3.29](#). The optical circulator enables us to select the path to measure, obtaining traces that are easier to interpret, while the fiber attenuator is used as a safety measure to protect the OFDR receiver.

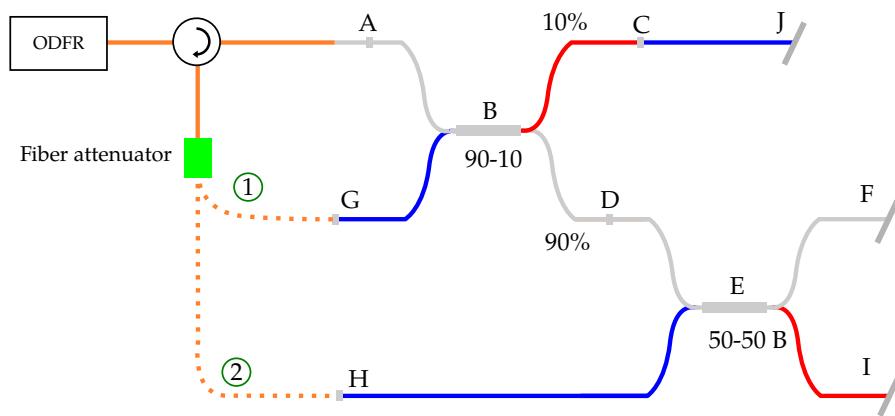


Figure 3.29: Time signal when the optical path difference is close to 0.

The acquired traces are illustrated in [Figure 3.30](#). From left to right we can observe:

- The reflection coming from connector C, when no patch cord is connected.
- The reflection coming from connector J, corresponding to the end of the fiber patch cord.
- The two reflections coming from connectors F and I. As we can see, the two are distanced by ~ 2 centimeters, confirming the measurements reported in [Table 3.5](#).

Using connector F to connect the reference system, the path length difference becomes $\Delta L \approx 70$ mm, as this path is travelled in the two directions. Since the interference between the two signals occurs in the fused section of coupler 50-50 A (point K in [Figure 3.28](#)), the length mismatch of its White and Red arms (≈ 6.2 mm) can be exploited to

bring the total **OPD** down to $\Delta L \approx 64$ mm. These 6 centimeters can be compensated with the translatable mirror in the reference arm.

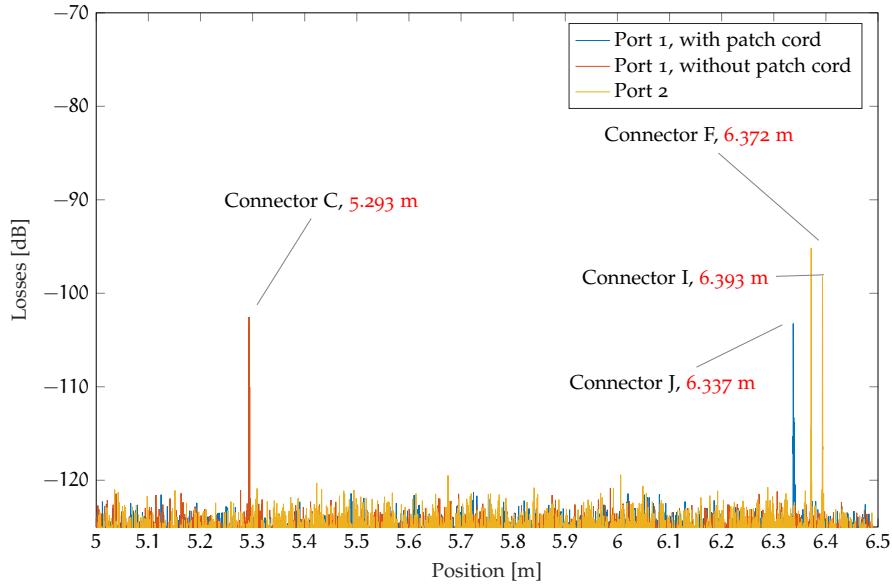


Figure 3.30: OFDR traces used to balance the interferometer.

After connecting the scanning system and the reference arm, a silver mirror was placed on the focal plane of the focusing lens and the interference signal detected by the photodiode was monitored with a high-speed oscilloscope. The reference mirror was translated until the beat frequency was close to 0. In [Figure 3.31](#) we can see the signal captured at three different instants and observe the low-frequency oscillation.

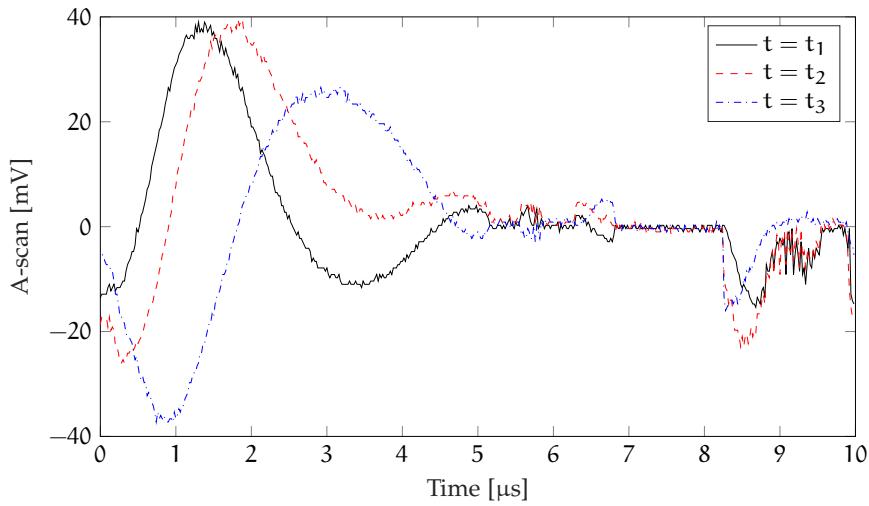
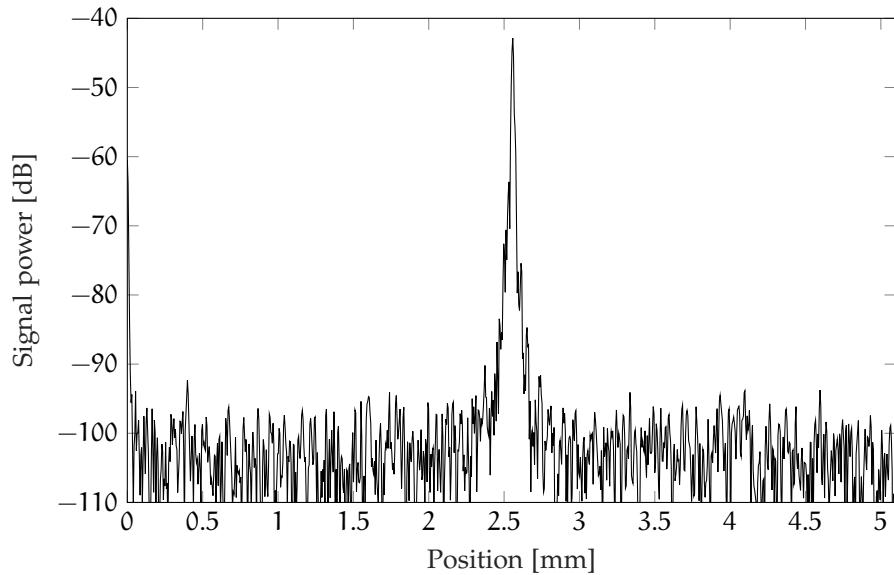


Figure 3.31: Time signal when the optical path difference is close to 0.

Finally, with the aid of the spectrum analyzer feature of the AlazarDSO software, the reference mirror was further adjusted in order to posi-

tion the focal plane of the lens in the middle of the imaging range. An A-scan of the silver mirror was acquired using the external clock and a Hamming apodization window, resulting in the plot in [Figure 3.32](#).



[Figure 3.32](#): Time signal when the optical path difference is close to 0.

3.4.6 Signal-to-noise ratio profile

In order to test the combined effect of sensitivity falloff and the limited DOV of the lens, multiple A-scans were acquired while placing the mirror at different imaging depths. This was accomplished by translating the scanning system on a vertical support with the help of micrometer screw. The whole imaging range was covered with 11 acquisitions at multiples of $\sim 500 \mu\text{m}$, resulting in [Figure 3.33](#). An additional acquisition has been acquired without placing the silver mirror on the sample arm to obtain the noise level of the system.

The signal-to-noise ratio ([SNR](#)) profile was then measured by subtracting the average noise level from the peak power of each A-scan, obtaining the graph in [Figure 3.34](#). The SNR dropoff factor was determined by fitting the data points to two lines using the Linear Least Squares method, obtaining $\alpha' = 5.6 \text{ dB/mm}$ for the first half of the imaging range and $\alpha'' = 7.1 \text{ dB/mm}$ for the second half, guaranteeing a 6 dB range of $\sim 1.86 \text{ mm}$.

3.4.7 Axial Resolution

The axial resolution of a OCT system is arguably its most important parameter. As discussed in [Chapter 2](#), the resolution of an SS-OCT

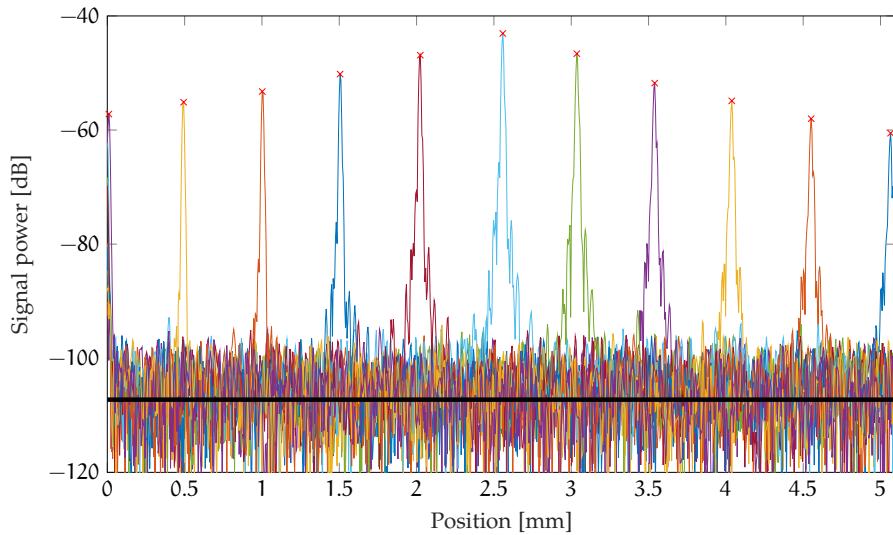


Figure 3.33: A-scans of a silver mirror at various imaging depths.

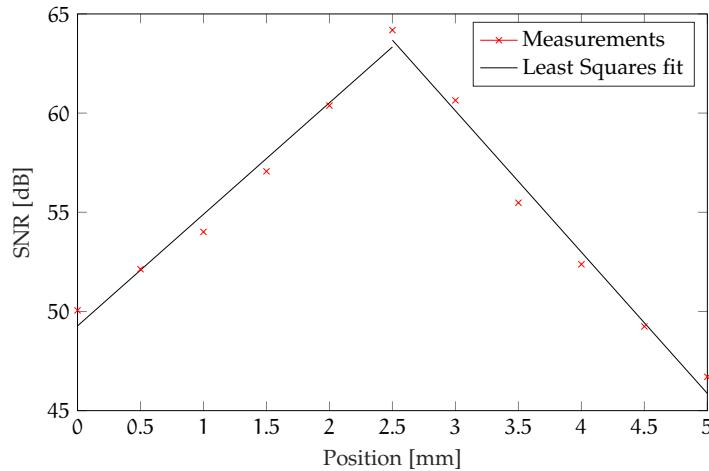


Figure 3.34: Signal-to-noise ratio falloff.

can be estimated in the case of a Gaussian source with [Equation 2.34](#). For our source, this value is found to be

$$\delta z = 0.75 \frac{\lambda_0^2}{\Delta\lambda} = 0.75 \frac{(1305 \times 10^{-9})^2}{140 \times 10^{-9}} \approx 9.12 \text{ } \mu\text{m}. \quad (3.41)$$

Since the spectrum of the laser is not Gaussian, the axial resolution has to be determined experimentally by measuring the FWHM of the A-scan generated by a mirror. This measurement will also take into account all the dispersion introduced by the optical components of the system. The acquired A-scans will first be spline-interpolated and then fit to a Lorentzian function of the form

$$L(x) = L_0 + \frac{A}{\pi} \frac{\frac{1}{2}\gamma}{(x - x_0)^2 + (\frac{1}{2}\gamma)^2}, \quad (3.42)$$

to determine the FWHM parameter $\gamma/2$. Figure 3.35 pictures the results of this method when using four different apodization functions. The Blackman-Harris window guarantees the worst result of the four, while the Hanning and Hamming windows improve the resolution by about 3 μm . The best result is instead achieved by the Kaiser window, which is just 2 μm short from reaching the theoretical value. A side effect of using the Kaiser window is the introduction of more pronounced side lobes which can result in distorted images. Further improvements to the axial resolution could be achieved by replacing the fiber patch cord connecting the interferometer to the scanning system. In fact, this patch cord is a Single Mode Fiber (SMF) centered at 1550 nm which is not designed to work in the 1300 nm range used by the laser, meaning that modal dispersion could impact the system's performance.

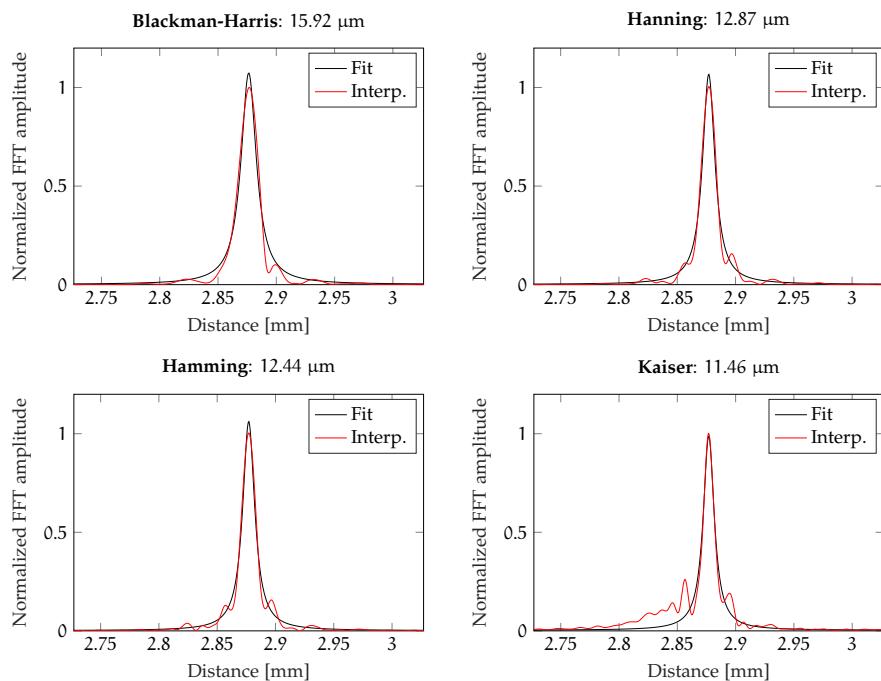


Figure 3.35: Estimate of the axial resolution by fitting a Lorentzian curve to the A-scan of mirror using a Nonlinear Least Squares method.

The same method was tested for different mirror depths in order to estimate the behaviour of the resolution over the whole imaging range. The obtained profile is presented in Figure 3.36. A few observations can be made

- The resolution worsens at higher imaging depths. This can be explained by the fact that light has to travel for longer distance in air, therefore enhancing the dispersion mismatch between the sample and reference arm.

- The Kaiser window exhibits a more irregular trend compared to the other three windows while still generally guaranteeing the best axial resolution.
- The variation in axial resolution is in the order of 2-3 μm over the entire imaging range for all three windows.

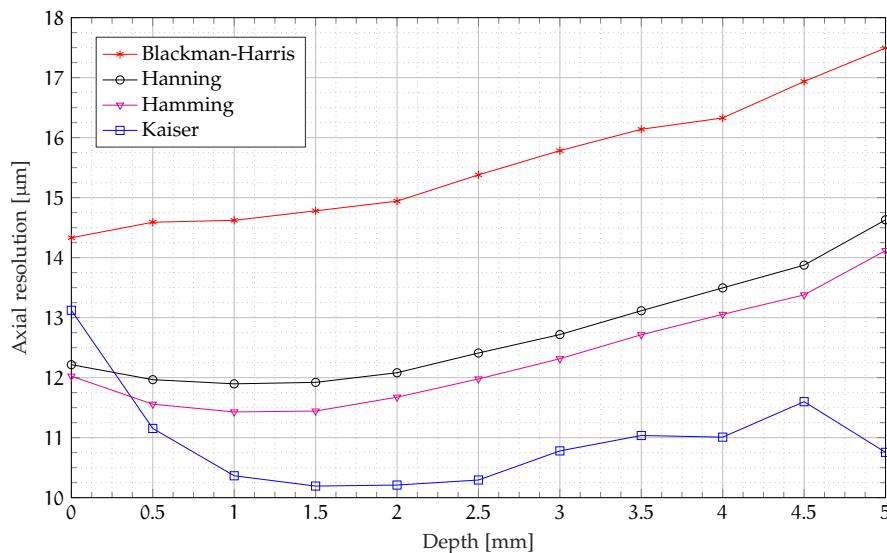


Figure 3.36: Axial resolution profile on the entire imaging depth using four different apodization windows.

3.4.8 Thickness measurements

One of the first

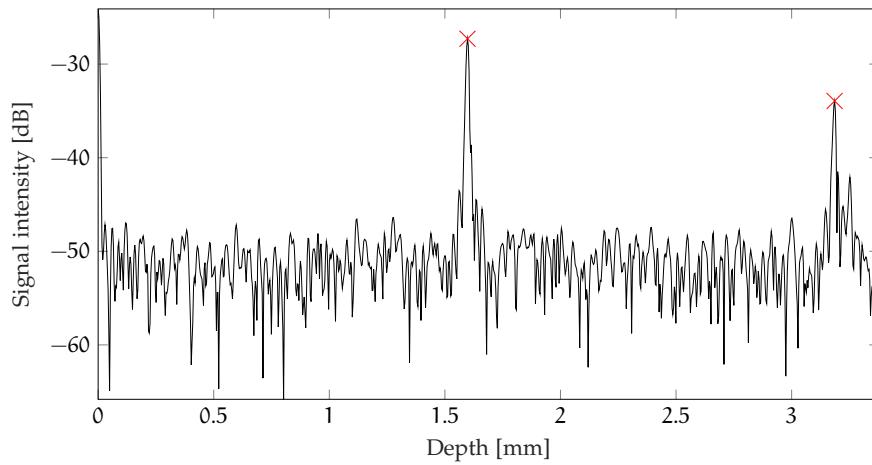


Figure 3.37: Axial measurement of the USAF target.

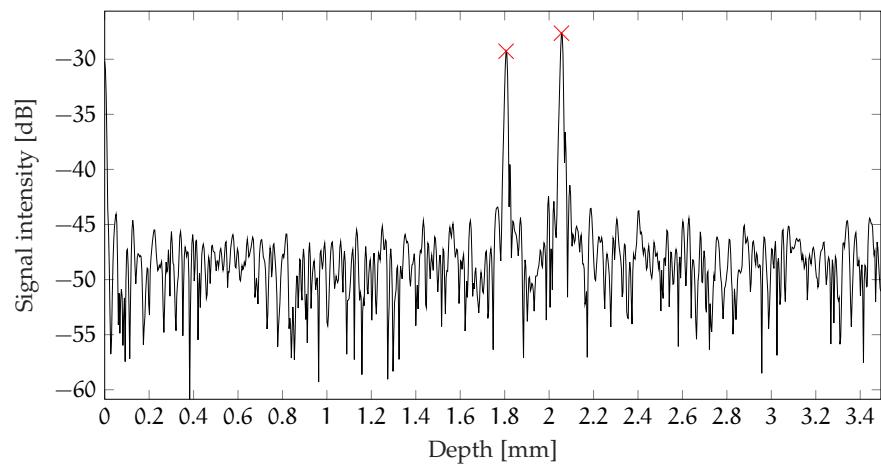


Figure 3.38: Axial measurement of an optical fiber.

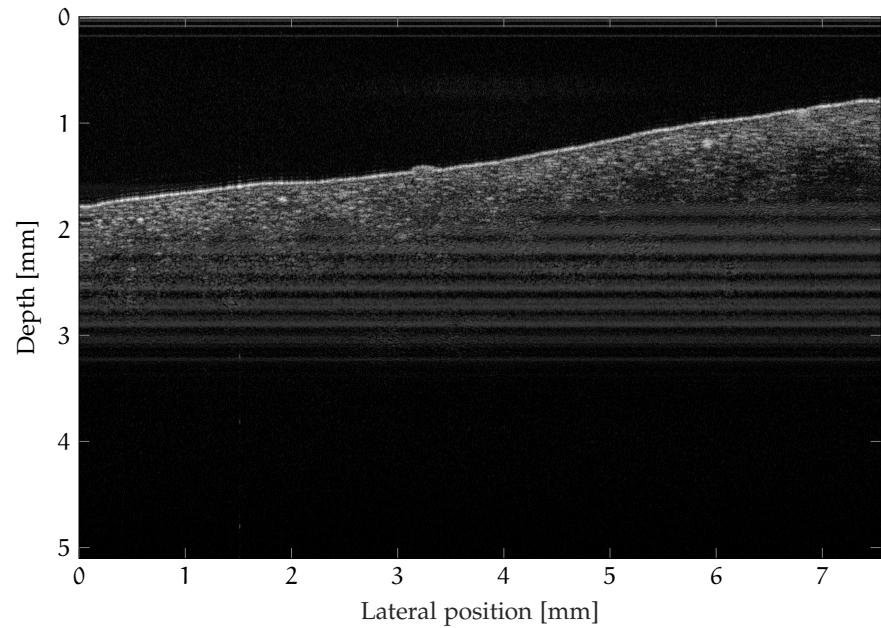


Figure 3.39: B-scan of banana peel.

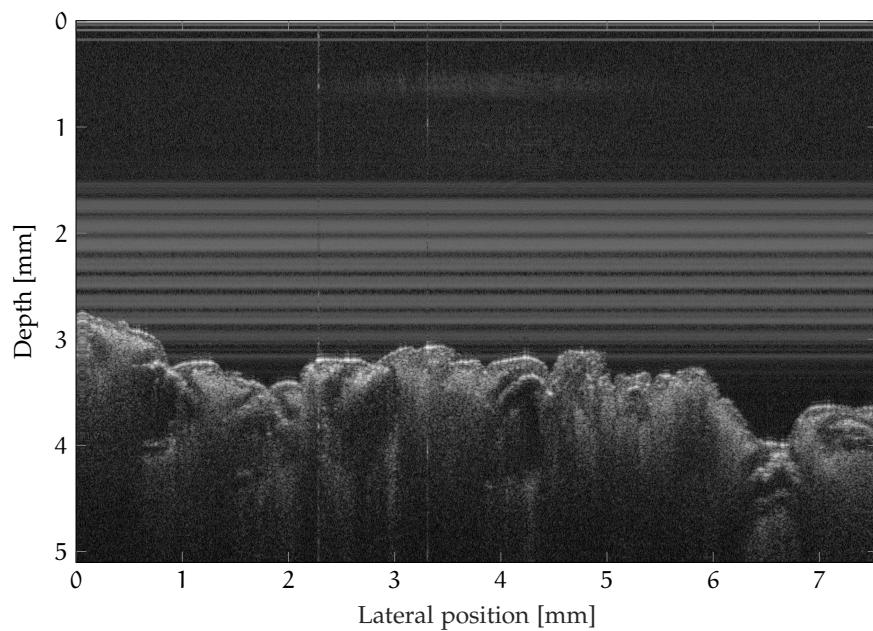


Figure 3.40: B-scan of dry orange peel.

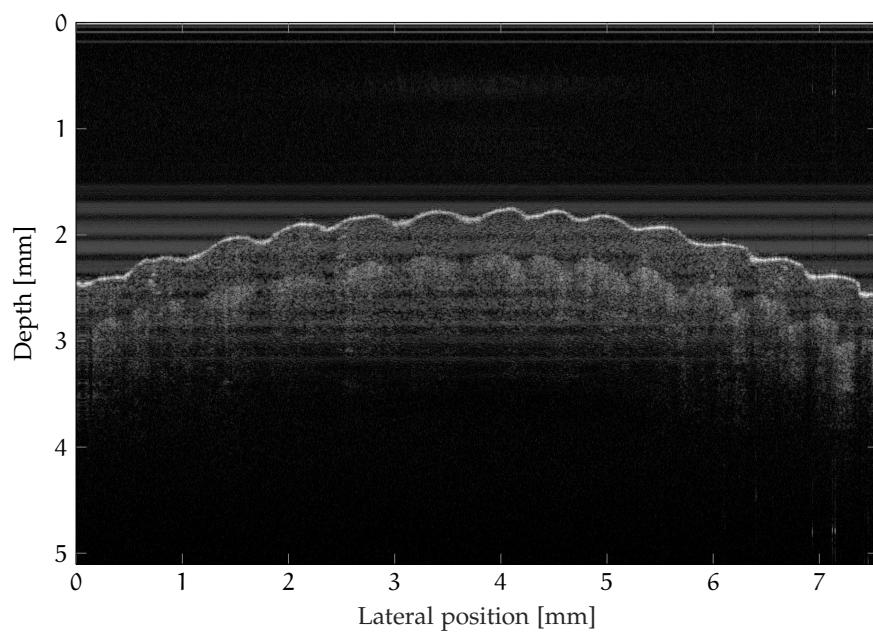


Figure 3.41: B-scan of a human finger.

4

RESULTS

4.1 DATA ACQUISITION SOFTWARE

Leave this chapter for the explanation of the software, technologies used, performance achieved and showcase of the measurements. B-scans, volumes and enface. Also include control of the galvo mirrors in here since it's about C++ programming.

Explain in detail the fact that for real time the average time to process has to be < frame time and explain buffer ring to get buffer from board using DMA. Check datasheet in the pdfviewer since there is a nice explanation of async drivers.

4.1.1 *Digital Signal Processing Chain for OCT*

4.1.2 *The Qt programming framework*

4.2 THE SIGNAL-SLOT PARADIGM

Blocking vs Non-blocking calls

4.2.1 *Displaying graphics with OpenGL*

4.2.2 *Parallel Computing with OpenMP*

OpenMP¹ is a multi-platform *Application Programming Interface* (API) for shared-memory parallel processing programming available for various programming languages, such as C, C++ and Fortran.

OpenMP stands for
Open
Multi-Processing

¹ <http://www.openmp.org/>

4.3 BASIC SETUP

4.3.0.1 Stability Analysis

4.3.1 New setup

4.3.2 Balancing the optical paths

4.3.2.1 Working mechanism of a OBR

4.3.2.2 Measuring path lengths

4.3.3 Axial Measurements

4.3.4 Amplified Photodiode

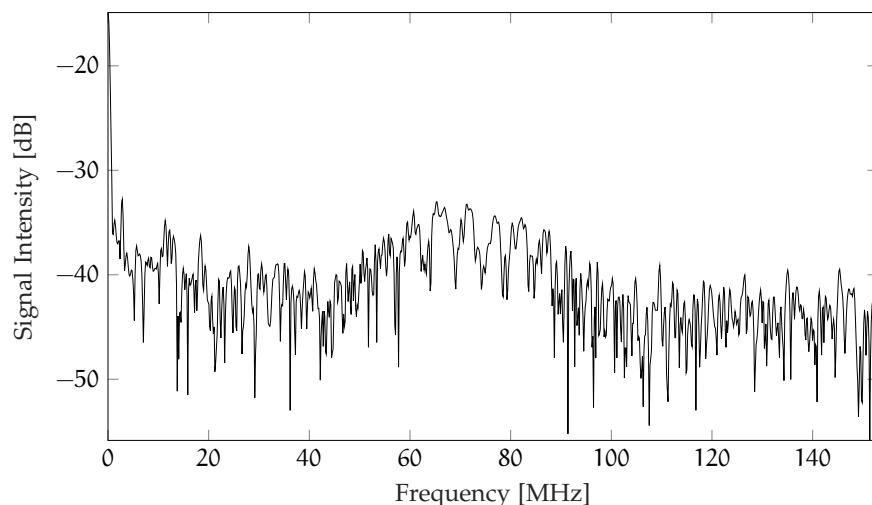


Figure 4.1: Spurious beat frequencies detected by the Exalos BPD.

Content

4.4 GALVANOMETRIC SYSTEM CONTROL

1. NI-DAQmx framework
2. C programming
3. Triggering and Clocking
4. Splitting Positive and Negative voltages
5. Conversion from volts to angles to surface area
6. Trigger enable
7. X-Motor -> Triangle Wave

8. Y-Motor -> Staircase for C-scans

4.5 USAF TARGET

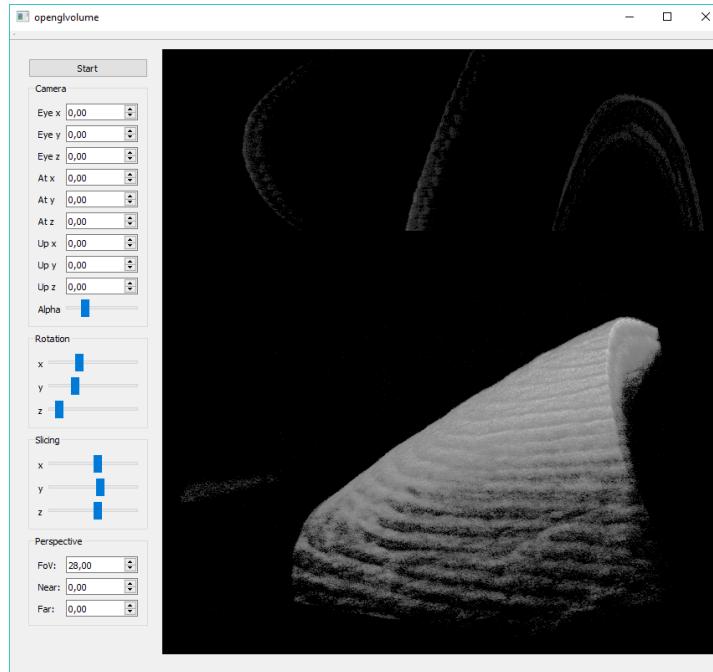


Figure 4.2: 3D rendering of a human finger.

1. What it is
2. How it works
3. Acquisitions in different conditions
4. B-scans
5. Surface images -> verifying transverse resolution

Content

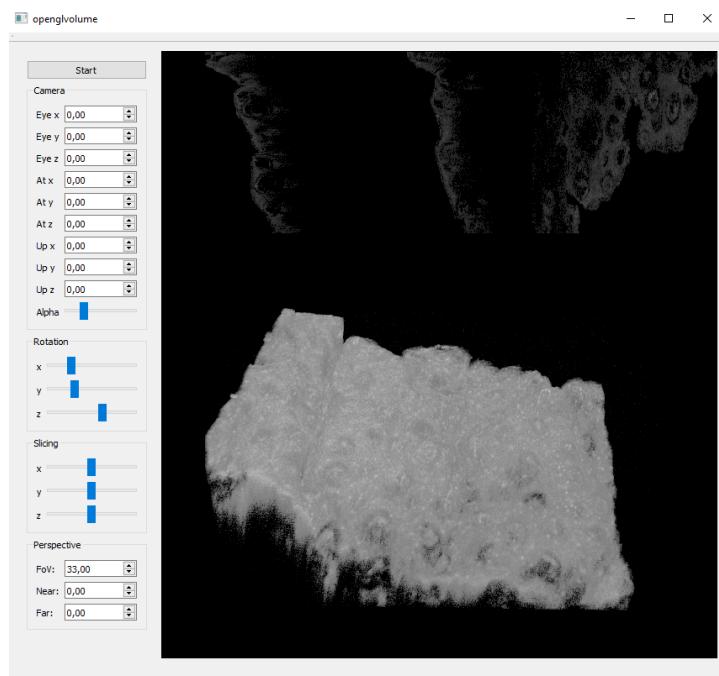


Figure 4.3: 3D rendering of a human finger.

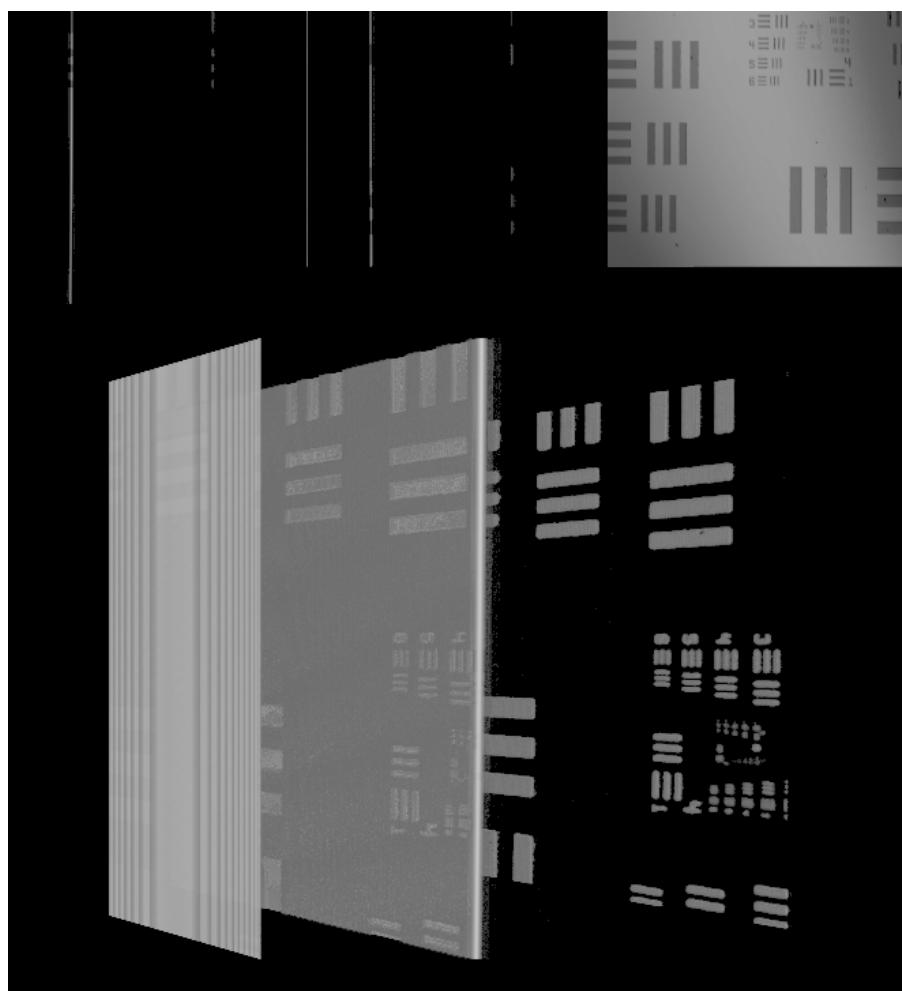


Figure 4.4: 3D rendering of the USAF target

5

CONCLUSIONS

Part I
APPENDIX

A

OFDR TRACES

[Figure A.1](#), [Figure A.2](#) and [Figure A.3](#) represent the OFDR traces acquired for determining the length of the arms of the three fiber couplers, as discussed in [Chapter 3](#). The title on top of each graph indicates the corresponding input port, while the color of the traces refers to the port which was connected to the patch cord (black is used instead of white).

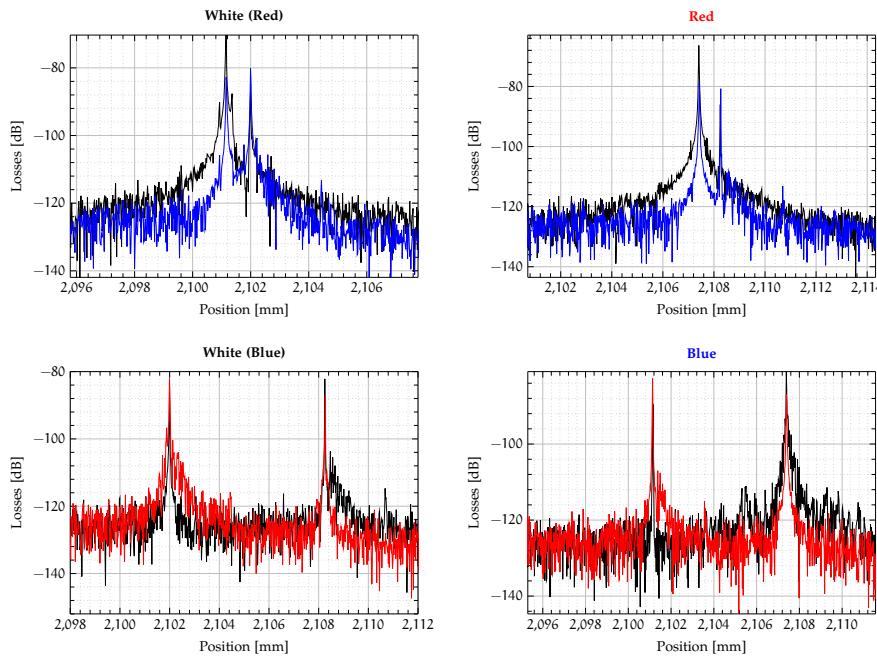


Figure A.1: OFDR traces of the first 50-50 coupler (Coupler A).

A.1 APPENDIX SECTION TEST

More dummy text

A.2 ANOTHER APPENDIX SECTION TEST

There is also a useless Pascal listing below: [Listing A.1](#).

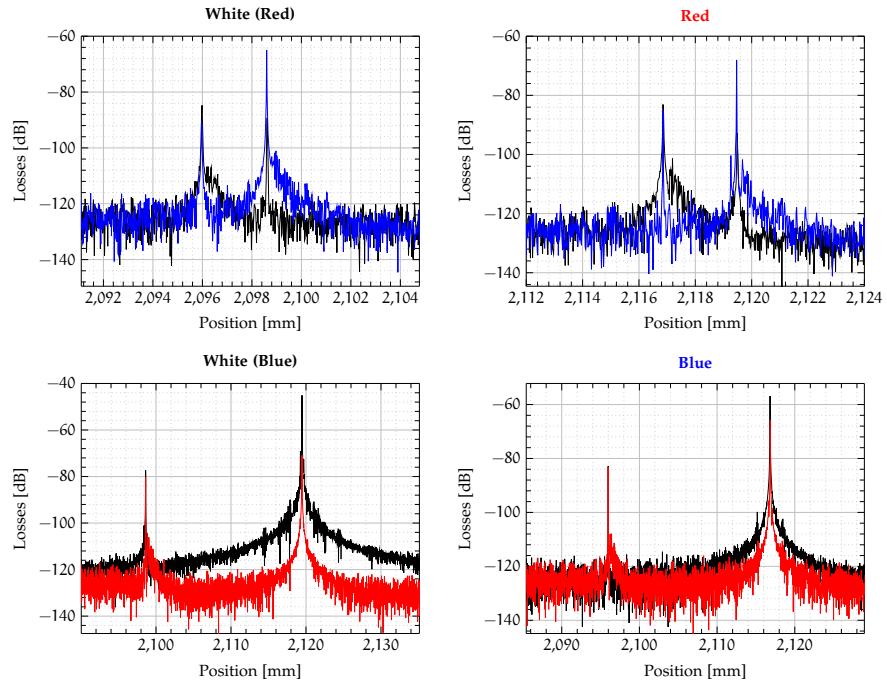


Figure A.2: OFDR traces of the second 50-50 coupler (Coupler B).

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suscipit instructior	titulo	personas
quaestio philosophia	facto	demonstrated

Table A.1: Autem usu id.

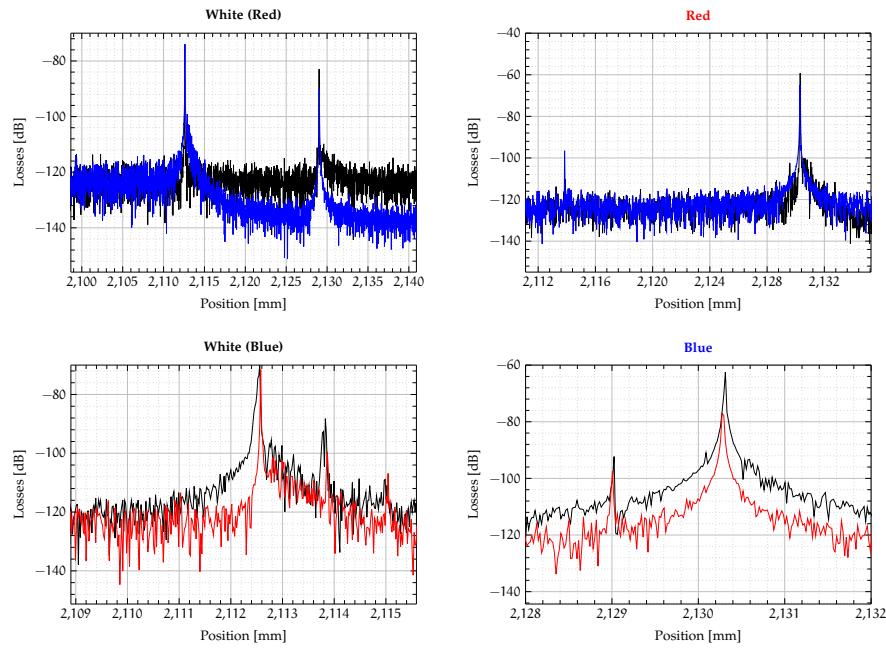


Figure A.3: OFDR traces of the 90-10 coupler.

Listing A.1: A floating example (listings manual)

```

int index = 0;
float timeout = 2000;
while (! stopAcquisition)
{
    index = index mod buffercount

    short integer* buffer = AlazarWaitBuffer(bufferArray[
        index], timeout);

    float* toDisk = processBuffer(buffer);
    byte* Bscan = createBitmap(buffer);

    emit newBuffer(toDisk);
    emit newFrame(Bscan);

    index++;
}

```

BIBLIOGRAPHY

- [1] Erkki Alarousu, Leszek Krehut, Tuukka Prykäri, and Risto Myllylä. «Study on the use of optical coherence tomography in measurements of paper properties.» In: *Measurement Science and Technology* 16.5 (2005), p. 1131. URL: <http://stacks.iop.org/0957-0233/16/i=5/a=012>.
- [2] B. T. Amaechi, S. M. Higham, A. GH. Podoleanu, J. A. Rogers, and D. A. Jackson. «Use of optical coherence tomography for assessment of dental caries: quantitative procedure.» In: *Journal of Oral Rehabilitation* 28.12 (2001), pp. 1092–1093. ISSN: 0305-182X. DOI: [10.1046/j.1365-2842.2001.00840.x](https://doi.wiley.com/10.1046/j.1365-2842.2001.00840.x). URL: <http://doi.wiley.com/10.1046/j.1365-2842.2001.00840.x>.
- [3] Johannes F. de Boer, Thomas E. Milner, Martin J. C. van Gemert, and J. Stuart Nelson. «Two-dimensional birefringence imaging in biological tissue by polarization-sensitive optical coherence tomography.» In: *Optics Letters* 22.12 (1997), p. 934. ISSN: 0146-9592. DOI: [10.1364/OL.22.000934](https://doi.osapublishing.org/abstract.cfm?URI=ol-22-12-934). URL: <https://www.osapublishing.org/abstract.cfm?URI=ol-22-12-934>.
- [4] Stephen A. Boppart, Juergen Herrmann, Costas Pitris, Debra L. Stamper, Mark E. Brezinski, and James G. Fujimoto. «High-Resolution Optical Coherence Tomography-Guided Laser Ablation of Surgical Tissue.» In: *Journal of Surgical Research* 82.2 (1999), pp. 275–284. ISSN: 00224804. DOI: [10.1006/jsre.1998.5555](https://linkinghub.elsevier.com/retrieve/pii/S002248049895555X). URL: [http://linkinghub.elsevier.com/retrieve/pii/S002248049895555X](https://linkinghub.elsevier.com/retrieve/pii/S002248049895555X).
- [5] Stephen A. Boppart, Wei Luo, Daniel L. Marks, and Keith W. Singletary. «Optical Coherence Tomography: Feasibility for Basic Research and Image-guided Surgery of Breast Cancer.» In: *Breast Cancer Research and Treatment* 84.2 (2004), pp. 85–97. ISSN: 1573-7217. DOI: [10.1023/B:BREA.0000018401.13609.54](https://doi.org/10.1023/B:BREA.0000018401.13609.54). URL: <https://doi.org/10.1023/B:BREA.0000018401.13609.54>.
- [6] Max Born, Emil Wolf, A. B. Bhatia, P. C. Clemmow, D. Gabor, A. R. Stokes, A. M. Taylor, P. A. Wayman, and W. L. Wilcock. *Principles of Optics: Electromagnetic Theory of Propagation, Interference and Diffraction of Light*. 7th ed. Cambridge University Press, 1999. DOI: [10.1017/CBO9781139644181](https://doi.org/10.1017/CBO9781139644181).
- [7] B E Bouma et al. «Evaluation of intracoronary stenting by intravascular optical coherence tomography.» In: *Heart* 89.3 (2003), pp. 317–320. ISSN: 1355-6037. DOI: [10.1136/heart.89.3.317](https://doi.org/10.1136/heart.89.3.317). eprint: <http://heart.bmjjournals.org/content/89/3/317.full.pdf>. URL: <http://heart.bmjjournals.org/content/89/3/317>.

- [8] Brian Cabral, Nancy Cam, and Jim Foran. «Accelerated Volume Rendering and Tomographic Reconstruction Using Texture Mapping Hardware.» In: *Proceedings of the 1994 symposium on Volume visualization - VVS '94*. New York, New York, USA: ACM Press, 1994, pp. 91–98. ISBN: 0897917413. DOI: [10.1145/197938.197972](https://doi.org/10.1145/197938.197972). URL: <http://portal.acm.org/citation.cfm?doid=197938.197972>.
- [9] Marco Calabrese. «Progetto di un Tomografo a Coerenza Ottica ad Alta Velocità.» MA thesis. Dipartimento di Ingegneria dell'Informazione, Padova (Italy): Università degli Studi di Padova, 2017.
- [10] Dong-hak Choi, Hideaki Hiro-Oka, Kimiya Shimizu, and Kohji Ohbayashi. «Spectral domain optical coherence tomography of multi-MHz A-scan rates at 1310 nm range and real-time 4D-display up to 41 volumes/second.» In: *Biomedical Optics Express* 3.12 (2012), p. 3067. ISSN: 2156-7085. DOI: [10.1364/BOE.3.003067](https://doi.org/10.1364/BOE.3.003067). URL: <https://www.osapublishing.org/abstract.cfm?URI=boe-3-12-3067>.
- [11] Tzu H. Chow, Khay M. Tan, Beng K. Ng, Sirajudeen G. Razul, Chia M. Tay, Tet F. Chia, and Wee T. Poh. «Diagnosis of virus infection in orchid plants with high-resolution optical coherence tomography.» In: *Journal of Biomedical Optics* 14.1 (2009), p. 014006. ISSN: 10833668. DOI: [10.1117/1.3066900](https://doi.org/10.1117/1.3066900). URL: <http://biomedicaloptics.spiedigitallibrary.org/article.aspx?doi=10.1117/1.3066900>.
- [12] Seyed Hamid Hosseiny Darbazi. «GPU-Accelerated Optical Coherence Tomography Signal Processing and Visualization.» PhD thesis. Universidade do Porto, 2016.
- [13] Anjul M. Davis, Michael A. Choma, and Joseph A. Izatt. «Heterodyne swept-source optical coherence tomography for complete complex conjugate ambiguity removal.» In: *Journal of Biomedical Optics* 10.6 (2005), p. 064005. ISSN: 10833668. DOI: [10.1117/1.2136147](https://doi.org/10.1117/1.2136147). URL: <http://biomedicaloptics.spiedigitallibrary.org/article.aspx?doi=10.1117/1.2136147>.
- [14] Al-Hafeez Dhalla, Derek Nankivil, and Joseph a. Izatt. «Complex conjugate resolved heterodyne swept source optical coherence tomography using coherence revival.» In: *Biomedical Optics Express* 3.3 (2012), p. 633. ISSN: 2156-7085. DOI: [10.1364/BOE.3.000633](https://doi.org/10.1364/BOE.3.000633).
- [15] Wolfgang Drexler and James G. Fujimoto, eds. *Optical Coherence Tomography*. Cham: Springer International Publishing, 2015. ISBN: 978-3-319-06418-5. DOI: [10.1007/978-3-319-06419-2](https://doi.org/10.1007/978-3-319-06419-2). URL: <http://link.springer.com/10.1007/978-3-319-06419-2>.

- [16] Klaus Engel, Martin Kraus, and Thomas Ertl. «High-quality pre-integrated volume rendering using hardware-accelerated pixel shading.» In: *Proceedings of the ACM SIGGRAPH/EUROGRAPHICS workshop on Graphics hardware - HWWS '01*. New York, New York, USA: ACM Press, 2001, pp. 9–16. ISBN: 158113407X. doi: [10.1145/383507.383515](https://doi.org/10.1145/383507.383515). URL: <http://portal.acm.org/citation.cfm?doid=383507.383515>.
- [17] P. F. Escobar, J. L. Belinson, A. White, N. M. Shakhova, F. I. Feldchtein, M. V. Karetz, and N. D. Gladkova. «Diagnostic efficacy of optical coherence tomography in the management of preinvasive and invasive cancer of uterine cervix and vulva.» In: *International Journal of Gynecological Cancer* 14.3 (2004), pp. 470–474. ISSN: 1048-891X. doi: [10.1111/j.1048-891x.2004.14307.x](https://doi.wiley.com/10.1111/j.1048-891x.2004.14307.x). URL: <http://doi.wiley.com/10.1111/j.1048-891x.2004.14307.x>.
- [18] Daniel Fried, John Xie, Sahar Shafi, John D. B. Featherstone, Thomas M. Breunig, and Charles Le. «Imaging caries lesions and lesion progression with polarization sensitive optical coherence tomography.» In: *Journal of Biomedical Optics* 7.4 (2002), p. 618. ISSN: 10833668. doi: [10.1117/1.1509752](https://doi.org/10.1117/1.1509752). URL: <http://biomedicaloptics.spiedigitallibrary.org/article.aspx?doi=10.1117/1.1509752>.
- [19] J G Fujimoto, C a Puliafito, R Margolis, A Oseroff, S De Silvestri, and E P Ippen. «Femtosecond optical ranging in biological systems.» In: *Optics Letters* 11.3 (1986), p. 150. ISSN: 0146-9592. doi: [10.1364/OL.11.000150](https://doi.org/10.1364/OL.11.000150). URL: <https://www.osapublishing.org/abstract.cfm?URI=ol-11-3-150>.
- [20] G. Loeb and J. K. Barton. «Imaging Botanical Subjects With Optical Coherence Tomography: A Feasibility Study.» In: *Transactions of the ASAE* 46.6 (2003), pp. 1751–1757. ISSN: 2151-0059. doi: [10.13031/2013.15616](https://elibRARY.asabe.org/abstract.asp?JID=3&AID=15616&CID=t2003&v=46&i=6&T=1). URL: <http://elibRARY.asabe.org/abstract.asp?JID=3&AID=15616&CID=t2003&v=46&i=6&T=1>.
- [21] T Gambichler, G Moussa, M Sand, D Sand, P Altmeyer, and K Hoffmann. «Applications of optical coherence tomography in dermatology.» In: *Journal of Dermatological Science* 40.2 (2005), pp. 85–94. ISSN: 09231811. doi: [10.1016/j.jdermsci.2005.07.006](https://doi.org/10.1016/j.jdermsci.2005.07.006). URL: <http://linkinghub.elsevier.com/retrieve/pii/S092318110500201X>.
- [22] Alexandre R. Hariri Lidaand Tumlinson, David G. Besselsen, Urs Utzinger, Eugene W. Gerner, and Jennifer K. Barton. «Endoscopic optical coherence tomography and laser-induced fluorescence spectroscopy in a murine colon cancer model.» In: *Lasers in Surgery and Medicine* 38.4 (2006), pp. 305–313. ISSN: 0196-8092.

- DOI: [10.1002/lsm.20305](https://doi.org/10.1002/lsm.20305). URL: <http://doi.wiley.com/10.1002/lsm.20305>.
- [23] David Huang et al. «Optical Coherence Tomography HHS Public Access.» In: *Science*. November 22.2545035 (1991), pp. 1178–1181. ISSN: 0036-8075. DOI: [10.1126/science.2545035](https://doi.org/10.1126/science.2545035). arXiv: [1533.4406](https://arxiv.org/abs/1533.4406).
 - [24] Ik-Kyung Jang et al. «Visualization of coronary atherosclerotic plaques in patients using optical coherence tomography: comparison with intravascular ultrasound.» In: *Journal of the American College of Cardiology* 39.4 (2002), pp. 604–609. ISSN: 0735-1097. DOI: [10.1016/S0735-1097\(01\)01799-5](https://doi.org/10.1016/S0735-1097(01)01799-5). eprint: <http://www.onlinejacc.org/content/39/4/604.full.pdf>. URL: <http://www.onlinejacc.org/content/39/4/604>.
 - [25] Yali Jia et al. «Quantitative Optical Coherence Tomography Angiography of Choroidal Neovascularization in Age-Related Macular Degeneration.» In: *Ophthalmology* 121.7 (2014), pp. 1435–1444. ISSN: 01616420. DOI: [10.1016/j.ophtha.2014.01.034](https://doi.org/10.1016/j.ophtha.2014.01.034). URL: <http://linkinghub.elsevier.com/retrieve/pii/S0161642014001043>.
 - [26] Vrushali R. Korde et al. «Using optical coherence tomography to evaluate skin sun damage and precancer.» In: *Lasers in Surgery and Medicine* 39.9 (2007), pp. 687–695. ISSN: 01968092. DOI: [10.1002/lsm.20573](https://doi.org/10.1002/lsm.20573). arXiv: [NIHMS150003](https://arxiv.org/abs/1500.003).
 - [27] Philippe Lacroute and Marc Levoy. «Fast volume rendering using a shear-warp factorization of the viewing transformation.» In: *Proceedings of the 21st annual conference on Computer graphics and interactive techniques - SIGGRAPH '94*. New York, New York, USA: ACM Press, 1994, pp. 451–458. ISBN: 0897916670. DOI: [10.1145/192161.192283](https://doi.org/10.1145/192161.192283). URL: <http://portal.acm.org/citation.cfm?doid=192161.192283>.
 - [28] Haida Liang, Marta Gomez Cid, R. G. Cucu, G. M. Dobre, A. Gh. Podoleanu, Justin Pedro, and David Saunders. «En-face optical coherence tomography - a novel application of non-invasive imaging to art conservation.» In: *Optics Express* 13.16 (2005), p. 6133. ISSN: 1094-4087. DOI: [10.1364/OE.13.006133](https://doi.org/10.1364/OE.13.006133). URL: <https://www.osapublishing.org/abstract.cfm?URI=oe-13-16-6133>.
 - [29] Wenchao Liao, Tianyuan Chen, Chengming Wang, Wenxin Zhang, Zhangkai Peng, Xiao Zhang, Shengnan Ai, Deyong Fu, Tieying Zhou, and Ping Xue. «Endoscopic optical coherence tomography with a focus-adjustable probe.» In: *Optics Letters* 42.20 (2017), p. 4040. ISSN: 0146-9592. DOI: [10.1364/OL.42.004040](https://doi.org/10.1364/OL.42.004040). URL: <https://www.osapublishing.org/abstract.cfm?URI=ol-42-20-4040>.

- [30] Monika Machoy, Julia Seeliger, Liliana Szyszka-Sommerfeld, Robert Koprowski, Tomasz Gedrange, and Krzysztof Woźniak. «The Use of Optical Coherence Tomography in Dental Diagnostics: A State-of-the-Art Review.» In: *Journal of Healthcare Engineering* 2017 (2017). ISSN: 20402309. DOI: [10.1155/2017/7560645](https://doi.org/10.1155/2017/7560645).
- [31] Mohammad Sultan Mahmud, David W. Cadotte, Barry Vuong, Carry Sun, Timothy W. H. Luk, Adrian Mariampillai, and Victor X. D. Yang. «Review of speckle and phase variance optical coherence tomography to visualize microvascular networks.» In: *Journal of Biomedical Optics* 18.5 (2013), p. 050901. ISSN: 1083-3668. DOI: [10.1117/1.JBO.18.5.050901](https://doi.org/10.1117/1.JBO.18.5.050901). URL: <http://biomedicaloptics.spiedigitallibrary.org/article.aspx?doi=10.1117/1.JBO.18.5.050901>.
- [32] MS; Michael R. Hee. «Quantitative Assessment of Macular Edema With Optical Coherence Tomography.» In: *Journal of Chemical Information and Modeling* 39.9 (1995), pp. 1689–1699. ISSN: 1098-6596. DOI: [10.1017/CBO9781107415324.004](https://doi.org/10.1017/CBO9781107415324.004). arXiv: [arXiv:1011.1669v3](https://arxiv.org/abs/1011.1669v3).
- [33] Michele Midrio. *Campi Elettromagnetici*. Via Lagrange 3, Padova - 35143: SGE Editoriali, 2006. ISBN: 88-86281-82-X.
- [34] T Prykari, J Czajkowski, E Alarousu, and R Myllyla. «Optical coherence tomography as an accurate inspection and quality evaluation technique in paper industry.» In: *Optical Review* 17.3 (2010), pp. 218–222. ISSN: 1340-6000. DOI: [10.1007/s10043-010-0039-y](https://doi.org/10.1007/s10043-010-0039-y).
- [35] Naresh Kumar Ravichandran, Eranga Wijesinghe, Seung-yeol Lee, Muhammad Faizan, Kibeom Park, Hee-young Jung, Mansik Jeon, and Jeehyun Kim. «Swept source optical coherence tomography for in vivo growth monitoring of capsicum annuum seeds treated with different NaCl concentrations.» In: 10323 (2017), pp. 1–4. ISSN: 1996756X. DOI: [10.1117/12.2267457](https://doi.org/10.1117/12.2267457).
- [36] M Sabuncu and M Akdogan. «Photonic Imaging with Optical Coherence Tomography for Quality Monitoring in the Poultry Industry: a Preliminary Study.» In: *Revista Brasileira de Ciência Avícola* 17.3 (2015), pp. 319–324.
- [37] Marinko V. Sarunic, Michael A. Choma, Changhuei Yang, and Joseph A. Izatt. «Instantaneous complex conjugate resolved spectral domain and swept-source OCT using 3x3 fiber couplers.» In: *Optics Express* 13.3 (2005), p. 957. ISSN: 1094-4087. DOI: [10.1364/OPEX.13.000957](https://doi.org/10.1364/OPEX.13.000957). URL: <https://www.osapublishing.org/oe/abstract.cfm?uri=oe-13-3-957>.
- [38] Carlo G. Someda. *Electromagnetic Waves*. 2nd ed. CRC, 2006.

- [39] Richard F. Spaide, James M. Klancnik, and Michael J. Cooney. «Retinal Vascular Layers Imaged by Fluorescein Angiography and Optical Coherence Tomography Angiography.» In: *JAMA Ophthalmology* 133.1 (2015), p. 45. ISSN: 2168-6165. DOI: [10.1001/jamaophthalmol.2014.3616](https://doi.org/10.1001/jamaophthalmol.2014.3616). URL: <http://archopht.jamanetwork.com/article.aspx?doi=10.1001/jamaophthalmol.2014.3616>.
- [40] Rong Su, Mikhail Kirillin, Ernest W. Chang, Ekaterina Sergeeva, Seok H. Yun, and Lars Mattsson. «Perspectives of mid-infrared optical coherence tomography for inspection and micrometrology of industrial ceramics.» In: *Optics Express* 22.13 (2014), p. 15804. ISSN: 1094-4087. DOI: [10.1364/OE.22.015804](https://doi.org/10.1364/OE.22.015804). URL: <https://www.osapublishing.org/oe/abstract.cfm?uri=oe-22-13-15804>.
- [41] Yuankai K. Tao, Mingtao Zhao, and Joseph A. Izatt. «High-speed complex conjugate resolved retinal spectral domain optical coherence tomography using sinusoidal phase modulation.» In: *Optics Letters* 32.20 (2007), p. 2918. ISSN: 0146-9592. DOI: [10.1364/OL.32.002918](https://doi.org/10.1364/OL.32.002918). URL: <https://www.osapublishing.org/abstract.cfm?URI=ol-32-20-2918>.
- [42] G. J. Tearney, S. A. Boppart, B. E. Bouma, M. E. Brezinski, N. J. Weissman, J. F. Southern, and J. G. Fujimoto. «Scanning single-mode fiber optic catheter–endoscope for optical coherence tomography.» In: *Opt. Lett.* 21.7 (1996), pp. 543–545. DOI: [10.1364/OL.21.000543](https://doi.org/10.1364/OL.21.000543). URL: <http://ol.osa.org/abstract.cfm?URI=ol-21-7-543>.
- [43] Karin Wiesauer, Michael Pircher, Erich Götzinger, Siegfried Bauer, Rainer Engelke, Gisela Ahrens, Gabi Grützner, Christoph Hitzenberger, and David Stifter. «En-face scanning optical coherence tomography with ultra-high resolution for material investigation.» In: *Optics express* 13.3 (2005), pp. 1015–1024. ISSN: 1094-4087. DOI: [10.1364/OPEX.13.001015](https://doi.org/10.1364/OPEX.13.001015).
- [44] Wolfgang Wieser, Wolfgang Draxinger, Thomas Klein, Sebastian Karpf, Tom Pfeiffer, and Robert Huber. «High definition live 3D-OCT in vivo: design and evaluation of a 4D OCT engine with 1 GVoxel/s.» In: *Biomed. Opt. Express* 5.9 (2014), pp. 2963–2977. DOI: [10.1364/BOE.5.002963](https://doi.org/10.1364/BOE.5.002963). URL: <http://www.osapublishing.org/boe/abstract.cfm?URI=boe-5-9-2963>.
- [45] Maciej Wojtkowski, Rainer Leitgeb, Andrzej Kowalczyk, Tomasz Bajraszewski, and Adolf F. Fercher. «In vivo human retinal imaging by Fourier domain optical coherence tomography.» In: *Journal of Biomedical Optics* 7.3 (2002), p. 457. ISSN: 10833668. DOI: [10.1117/1.1482379](https://doi.org/10.1117/1.1482379). URL: <http://biomedicaloptics.spiedigitallibrary.org/article.aspx?doi=10.1117/1.1482379>.

- [46] Tuqiang Xie, Huikai Xie, Gary K. Fedder, and Yingtian Pan. «Endoscopic optical coherence tomography with a modified microelectromechanical systems mirror for detection of bladder cancers.» In: *Applied Optics* 42.31 (2003), p. 6422. ISSN: 0003-6935. DOI: [10.1364/AO.42.006422](https://doi.org/10.1364/AO.42.006422). URL: <https://www.osapublishing.org/abstract.cfm?URI=ao-42-31-6422>.
- [47] Kang Zhang and Jin U. Kang. «Real-time 4D signal processing and visualization using graphics processing unit on a regular nonlinear-k Fourier-domain OCT system.» In: *Opt. Express* 18.11 (2010), pp. 11772–11784. DOI: [10.1364/OE.18.011772](https://doi.org/10.1364/OE.18.011772). URL: <http://www.opticsexpress.org/abstract.cfm?URI=oe-18-11-11772>.