1550 nm Superluminescent Diode and Anti-Stokes Effect
CCD camera based Optical Coherence Tomography for Full-Field Optical Metrology

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Abstract

1550 nm Superluminescent Diode and Anti-Stokes Effect CCD camera based Optical Coherence Tomography for Full-Field Optical Metrology
by Lukasz Kredzinski.

Optical Coherence Tomography (OCT) is a well-established non-invasive imaging technology capable of carrying out 3D high-resolution cross-sectional images of the internal microstructure of examined material. However, almost all of these systems are expensive, requiring the use of complex optical setups, expensive light sources and complicated scanning of the sample under test. In addition, most of these systems have not taken advantage of the competitively priced optical components available at wavelength within the main optical communications band located in the 1550 nm region. A comparatively simple and inexpensive full-field OCT system (FF-OCT), based on a Superluminescent Diode (SLD) with an Erbium-Doped Fiber Amplifier (EDFA) light source and anti-stokes imaging device was constructed, to perform 3D cross-sectional imaging. This kind of inexpensive setup with moderate resolution could be applicable in low-level biomedical and industrial diagnostics. This work involves assembly, calibration of the system and determines its suitability for imaging structures of biological tissues such as teeth, which has low absorption at 1550 nm. The first ever 1550 nm full-field OCT system is presented.
Declaration

I hereby allege that this Thesis, which I now submit to meet the requirements for the degree of Doctorate of Philosophy, is entirely my own work and no portion of it has been submitted for assessment for another degree or qualification of University of Limerick or any other institution of learning. Where the work of other people has been used, it has been fully referenced and acknowledged.

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Michael J. Connelly
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1. Introduction.

1.1. Motivation.

Optical Coherence Tomography (OCT) is a well-known and powerful technique for carrying out 3D cross-sectional imaging for biomedical and industrial applications. The major part of work on OCT up to date, has aimed towards improving its functional parameters, such as resolution or dynamic range. However, almost all of these systems are expensive requiring the use of complex optical setups, expensive light sources and complicated scanning of the sample under test. In addition most of these systems have not taken advantage of the competitively priced optical components available at wavelengths within the main optical communications band located in the 1550 nm region. It is believed, there is a need for a cost effective OCT system to perform 3D topographic measurement with moderate resolution, which would have applications in low-level biomedical and industrial diagnostics. Such a system would allow the more widespread use of the OCT technique in a range of technology sectors (medical and industrial) where the cost of the diagnostic equipment is critical.

This project addresses the question of how to achieve a comparatively simple and inexpensive OCT system, based on a superluminescent diode (SLD) and erbium-doped fiber amplifier light (EDFA) sources and anti-Stokes imaging device, to perform 3D cross-sectional imaging. The principal novelty in the project is the use of a new OCT system, based on components readily available in the main optical telecommunications band, to carry out 3D imaging of an object under test. The use of the anti-Stokes camera in low-coherence interferometry is particularly novel. The
experimental work will demonstrate the use of our novel OCT system in biomedical diagnostics.

The principal parameters of interest in the OCT system design are the practicality of the optical setup (stability, ease of construction), dynamic range, depth of penetration, with a particular aim of achieving depth and lateral resolutions of the order of approximately 20 μm. The proposal uses wide experience of the Optical Communications Research Group in optical metrology, signal processing, optoelectronics and optical communication systems.

1.1. Thesis structure.

This thesis is organized as follows. Chapter 1 is an introductory section. Chapter 2 will outline all important, theoretical aspects of low coherence interferometry, principles and features of Optical Coherence Tomography (OCT) and eventually take on full-field modality of OCT. The particular aspects of OCT operating in the near-infrared part of the optical band are also explained, with special focus on the full-field imaging. Chapter 3 is an analysis of the anti-Stokes imaging device and light emitting properties of light sources. The SLD operation and SLD-pumped EDFA operation are described, as well as attempts of destroying spatial coherence of produced optical beams. This is achieved by using a mode scrambler and a long, multimode, optical fiber. The next section is a comprehensive description of the proposed OCT system. This contains an explanation and an evaluation of system optical setup, principle of operation and its performance specification, like resolution, dynamic range and data acquisition speed. This is followed by testing the OCT in terms of its capability of retrieving sample depth information and as a profilometric scanner. Chapter 5 outline further testing the OCT, with tooth samples as objects of
examination. Those tests prove system usefulness in imaging dense, biological tissues. Chapter 6 summarizes all previous chapters, describing system advantages and limitations and points out paths of further development.
2. Theory of low-coherence interferometry and OCT.

This section presents an introduction to Low-Coherence Interferometry (LCI) and optical coherence tomography (OCT). Starting with the theoretical fundamentals of closely related interference and coherence phenomena, low-coherence interferometry will be outlined. Then, using LCI theory, focus will be moved to time-domain optical coherence tomography and its features. This will include general principles of operation and methods of achieving optimal OCT system performance. Finally, an OCT modality for parallel area imaging, a Full-Field Optical Coherence Tomography (FF-OCT) and its special features are described.

2.1. Low-coherence interferometry.

Interferometry describes a family of methods based on interference effects, where waves are superimposed in a way to retrieve some meaningful information about the original state of those waves [3]. It is an investigative technique used widely in science and industry in fields like astronomy, engineering and optical metrology, seismology, oceanography, spectroscopy, mechanical stress measurement, velocimetry, particle and nuclear physics, etc. [3, 4].

Interference is a phenomenon that applies to all kinds of waves, including electromagnetic fields, water surface waves, etc., but we will pay attention only to the optical region of the electromagnetic spectrum. From this point of view, interference is the mechanism by which light interacts with light. Other phenomena, such as diffraction, refraction or scattering, describe how light interacts with elements of physical environment where that light propagates. Optical fields require certain conditions for interference to occur. To interfere, two or more light beams must be
temporally and spatially coherent with one another and their polarizations must not be strictly orthogonal. These requirements are generally fulfilled when beams originate from the same source. In this subchapter a theoretical presentation of interference and its association to coherence phenomena is given. Thus an introduction to low-coherence interferometry will be presented and most important terms and issues explained.

2.1.1. Interference phenomena.

Interference [5] is a result of the superposition of two or more mutually coherent electromagnetic waves. Our discussion is limited to the case of optical beams. When two beams are produced from a single monochromatic point source and then recombined, the resulting intensity distribution is determined by the phase difference
between the two beams. As a result, the light intensity in the superposition area can be observed as a variation pattern with well-defined maxima and minima. The former can greatly exceed the sum of both beams intensities, while the latter can get close, or even reach zero. That pattern can be referred as “fringes”. The fringe pattern depends on the type of interferometer used. A well-adjusted interferometer with a quasi-monochromatic light source can yield a clear and sharp fringe pattern. Distortions in the fringe pattern can be caused by optical aberrations arising from component imperfections and optical setup misalignment. The interferometric pattern may be ill-defined when the input light spectrum is wide or comes from a spatially extended source.

There are two general ways of getting multiple beams from a single light source. In the first method, called division of amplitude, beams are divided by one or more beam splitters, and then recombined. Division of amplitude is utilized in Michelson interferometers (Fig. 2.1a), Mach-Zehnder interferometers, Fizeau interferometers, interference in thin films, etc. In the other method, a division of wavefront, source beam passages through a number of apertures, or pin-holes, placed close to each other. This method is used in Fresnel’s biprism and mirror, Billet’s split lens, in Lloyd’s mirror and in Young’s experiment (Fig. 2.1b), the first ever experiment demonstrating interference and the wave nature of light [1]. In Young’s experiment, an input light beam passages through twin slits S1, S2 and is imaged on the screen. Both slits are considered as two new point sources that produce mutually coherent, spherical waves that interfere with each other and give rise to a fringe pattern on the screen. The variation of observed fringe intensity is a result of difference in phase between the light striking on the screen, which is determined by the path differences $d_1$ and $d_2$. The maximum intensity is obtained when the phase difference is either zero
or a multiplication of $2\pi$. Thus path length difference must be equal to a multiple of a wavelength.

$$\frac{2\pi}{\lambda} |d_1 - d_2| = 2n\pi, \quad n = 0, 1, 2 \ldots$$

(2.1)

Since wavelength of light is in range of hundreds of nanometers, phase fluctuations and therefore the phase sensitivity can reach fraction of nanometer precision.

Mathematical presentation of interference is started with defining the net complex amplitude as the sum of all component fields that undergo interference:

$$E(x, y, z, t) = \sum_i E_i(x, y, z, t)$$

(2.2)

The intensity is a time average over an optical period of the complex amplitude squared,

$$I(x, y, z, t) = \langle |E(x, y, z, t)|^2 \rangle$$

(2.3)

If only two interfering waves $E_1, E_2$ are assumed, then the intensity is,

$$I(x, y, z, t) = \langle |E_1|^2 \rangle + \langle |E_2|^2 \rangle + \langle E_1^* E_2^* \rangle + \langle E_1^* E_2 \rangle =$$

$$= I_1 + I_2 + \langle E_1^* E_2^* \rangle + \langle E_1^* E_2 \rangle$$

(2.4)

Assuming linearly polarized, monochromatic waves of the following form:

$$E_i(x, y, z, t) = A_i(x, y, z, t) \exp \{i[\omega_i t - \phi_i(x, y, z)]\}$$

(2.5)

with $A_i(x, y, z, t)$ and $\phi_i(x, y, z)$, as the envelope and phase of the function representing the wave, respectively. Then the interference pattern intensity is:
\[ I(x, y, z, t) = I_1 + I_2 + 2(I_1 I_2) \cos\left\{ (\omega_1 - \omega_2) t + [\phi_1(x, y, z) - \phi_2(x, y, z)] \right\} \]  

(2.6)

The interference fringes are contained in the third term on the right-hand side of (2.6). It can be easily noticed that to produce a fringe pattern, interfering waves must not be polarized orthogonally, because the cosine function containing phase terms in (2.6) would be equal to zero. If both waves differ in frequency, the product of interference will be modulated at a temporal beat frequency of the value equal to the difference frequency.

If the waves have identical optical frequencies and their linear polarization vectors are parallel, the interference equation becomes:

\[ I(x, y, z) = I_1 + I_2 + 2\sqrt{I_1 I_2} \cos[\Delta \phi(x, y, z)] \]  

(2.7)

where \( \Delta \phi = \phi_1 - \phi_2 \) is the phase difference that is related to the difference of optical paths. Figure 2.2 shows an intensity variation, an idealized product of interference of
two optical waves. The hills and valleys of detected intensity are referred as interferometric fringes.

2.1.2. Interference and Coherence.

If waves originate from the same source, they are usually correlated in terms of their amplitude and phase fluctuations at the atomic level [6]. Hence are said to be completely coherent, if the correlation is complete, or partially coherent, when correlation is partial. On the other hand, waves originating from different sources are mutually incoherent, because no correlation exist between their phase and amplitude fluctuations.

However, discussing interference phenomenon, two facts have to be noted. Classical coherence theory says the light obtained from a real light source is, firstly, never monochromatic, which means it is a sum of individual wavelengths (frequencies) radiating independently from one another and together they can be presented as a spectrum of extent that can be measured. So, the temporal average of the source is the sum of separate temporal averages of single wavelengths. Secondly, a real source is never a point source, but has certain spatial dimensions. So it is assumed the light is a sum of many point sources that radiate independently and hence produces its own interference pattern. So, the final product of the interference is the sum of those individual patterns.

Assuming interference by division of amplitude using an extended source, every point on that source interferes with images of the same point and other points at the plane of observation. But waves originating from different points on the source, may not exhibit any mutual correlation and interference occurs in the region, where
the correlation between the interfering fields is significant. The size of that region, where fringes are localized, is related to the spatial coherence of the illumination. Similarly, the correlation between waves originating from the same point on the source, at different times, is a measure of the temporal coherence of the light and is related to its spectral bandwidth $\Delta \omega$. The maximum value of the Optical Path Difference (OPD) at which interference fringes can be observed, the coherence length (Fig. 2.3b), is a measure of the temporal coherence of the illumination (2.8). Therefore the fringe pattern visibility is a function of temporal and spatial coherence of the interfering beams [4, 5].

$$l_c = \frac{OPD_{\text{max}}}{c} = \frac{\lambda^2}{\Delta \lambda}$$  \hspace{1cm} (2.8)

Figure 2.3. Typical interference intensity pattern for (a) perfectly monochromatic light source and (b) for broadband (low-coherence) source. The green, dashed line represents the envelope of the low-coherence signal. The FWHM of the low-coherence signal envelope is often referred as the coherence length – the maximum path difference over which the interference can be obtained.
2.1.3. LCI signal.

Low-coherence or white-light interferometry employs a broadband light source. As a result, interference is only achieved when the sample and reference paths lengths difference is matched within the coherence length of the source, due to its low temporal coherence. Therefore, by translating one arm of the interferometer, depth information of the sample can be acquired in a coherence envelope (Fig. 2.3b) and a 3D image of the sample can be constructed.

Assuming polarization effects are negligible, the mathematical treatment of the LCI signal [7] is started with the Fourier Transform (FT) representation of the electric field $E(t)$:

$$E(\omega) = FT\{E(t)\} = \int_{-\infty}^{\infty} E(t) \exp(2\pi i \omega t) \, dt \quad (2.9)$$

With the corresponding analytical signal,

$$V(t) = 2 \int_{0}^{\infty} E(\omega) \exp(-2\pi i \omega t) \, d\omega = A(t) \exp\{-i[2\pi \bar{\omega} t - \Phi(t)]\} \quad (2.10)$$

$A(t) \exp[i\Phi(t)]$ is the complex envelope of $V(t)$, $A(t) = |V(t)|$ is the real envelope and $\bar{\omega}$ is the center frequency of the power spectrum. The instantaneous intensity of $V(t)$ is defined as convolution between $V(t)$ and its complex conjugate:

$$I(t) = V^*(t) * V(t) \quad (2.11)$$
The interference phenomena of light waves can be described as a second-order correlation phenomena. The mutual coherence function of a sample wave $V_S$ and a reference wave $V_R$ is a convolution function between the two,

$$\Gamma_{SR}(\tau) = \langle V_S^*(t) * V_R(t + \tau) \rangle \quad (2.12)$$

where brackets mean ensemble average. Since only stationary and ergodic waves are discussed, all ensemble averages can be substituted by time averages. Therefore, the averaged intensity $\bar{I}$ is the autocorrelation function at $\tau = 0$:

$$\bar{I} = \langle I(t) \rangle = \langle V^*(t) * V(t) \rangle = \Gamma(0) \quad (2.13)$$

The relative time delay $\Delta t = \Delta z/c$, where $\Delta z$ is the path difference between the interfering waves and $c$ is the speed of light, is introduced. The sample and reference beams interfere at interferometer output, give an analytical signal

$$V_E(t, \Delta t) = V_S(t) + V_R(t + \Delta t) \quad (2.14)$$

Thus the averaged intensity at interferometer output is:

$$\bar{I}_E(\Delta t) = \langle I_E(t, \Delta t) \rangle = \Gamma_{EE}(0, \Delta t) = \langle V_E^*(t, \Delta t) * V_E(t, \Delta t) \rangle =$$

$$= \langle I_S(t) \rangle + \langle I_R(t) \rangle + G_{SR}(\Delta t) \quad (2.15)$$

$G_{SR}(\Delta t)$, the interferogram, is twice the real part of the cross-correlation of the analytic signals of both interfering beams:

$$G_{SR}(\Delta t) = 2Re\{\langle V_S^*(t) * V_R(t; \Delta t) \rangle\} = 2Re\{\Gamma_{SR}(\Delta t)\} \quad (2.16)$$
\[ \gamma_{SR}(\Delta t) \] is the complex degree of coherence of both interfering signals and \( |\gamma_{SR}(\Delta t)| \) is their degree of coherence. The cosine argument is split into two parts: \( \alpha_{SR} \), a constant phase, and \( \delta_{SR}(\Delta t) = 2\pi \overline{\phi} \Delta t \), the phase delay. The interferogram \( G_{SR}(\Delta t) \) is called the LCI signal and \( |\gamma_{SR}(\Delta t)| \) is called the LCI signal envelope. An example of the LCI signal envelope is shown in figure 2.3b.

Corresponding spectral relations can be obtained using the Wiener-Khintchine theorem [5]. The theorem says, firstly, the power spectrum of the light wave can be obtained by Fourier transforming its autocorrelation (Fig. 2.4):

\[ S(\omega) = FT\{\Gamma(\tau)\} \quad (2.17) \]

And secondly, the cross-spectral density function of two interfering waves \( V_S \) and \( V_R \) can be calculated from their cross-correlation function:

\[ W_{SR}(\omega) = FT\{\Gamma_{SR}(\tau)\} \quad (2.18) \]

Therefore, the spectral interference law is as follows:

\[ S(\omega, \Delta t) = S_S(\omega) + S_R(\omega) + 2Re[W_{SR}(\omega)]cos(2\pi \omega \Delta t) \quad (2.19) \]
2.1.4. Application of LCI.

Low-coherence or white-light interferometry has been used for years in industrial metrology, e.g., in measurements of thin films thickness variations and the birefringence of optically anisotropic materials [8]. LCI was successfully employed as an absolute position-distance sensor [9] and a surface contour mapping in integrated circuits [2]. It is used in profilometry to estimate roughness of examined surface with sub-micrometer precision. Use of LCI in profilometry makes it possible to overcome the issue of phase ambiguities at discontinuities and steps. Furthermore, in contrary to confocal microscopy, lateral and longitudinal resolutions are decoupled from each other, so longitudinal resolution can be furtherly enhanced by white-light interference [10]. Another very important application is Optical Coherence-Domain Reflectometry (OCDR). It is a one dimensional optical ranging technique, developed for finding

Figure 2.4. Illustration of Fourier transform relationship between a Gaussian shaped coherence function (a), characterized by the coherence length $l_C$, and the light source spectrum (b) with frequency bandwidth $\Delta \omega$. Relation between $l_C$ and $\Delta \omega$ is shown in (2.8).
positions and magnitude of reflecting sites, representing faults, within optical assemblies, fiber optic waveguides and therefore in fiber optic networks [11, 12]. Later on, its feasibility of probing and retrieving images of an eye and other biological tissues has been recognized. As an example, in [13, 15], LCI was used to measure intra-ocular distances. In [14] a modified laser Doppler interferometry setup was employed to measure the corneal thickness of the human eye in vivo. In [16, 17] optical parameters of human and non-human dense, biological tissues, such as skin, arteries, etc.

LCI is a foundation of Time-Domain Optical Coherence Tomography (TD-OCT) and was pioneered by J. G. Fujimoto and his co-workers [2]. They extended a fiber based, low-coherence reflectometer system with 830 nm SLD, to perform high speed two dimensional scanning of human retina and coronary artery. OCT uses series of adjacent LCI depth scans to create cross-sectional images of examined sample with better resolution than other tomographic techniques, such as ultrasound imaging or magnetic resonance. This technology will be presented much more closely in the following subchapter.

2.2. Fundamentals of Optical coherence tomography.

Optical technologies are of great importance in medical diagnostics and have a long history. It was started in 17th century, when the microscope became an essential equipment for biologists and scientists [18]. The next step in its evolution was made in 1960, when a first working laser was invented [19], giving rise to a new potential surgical instrument. In parallel, the development of fiber optics led to the manufacture of endoscopes that allow direct viewing of internal organs deep inside the body [20].
In the modern clinical laboratory, optical technologies facilitate many tasks, such as chemical analysis of tissue samples, the counting and sizing of blood cells, etc. In spite of many advances, very few of the optical instruments used currently in medicine take advantage of the coherent properties of light. This is why optical coherence tomography has particularly attracted attention of scientists working in the photonics field. It has the potential to become the first medical diagnostic technology in which coherent optics features prominently.

2.2.1. Introduction to OCT.

OCT is an imaging technology capable of producing high-resolution, three-dimensional images of internal microstructure of inhomogeneous samples, such as biological tissues, non-invasively. It is based on a Low-Coherence Interferometry, an absolute measurement technique, and employs an interferometer similar to those used in thin films thickness measurements in industrial metrology [8]. OCT performs three-dimensional imaging by measuring the magnitude and time delay of light reflected from the sample. The delay information is then used to approximate the axial location of reflecting interfaces and backscattering sites within the specimen. This particular feature is a foundation for OCDR [11], a technique used in short range fault findings and characterization of optical fibers. Soon after OCDR was discovered, the possible application of LCI to perform cross-sectional imaging of biological tissues was reported. The very first realization of this idea was measuring an optical length of an eye [13]. Later, a fiber-based OCT system capable of taking images of human retina, optical disk and coronary artery was presented [2].

OCT is an interferometric technique, relying on interference between split and later recombined broadband optical beams. A generic free-space optics as well as fiber
Based systems are possible. OCT schematics of both time domain and Fourier/spectrum domain systems are shown in figure 2.5. In both variations, the incident optical power entering the Michelson interferometer is split into two arms. One beam travels in a reference path, reflecting from a reference mirror, and the other one propagates in a sample path where it is reflected from multiple interfaces within a sample. Eventually, both reflected beams are recombined at the interferometer exit. Due to the broad spectrum of the input light, they interfere only when the reference and sample arm optical path lengths are matched to within the coherence length of the source. Therefore, the depth (axial) resolution of an OCT system is determined by the temporal coherence of the light source. Sharp refractive index variations in the sample medium correspond to certain intensity peaks in the interference pattern. A time domain interference pattern can be recorded by translating the reference mirror to

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change the reference path length to match optical paths of each successive reflecting or scattering site within the sample at varying depths. The sample depth resolved reflectivity profile obtained in such procedure at the focal spot of the sample beam at a fixed lateral position, is called an A-scan (axial scan). For Time Domain OCT (TD-OCT), light source used is broadband and continuous wave. Fourier domain systems are divided into Spectral Domain OCT systems (SD-OCT) and Swept Source OCT systems (SS-OCT). In SD-OCT the source is broadband and continuous wave and the reference arm length remains fixed, in a position approximately matching position of the sample. The interference pattern containing information from all depths in the sample is dispersed by spectrometer and collected by an array detector. In the case of SS-OCT, the reference arm is also fixed and the source has narrow instantaneous line-width but is rapidly swept in wavelength, and the spectral interference pattern is detected on a single or small number of photodetectors as a function of time. In both cases, the recorded spectral interference pattern contains an entire depth-resolved sample optical structure at the position of the focal spot. A-scans are then obtained by using an inverse Fourier transform.

In OCT, three-dimensional images are built by performing multiple adjacent A-scans by scanning the incident optical beam in lateral direction (Fig. 2.6). This procedure gives a two-dimensional data set, which represents the optical backscattering in a cross-sectional plane through the tissue. These images, or B-scans, can be displayed to visualize tissue internal structure. A number of B-scans acquired adjacently generates a volumetric data set that represent a comprehensive, three-dimensional structural detail about the sample under test. That kind of data can be manipulated and visualized in many ways, for instance in grey scale or various false colors patterns, to retrieve any meaningful information about the specimen.
Due to the ability of performing tissue microstructure visualization in real time, without the need of removing or processing samples, high depth and lateral resolution independent from each other, OCT has found applications in many medical fields. Since it is a noninvasive and contact-free method, it can be safely used to image fragile tissues. Initially, OCT was applied in ophthalmology [14, 15] to provide structural information that could not be obtained by any other technology. Later on, advances in OCT technology have made it possible to apply OCT in a wide variety of other medical and non-medical applications. OCT scanners have been extended to image higher scattering tissue, such as skin, collagen, dentin and enamel, etc. It resulted in a number of various function dependent image contrasting methods and technical extensions. There are Doppler OCT [21], a modality able to measure flow dynamics, Polarization

Figure 2.6. Procedure of creation a three-dimensional OCT image [29]. A-scan is a volume of backreflected or backscattered light intensity as a function of depth (a). B-scan is a set of adjacent A-scans building in a two-dimensional plane (b), while set of successive B-scans produces a cross-sectional three-dimensional image (c).
Sensitive OCT (PS-OCT) [22, 23], able to extract the birefringence properties of biological tissues and wavelength dependent OCT modalities, like refractometric OCT [24] or spectroscopic OCT [25, 26]. What is more, OCT can be interfaced to a wide range of commercially available and customized instruments such as endoscopes, catheters, or laparoscopes, which enable the imaging of internal organ systems [27, 28].

OCT has features of other well-known imaging technologies: ultrasonic pulse-echo imaging (or ultrasound) [31, 32] and confocal microscopy [33-36]. Performing a number of axial scans in series of adjacent locations to create a two-dimensional map of reflections within the sample is analogous to ultrasound (OCT A-scan and B-scan correspond to ultrasound A-mode and B-mode respectively). Instead of optical waves, ultrasound uses high-frequency sonic waves to image deep inside examined bodies. It is possible due to low absorption of sonic waves in biological tissues, but with expense of resolution (~ 0.1 – 1 mm). There are high frequency modalities with greatly improved resolution (tens of μm), but with a severely reduced penetration depth. The ability of OCT to distinguish separate reflectors in axial scans, the optical sectioning, is analogous to confocal microscopy. However, the axial resolution of OCT depends on the coherence length of the source only, while in confocal microscopy it depends on the numerical aperture of scanning objective. This technique allows to build up three-dimensional images of biological samples with very high resolution, but on depths not greater then hundreds of microns. All mentioned techniques are compared in terms of resolution and image penetration in figures 2.7 and 2.8.
Optical coherence tomography systems are still being developed and improved since new potentially useful ideas, techniques and components are being invented.
New light sources with broader and smoother spectrum, more efficient and faster detectors, better beam focusing systems and probes, more effective image processing and display methods are among the technological advances that are constantly improving the quality of images produced by new OCT systems. The following subchapters outline the fundamental features and basics of OCT imaging. The emphasis will be put on Full-Field Optical Coherence Tomography (FF-OCT), a technical modality able to acquire interferometric images in en-face orientation with high resolution.

2.2.2. Resolution.

Resolution is a crucial parameter for any diagnostic system that aims at imaging the internal structure of examined subject. One of the greatest advantage of OCT compared with other imaging technologies is independency between depth resolution and sample beam scanning optics, so the probing depth in scattering media is greatly increased. Hence, the optical design of the system can be optimized for lateral scanning, with no effect on the axial resolution. Figure 2.9 presents the characterization of parameters important from the perspectives of lateral and axial directions. The optical system is assumed to be cylindrically symmetric, and so only one lateral dimension is described. Numerical definitions for depicted parameters will be derived in following subchapters.
2.2.2.1. Axial resolution and coherence gating.

As mentioned in 2.1.3, LCI measures the occurrence of fringe bursts in the so-called coherence gate, which can be described as the size of the round trip coherence length of the light source and is directly related to its power spectrum [7]. As indicated by (2.17) and (2.18), the power spectrum of the light wave can be obtained by Fourier transforming its autocorrelation, and the cross-spectral density function of two interfering waves can be calculated from their cross-correlation function,

\[ S(\omega) = FT[\Gamma(\tau)] \]  \hspace{1cm} (2.20)

\[ W_{SR}(\omega) = FT[\Gamma_{SR}(\tau)] = S(\omega)H(\omega) \]  \hspace{1cm} (2.21)
where $H(\omega)$ is the sample transfer function. The light power spectrum can be approximated by a Gaussian function with high accuracy, since many low coherence sources used in LCI and OCT, like SLDs, exhibit nearly Gaussian shape of the spectrum envelope [7, 30]. The Gaussian spectrum is denoted as:

$$S(\omega) \propto \exp \left[ -4\ln 2 \frac{(\omega - \omega_0)^2}{\Delta\omega^2} \right] \quad (2.22)$$

Where $\omega_0$ and $\Delta\omega$ are source center frequency and bandwidth, respectively. Hence, a depth Point Spread Function (PSF) is equal to:

$$\text{Re} \left[ \Gamma(z) \right] \propto \exp \left[ -4\ln 2 \left( \frac{z}{l_{\text{FWHM}}} \right)^2 \right] \cos \left( \frac{2\pi}{\lambda_0} z \right) \quad (2.23)$$

It is necessary to keep in mind that the light travels back and forth in the interferometer, so the axial resolution can be expressed by the following formula:

$$\Delta z = \frac{l_{\text{FWHM}}}{2} = \frac{2\ln 2}{n\pi} \frac{\lambda_0^2}{\Delta\lambda_{\text{FWHM}}} \quad (2.24)$$

where $\lambda_0$ describes the central wavelength, $\Delta\lambda_{\text{FWHM}}$ and $l_{\text{FWHM}}$ are the full width at half-maximum of the source spectral profile (bandwidth) and autocorrelation function (coherence length), respectively. $n$ is the refractive index of the medium where light propagates. Figure 2.10 visualizes the axial resolution dependency of bandwidth of the light source.

The axial resolution in both the time domain and Fourier domain OCTs is determined by the coherence time or the coherence length of the light source. FD-OCT does not perform coherence gating by scanning the delay line in a reference arm, but uses spectral interferometry or wavelength tuning techniques [7]. Since the
interferogram and cross-spectral intensity are mutual Fourier transforms (2.21), the inverse Fourier transform of the spectral interferogram intensity yields the same signal as obtained by LCI [38].

From (2.24) the broader the source spectrum, the better the depth resolution. However, separate wavelengths are propagating with slightly different velocities, so when broad light spectrum travels through an interferometer, dispersion is introduced. It leads to degradation of the traveling wave front and therefore, axial resolution. According to [39, 40], first order dispersion improves resolution, but second and higher order dispersion brings even greater degradation. To overcome this problem, usually a dispersive material is placed in the reference arm to balance the sample arm dispersion [41]. There are several numerical methods that help remove this negative effect impact during post processing. Examples of such algorithms are given in [42-44]. A technique improving axial resolution by step-frequency encoding both in the time domain and the Fourier domain was presented in [45]. In [46] authors prove that the polarization mismatch of the sample and reference arms in optical fiber based OCT has a critical effect on its depth resolution. A disadvantage of using non-Gaussian

![Figure 2.10. OCT axial resolution versus optical source bandwidth for three different central wavelengths.](image1)

![Figure 2.11. OCT lateral resolution versus scanning lens numerical aperture for three different central wavelengths.](image2)
sources is worth mentioning, because they may give rise to side lobes in the output interferogram that appear as shadows or false artefacts in the final image. These can be corrected to some degree by numerical methods [47-49], but it is preferable to use a Gaussian spectrum source. To suppress false artefacts, spectral shaping of the input spectrum can be successfully applied [50, 51].

2.2.2.2. Lateral resolution and confocal gating.

The sample arm of an OCT system can be treated as a reflection-mode scanning confocal microscope, where the single mode optical fiber serves the purpose of a pinhole for both illumination and collection of light from the sample. For strictly free-space optics OCT systems, the response function of the interference can be shown to be equivalent to confocality [30]. Confocal microscopes [33-36] have the advantage of slightly improved transversal resolution over typical bright-field microscopes and the ability to perform optical sectioning, removing the contribution of out of focus light in each image plane, due to their peaked axial response. The intensity detected from a point reflector located in the focal plane of an ideal reflection confocal microscope as a function of lateral position, is given by:

\[
I(v) = \left( \frac{2J_1(v)}{v} \right)^4, \quad v = 2\pi x \frac{NA}{\lambda_0}
\]  

(2.25)

where \( J_1(v) \) is a first-order Bessel function of the first kind, \( v \) is the normalized lateral range parameter, \( x \) is the lateral distance from the optical axis, \( NA \) is the Numerical Aperture of the scanning objective, and \( \lambda_0 \) is the center wavelength of the light source. \( I(v) \) can be interpreted as the lateral point spread function of an OCT system at the
position of its focal plane. Therefore, the lateral resolution $\Delta x$ is defined as a full width at half maximum power of the PSF function:

$$\Delta x = 0.37 \frac{\lambda_0}{NA}$$  \hspace{1cm} (2.26)

The lateral field of view for an OCT system depends on the optical system facilitating lateral scanning (Fig. 2.11). A particularly simple scanning system is able to tilt the input aperture of the objective lens to a maximum scan angle $\phi$ (Fig. 2.9). Hence, the lateral field of view of the OCT system employing light source with frequency $f_0$, is simply given by:

$$\text{FOV}_{\text{lateral}} = 2f_0\phi.$$  \hspace{1cm} (2.27)

Using the same convention of confocality, the axial response of the OCT sample arm optics can be interpreted as the confocal response to a planar rather than point reflector. The detected intensity of an ideal confocal microscope from a planar reflector as a function of the reflector position $z$ along the optic axis is given by

$$I(u) = \left( \sin\left( \frac{u/2}{u/2} \right) \right)^2, \quad u = 8\pi z \frac{\sin^2(\alpha/2)}{\lambda_0}$$ \hspace{1cm} (2.28)

where $u$ is the normalized axial range parameter and $\alpha = \sin^{-1}(NA)$. The axially peaked response of a confocal microscope gives it its depth sectioning capability. This is also the response that would be expected by translating a mirror axially through the focus of an OCT sample arm and comparable to the axial response of the OCT system arising from low-coherence interferometry. OCT systems operating in this fashion are referred to as Optical Coherence Microscopy (OCM) systems [52-54]. The full width at half maximum power of the confocal axial response function is defined as the axial Depth.
Of Focus (DOF) of the OCT system. It is a confocal parameter equal to twice the Rayleigh range $\Delta z_R$,

$$DOF = 2\Delta z_R = \frac{\pi \Delta x^2}{2\lambda_0} \quad (2.29)$$

Definitions of DOF and lateral resolution (2.26 and 2.29) state clearly, that when picking an objective with high NA to achieve high lateral resolution, the DOF is decreased at the same time, as depicted in figure 2.12, so it limits the axial range over which the LCI depth scanning may operate without yielding blurred and unreadable images. Furthermore, by inspection of Abbe’s rule for the lateral resolution, it can be noticed that maximum penetration depth is reduced as well [55]. One way to overcome these issues is to introduce dynamic focusing into sample arm [56].

Figure 2.12. Relationship between focus spot size and depth of focus for objective lens with high and low numerical aperture.
2.2.3. Probing depth.

OCT images are synthesized from a series of adjacent interferometric depth scans performed by a low-coherence probing beam. Hence, the penetration depth is one of the key parameters determining system usefulness and possible applications. In OCT it depends on several factors: source optical power and wavelength, optical scanning setup (NA of imaging lens, beam focal spot diameter) and the absorption and scattering properties of sample under test. The latter is very important, since only single scattered light contribute to the object’s Fourier spectrum. Multiple scattered photons are not just a loss of useful illumination, but can cause reduction of imaging sensitivity and resolution by forming noise [7]. Figure 2.13 shows how sample beam is focused at the specimen, assuming the coherence matching point is coinciding with the imaging lens focal plane. The OCT detector detects all photons that reach it, but only those scattered within the coherence probe volume and travelled same distance.

Figure 2.13. Geometry of sample and probing beam. A, B are trajectories of multiple scattered photons rejected by the coherence gate (A) and confocal gate (B). C presents an example of single scattered photon that contribute in creation of an image [7].
as photons reflected from the reference mirror contribute to interference image creation. Therefore, even if a photon scattered within the probing beam cone, gets scattered again within the coherence volume and travels back to the interferometer, it may be rejected by the coherence gate, due to too long path travelled.

When imaging at larger depths, the mean number of scattering events for single photons increases, reducing their temporal and spatial coherence properties. Eventually, diffuse scattered, incoherent photons start dominating. It has been shown in [59], that in the case of photon random walks, the probability of a photon to get scattered $n$ times and return to the surface scales with:

$$P_{ph} = n(1 - g)(1 + g)$$

(2.30)

where $g$ is sample scattering anisotropy. When scattering anisotropy of the sample increases, the smaller number of photons experienced $n$ number of scattering events is detected. Experimental studies [57] showed, that the photons transition from single to scattering to diffusion occurs at longer path length when sample scattering anisotropy is low (Fig. 2.13) or when using sample lens with a relatively small $NA$ [57, 58].

Generally, light scattering becomes weaker as the incident wavelength is increased [60]. Absorption in tissues decreases with increasing wavelength in the visible spectrum of light. At the red end of the spectrum and near infrared, biological tissues have scattering anisotropies of approximately 0.9 [7], so forward scattering dominates and the probing depth is of the order of a few mean free random walks. Therefore, those wavelengths are preferred in OCT.

In highly scattering tissues penetration depth of OCT systems reaches 1-3 mm [30], depending on the kind of tissue under test and imaging OCT system properties. It is approximately an order of magnitude worse than mentioned ultrasound techniques.
(Fig. 2.4). But, there have been attempts to extend probing depth by matching refractive indices of samples [61]. In OCTs using sources generating near infrared illumination, it is common to use water immersed microscope objectives as scanning lenses to match light propagating medium refractive index with the sample [62].

A comprehensive analysis of single scattering and light-tissue interactions can be found in [30]. While multiple scattering has been extensively investigated and modelled in [63].

2.2.4. Signal-to-noise ratio of OCT systems.

Signal-to-noise ratio (SNR) of the OCT system, or the optical dynamic range, is defined as the ratio of the signal power back-reflected from a perfectly reflecting mirror to the noise generated by the system (2.31). That noise is equal to the signal power produced by the weakest reflective sample detectable by the system [7].

\[
SNR = \frac{i_s^2}{i_n^2} \tag{2.31}
\]

The total source power \( P \) is defined, the signal photocurrent \( i_s \) with an assumption that system is examining a perfectly reflecting sample and reference mirror. Hence, according to [64], \( i_s \) is equal to:

\[
i_s = \frac{\eta q}{\hbar \omega_0} P \tag{2.32}
\]

where \( \eta \) is the quantum efficiency of a detector, \( q \) the electron charge, \( \hbar \) is a normalized Planck’s constant and \( \omega_0 \) is the signal central frequency.
In optical detection, three particular sources of noise can be distinguished [65, 66]:

- **Shot noise** $i_{sh}$, effect of a photocurrent variance, due to random electron arrival of electrons comprising the photocurrent.

$$i_{sh} = \sqrt{2qi_{DC}B} \tag{2.33}$$

where $i_{DC}$ is the DC detector photocurrent, due to reference arm signal. Assuming a perfect 50:50 beam splitter, $i_{DC}$ is equal to the half of total photocurrent $i$. $B$ is the electronic bandwidth [64], and is a function of the acquisition time $\tau$ for a single photodetector measurement.

- **Optical excess intensity noise** $i_{ex}$, a result of beating between single light waves making up the broadband beam:

$$i_{ex} = i_{DC}(1 + P)\sqrt{\frac{B}{2\pi\Delta\omega}} \tag{2.34}$$

where $P$ is the source degree of polarization and $\Delta\omega$ is the optical bandwidth.

- **Receiver noise** $i_{re}$, a thermal noise generated in the detector amplifier and load resistor (both represented as effective load resistor $R_{eff}$):

$$i_{re} = \sqrt{\frac{4k_BT}{R_{eff}}} \tag{2.35}$$

$k_B$ is the Boltzmann’s constant, $T$ is the temperature.

The total noise current is the sum of squares of all described noise contributions, so general derivation for SNR of any OCT can be written as:
If the optical power from the reference arm is much greater than the power from the sample arm, the upper derivation can be simplified to only shot noise limited detection. This statement is true in many cases, especially when OCT is imaging highly scattering and absorbing samples. Hence, the shot noise limited SNR is defined as:

\[
SNR_{sh} = \frac{\frac{\eta P}{\hbar \omega_0} B}{\frac{\eta \tau}{\hbar \omega_0} P} = 2 \frac{\eta \tau}{\hbar \omega_0} P
\]  

(2.37)

And in logarithmic units as:

\[
SNR_{log} = 10 \log_{10}(SNR_{sh})
\]  

(2.38)

It is shown in (2.37) that SNR depends linearly on source power and is inversely proportional to electronic bandwidth \( B = 1/2\tau \), where \( \tau \) is the acquisition time for photodetector measurement. Longer acquisition time improves dynamic range, as
shown in figure 2.14, but limits OCT capability of fast image capture rate. The shot noise limited detection for OCT may be treated as optimal performance regime. If the input source power is low, the SNR is limited by the receiver noise. On the other hand, with increasing source power, optical intensity excess noise increases greatly and limits the dynamic range. Analysis of SNR for unbalanced optical power division in OCT interferometer has been presented in [67]. This work states that balanced configuration offers better SNR in most cases and is more suited for fast OCT imaging.

The above applies to time domain OCT. Fourier domain OCT uses a number of photodetectors or a detector array [7, 69] to record the optical signal. In the time domain, a parallel detection scheme has been applied in linear-OCT [70, 71], OCM [52-54] and FF-OCT [72, 73]. It is important to mention how SNR changes when detection is performed by a detector array due to FF-OCT which will be described in details in further part of this work. When there are $M$ equivalent detectors in the FD-OCT system, the mean frequency noise photocurrent can be shown in the time domain, if recalculated by a factor of $M/2$ data points brought by the Fourier transform:

$$\bar{i_n}^2 = 2 \frac{i_n^2}{M}$$

(2.39)

Still using the assumption of a perfectly reflecting sample and reference mirror, as well as a source Gaussian spectrum, the detected signal peak transforms into the time domain as the total input power $P$. Hence, the Fourier domain signal photocurrent $\bar{i_s}^2$ is equal to time domain current $i_s^2$. Therefore, the formula for SNR for FD-OCT can be written as:
\[ SNR_{FD-OCT} = \frac{i_S^2}{i_n^2} = M \frac{\eta \tau}{\hbar \omega_0} P. \] (2.40)

It is clear that the theoretical SNR for FD-OCT grows with M. It is related to the reduction of shot noise obtained by using a multiple element (an array) detector. In a single detector time domain OCT, the shot noise generated by the optical power from each individual frequency affects all other frequencies. Hence it affects the SNR of individual frequencies across whole spectrum of the source. This particular advantage of FD-OCT over time domain systems employing single detector is depicted in figure 2.15. Another feature that makes FD-OCT better is terms of SNR and not involved in this analysis, is its principle of operation. In FD-OCT, the reference arm does not contain any scanner or a delay line, but is fixed, and therefore requires fewer moving parts, and hence is more stable. Differences in dynamic range between Fourier and time domain OCTs have been investigated much more closely in [64, 68].

2.3. Full-field Optical Coherence Tomography (FF-OCT).

The early generation of time-domain OCT systems involved point by point scanning of the reference arm to obtain a depth reflection profile of the sample at single location. Scanning the optical beam in the lateral direction and performing axial scans allows creation of two dimensional images. To make a 3D image, another lateral scan is required. A great limitation to that approach are motion artifacts arising due to mechanical scanning and lack of repeatability. In this technique typically low-NA imaging lenses were used, to achieve large depth of field, but with the expense of lateral resolution. Full-field optical coherence tomography is an alternative technique
that captures two dimensional tomographic images orthogonal to the optical axis (the en-face orientation) in single exposure, without transverse scanning. A sequence of en-face images can be incorporated into a three dimensional image with ease. The FF-OCT body is based on an interferometric microscope [54] with a broadband source producing low spatial coherence illumination [72, 74]. Images are recorded using a well-known area imaging sensor, like Charge Coupled Device (CCD) or Complementary Metal-Oxide Semiconductor (CMOS) camera [76]. This setup allows both high axial and lateral resolution while simplifying the scanning optics.

2.3.1. Instrumentation.

As mentioned, FF-OCT is based on a low-coherence interference microscopy. The experimental arrangement, depicted in figure 2.16, is based on a Michelson interferometer with identical microscope objectives in the sample and reference arms, known as the Linnik configuration [72-74]. This configuration allows separate adjustment of path lengths and focusing for both interferometer arms [76]. Moreover, the reference arm length can be changed without modifying the focus on the reference mirror by mounting both elements together on a single motorized translation stage. Obtaining high lateral resolution requires high-NA microscope objectives [75] and systems with very high NA objectives are referred to as OCM systems. To keep the focus plane and coherence plane matched during measurements of high scattering samples, water- or oil-immersion microscope objectives can be used. Another solution is to employ another motorized translation stage to translate the objective in the sample arm with respect to the specimen, as depth scanning progresses through the specimen.
This method is called dynamic focusing and will be described in more detail later in this chapter.

![Figure 2.16. Example optical arrangement of a FF-OCT system. BS – broadband beam splitter, MO – water immersion microscope objectives, PZT – piezoelectric actuator, MTS – motorized translation stage.](image)

To capture a 2D en-face image in a single exposure, CCD or CMOS camera is used. CCD devices provide linear intensity response, but suffer from relatively low frame rate (typically ~100Hz), which is a limiting factor for a commercial use of FF-OCT. For imaging in infrared region, Indium Gallium Arsenide (InGaAs) cameras are used [101]. In two dimensional detector arrays it is possible to use a pixel binning feature. It is a very effective way to increase the dynamic range of a detector.

FF-OCT employs a broadband light source, ideally producing low spatial coherence illumination. It is due to multiple scattered light from the specimen that can result in strong cross-talk generated noise between adjacent detection channels in a detector array, which yields severe image degradation [63, 78]. Using spatially
incoherent source or decreasing spatial coherence properties of the illumination can avoid this coherent cross-talk effect. Furthermore, a spatially incoherent source provides uniform illumination over the whole microscope objective aperture. The most popular light sources for early FF-OCTs were SLDs [54, 86], but tungsten halogen lamps incorporated in a Köhler illumination systems are currently the sources of choice [79]. Halogens provide high power, broadband light resulting in sub-nanometer depth resolution, but have low brightness per spatial mode.

2.3.2. Principle of operation.

Image reconstruction in FF-OCT is realized by Phase Shifting Interferometry (PSI) [80, 81]. A discrete or continuous phase shift is realized and several interferometric images of every volume within the sample are taken. The en-face image is then extracted by combining those phase shifted images using various algorithms. The first reported FF-OCT [54] used a multiplexed lock-in detection technique and a photoelastic birefringence modulator to implement a periodic phase shift between reference and sample signals, and to take four images of every volume, each corresponding to different phase shift. In the phase stepping technique, fundamental for PSI [80], and so called “integrating buckets” technique [72, 82], the phase is changed by mechanical displacement of a reference mirror using a high precision actuator, such as piezoelectric transducer [73, 74]. In the former method, phase is mechanically changed by a certain extent between each intensity measurement. Hence, inertia limits image maximum acquisition rate. Another issue is a system requiring stable operation conditions and stationary sample throughout the measurement time, because any movements can severely distort en-face image quality. In the “integrating buckets” technique, the intensity is integrated while the phase is
continuously changed, shown in figure 2.17. The phase can be shifted linearly, by controlling the reference mirror movement with a sawtooth signal, and few integrated intensity values, or “buckets”, are recorded. Continuous mirror movement is a cause of signal modulation. This method offers far better operation speed that phase stepping, especially if sinusoidal modulation, which is more steady at high frequencies, is applied [82]. Further image restoration algorithms aimed at improving FF-OCT operation speed were developed. Those methods utilized phase shifting and a numerical Hilbert Transform (HT) [83, 84].

In any of mentioned methods, a variable number of phase steps or “integrating buckets” to reconstruct en-face images can be used. Depending on the algorithm, mostly two [62, 84-86] or four [54, 72, 73, 83, 87] interferometric images, with phase shifts of π and π/2 respectively, are gathered. Using just two steps is the best option in terms of OCT speed of operation, however this algorithm might yield residual fringe pattern distorting the output en-face images. The latter drawback does not occur for image restoration algorithms based on the Hilbert transform [84]. Additionally, an

![Figure 2.17. Illustration of four frame “intergration buckets” image acquisition technique. Each frame \(S_0, S_1, S_2, S_3\) is a result of integration of modulated signal intensity \(I(t)\) over one quarter of modulation period. Series of four frames \(S_0, S_1, S_2, S_3\) are recorded continuously and the en-face image is a result of combination of these frames [82].](image-url)
adjustable number of each interferometric image can be recorded and averaged to improve signal to noise ratio, but with expense of operation speed.

Assuming four frames algorithm, N image accumulations and a phase stepping technique, an interferometric signal intensity from a pixel from \((x, y)\) location, can be described as:

\[
S_i(x, y) = \frac{1}{N} \sum_{j}^{N} \left[ I_{Sj}(x, y) + I_{Rj} + 2 \sqrt{I_{Sj}(x, y)I_{Rj}} \cos(\Phi(x, y) + i \frac{\pi}{2}) \right]
\]  

(2.41)

,where \(i = 0, 1, 2, 3\). \(I_{Sj}(x, y)\) and \(I_{Rj}\) are intensities originated from the sample and reference arms, respectively, and \(\Phi(x, y)\) is the initial phase representing optical path difference between the two interferometer arms. Example of four phase-shifted en-face images is shown in figure 2.18. An en-face reflectivity image is reconstructed by applying the following formula:

\[
S = (S_0 - S_2)^2 + (S_1 - S_3)^2
\]

(2.42)

Using the same conditions to “integration buckets” technique, a time varying intensity incident upon pixel from \((x, y)\) location is expressed as:

\[
I(x, y, t) = I_S(x, y) + I_R + 2\sqrt{I_S(x, y)I_R} \cos(\phi(x, y) + \psi \sin(2\pi ft + \theta))
\]

(2.43)

,where \(\psi, f\) and \(\theta\) are amplitude, frequency and time origin parameter of sinusoidal phase modulation. If \(\eta\) is the quantum efficiency of a detector array, “integration buckets” are described as:

\[
S_i(x, y) = \eta N \int_{iT/4}^{(i+1)T/4} I(x, y, t) dt
\]

(2.44)
where $i = 0, 1, 2, 3$. The output en-face image can be constructed using the same formula as in phase stepping (2.42).

![Four phase-stepped interferometric images captured by the CCD camera [76], each shifted by $\pi/2$ with respect to another.](image)

Figure 2.18. Four phase-stepped interferometric images captured by the CCD camera [76], each shifted by $\pi/2$ with respect to another.

A latest concept of FF-OCT, basing on single-shot PSI, named Single-Shot Full-Field OCT (SS-FF-OCT), aims at improving acquisition speed by recording two or four phase shifted interferograms on single CCD camera at the same time [76, 88]. It is possible by employing a complex two or four channel phase stepper to divide the
output signal, perform phase shifts and illuminate the detector. Changes of phase are achieved without mechanical adjustments of interferometer arm length, and thus provide more stable OCT operation.

2.3.3. Performance of FF-OCT.

Axial resolution of any OCT system depends on the coherence length of the light source and therefore on the source bandwidth (see section 2.2.2.1). The theoretical value can be roughly calculated from (2.24). However, axial resolution can be affected by the spectral response of the detector and possible dispersion mismatch in the two arms of interferometer. If a FF-OCT system is designed to image biological tissues, water immersion objectives are often used, to help match the refractive indices of the sample and light propagating medium. As water is the main constituent of biological tissues, light reflections from sample surface and dispersion mismatch are reduced. To compensate for residual dispersion mismatch, glass plates can be put in both interferometer arms [43].

In Full-Field Optical Coherence Microscopy (FF-OCM), large NA objectives are used to achieve high lateral resolution (see section 2.2.2.2). In this case, the axial resolution depends not only on source coherence length, but on the objective NA as well [72]. Assuming a source with Gaussian shape of its spectrum, the theoretical axial resolution at the surface of the sample is defined as [72]:

\[
\Delta z = \left[ \frac{NA^2}{n_{im} \lambda} + \frac{n_{im} \pi (\Delta \lambda)}{2ln2 \lambda^2} \right]^{-1}
\]

(2.45)
where $n_{im}$ denotes refractive index of the immersion medium. For dry objectives it is equal to unity.

Since FF-OCT collects en-face images, like a conventional microscope, lateral resolution depends only on the objective numerical aperture and wavelength, and it is decoupled from axial resolution. Physically, it is the width of the transverse point spread function (PSF). And, in diffraction limited optical system, the PSF is approximated by the square of an Airy function. For FF-OCT, lateral resolution is identical to that of confocal microscopy, as mentioned in section 2.2.2.2. Calculations apply to the same formula, as for a standard OCT system (2.26). Typically used objectives with NA values between 0.25 and 0.5 yield transverse resolution of approximately 1 μm, depending on the central wavelength of the source. However, experimental measurements of transverse resolution might slightly differ from theoretical considerations due to optical aberrations of microscope objectives used, especially when objectives are not made for the spectral region employed in the system [89].

The general derivation for OCT signal-to-noise ratio (SNR), or dynamic range, has been shown in section 2.2.4. It is one of the most important factors describing OCT systems, affecting the imaging contrast and maximum probing depth. High dynamic range is a key to detect weakly backscattering structures and interfaces. To determine the dynamic range for FF-OCT, noise sources have to be recognized. The most important ones are shot noise and noise originating in the detector array, such as read out noise and dark noise. Both can be described as electrical noise. Now assuming that intensity noise can be neglected and illuminated camera pixels wells are close to saturation and system uses four frames algorithm, minimal detectable reflectivity can be calculated from [72, 90]:

- 44 -
where $\xi_{sat}$ is the full well capacity of the camera pixels, $\eta$ represents total electrical noise, $R_{inc}$ is the amount of incoherent light incident on the camera, $R_{ref}$ the reference mirror reflectivity and $N$ is the number of image accumulations. $R_{min}, R_{inc}$ and $R_{ref}$ are expressed in percentages.

From (2.46) it can be seen that increasing camera full well capacity $\xi_{sat}$ can decrease the value of minimum reflectivity and therefore less reflective sites could be measured. This can be achieved by custom adjustment of camera gain. On the other hand, increasing camera gain, cause electrical noise to increase in similar proportion [90, 91], limiting customizing options. Another parameter that limits sensitivity is $R_{inc}$, amount of light not taking part in interference, originating from multiple scattering, backreflection by interfaces and structures located outside the coherence volume and reflections from optical components. Therefore, all optical components in the interferometer must be antirefection coated. From (2.46), the SNR can be derived:

$$SNR = 10\log\left(\frac{1}{R_{\min}}\right)$$  \hspace{1cm} (2.47)

SNR of the FF-OCT can be improved by increasing the number of image accumulations when performing an FF-OCT scan. Basing on (2.46), (2.47) and following data of a certain CCD camera [92]: $\xi_{sat} = 10^6$, $\eta = 0$, SNR versus $N$ dependency was plotted and shown in figure 2.19. Reference mirror reflectivity is assumed to be 99% and amount of incoherent light $R_{inc}$ to be 0.5% of total light incident on a camera. Despite significant SNR improvement, $N$ cannot be increased without...
limit, because system using \( N \) image accumulations is \( N \) times slower in terms of image acquisition rate.

Figure 2.19. Theoretical analysis of SNR versus number of image accumulations in FF-OCT.

2.3.4. Advantages and drawbacks of parallel data acquisition.

The biggest advantage of full-field OCT is the ability of recording image of a whole plane orthogonal to optical axis in single exposure. Compared to typical OCT, this technique does not need lateral beam scanning which makes the optical arrangement a lot easier, due to less moving parts in the system. Since spatially incoherent illumination is required, a simple and inexpensive thermal source, like a tungsten-halogen lamp is totally sufficient for full-field illumination. This kind of source offers large bandwidth which yields ultra-high (\( \sim 1 \) \( \mu \)m) axial resolution. This, combined with lateral resolution depending only on microscope objective properties and source wavelength, makes FF-OCT a possible replacement for current methods used for histology [93].

However, acquiring images in en-face orientation has certain disadvantages compared to standard OCT. In FF-OCT the light beam under detection is spread over
a large number of single detectors or pixels, so a number of photons received by every pixel is small comparing to a photodetector in point scanning OCT. Therefore much greater sensitivity can be achieved in conventional OCTs. Full field type of image acquisition does not necessarily bring better speed of operation. The necessity of taking a number of frames of single volume sequentially in time, due to phase shifting algorithms, and adjustable dynamic range by image summing or averaging, plus limitations of frame rate of typical CCD and InGaAs cameras (~100 Hz) yields FF-OCT frame rate of several Hz or less. This makes FF-OCT very vulnerable to sample motion, as any changes of incident flux received by the camera pixels between successive images will appear in the calculated interferometric image. The speed of operation issue might yet change due to invention of very high frame rate cameras, like the CMOS camera used in [97].

2.3.5. Infrared FF-OCT.

There are several extensions of the full-field OCT technique developed up to date. A spectroscopic full-field OCT offers improved imaging, comparing to conventional FF-OCT, by providing enhanced image contrast at the expense of decreased frame acquisition rate [29]. Polarization-sensitive FF-OCT (PS-FF-OCT) [87] uses polarizing optics and two CCD cameras to measure phase retardation and reflectivity of biological media. A complex, multimodal system combining spectroscopic and polarization sensitive features was presented in [94]. A FF-OCT system, using pulse illumination, suppressing motion artifacts originating from sample lateral movements, was designed for in vivo imaging [95]. A single-shot FF-OCT is a technique able to record all phase-shifted interferograms simultaneously to increase speed of operation and a potentially suitable for in-vivo imaging [76, 88]. There are
successful approaches of full-field type of image acquisition in the frequency domain, either by using frequency tuned laser [96, 97] or an SLD with an acusto-optic tunable filter [98, 99].

By using an InGaAs camera and dry microscope objectives, FF-OCT can operate in the infrared wavelength region which permits enhanced imaging penetration depth in highly scattering tissues [100]. The optical arrangement is essentially the same as for standard FF-OCT (Fig. 2.61). Another multimodal system, capable of performing imaging in 800 nm and 1200 nm range at the same time by employing an InGaAs and a CCD cameras was described in [101]. Since FF-OCT employing infrared illumination is a key aspect of this thesis, its important features are described in following subchapters.

2.3.5.1. Imaging for deeper penetration.

As explained in section 2.2.3, the probing depth depends on scattering and absorption of light within biological tissue samples. It is known that scattering of light decreases with increasing incident wavelength [60] and light absorption within biological tissues is dominated by absorption of water (Fig. 2.20). Therefore, to pick optimal wavelength band maximizing OCT depth of penetration, these two factors have to be considered.

It was shown in [102], that probing depth within a dense, highly scattering tissue was greater when using 1.3 μm wavelength than 0.8 μm. However if choosing operation with longer wavelengths, the whole system suffers from depth resolution degradation, because its value is proportional to the source center wavelength squared and inversely proportional to the width of the source spectrum (2.24 and 2.45). Therefore, increasing wavelength can only be compromised by increasing the operating bandwidth of the source, if axial resolution is to be maintained. The easiest
available option is thermal source [103], due to its very broad and smooth spectrum. In [104] a compact fiber laser with a broad emission bandwidth (Δλ=470nm) centered at λc = 1375nm was used to achieve sub 2 μm axial resolution in OCT. Another useful source is a solid-state Kerr-lens mode-locked Cr:forsterite laser with 250 nm bandwidth generating 80 mW pulses [105].

Figure 2.20. Optical absorption of water [100].

2.3.5.2. Dynamic focusing feature.

When imaging in the infrared, standard dry microscope objectives are used. Water cannot be used as an immersion medium, because of its increased absorption in this band (Fig. 2.20). Hence, an incident light beam passes through an air and sample surface interface, where a transition between two mediums differing largely in refractive index. In FF-OCT imaging, the plane of coherence must coincide with the focal plane of the microscope objective. As imaging progresses deeper within the sample, coherence plane and objective focal plane significantly move away from each other. It is an effect of refractive index difference, as the optical path within the sample
is longer than in air. The resulting image may then be produced blurred and unreadable or even vanish at certain depth. Therefore, a compensation adjustment has to be done before acquiring images of specimen internal structure. In order to image a volume inside the sample, at depth $z$, of the object with refractive index $n$, the sample (or the reference arm, depending on the optical arrangement) has to be moved by a distance $nz$ and scanning objective has to be moved by distance $Z$:

$$Z = z \left( \frac{n^2 - 1}{n} \right).$$  \hfill (2.48)

To change the position of the objective, an additional computer controlled translation stage needs to be employed. This process referred to as dynamic focusing is depicted in figure 2.21.

![Figure 2.21. The interferometer is adjusted so that the focal plane and coherence plane coincide at the surface of the object (A). When the object is elevated of a distance $e$, the coherence plane is then located at $e/n$ below the object surface, and the focal plane at $n*e$ (B). The focal plane matches the coherence plane again if the objective is elevated of a distance $Z = e(1-1/n^2)$ (C) [101].](image-url)
3. 1550 nm SLD and anti-Stokes imaging device for full field metrology.

In this chapter, two key components, around which this work is built on, will be presented and characterized in details. These are a broadband SLD with spectrum in 1550 nm range and a phosphor-coated CCD camera. The SLD source power will be extended by an EDFA amplifier and components to decrease spatial coherence of the produced light. The light source system and detector suitability for OCT applications will be analyzed and their technological limitations discussed.

3.1. Light source requirements for OCT.

Several important factors must be considered when choosing a light source for optical coherence tomography.

**Wavelength range.** Picking the operating wavelength range is closely related to OCT system application. It is necessary to take under consideration what kind of tissues or materials are going to be examined. It was mentioned in section 2.2.3, that OCT probing depth is dependent on scattering and absorption of the sample under test. And since scattering of materials and absorption of most tissue chromophores decrease with increasing wavelength [7], longer wavelength sources seem to be an answer. However, the main constituent of biological tissues is water and its absorption of light increases nearly exponentially with wavelength (Fig. 2.19) for infrared. According to [7] the spectral window for medical applications is between 600 and 1300 nm due to high occurrences of forward directed scattering events in tissues and relatively low water absorption coefficient. Longer wavelengths could possibly be used for imaging
objects containing very little or no water. On the other hand, the axial resolution value increases with the square of the wavelength (see section 2.2.2.1, equation 2.24).

**Spectral properties.** The depth resolution is inversely proportional to source bandwidth (2.24), which means the broader the spectrum, the better the resolution. Usually, a single broadband light source is used, but it is possible to combine emissions from a number of similar sources with adjacent bandwidths to obtain wider spectrum with greater power, like in [53], or synthesize several quantum well emitters on a single substrate [109]. However, the wider the spectrum, the greater the dispersion mismatch between sample and reference beams in interferometer [39, 40]. For large bandwidth sources, a compensation method is required, either by introducing a dispersive material in interferometer arm or arms [41], or using post-processing algorithms [42, 43].

The shape of the spectrum is an important factor, as OCT depth point spread function is defined as twice the real part of the coherence (autocorrelation) function of the source spectrum envelope (see section 2.1.3). Therefore, it is suggested for spectrum to have smooth and symmetric shape and have no sub-peaks. Ideally it is a Gaussian shape, due to Fourier uncertainty relation, which says that the product of the variances of a Fourier transform pair (in this case it is a width of a coherence function) is minimal for Gaussian functions [7]. Figure 3.1 shows the spectrum and coherence function of Superlum SLD-38-HP diode. Coherence function exhibits distortions, in form of additional side-lobes, due to spectral envelope asymmetric shape. Spectral envelope can be corrected to some level using spectral shaping techniques [50, 51, 110], but at the expense of losing some optical power.

Another crucial parameter for OCT performance is **high optical power,** which is necessary to get high sensitivity, wide dynamic range and reasonable depth of penetration, especially when scanning weakly backscattering samples (2.36).
Furthermore, the demand for optical power increases proportionally to system scanning speed, due to shorter signal integration time on photodetector or camera surface and loss of SNR.

Spatial coherence of the source plays an important role as well. While temporal coherence of the source determines OCT axial resolution, spatial coherence

Figure 3.1. (a) presents spectra of Superlum SLD-38-HP and MQW SLD-37-HP diodes. The former exhibits smooth and slightly asymmetric shape (red). The latter is much broader due to MQW heterostructures, has two peaks and is much more distorted (blue). (b) show coherence function of Superlum SLD-38 diode and indicates side lobes, a consequence of spectrum envelope asymmetry [107].
can affect lateral resolution and image quality [111]. High spatial coherence is desired for point scanning OCT, where lateral resolution depends on the beam spot size. However in parallel OCT, like linear OCT or full-field OCT, reduced spatial coherence is a key to obtaining a uniform irradiation across the microscope objective aperture and to suppressing cross-talk between adjacent detection channels in CCD camera or other detector array [63, 78]. A complex analysis of spatial and temporal coherence effects in FF-OCT can be found in [89].

An OCT light source must exhibit short-term and long-term stability in terms of irradiation amplitude and produced spectrum. Intensity fluctuations are determined by intensity noise (see section 2.2.4), which is a result of bilateral reflections in a laser cavity or in the SLD active region.

For various OCT modalities other requirements might arise. For example, swept source OCT employs a specific type of source, a tunable laser, where parameters like instantaneous linewidth, swept repetition rate or tuning range have to be considered.

3.2. SLD modelling.

SLDs combine properties of laser diodes (LDs), for their output power and brightness, and light emitting diodes (LEDs), because they produce a broadband spectrum. It is possible due to the high optical gain in semiconductor laser material and its broad optical spectrum. Initially SLDs became popular as the real “light sources of choice” for fiber optic gyroscopes in early 1980s [106]. Later on, SLDs were used in fiber optic telecom components and an OCT technique employing a SLD was presented [2]. Hence, a new generation of more powerful (approximately 5 to 10 mW from a single mode fiber) SLD devices was developed. Currently, SLDs are used in
Figure 3.2. Net gain and optical power versus pumping current characteristics of a high power Superlum SLD-38-HP diode [107].

Figure 3.3. Distribution of photon density along active channel in high power Superlum SLD-38-HP diode [107].

Figure 3.4. Distribution of current density along active channel in high power Superlum SLD-38-HP diode [107].
variety of applications, due to their small size, easy use, competitive price compared to alternative sources, relatively high optical power as well as their broad and flat spectrum. Based on Superlum document [107], the main principles of SLD operation will be outlined. Then, a Superlum SLD-761-HP1-DIL-SM diode performance will be analyzed, alone and with an optical amplifier, as a potential 1550 nm range OCT light source.

3.2.1. Principles of SLD operation.

A SLD can be considered as an optimized traveling wave laser amplifier with ideally no reflections from the ends of the active channel, where two counter propagating beams of spontaneous emission travel along an active region. Taking assumption of uniform distribution of carriers inside the active region and zero reflections from its ends, as well as neglecting spectral effects, the distribution of photon density inside the active channel can be described as:

\[
S^+(z) = R_{sp} \frac{\exp[(g - \alpha)z] - 1}{c(g - \alpha)} \tag{3.1}
\]

\[
S^-(z) = R_{sp} \frac{\exp[(g - \alpha)(L - z)] - 1}{c(g - \alpha)} \tag{3.2}
\]

where \( R_{sp} \) is the spontaneous emission rate into the guided mode, \( g \) – modal gain, \( L \) – length of the active channel, \( \alpha \) – non resonant optical losses and \( c \) – velocity of light.

Therefore the SLD output power can be expressed as:
\[ P_{out}^+(L) = h \nu \Pi S^+(L) = h \nu \Pi \left\{ R_{sp} \frac{\exp[(g - \alpha)L] - 1}{c(g - \alpha)} \right\} \]  \hspace{1cm} (3.3)

if \( h \) is Plank’s constant, \( \nu \) the optical frequency and \( \Pi \) is the size of optical mode. From above equation, a get gain can be distinguished:

From (3.3) it is seen clearly that high modal gain is essential to achieve high output power, due to its exponential dependence. According to estimations made by Superlum Diodes, the modal gain in SLDs is at least twice greater than the threshold modal gain in laser diodes. Figure 3.2 presents a net gain and optical power characteristics versus forward current for a typical high power SLD from Superlum. Figures 3.3 and 3.4 show distributions of photon and current density inside active region of the same diode. Comparing to laser diodes, SLDs have higher density of current along active channel. What is more, distributions of photons and current along active region exhibit much greater non-uniformity (Fig. 3.3 and 3.4). Optical gain depends highly on temperature, therefore, output power (Fig. 3.5), spectrum width and central wavelength are temperature dependent as well.

Figure 3.5. Optical power versus temperature for Superlum SLD-26-HP diode [107].
Typical SLDs have a smooth and a slightly asymmetric spectrum that can easily be approximated with a Gaussian function. Non-zero reflections from both ends of active region are the source of residual spectral modulation, but have negligible effect in most applications. The width of the spectrum depends on optical gain spectrum width, but can be broadened by implementing Quantum Well (QW) and Multiple Quantum Well (MQW) heterostructures [108]. However, this cause additional distortions, of the optical spectrum. Comparison of spectra of two corresponding Superlum diodes are shown on figure 3.1a.

The dominating noise in SLDs is Relative Intensity Noise (RIN). It has a white spectrum and its origin is due to beating of independent photons arising from spontaneous emission. There is shot noise present as well, but weaker from RIN by orders of magnitude. However, no dependence of SLD noise on output power has been recognized.

3.2.2. Performance of the Superlum SLD-761-HP1 diode.

The Superlum SLD-761-HP1 superluminescent diode is the light source chosen in this project. It is a single mode fibre pigtailed SLD producing approximately 4.5 mW of optical power (Fig. 3.6). The spectral width and central wavelength were measured several times and are estimated to be 44.5 nm and 1564.5 nm respectively. Therefore, coherence length of this source equals to:

\[ l_{c1} = \frac{2 \ln 2}{n \pi} \frac{\lambda_0^2}{\Delta \lambda_{FWHM}} \cong 24.3 \mu m \]  

(3.4)

As seen on figure 3.8a, SLD spectrum is very smooth (no more than 4.3 % spectral ripple, according to manufacturer) and nearly Gaussian shaped, hence (2.24) can be
used in the above calculation. No excessive rippling in the spectrum envelope yields a smooth autocorrelation function (Fig. 3.8b and 3.8c). However, it is possible that the resolution of spectral data from figure 3.8a might not have been sufficient to discover distortions in calculated autocorrelation function of the spectrum. Since the SLD is single mode fibre pigtailed, its output beam exhibits high spatial coherence [111] and Gaussian power distribution across its beam waist.

Figure 3.6. Optical power versus current (a) and voltage versus current (b) characteristics for SLD-761-HP1 diode taken in temperature 20°C. Both characteristics provided by the manufacturer.

Figure 3.7. Superlum SLD-761-HP1 superluminescent diode mounted on a Newport 740 series.
Figure 3.8. Measured spectrum of Superlum SLD-761-HP diode (a) and its calculated autocorrelation function in linear (b) and logarithmic scale (c). Figure (d) is focused on the peak of the spectrum to determine its FWHM and central wavelength. Measurements were made in 20°C, for maximum diode forward current 315 mA.
3.2.3. Performance of the Superlum SLD-761-HP1 with an EDFA amplifier.

The subject of this subchapter is the performance of an Erbium Doped Fiber Amplifier (EDFA) with an input signal coming from previously described Superlum SLD. The EDFA model available for this test was SRFA1590BDR1SC-MUX from AFC Technologies Inc (AFC-EDFA). EDFAs [112] are fiber-based optical signal amplifiers where gain is created by stimulated emission due to presence of dopant ions in fiber core. They are frequently used as repeaters in fiber telecommunication in the 1550 nm wavelength range. Additionally EDFAs can serve as the gain media for fiber lasers or standalone light sources.

In the project, an AFC-EDFA is used in the saturation power regime, where amplifier net gain is compressed by up to 3 dB, due to high input signal power [112]. Figure 3.9 shows EDFA output power dependence of SLD pumping current. It is clear that gain gets weaker with increasing input signal power, and output power saturates at some point. Experimental studies showed output signal spectral fluctuations, in terms of spectral width and stability, prior to reaching saturation mode.

Manufacturer data sheet provides most important parameters of AFC EDFA:

- saturation power: 59.2 mW at 1590 nm,
- spectral bandwidth: 1570 nm to 1603 nm $\rightarrow \Delta \lambda = 33$ nm.

Noise figure and gain values were published for low input signal (-8 dBm), which makes them irrelevant for this work.

The measured AFC-EDFA output signal spectrum and corresponding autocorrelation function are shown in figure 3.10. The spectral envelope (figure 3.10a) loses its symmetrical shape, but it is still smooth and lacks extensive rippling. Measured optical power and spectrum differs from manufacturers data sheet, as follows:
- measured optical power: 20.85 mW,
- measured spectral bandwidth: 1567.9 nm to 1598 nm \( \Delta \lambda = 30.1 \) nm.

For given spectrum, central wavelength is 1583.4 nm, hence coherence length for AFC-EDFA equals to:

\[
l_c^2 = \frac{2 \ln 2}{n\pi} \frac{\lambda_0^2}{\Delta \lambda_{FWHM}} \approx 36.7 \mu m
\]

(3.5)

Figure 3.9. Interpolated characteristic of AFC-EDFA output power versus Superlum SLD input current. AFC-EDFA power is shown in arbitrary units.
Figure 3.10. Measured spectrum of AFC-EDFA powered by Superlum SLD-761-HP diode (a) and its calculated autocorrelation function in linear (b) and logarithmic scale (c). Figure (d) is focused on the peak of the spectrum to determine its FWHM and central wavelength. Measurements were made in 20°C, for near maximum diode forward current 310 mA.
3.3.4 Discussion.

The Superlum SLD was initially chosen as a light source for this project. However, during laboratory tests on weakly reflecting samples, its optical power was found to be insufficient for area imaging with FF-OCT system in this, particular components configuration. Hence, a high output power, alternative light source system was proposed. It is comprised of EDFA pumped by the SLD diode, which generates 21 mW of optical power, compared to 4.5 mW of SLD’s. Unfortunately SLD and EDFA produced bandwidth is more narrow than pure SLD illumination and is shifted towards longer wavelengths which results in longer coherence length of the source and ultimately, worse OCT axial resolution. Comparison of parameters for both source systems is presented in Table 1.

Both SLD and SLD+EDFA generated light is characterized by high spatial coherence properties. It is unacceptable for FF-OCT application, due to heavy cross-talk noise created by multiple scattered photons in adjacent detection channels in a detector array [63, 78]. Methods of decreasing degree of spatial coherence of optical beam will be described in next subchapter.

<table>
<thead>
<tr>
<th></th>
<th>SLD</th>
<th>SLD + EDFA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Optical power [mW]</td>
<td>4.5</td>
<td>20.85</td>
</tr>
<tr>
<td>Bandwidth [nm]</td>
<td>44.5</td>
<td>30.1</td>
</tr>
<tr>
<td>Center wavelength [nm]</td>
<td>1564.5</td>
<td>1583.4</td>
</tr>
<tr>
<td>Coherence length [μm]</td>
<td>24.3</td>
<td>36.7</td>
</tr>
</tbody>
</table>

Table 1. Comparison of SLD and SLD + AFC-EDFA light source systems.
3.3. Reducing degree of spatial coherence in SLD based light source.

Low spatial coherence of the illumination source is an important factor for any type of full-field optical diagnostics. Karamata in [63, 78] proved highly spatially coherent light causes cross-talk at detector array plane, due to interference between multiple scattered photons entering adjacent detection channels. This speckle related noise leads to severe degradation of both axial and lateral resolution and therefore en-face image quality.

Theoretical analysis states that light propagating in free space over certain distance will have its degree of spatial coherence degraded [113]. However, the most popular way to do it is by using a high-NA, large core diameter Multimode Optical Fiber (MMF). Variety of such fibers available makes it a popular solution for speckle suppression and as a light delivery systems [115, 116]. Authors of [96] enhanced this method by adding an acoustic mode mixer to introduce micro bending at acoustic frequencies to the MMF and cause mixing of transversal modes propagating down the fiber core. But the author of this work never encountered using a mode scrambler in similar work. It is a device that creates micro bendings to the MMF by applying mechanical stress and hence forces mode coupling in the fiber.

3.3.1 Achieving spatially incoherent illumination within a multimode fiber.

The simplest and most cost-effective way to reduce the degree of spatial coherence of the light beam is by launching it into a long, large core diameter, high-NA, step-index MMF. Light travelling down the fiber is systematically dispersed and scattered, due to the inhomogeneity of the fiber core diameter, the core refractive index or fiber bendings [117]. Then optical power is redistributed among a number of rays.
propagating under different angles and paths (an effect known as mode coupling or mode mixing). Hence, many variations of intensity, or propagation modes, can be distinguished inside the waveguide. A main parameter describing step-index fibers is its normalized frequency $V$:

$$V = \frac{2\pi a}{\lambda} \sqrt{n_1^2 - n_2^2} = \frac{2\pi a}{\lambda} NA$$

(3.6)

where $a$ is the core diameter, $NA$ the fiber numerical aperture and $n_1$, $n_2$ are core and cladding indexes of refraction. For fibers characterized by large $V$ values, the approximated number of modes a step-index fiber is able to support can be calculated from:

$$M \approx \left( \frac{2V}{\pi} \right)^2$$

(3.7)

Higher order modes travel over longer paths in the core, so are more likely to be scattered or absorbed at core-cladding interfaces. This means larger light intensity attenuation for multimode fibers supporting great number of modes. On the other hand, for long enough MMFs, it is possible to obtain a uniform light intensity pattern across core diameter, independent of the initial modal distribution. Hence, MMFs can be effectively used as an illumination delivery system.

In full field detection, to avoid measuring cross-talk degraded signals, the input light has to be spatially incoherent. It is necessary to obtain an effect equivalent to confocal spatial filtering, where interference occurs only between light originating from conjugated areas in reference and sample fields [78]. This implies that the cross-sectional plane of light beam leaving the MMF has to be made of mutually incoherent coherence areas, represented by mutually incoherent propagation modes. This
condition is met for sufficiently long fibers, where each mode travels distance differing from other modes by at least a coherence length [118]. The formula for minimum distance for each mode to travel was derived in [119]:

\[ L_{\text{min}} = \frac{l_c M}{n_1/n_2 - 1} \]  

(3.8)

where \( M \) is the number of modes a MMF is able to support, \( l_c \) is the source coherence length and \( n_1, n_2 \) are core and cladding refractive indexes.

Rejection of crosstalk is most efficient when the spatial extent of coherence areas of the light leaving the MMF is no greater than OCT system lateral resolution. Assuming the length condition for MMF, explained above, is fulfilled, mutual degree of coherence for two points at the tip of step-index MMF, spread by distance \( \Delta r \), is equal to [118]:

\[ G(\Delta r) = \frac{2J_1(k_0 NA\Delta r)}{k_0 NA\Delta r}, \]  

(3.9)

where \( J_1 \) is the Bessel function of the first kind, \( k_0 \) is the light wavenumber in vacuum, \( NA \) is the MMF numerical aperture. Sharma in [120] defines coherence area as a region where degree of mutual coherence (3.9) varies between 0.88 and 1, which means the argument of \( J_1 \) needs to be less than unity.

<table>
<thead>
<tr>
<th>Normalized frequency</th>
<th>( V )</th>
<th>309.5</th>
</tr>
</thead>
<tbody>
<tr>
<td>No. of modes supported</td>
<td>( M )</td>
<td>( \sim 38826 )</td>
</tr>
<tr>
<td>Minimal length needed</td>
<td>( L_{\text{min}} )</td>
<td>36.98 m</td>
</tr>
<tr>
<td>Coherence area diameter</td>
<td>( \Delta r )</td>
<td>( &lt; 0.65 \mu m )</td>
</tr>
</tbody>
</table>

Table 2. Calculated properties of FT200EMT multimode fiber.
Taking a step-index MMF chosen as the source illumination delivery mechanism, calculations for Thorlabs FT200EMT, with 200 μm core diameter and 0.39 NA, are presented in Table 2. These results indicate the need is a nearly 40 meters of FT200EMT MMF to obtain spatially incoherent illumination. The theoretical lateral resolution for the OCT system is approximately 2.35 μm (equation 2.25, microscope objective NA = 0.25) which is much more than the achievable coherence area diameter.

More detailed information about the spatial degree of coherence of the optical field in a MMF can be found in [114, 118].

3.3.2. Achieving spatially incoherent illumination by mode scrambling.

Mode scrambling is a process of inducing higher order modes from low order guided modes in an optical fiber to achieve an approximate of stable modal distribution independent of launch conditions. It is a state where power in each mode is constant with respect to other modes as the signal propagates down the fiber. Stable modal distribution is very useful for testing fiber components properties and losses as measurements tend to be dependent on light intensity distributions among modes [117]. Stable distribution is obtainable by propagating light through long fibers, from tens of meters to even kilometers of length, which is not always practical in laboratory applications. Hence, increasing mode coupling using external means, like a mode scrambler, is usually easier and helps focusing on measuring actual fiber properties and losses instead of light launch conditions.

Newport’s FM-1 mode scrambler (Fig. 3.11) uses a precision mechanism to introduce a longitudinal perturbation to the fiber waveguide placed between
corrugated, periodical, jaw-like surfaces. Applied force greatly increases coupling among guided and radiation modes and results in a wide spectrum of spatial frequencies. FM-1 is controlled by a manual knob to achieve a desired output light pattern. Progress in modal redistribution can be observed in real-time using a microscope or camera, however the more pressure is applied, the more time is needed to stabilize the modal pattern within the fiber. Stable modal distribution is achieved in two ways, depending on launch conditions. Mode scrambling requires launching low-order guided modes (underfilled launch – input beam NA < fiber NA, input beam diameter < fiber core diameter), but mode filtering couples light from higher order guided modes to radiation modes (overfilled launch – input beam NA > fiber NA, input beam diameter > fiber core diameter). In this work, scrambled fiber is always underfilled, due to light delivered either from a pigtailed SLD or a single mode fiber output EDFA. The tested device has two microbend periods optimized for 50 μm and 100 μm core diameter fibers.

Manufacturer tested FM-1 device in terms of losses and output modal distribution as an effect of mode scrambling applied. In the latter, a light intensity detector scanned across the fiber end-face in near field. As seen in figure 3.12, mode scrambled distribution is very close to theoretical fully-filled distribution, however
higher order modes intensity tend to be weaker. Figure 3.13 illustrates insertion losses caused mainly by transferring energy from guided to radiation modes when applying lateral pressure during mode scrambling. Overfilled fibers have higher losses, single
dBs of value. Theoretical losses for underfilled fibers are negligible, but in laboratory practice obtaining stable and uniform light distribution on mode scrambled MM fiber end tip requires 1 – 2 dB losses.

3.3.3. Results and discussion.

To test the FM-1 mode scrambler performance, the light beam from the SLD was coupled into a stripped, 1 meter long, 100 µm diameter core, step-index MMF. The fiber was placed in a FM-1 slot to apply point perturbations. Light leaving the fiber was coupled to a 10 meter long, 200 µm core, step index MMF, which serves as an illumination source delivery system and defines the diameter of an output beam. Spatially incoherent light is then collimated and pointed onto a phosphor-coated CCD camera. While changing pressure applied to the stripped fiber, several images were obtained and are shown in figure 3.14. The pressure is represented by displacement of mode scrambler translation stage with corrugated, periodical surfaces attached. Figure 3.14a shows a light spot leaving the end tip of the MMF without mode scrambling. The SLD Gaussian power distribution is distorted with many higher order spatial frequencies arisen. This is a confirmation that with sufficiently long MMF, mode scrambler would not be necessary. However, even a little force applied to the stripped fiber greatly induces mode coupling resulting in more even light distribution across the fiber aperture (Fig. 3.14b, c, d) and further degradation of its spatial coherence. The contribution of mode scrambler to destroying the spatial coherence of the source is shown in figure 3.15, where the USAF 1951 resolution test chart is illuminated by both spatially coherent (SLD with MMF attached) and spatially incoherent light (SLD with MMF and mode scrambling). The backreflected signal is then recorded by the phosphor-coated CCD camera. The image obtained with a spatially incoherent light is
broader and has sharper edges. It is an effect of evenly spread illumination across the beam waist and reduced coherent crosstalk [63, 78]. Figure 3.16 presents interferometric images of the boundary between the reflecting and non-reflecting surface on the USAF 1951 test chart. Images were taken with an assembled full-field OCT system, with SLD and phosphor-coated CCD camera. Spatially coherent signal produces blurred image with undiscernible interfaces boundary. Gaussian-like optical power distribution across its diameter, causes camera signal saturation in the center of beam spot. On the other hand, spatially incoherent signal produces sharp intensity
profile and difference between two surfaces is clear. Interferometric signal intensity is lower due to more evenly spread optical power across beam waist.

Figure 3.15. Images of backreflected light from USAF 1951 resolution test chart recorded by the phosphor-coated CCD camera by illuminating the target with spatially coherent (left) and spatially incoherent light (right).

Figure 3.16. En-face images and intensity profiles of boundary between reflecting and non-reflecting surface on USAF 1951 test chart. Images acquired using spatially coherent (left) and spatially incoherent illumination (right).
It was shown that a combination of multimode fiber and a mode scrambler is a way of reducing spatial degree of coherence of SLD illumination. Further improvement can be achieved by placing an imaging lens after collimating the beam in the interferometer input. Mode scrambling generates losses which are difficult to measure, due to mismatch between core diameters of the scrambled MMF (at least 50 μm) and standard optical power meter inputs (single mode fiber). The CCD camera used in measurements is not calibrated to directly measure the energy of light pointed onto the detector array, however its software displays processed digitizer values as a measure of recorded energy. Therefore it recognizes subtle falls of acquired intensity when increasing pressure on the scrambled MMF. According to those readings, losses are negligible up to 100-140 μm of mode scrambler translation stage displacement. Then grow in more or less quadratic manner, which agrees with logarithmic scale on graph shown in figure 3.13.

3.4. Analysis of anti-Stokes imaging device.

Time domain and classical swept-source OCTs employ single channel photoreceivers, usually avalanche or PIN photodiodes optimized for various bandwidths [29]. In spectral domain OCT, the signal is dispersed by a spectrometer and collected by an array detector, such as photodiode array or CCD (charged-coupled device) camera [29]. Wide field systems require an array type of detector, such as CCD, CMOS [97] or InGaAs [101] cameras, to capture en-face images without lateral scanning. Imaging in the 1550 nm range is generally limited to InGaAs detectors, which are expensive, especially as one or two dimensional arrays. A cost effective alternative is a phosphor-coated CCD camera, which was previously used in
interferometric measurements [122]. The coating is an anti-Stokes material that upconverts radiation in 1460 to 1625 nm range to visible light detectable by the CCD. In this subchapter, the anti-Stokes imaging device will be reviewed as a potential detector for the 1550 nm range full-field OCT.

3.4.1. Detection system requirements for OCT purposes.

There are several decisive factors governing detector suitability for interferometric imaging. Since a CCD camera is used in the project, the focus will be put on array detector parameters.

- Frame rate. Limits imaging system maximum operation speed. Ideally, imaging systems should work in real time and produce instant results. In medical imaging, probe-patient contact ought to be as brief as possible to avoid undesirable movements of examined body part. Related to camera signal exposure time, camera architecture and camera electronics speed.

- Dynamic range. It is ratio between maximum and minimum resolvable signal power. It determines the maximum number or resolvable steps at which measured signal can be distinguished, and hence high dynamic range is desired. Camera dynamic range can limit whole system dynamic range. Dependent on camera quantum efficiency.

- Linear wavelength response. Or more precisely, it is desired that detector’s quantum efficiency value remains stable for wavelengths that are consisted within the light source spectrum. If a sub-band within source bandwidth is weakly detectable or totally rejected by the camera, it results in loss of useful
signal illumination, and might narrow the detectable source bandwidth and ultimately degrade the OCT axial resolution.

- Spatial resolution. Is determined by the size of an individual element of a detector array. Imaging with high lateral resolution requires small pixel area, so highly magnified images can be recorded. But there is a tradeoff between pixel size and its full-well capacity and therefore with camera dynamic range, as smaller pixels can hold lesser maximum charge. Camera pixel size should be matched with OCT signal lateral resolution.

- Linear power response. Camera with linear response over a range of light intensities has better quantitative capabilities. Hence collected images are easier to interpret due to correct digital representation of detected intensity.

3.4.2. Characteristics of CCD technology.

Cameras using CCD (charge-coupled devices) readout elements are the most widely used area detection devices in commercial, professional and scientific applications due to their low noise and high dynamic range. A CCD is formed of overlapping light-sensitive MOS capacitors on a thin, silicon substrate. These photosensitive elements create charge packets from collected photons during certain exposure (or integration) time, store and pass them via shift register to a floating diffusion output and an amplifier to create a voltage signal. The latter is transmitted to an n-bit analog-to-digital converter and converted to a binary code interpretable by computer [123]. The major parameters of CCD readout elements are charge-handling capacity (or full-well capacity), charge-transfer efficiency, charge-to-voltage conversion and noise. Noise
sources in all type of CCD image sensors are dark current pattern, shot noise and amplifier noise. Total contribution of noise patterns is negligible compared to maximum charge capacity (Fig. 3.17) and increases with temperature.

However, low noise and high dynamic range are compromised by moderate operation speed compared to CMOS cameras or Photomultiplier Tubes (PMT). The readout rate depends on CCD image sensor architecture, camera electronics, and varies from tens of Hz to tens of MHz for devices made for sophisticated scientific purposes. Faster operation requires shorter exposure times, faster electron transfer and pixel value digitizing, and hence generates higher noise volumes and worse signal-to-noise ratio.

Technological improvements have enabled the manufacture of smaller CCD pixels, which brought imaging sensors characterized by outstanding spatial resolution comparable or even smaller than typical film grain size (~ 10 μm). High resolution

Figure 3.17. Related signal and noise volumes as a function of CCD exposure time [123].
imaging usually requires large pixel arrays to cover larger area, but with the expense of operation speed.

In all CCDs, output voltage signals grow in a linear manner with respect to measured illumination, with the exception of near saturation levels. On the other hand, quantum efficiency, the parameter representing effectiveness of a CCD sensor to generate charge from incident photons is wavelength dependent. The detectable spectrum lies within 400 to 1100 nm with peak sensitivity in range of 550 – 800 nm. As an example, figure 3.18 shows how quantum efficiency changes with wavelength for a typical full-frame CCD sensor. Quantum efficiency for a standard CCDs reaches 50 – 60%, but for some high-performance sensors, it even approaches 90%.

CCD cameras are a large and diverse group of imaging devices. Versatility of this technology makes inexpensive cameras available for commercial use, as well as
high resolution and/or high speed devices available for microscopy and other scientific purposes. More detailed technical information about CCD sensors can be found in [123].

3.4.3. Anti-Stokes coating technology.

The anti-Stokes coating is a phosphor thin film bound directly to a CCD array. It serves a purpose of absorbing emission between 1460 nm and 1625 nm and emitting in the range of 950 nm to 1075 nm. Up converted light can be detected by CCD and therefore it is possible to use the coated camera as an inexpensive and efficient array detector for infrared. Such a device was patented by Applied Scintillation Technologies Ltd. as an alternative to bulky and expensive InGaAs cameras [124]. Similar techniques using different coating materials were previously used to adapt a CCD to x-ray imaging [125].

![Figure 3.19. Phosphor coated CCD camera nonlinear response to light intensity at 1550 nm and 633 nm [122].](image-url)
Phosphor sensitive to the 1550 nm band may be $\text{Y}_2\text{O}_2\text{S}:\text{ErYb}$, $\text{YF}_3:\text{ErYb}$, $\text{NaYF}_4:\text{ErYb}$, $\text{La}_2\text{O}_2\text{S}:\text{ErYb}$ or a related material [124]. It is a crystalline host doped with ionized rare earth ions $\text{Er}^{3+}$ and $\text{Yb}^{3+}$ to trigger the material’s fluorescence emission with photon energy greater than energy of absorbed photons. Additional energy comes from doping ions of the emitting material. This phenomenon is in contrary to Stoke’s second law and hence is described as an anti-Stokes effect. The coating may be attached to CCD using an isobutyl/butyl acrylic copolymer adhesive [124].

Additional energy released when anti-Stokes effect occurs is the cause of nonlinear up-conversion from infrared to CCD detectable light. Figures 3.19 and 3.20a present measured and data sheet camera response characteristics, respectively. Both show roughly quadratic response to 1550 nm wavelength and hence greater camera sensitivity to light of higher intensity. Comparing with visible light, the measured curve shows absolute rejection of low incident intensity for 1550 nm, as shown in

![Graphs from L070-1550 camera data sheet](image-url)
Figure 3.19. Figure 3.20b shows a cross-section of a fiber beam incident on a phosphor coated camera. The beam width is detected smaller than actual as a result of signal wings suppression due to their lower intensity.

Egan reported in [122] that pixel blooming occurs for high intensity measurements. In fact, image blurring happens for any measurements with anti-Stokes CCD, due to diffusion of the light by the phosphor coating. According to the manufacturer, camera spatial resolution will not exceed 50 μm, despite the small pixel size (7.4 μm). Figure 3.21 illustrates an image of a digit labeling fifth group on the USAF 1951 test chart. Despite positioning the test chart surface precisely within microscope objective focal plane, the object does not have sharp edges due to diffusion of up-converted light.

Another issue is a non-linear wavelength dependency of anti-Stokes up-conversion. Figure 3.22 shows characteristic of signal intensity required to saturate the camera over entire phosphor sensitive range, as well as its distinguished parts for both SLD and AFC-EDFA bandwidths. Wavelength dependency is very irregular and for

Figure 3.21. Image of a digit denoting a group 5 on a USAF 1951 test chart. Shape does not have sharp edges due to light diffusion by phosphor coating.
Figure 3.22. Signal required vs. wavelength to achieve camera full signal illumination by anti-Stokes up-conversion material. (a) shows response for entire sensitive spectrum of phosphor coated CCD. (b) and (c) present phosphor coated CCD response for SLD and SLD+EDFA spectra, respectively. Original graph comes from Spiricon L070-1550 camera data sheet [126].
some wavelengths differ by an order of magnitude. For instance, for 1561 nm and 1586 nm, 5 mW/cm$^2$ is required to achieve camera full signal illumination. For 1575 nm, it is nearly 50 mW/cm$^2$. These wavelengths are within SLD and AFC-EDFA bandwidths, so it means excessive losses during OCT detection.

The anti-Stokes imaging device is an inexpensive alternative for infrared area imaging. It cannot be used as a measure of incident power and due to its degraded spatial resolution, high resolution imaging is impossible. Nonlinear power response and limited sensitivity for low intensity signals may cause problems in OCT detection of weakly reflecting interfaces. The phosphor coating wavelength dependency impact on OCT axial resolution will be investigated in next chapter, while characterizing the assembled OCT system. On the other hand, anti-Stokes camera can be successfully employed in telecom measurements and moderate resolution imaging, as will be shown in the next chapter. Phosphor nonlinear up-conversion can be corrected to some level using camera software algorithms (Fig. 3.20), but with expense of severe degradation of dynamic range. According to camera data sheet, SNR can drop even by 30 dB when using non-linearity correction [126].


L070-1550 (Fig. 3.23) is an interline transfer progressive scan, phosphor coated CCD camera initially intended for telecom measurements. Applications include evaluating single and multimode fiber output, analyzing beam profiles and output of light sources emitting within 1460 nm to 1625 nm region. It is useful in aligning optical systems, can measure X and Y divergence of devices, and analyze the pointing direction anomalies created by improper fiber end polish, and many others. It is a 12
bit A/D, PC-based device with a USB 2.0 interface and LBA-USB PC software provided. The device can be equipped with variety of spectral filters and imaging lenses. The most important parameters provided by the manufacturer are outlined in Table 3.

Table 3. Technical properties of L070-1550 camera.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of elements</td>
<td>640 x 480</td>
</tr>
<tr>
<td>Element pitch</td>
<td>7.4 x 7.4 μm</td>
</tr>
<tr>
<td>Detector area</td>
<td>4.7 x 3.6 mm</td>
</tr>
<tr>
<td>Spectral response</td>
<td>1440 – 1605 nm</td>
</tr>
<tr>
<td>Full video in continues wave operation</td>
<td>7 mW/cm² at 1550 nm</td>
</tr>
<tr>
<td>Dynamic range</td>
<td>61 dB (30 dB with nonlinearity correction)</td>
</tr>
<tr>
<td>Frame rate</td>
<td>60 fps at full resolution, 100+ fps at 320 x 240</td>
</tr>
</tbody>
</table>

Figure 3.23. Ophir-Spiricon L070-1550 camera (left) and technical drawing (right).
The LBA-USB software is a versatile application designed for Ophir-Spiricon cameras allowing various detection modes, advanced numerical operations and post-processing over detected beam. Some of its features are:

- False color beam intensity profile displays in 2D and 3D (Fig. 3.24),
- Numerical analysis employing advanced calibration algorithms,
- SNR improvement through pixel binning, frame summing and averaging, background subtraction,
- Statistical analysis of all measured parameters
- Beam stability analysis,
- Post processing capabilities,
- Pass/fail testing,
- User selectable choices for beam width measurements,
- Whole beam and Gaussian fits to beam data,
- Drawn and Auto aperture for isolating data,
- Results and data logging capabilities,

LBA-USB uses ActiveX server to exchange frame data, frame images, results and statistics and to perform basic controls over the camera with external PC applications. These include National Instruments LabVIEW, Microsoft Excel, programming languages like Visual C++, Borland C++ Builder and Visual Basic.

LBA-USB software communication with camera and the external subjects is performed in an event-driven environment. When the camera software collects a new frame or data, a new event is triggered and any operation over that data can be performed. But, when a new frame is collected when last event is still being processed,
the new event is not fired and the new frame is discarded. Laboratory tests showed discarding data even for low frame rates and its probability of occurrence grows when defined frame rate parameter increases. Thus the number of collected frames within a unit of time varies and perfect timing synchronization is impossible. This drawback is significant from the OCT point of view, where various devices require mutual synchronization. It can be overcome by incorporating the camera functions into a queuing system, where other devices stay idle until camera event is fully processed and a new one triggered.

The actual frame rate of the tested device does not reach values published in specification by manufacturer. For full resolution (640 x 480) and 12 bits per pixel,
frame rate varies between 40 – 45 Hz. However this can be easily improved by enabling a camera function that groups neighboring detecting elements, averages their intensity value and processes them as one pixel (pixel binning) to handle less data in a unit of time. For example when 2x2 binning is activated, every pixel on an acquired image is constituted of 4 detecting elements contribution. This is a way of improving SNR as well, but happens with the expense of degradation of spatial resolution of the camera array. When 2x2 binning is enabled, resolution is 320 x 240 and for 12 bit per pixel, the acquisition rate changes from 80 to 95 Hz. All technical requirements regarding a test PC connected to the camera, outlined in the camera manual, were met or exceeded requirements.

Despite using phosphor coating, the camera responds efficiently to radiation within typical CCD detection band. It was confirmed by using a red laser and a broadband halogen lamp. This means phosphor is transparent or at least semi-transparent to those wavelengths, but its spectral efficiency has not been reviewed.

The technical specification of the described camera originates from its manufacturer sources, including the camera software manual [126].

3.4.5. Discussion.

The anti-Stokes phosphor coating used in the described camera allows an easy and cost-effective method of detecting light in range of 1440 nm to 1625 nm, but at the expense of a few important parameters characteristic for CCDs. Limited sensitivity for low intensity signals and highly non-linear wavelength response suggest possible issues with OCT imaging of weakly reflecting samples. Diffusion of light caused by the phosphor, considerably limits camera spatial resolution which is an important
parameter for en-face imaging. Varying frame rate can be overcome by proper design of OCT operating software, but will limit system’s speed of operation. LBA-USB software provides an easy and comprehensive way of processing and exchanging recorded data with external computer or network applications. It brings tools like pixel binning, automatic frame summing and averaging that improve the camera dynamic range. Averaging or summing of up to 256 frames improves the dynamic range by up to 24 dB.
4. Full-field Optical Coherence Tomography with a 1550 nm SLD and anti-Stokes imaging device.

4.1. System description.

This chapter will outline technical aspects of the assembled FF-OCT system, starting with architecture of optical setup and explanation of each function component. A LCI signal envelope retrieving formula will be described, alongside with motivation of using this particular algorithm. The next part will be devoted to the realization of OCT operation process. Eventually, functional OCT parameters are analysed. It includes axial and lateral resolution, signal-to-noise ratio analysis, system speed and test of a dynamic focusing feature.

4.1.1. Optical setup.

The FF-OCT system is based on a free-space Michelson interferometer incorporated into a Linnik configuration [72]. Its schematic and picture are shown in figures 4.1 and 4.2, respectively. The light source is AFC-EDFA supplied by Superlum, SLD-761-HP1 superluminescent diode. A mode scrambler and a 10 meters long multimode fiber are used to destroy the spatial coherence of the source illumination. The latter delivers spatially incoherent light to the system and is plugged into collimator \( L_C \). The collimated beam is conjugated by \( L_1 \) lens and divided into sample and reference beams by a 50/50 beam-splitter (BS). The interferometer reference arm consists of a dry microscope objective (MO – Olympus 10X plan achromat objective, 0.25 NA) and a reflecting mirror attached to a high precision
piezoelectric actuator PZT (PI S-303.CD, maximum travel range 2 μm, closed-loop operation, resolution 0.03 nm). The PZT is mounted on a custom holder and its position and objective position remain fixed. The sample arm is built using another MO objective mounted on a motorized piezoelectric translation stage O_MTS (Agilis AG-LS25, open-loop operation, travel range 12 mm, approximated resolution 50 nm) and sample holder placed on a motorized translation stage S_MTS (ThorLabs MTS25-Z8, closed-loop operation, travel range 25 mm, bidirectional repeatability 1.6 μm). The S_MTS performs OCT depth scanning. Its large load capacity allows mounting a sample holder and three manual translation stages to facilitate X-Y and rotation adjustments of the mounted specimen. The O_MTS is used for dynamic focusing purposes (see section 2.3.5.2) to keep interferometer coherence plane and objective

Figure 4.1. Experimental arrangement of a 1550 nm range FF-OCT. It is consisted of following elements: BS – beam splitter, MO – microscope objective, PZT – piezoelectric actuator with reference reflector attached, S_MTS, O_MTS – sample and objective motorized translation stages, L_C – collimator, L_1, L_2 – conjugating lenses.
focal plane coincident while imaging the internal structures of samples. O_MTS works in an open loop system, thus it is necessary to align its position after several measurements. The beams from reference and sample arms propagate back to beam-splitter and are imaged onto detector array by the large-focal length tube lens L2 (focal length 250 mm), where they undergo interference. The detector surface is coincided with the L2 focal plane. A selection of L2 with this particular focal length will be discussed later in this chapter.

All optical elements, are anti-reflection coated and designed for 1050 – 1620 nm band. L1 and L2 are achromatic doublets placed in the system so sides with lesser radius of curvature are facing the collimated beam to minimize spherical aberrations. Both objectives come from the same manufacturer and have the same properties, although they are not perfectly matched with each other. Experimental measurements
revealed small difference between their working distances (~ 30 μm), but no impact on OCT performance was observed. Sample and reference arms of the interferometer are telecentric, are kept as short as possible and do not exceed 10 cm. Two beamsplitters were tested and neither was perfectly cubic nor was refracting beams under exactly 90°, which made the free-space optical system much more difficult to align. Beam-splitter imperfection and not perfectly matched objectives can be an additional source of dispersion [127]. But due to relatively narrow light source spectrum width (maximum of 44 nm in case of SLD, 31 nm for AFC-EDFA) and short interferometer arms, it was decided that additional dispersion compensation was not necessary. During OCT measurements, the interferometer was covered to minimize possible phase changes resulting from particle movements in air.

The OCT system was automated and controlled by PC based application written in NI LabVIEW. This custom software collects scan images from LBA-USB application, performs the phase shifting algorithm and saves the data. Image visualization and post-processing is performed using MatLab or ImageJ software.

4.1.2. Phase-stepping algorithm description.

OCT systems should operate fast enough to achieve real-time image acquisition rate, and to minimize probe-sample contact time. In high axial resolution FF-OCT imaging, the detector is usually a factor limiting the speed of operation. What is more, averaging or summing a number of frames of the same volume within the specimen, to increase system signal-to-noise ratio, is a standard procedure. Most FF-OCTs employ Phase Shifting Interferometry (PSI) methods to extract en-face images [80, 81]. A discrete or continuous phase shift is realized by precise mechanical
displacement of a reflector in the reference arm [73, 74] and \( M \) (two or more) phase shifted images are recorded. This allows using various PSI algorithms to obtain sample reflectivity map at certain depth. Hence, high camera frame rate is important, as multiple images of the same region in the sample need to be collected. For maximum time efficiency, a precise synchronization of camera, phase shifting actuator and depth scanning translation stage is required.

Data transfer in Ophir-Spiricon phosphor-coated camera is performed in an event-driven environment, where frame rate varies (see section 3.4.4). Thus it is impossible to synchronize timing between the camera and PZT and implement continuous phase-shifting techniques [82] to minimize data acquisition time. Instead, to preserve a reliable data collection, a slower phase stepping technique [80] is used. It requires forcing PZT displacement to obtain phase shift and then recording a frame or a number of frames for averaging or summing purposes in a sequential manner. This procedure is repeated \( M \) times for every measurement step, where \( M \) is a number of phase shifts in the algorithm.

The numerical description of the employed method is analogous to the one presented in section 2.3.2. It contains a \( M = 4 \) frame algorithm and variable number \( N \) of averaged images. Thus, an averaged interferometric signal intensity at \((x, y)\) location at array detector surface can be written as:

\[
S_i(x, y) = \frac{1}{N} \sum_{j}^{N} \left[ I_{SRj}(x, y) + 2 \sqrt{I_{Sj}(x, y)I_{Rj}} \cos(\varphi(x, y) + i \frac{\pi}{2}) \right] \tag{4.1}
\]

where \( i = 0, 1, 2, 3 \) and denotes corresponding phase-shifted images. These are combined by the following formula to construct the en-face sample reflectivity image:
\[ S = (S_0 - S_2)^2 + (S_1 - S_3)^2 \] (4.2)

This simple algorithm requires minimum computations, provides instant image reconstruction and elimination of DC background signal component \( I_{SR}(x, y) \). However, this method requires stable measurement conditions as any external vibrations may distort the phase of interfering waves and cause residual fringe pattern in calculated images.

The collection of en-face images is saved as a 3D numerical array and can be visualized in MatLab or ImageJ. Further post-processing can be applied to enhance image contrast and details visibility. In this work, different types of data processing was used, i.e. Wiener or Gaussian smoothing, band-pass filters, despeckle algorithm, etc. These operations are optional and depend on type of sample imaged and the resulting clarity.

4.1.3. Method of operation.

The FF-OCT system uses custom operation software created in NI LabVIEW. Its tasks include communication with the camera, PZT and two motorized stages, automation of scanning process and managing acquired data.

The motorized translation stages are linked to the PC by an USB interface to receive control information and to send back status data. S_MTS is equipped with comprehensive ActiveX LabVIEW drivers regulating translation velocity, acceleration and providing precision information over its absolute and relative position. O_MTS communicates using a simplified set of LabVIEW VIs and command line instructions. To control the PZT actuator, a USB connected National Instruments
Data Acquisition Card (NI USB-6221) generates the desired analog voltage level, which is amplified by the PZT amplifier/controller and applied to the actuator to cause its displacement and perform a phase shift. For the camera, a recorded frame is acquired directly by LBA-USB software using the USB 2.0 interface. The frame can be extracted in LabVIEW. This specific solution requires the LBA-USB to be active during measurements and limits operation speed, but allows automatic pre-processing, like frame averaging, background subtraction, gain control, and many more. Frame capture is triggered by a raising slope of a square voltage signal created by the NI DAQ.

The scanning process begins with activating the OCT custom application and defining several scan parameters:

- Camera trigger signal frequency \( f \),
- Scanning range \( Z \),
- Scanning step \( \Delta z \) (absolute distance between successive measurement volumes),
- Approximated refractive index of the sample \( n \) and air-sample interface position related to S_MTS position \( z_n \) (optional, for dynamic focusing purpose only),
- Light source selection (purpose of picking right phase shifts),
- Path choice for data saving,
- Inside LBA-USB camera software, an ‘external trigger’ option has to be enabled and number of images to average \( N \) chosen.

After defining the input parameters, the OCT measurement operation performs as depicted in the block scheme in figure 4.3. Tasks within a single block, denoted with
1. Activating camera trigger
2. Acquiring image made of $N$ averaged frames and saving it to buffer
3. Changing PZT voltage level to obtain adequate phase shift
4. Repeating steps 2 and 3 three more times.
5. Setting PZT to zero displacement. Deactivating camera trigger.

- Combining four acquired frames using 4-step algorithm and saving resulting en-face image to buffer
- Moving S_MTS by $\Delta z$ distance
- If sample refr. index $n > 1$ and $z > z_n$, O_MTS position is adjusted

- If $z \neq z_0$, S_MTS is returned to the starting position
- If $n > 1$, O_MTS is returned to the starting position
- Camera software standby mode is disabled
- Data buffer is saved to specified disk location and can be opened in MatLab or ImageJ for visualization and further post-processing

Figure 4.3. Simplified block scheme outlining FF-OCT operation details.
pointers, can be done simultaneously. A block with tasks requiring certain order of execution, uses numbering.

After the measurement is finished, application returns translation stages to their initial positions, disables camera recording and saves collected data on disk for further visualization and/or post processing.

4.1.4. System performance.

4.1.4.1. Resolution.

In FF-OCT, the axial resolution depends on the coherence length of the source and on the microscope objective numerical aperture and is governed by (2.45). The best achievable lateral resolution is a function of the objective NA and wavelength. Both theoretical values for the presented system with AFC-EDFA light source and 0.25 NA microscope objectives are 36.7 μm and 2.3 μm respectively. However, the real achievable lateral resolution is worse due to optical aberrations introduced by microscope objectives in the near infrared. From image visualization point of view, these values are still highly disproportional.

The anti-Stokes camera has pixels of size 7.4 x 7.4 μm, but real spatial resolution is approximately 50 μm, due to light diffusion by the phosphor coating. Hence, lateral resolution can be adjusted by picking imaging lens \( L_2 \) (Fig. 4.1). \( L_2 \) with relatively short focal length means small imaging field and therefore greater optical power density on the detector array (which may result in improvement of the system dynamic range) and poorer lateral resolution. Longer focal length of \( L_2 \) enables acquisition of more detailed en-face images, but the optical power density across detector array is lower, as the imaging field is larger. A 250 mm focal length \( L_2 \) offers
lateral resolution of 12.5 μm and an imaging field of approximately 2.75 mm in diameter. It covers large area of the detector, and facilitates approximately 10X magnification with respect to the actual sample dimensions. The lateral resolution was estimated by finding the smallest resolvable element on a USAF 1951 test chart. It was a short edge of third element in fifth group, equal to 12.5 μm, as depicted in figure 4.4.

![Image](image.png)

**Figure 4.4.** An en-face image of the USAF 1951 resolution chart, group 5, elements 2 and 3.

Axial resolution was measured indirectly by performing OCT scans across an interface between air and a flat, weakly reflective surface and estimating its width at half-maximum value related to noise level. A soda-lime glass plate with refractive index of 1.52 and reflectivity of approximately 4.2 % [128] was chosen. Scans were made for various number $N$ of averaged images and are shown in figure 4.5 in linear and logarithmic scales. For $N=1$, interface profile is severely distorted, due to optical signal phase fluctuations, and it is difficult to resolve its FWHM. With increasing $N$,
profiles are smoother and estimated FWHMs vary between 35 μm and 38 μm. Hence, experimental results are close to the theoretical value of 36.7 μm.

A similar experiment was performed for a highly reflective surface, in this case mirror optimized for infrared illumination (R ~ 99%). Results, presented in figure 4.6, confirm that the phosphor-coated camera has a nonlinear signal response (section 3.4.3, figure 3.19) and due to greater sensitivity for higher intensity signals, camera pixels saturate. This cause the interface axial profiles to be wider than they should be, as the camera is not able to distinguish intensity changes after full-well capacity of pixels reaches its limit. This effect will occur when imaging solid samples with well-polished highly-reflective surfaces. Depth scans will show disproportionally wide profile of the surface with bright side lobes on both sides, as illustrated in figure 4.6.

Experiments with measuring profiles of air-glass and air-mirror interfaces confirm that the axial resolution does not depend on image averaging.
Figure 4.5. Profiles of an air-glass interface in linear (left column) and logarithmic scales (right column) to estimate axial resolution and signal-to-noise ratio. Scans were made across total optical distance of 120 μm, 1 μm steps and for various number of averaged images N.
4.1.4.2. Signal-to-noise ratio analysis.

Dynamic range or SNR of OCT system was previously described in section 2.2.4 and can be defined as the ratio of the signal power back-reflected from a perfectly reflecting mirror to the noise generated in the system. The lowest detectable signal is equal to the noise level. Theoretical sensitivity for OCTs using a detector array can be calculated from (2.46). The camera manufacturer does not provide detailed technical information about the CCD array used in the camera, such as full well capacity of pixels or electric noise, necessary to use (2.46). However, the manufacturer provides estimated values for the lowest measurable signal intensity and saturation intensity, \(~70\) nW/cm\(^2\) and \(7\) mW/cm\(^2\) respectively. The ratio of these values is equal to 50 dB, which is in contrary to the camera dynamic range (61 dB, without non-linearity correction, see section 3.4.4) claimed by the manufacturer. Therefore, it was decided to rely only on experimental data.

The sensitivity was estimated from the same experiment as the axial resolution, using measured profiles of air-glass and air-mirror interfaces for various number of averaged images N.

Figure 4.6. Profiles of an air-mirror interface in linear (left) and logarithmic scales (right) to estimate axial resolution and signal-to-noise ratio. Scans were made across total optical distance of 110 μm, 1 μm steps and for various number of averaged images N.
images $N$. Results are illustrated in figures 4.5 and 4.6. For a weakly reflecting sample, the SNR changes from 30 dB to 35 dB, with increasing $N$. When imaging a highly reflecting surface of a mirror, system SNR increases by 16 dB for $N=1$, compared to air-glass measurements. This shows again, that the phosphor coated camera nonlinear response affects whole OCT system.

A comparison of results from performed measurements is outlined in Table 4. For a weakly reflecting sample, system sensitivity does not improve much when increasing $N$. But when imaging highly reflective surface, SNR improvement is apparent when increasing $N$ and its manner of change is roughly convergent with the curve from figure 2.19.

<table>
<thead>
<tr>
<th>SNR [dB]</th>
<th>N</th>
<th>1</th>
<th>10</th>
<th>20</th>
<th>50</th>
<th>100</th>
</tr>
</thead>
<tbody>
<tr>
<td>for an air-glass interface</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(R ~ 4.2 %)</td>
<td>30</td>
<td>33</td>
<td>33</td>
<td>35</td>
<td>32</td>
<td></td>
</tr>
<tr>
<td>Air-mirror interface</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(R ~ 99 %)</td>
<td>46</td>
<td>55</td>
<td>56</td>
<td>64</td>
<td>65</td>
<td></td>
</tr>
</tbody>
</table>

Table 4. Signal-to-noise ratio of FF-OCT system measured for weakly and highly reflecting sample for various numbers of averaged images $N$.

4.1.4.3. System speed.

The operation speed of any electronic or electro-mechanical system is limited by its slowest components. The PZT is limited by its electronic delay, while translation stages are limited mainly by mechanical issues, such as inertia. All are negligible, compared with the theoretical camera image acquisition rate. The L070-1550 camera operates with varying frame rate of 80 to 95 Hz for 2x2 binning. Rejecting frame data
as a result of event driven environment (see section 3.4.4 for more details) reduces the speed of OCT as well.

The OCT speed was tested by measuring total operation time of a scan across 1.8 mm in depth with 15 μm step, which means 120 iterations. Measurement was repeated several times for scans with different number of averaged images $N$. Assuming a camera frame rate of 80 Hz and no rejection of frame data, use of the four frame algorithm, 120 iterations and $N=1$, the data acquisition of a single volume should take approximately 6 s. Based on the results from Table 5 for $N=1$, total time for components initialization, software data operations, components control and data saving takes 76 s. This means 76 s is a estimated time for all mentioned tasks for any $N$ value. When $N=2$, total operation time increases by calculated theoretical value of 6 s. However, when $N>2$, data acquisition time per additional averaged image increases linearly with $N$ (Table 5, figure 4.7). This is an effect of the event-driven system that the camera operates in. For longer data acquisition, discarding data occurs statistically more often and idle time, when system is awaiting for a “new frame event” to be fired, accumulates, and extends total measurement time.

The camera is the main factor limiting speed of the system and achieving real-time imaging is impossible. However, more efficient LabView programming could slightly improve the FF-OCT operation speed by reducing the 76 s time for performing initial tasks, data operations and components control.

<table>
<thead>
<tr>
<th>$N$</th>
<th>1</th>
<th>2</th>
<th>4</th>
<th>10</th>
<th>20</th>
<th>50</th>
<th>100</th>
</tr>
</thead>
<tbody>
<tr>
<td>$T$</td>
<td>1'22''</td>
<td>1'28''</td>
<td>1'54''</td>
<td>3'15''</td>
<td>5'6''</td>
<td>12'31''</td>
<td>25'17''</td>
</tr>
<tr>
<td>$(T - 76'')/N$</td>
<td>6''</td>
<td>6''</td>
<td>9.5''</td>
<td>11.9''</td>
<td>11.5''</td>
<td>13.5''</td>
<td>15.9''</td>
</tr>
</tbody>
</table>

Table 5. Total operation time $T$ for different number of averaged images $N$. 

- 103 -
Dynamic focusing test.

Dynamic focusing is used in infrared FF-OCT to keep the focal plane of microscope objective, placed in sample arm, coincident with the interferometer coherence plane without using immersion medium objectives. More detailed information can be found in section 2.3.5.2.

In the presented FF-OCT system, the microscope objective in the sample arm is placed on a separated motorized translation stage O_MTS. System software enables entering the estimated sample refractive index $n$ and position of the sample surface. The latter parameter is a position of the sample translation stage S_MTS, noted when reference field interferes with light back-reflected from the sample surface. Hence, imaging with dynamic focusing requires a pre-scan to estimate its surface position, every time a sample is changed. If entered value of $n$ is greater than unity and scanning progresses through the sample, position of microscope objective is adjusted by O_MTS. The adjusting translation is calculated from (2.47).

The significance of dynamic focusing was confirmed in the experiment, where four OCT scans were performed across a USAF 1951 test chart. The target was imaged...
from the back side and the test pattern was reached after going through 1.54 mm of soda-lime glass. All scans were made in identical conditions with the same parameters, except for sample refractive index parameter \( n \), set to 1 (no dynamic focusing), 1.3, 1.5 (soda-lime glass refractive index) and 1.7. Obtained test patterns are shown in figure 4.8. For \( n=1 \), it is impossible to identify the imaged object. For \( n=1.3 \) and \( n=1.7 \), shapes are resolvable, but blurred. When \( n \) is equal to the sample refractive index, the obtained image is sharp and has better signal-to-noise ratio, compared to other scans.

Figure 4.8. En-face images of label and first element of 5th group on USAF 1951 resolution chart. Scans were performed across glass substrate of test chart (soda-lime glass, 1.54 mm thickness, refractive index 1.52) to reach resolution pattern. Refractive index of the sample was set in OCT custom software to: \( n=1 \) (a), \( n=1.3 \) (b), \( n=1.52 \) (c), \( n=1.7 \) (d).
4.2. Principle of operation – glass plate thickness.

This part of the chapter aims at proving that the presented FF-OCT system is capable of retrieving depth information of examined samples. Hence FF-OCT was used to perform a depth scan across a glass plate of known refractive index and thickness to uncover air-glass interfaces. Then, experimental results will be compared to known data. The same experiment was performed for a stack of two sapphire windows.

4.2.1. Experiment.

The experiment was designed to confirm the LCI principle of operation, i.e. sample internal structure measurement capabilities, of assembled FF-OCT system. A soda-lime glass plate and sapphire windows were scanned throughout their thickness in order to obtain the samples’ internal structure information with indicated boundaries between mediums differing in refractive index.

The OCT setup is exactly as presented in section 4.1.1, figure 4.1, with the SLD and AFC-EDFA combination as a light source (36 μm coherence length, ~20 mW optical power). S_MTS position was adjusted, so that the distance between sample front surface and beam-splitter was 150 – 200 μm greater than the length of the reference arm. Scan parameters are as follows:

- scan range 3 mm,
- scan step 10 μm (which gives total number of 301 iterations),
- number of averaged images, \( N = 50 \),
- dynamic focusing on (\( n \) equals refractive index of examined material).
Samples are a soda lime glass plate and a stack of two sapphire optical windows. Their thickness and optical properties are outlined in Table 6.

<table>
<thead>
<tr>
<th>Material</th>
<th>Thickness S [mm]</th>
<th>Refractive index n (for 1585 nm)</th>
<th>Air-material reflectivity R [%]</th>
<th>Transmission (for 1 mm thickness) T [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td>soda lime glass</td>
<td>1.54</td>
<td>1.506</td>
<td>4.2</td>
<td>84</td>
</tr>
<tr>
<td>sapphire</td>
<td>0.5, 1</td>
<td>1.746</td>
<td>7.4</td>
<td>86</td>
</tr>
</tbody>
</table>

Table 6. Properties of soda lime glass and sapphire [129] samples.

4.2.2. Results and discussion.

The first scanned sample was a soda lime glass plate. Measurement results, in form of two dimensional X-Z reflectivity map (B-scan) and one dimensional profile of one chosen pixel intensity (1D A-scan), are presented in figure 4.9. Both B-scan and A-scan have very well identified air-glass interfaces, however not without parasitic side-lobes. Side-lobes are the straight, parallel lines placed on both sides of the main peak, produced as a result of strong light reflection from the specimen surface, which interferes with reference optical field (see section 2.1.1). The entire B-scan contains irregular signal intensity variations representing either discontinuities in glass structure, or ghost artifacts [48, 49]. Using Hamming window in the frequency domain, helps narrowing the interface main peaks, reduces intensity of side-lobes and unknown artifacts, but with expense of decreasing whole signal amplitude by 7 – 8 dB, which decreased the image contrast. The speckle removal algorithm performed in ImageJ software greatly increases contrast of B-scans, but causes spreading of interface peaks. The signal to noise ratio of both raw, and post-processed data is approximately 65 dB and are close to the reach of the maximum of the camera capabilities. Front and back
interface intensity peaks are found in iterations 17 and 269, respectively. An iteration number is understood as a data point recognizing the optical distance in figures 4.9 and 4.10, and in fact, a particular en-face image within a 3D data volume. Hence, the experimental glass plate thickness equals to:

$$s_0 = \frac{(269 - 17) \times 10 \mu m}{n} = 1.673 \text{ mm}$$  \hspace{1cm} (4.1)

Another sample was a stack of two sapphire windows, 0.5 mm and 1 mm thick. Results are presented in the same way as previously, in figure 4.10. The higher reflectivity of sapphire is a cause of high intensity and broadness of the first peak, indicating the air-sapphire boundary. It is nearly twice greater in reflectivity, than the soda lime glass, and the phosphor nonlinear response results in saturation of intensity signal on the CCD surface. The remaining interfaces peaks are much weaker, but easily seen. Unidentified reflectivity variations occur, like in the case of the glass plate, but their intensity lowers as scanning progresses deeper into the sample. The SNR of raw data is 40 dB greater than in previous measurement, but only due to the saturated signal originates from the very first air-sapphire interface and should not be compared with glass plate scan in terms of quantity. The first boundary is indicated by the broad peak between 14 and 22 iteration (middle iteration is 18) in A-scan from figure 4.10. The remaining two peaks are iterations 103 and 278, respectively. Hence, the experimental thicknesses of both sapphire windows are:

$$s_1 = \frac{(103 - 18) \times 10 \mu m}{n} = 0.487 \text{ mm}$$  \hspace{1cm} (4.2)

$$s_2 = \frac{(278 - 103) \times 10 \mu m}{n} = 1.002 \text{ mm}$$  \hspace{1cm} (4.3)
Figure 4.9. Reflectivity B-scan and axial profile of soda lime glass plate. Upper set presents raw data in dB scale, middle one shows in dB scale the same data with Wiener smoothing and Hamming window applied. Bottom charts present raw data with despeckle algorithm applied in ImageJ and its profile shown in linear scale.
The thicknesses of sapphire windows were estimated with very good accuracy, but the glass plate thickness differs by 130 μm from the value obtained by using a 10 μm resolution calliper. Considering the 10 μm OCT scan step, the calculated thickness should be somewhere within $S \pm \frac{20 \mu m}{n}$. However, difference in plate measurements could be explained by possible glass thickness variation.

Both measurements confirm that this FF-OCT is capable of retrieving internal structure information of measured samples. In this case, low absorbing, low scattering, and with flat surfaces, soda lime glass and sapphire optical plates. All air-material interfaces were uncovered by FF-OCT, which confirms its ability of depth imaging. But system performance with dense, high scattering samples, such as teeth, is yet to be tested. The main issue is the detection threshold of the camera intensity response, that causes rejection of weakest intensity signals. This will likely prevent detection of light back-scattered from weakly reflecting interfaces. Furthermore, presented B-scans contain artifacts of unknown origin. Successive scanning of the same region of the glass plate reveal repeatability in occurrence of the artifacts pattern in certain areas, which suggest they might represent either variation of refractive index within glass, or interference side-lobes. The latter may explain the fact, that those artifacts occur beyond the sample, as seen on figure 4.9. In future measurements, this phenomenon might cover the true refractive index discontinuities within the samples, and make them difficult to identify.

All presented measurements were made by a system employing EDFA optical amplifier pumped by SLD diode. Analogous scans were performed for bare SLD with exactly the same optical setup, mode scrambling and light delivery system. The light spot entering the interferometer had relatively low power density, but, despite camera phosphor coating nonlinearity, it was possible to retrieve both air-glass boundaries in
Figure 4.10. Reflectivity B-scan and axial profile of a stack of 0.5 mm and 1 mm thick sapphire optical windows. Pictures on the top present raw data in dB scale, middle ones show in dB scale scan data with Wiener smoothing and Hamming windowing applied. Bottom charts present raw data with despeckle algorithm applied in ImageJ software and its profile shown in linear scale.
glass plate OCT scan. In sapphire windows measurement, only the first air-sapphire interface was revealed. Hence, the overall result was rather poor. However, neither of performed measurements employing the SLD as the only light source, recorded any unidentified artifacts. It may be due to low optical power distribution across the detector array and rejection of weak intensity signals by the camera phosphor coating. But it may as well mean that origin of those artifacts is different, like, for example, internal reflections within EDFA.

4.3. Principle of operation – profilometric measurements.

One of the applications of LCI is profilometry [10], which relies on precise estimation of the roughness of examined surfaces. LCI makes it possible to overcome the issue of phase ambiguities at discontinuities and steps. Profilometric measurement can be fully realized by an analog phase stepping technique which the presented FF-OCT is based on. Hence, the OCT system was tested by scanning the surface of a known object, a coin.

4.3.1. Experiment.

For comparison, profilometric measurements were performed with and without an AFC-EDFA optical amplifier. The OCT setup is as presented in figure 4.1. Mode scrambling and the light delivery system remain the same for both SLD (24.3 μm coherence length, ~4.5 mW optical power) and SLD-pumped EDFA (36 μm coherence length, ~20 mW optical power). S_MTS position was adjusted, so distance between
sample front surface and beam-splitter was 150 – 200 μm greater than length of the reference arm. Scan parameters are as follows:

- scan range 0.4 mm,
- scan step 10 μm (which gives total number of 41 iterations),
- number of averaged images, $N = 50$,
- dynamic focusing off.

The sample surface is an Irish Eurocent coin, made of copper-plated steel [130], and a Polish 10 groszy copper-nickel coin [131]. Measurement was focused on the strings of a harp on the obverse side of the Eurocent and on the beak of an eagle of 10 groszy coin (Fig. 4.11). All performed profilometric scan images were despeckled in ImageJ.

4.3.2. Results and discussion.

An obverse side of an Eurocent coin was chosen for profilometric measurement, due to the periodical pattern of harp strings that are easy to identify. Results of scans made with different light source configurations are shown in figure 4.12. The AFC-EDFA based system produced an image with well pronounced strings pattern and coin surface. But, strong reflection from the coin causes occurrence of
additional artifacts on both sides of the air-metal interface, as happened in the glass plate thickness test (section 4.2). They can be identified as side-lobes of interferometric signal or ghost images of the main interference peak. To get better insight of the artifacts pattern, a similar scan was performed, across the range of 1 mm, with coin surface positioned approximately in the middle (Fig. 4.13). The produced three dimensional image is difficult to read as artifacts spread across entire measurement range (z-axis). Cross sectional B-scan slices (Fig. 4.13 c, d, e) made at different positions (front, middle and back side of the 3D volume) show good approximation of the specimen surface and strings pattern. Again, artifacts exhibit side-lobes features,
Figure 4.13. Profilometric scan of harp’s string from an eurocent coin. (a), (b) show a 3D image, (c), (d), (e) present cross sectional slices from three different positions in the same orientation. Measurement performed with an EDFA based system.
as they appear on both sides of the air-metal interface. However, they are more visible
where back-reflected light intensity signal is the strongest, i.e. in the middle of the
scanning beam, and are vanishing on the sides of the 3D volume (Fig. 4.13 c, e).
Images from figure 4.13 b, d, e show artefact pattern spreading across entire x-axis,
not only above strings, and possibly forming a ghost image of the main interference peak, representing the coin surface.

Analogous profilometric scan of a harp strings made by the SLD light source based OCT is presented in figure 4.12. Due to lesser optical power density, smaller area is covered by the detectable illumination. The string pattern is very well pronounced, but other parts of coin surface were not detected, probably due to its worse flatness. No additional artifacts are present on the image.

Figure 4.14 shows profilometric scan results of an eagle’s beak from a 10 groszy coin, 3D images and Y-Z cross sectional image. Shape of a beak is very well resolvable, but the signal intensity is close to noise level. Cross sectional image (Fig. 4.14 c) indicates the beak sides, but not its surface. Relying only on B-scans instead of semi-transparent 3D images would make it very difficult to identify the measured shape. Measurement of a beak, using AFC-EDFA based OCT, yielded flat-like surface with irregular artifacts on both sides of air-metal interface. It means, this large value axial resolution OCT is not well suited to resolve rounded shapes.

4.4. Conclusions.

To summarize the information gathered about assembled FF-OCT system, all important results, observations, system merits, demerits and author opinions are pointed out.

- A fully functional, cost-effective, free-space optics full-field OCT operating in 1550 nm range is presented. Axial and lateral resolution are 36.7 μm and 12.5 μm, respectively. The system dynamic range is limited by phosphor coated CCD camera, used in the project, and is equal to approximately 60 dB.
• The camera operates in software event driven environment, which considerably limits system speed of operation, due to unreliable data acquisition, and requires using simplified phase-stepping algorithm to retrieve LCI signal envelope.

• Dynamic focusing is essential to perform depth scanning within samples if system is equipped with dry microscope objectives and operates in near-infrared region.

• Free-space optics FF-OCT is very sensitive to external vibrations occurring when OCT is operating. Vibration cause phase changes to propagating beams and distort acquired en-face images by introducing a residual fringe pattern, as depicted in figure 4.15. Despite using optical table with active isolator leg kit, external vibrations are the issue. Movement of particles in air has some influence as well, hence the interferometer is always covered when operating.

• SLD optical power is not sufficient for this particular OCT setup. The camera phosphor coating is characterized by nonlinear intensity and wavelength response, which rejects great amount of useful illumination. The glass plate thickness experiment and profilometric measurements were able to record weak back-reflected signals, but it is assumed that for high scattering samples SLD output power will not be enough. Hence, SLD was substituted by the AFC-EDFA light source and SLD was used as a pumping source for EDFA.

• Depth images obtained by FF-OCT employing the AFC-EDFA light source, experience distortion from artifacts of unknown origin. Artifacts exhibit features of interference side-lobes, as they appear on both sides of boundary between two mediums, but they tend to be located irregularly. This phenomena is not observed when SLD is the only source of light for OCT, so the origin of
those distortions lies within EDFA. This is most likely due to internal reflections of the signal within the amplifier or between the amplifier and other optical elements delivering light to the OCT system. Reflections cause creation of time-delayed signals that interfere with original signal. This so-called multipath interference is a known source of a noise that is especially visible in high-SNR systems (50+ dB) employing EDFA amplifiers [141, 142]. It is suspected that ghost images of revealed interfaces contribute in the creation of image distorting artifacts as well, as OCT B-scan slices sometimes contain shapes similar to the interface peak. However, it is likely that all of mentioned factors influence the image quality.

- Profilometric applications of the presented FF-OCT is very limited due to low axial resolution and artifacts distorting acquired imaged. A system with SLD as the only light source has better axial resolution and is able to retrieve shapes without artifacts, but has low optical power and might not detect non-flat surfaces.

Figure 4.15. Image of a reflecting, flat surface distorted by a residual fringe pattern arisen due to external vibrations occurring during measurement.
5. 1550 nm FF-OCT in teeth imaging.

More than a decade ago, OCT was recognized as a technology useful in imaging hard and soft tissues in the oral cavity [132]. With resolution advantage over other available imaging techniques, OCT enabled acquiring recordings of microstructural details that had not been possible before, like periodontal tissue contour, connective tissue attachment, etc. [133]. Its ability to in vivo imaging both hard and soft tissues of dental cavity enables early detection and intervention with dental and periodontal diseases. What is more, an OCT modality, a polarization sensitive OCT (PS-OCT) is able to record not only magnitude of light backscattered from the sample, but can reveal polarization properties of tissues, that are difficult or impossible to obtain by using other diagnostic methods [134-136]. This includes i.e. strong birefringence of enamel and irregular dentin structure. The extensive review of current usage of OCT in dentistry, views on its future potential in this field and discussion of other diagnostic methods used in dentistry can be found in [137].

In chapter 4, the FF-OCT system was proved fit for resolving discontinuities of refractive index inside the sample. However, instead of glass plates and optical windows, it is necessary to test how the system will perform with high scattering, biological samples. Teeth are good test subjects, as they contain little water [137, 138] which dominates absorption above 1400 nm region (> 10 cm\(^{-1}\), see figures 2.20 and 5.2).

This chapter will start with outlining the structure and optical properties of two outer teeth tissues, enamel and dentin. Then, goals and expectations regarding the tested FF-OCT system are presented. Finally, based on acquired images of tooth samples, the system capability of retrieving internal, structural information of dense,
biological samples will be evaluated. This discussion will be supported by scans of the same samples performed by a commercially available swept-source OCT system.

5.1. Optical properties of teeth tissues.

The bulk of a tooth is made of a calcified tissue, dentin. The visible part of the tooth, the crown, is covered and protected by highly mineralized and hard enamel substance. Its thickness varies over the tooth surface, up to 2.5 – 3 mm at the crown’s top and is the thinnest at the cervical margin. Very high brittleness of enamel requires less mineralized and less brittle dentin as a support material. The boundary between both, called dentinoenamel junction, is a mix of both tissues. Pulp, placed in pulp cavity and surrounded by dentin, is embryologically, histologically and functionally very similar to dentin, but contain nerves and blood vessels [137]. A schematic sketch of a tooth is shown in figure 5.1.

In this work, tooth imaging is performed in enamel and dentin, as their repair and maintenance is the major concern for dentistry. Optical properties of these tissues are outlined further in this chapter.

Figure 5.1. Cross sectional sketch of a tooth [48].
5.1.1. Enamel analysis.

Enamel is the hardest tissue in human body, consisting of 95% mineral, 4% organic material and 1% water by weight \([137]\) (8-12% by volume \([138]\)) and covers the crown of a tooth with a thin layer. Its structure is characterized by strong birefringence and is mostly formed by ordered array of rods oriented normally to the tooth surface. Rods are built of clustered calcium salt crystals and have 4-6 μm of lateral dimensions and are roughly 10 μm long \([134, 140]\). Space between rods is approximately 2% of enamel volume and is filled with proteins, water and lipids.

Enamel penetration by light is limited by scattering. Due to its complex structure, its distributions in enamel are mostly anisotropic and depend on tissue orientation relative to incident light beam direction \([140]\). The general relation between light scattering in dental enamel and wavelength is estimated as \(\lambda^{-3}\) \([137]\) and can be modelled by a combination of Mie and Rayleigh scattering \([140]\). Hence, the transmission of optical field from visible range is especially limited. For wavelengths exceeding 1400 – 1500 nm, Rayleigh scattering is negligible and absorption by water is the limiting factor for light attenuation, as shown in figure 5.2. In \([138]\) the authors, using measurement results taken from various sections of the sound enamel sample and a Beer-Lambert plot (Fig. 5.3), estimated a mean attenuation coefficient value in enamel for 1550 nm to be \(3.8 \pm 0.17\) cm\(^{-1}\). However, without employing refractive index matching, light attenuation is dominated by surface scattering, which is greater by an order of magnitude. According to this measurement, the mean free path an average photon travels before getting scattered or absorbed is \(\sim 3\) mm for 1550 nm.

Authors of \([139]\) estimated enamel refractive index be \(n_e = 1.631 \pm 0.007\). An all fiber 1310 nm OCT was used in that work, due to light maximum penetration in enamel at 1300 nm wavelength region.
Figure 5.2. Attenuation coefficient of sound enamel and water absorption vs. wavelength [138].

Figure 5.3. Beer-Lambert plot at 1550 nm through refractive index-matched enamel samples for various sample thickness and multiple measurements per point [138].
5.1.2. Dentin analysis.

The bulk of a tooth is a hard and elastic dentin tissue. Similar to bones, it consists, by weight, of approximately 70% inorganic material (50% of volume) – incorporated into hydroxyapatite crystals, 20% organic material (30% of volume) – such as collagen and fibrils, and 10% water (20% of volume) [137, 140]. The dentin structure is built of closely adjacent dentinal tubules extending entire thickness of dentin, from the pulp cavity to the periphery, with an irregular S-shaped curve [137]. Tubules diameter is the largest in the most inner part of the tooth, approximately 3 – 4 μm near the pulp, and narrows getting closer to the dentinoenamel junction. Hence, density of tubules within tissue varies and is the greatest near the pulp [134].

Despite containing birefringent fibrils, dentin exhibits no birefringence properties, due to its irregular structure, however anisotropic light propagation through dentinal tubules is observed [134]. Light attenuation in dentin is by far dominated by scattering and its value for 543 nm, 632 nm, 1053 nm was roughly constant and equalled approximately 280 ± 84 cm\(^{-1}\) [134]. There are no such data available for 1550 nm region, but assuming this tendency stays for longer wavelengths, it would be much more than light absorption by water (Fig. 5.2). However, as in case of enamel, if the dentin sample is not index matched, surface scattering and internal interfaces reflections dominate total attenuation in dentin tissue.

Authors of [139] estimated mean dentin refractive index be \(n_d = 1.540 \pm 0.013\). An all fiber 1310 nm OCT was used in that work, due to light maximum penetration in enamel at 1300 nm wavelength region.
5.1.3. Near-infrared teeth imaging – expectations.

According to Schmitt and other researchers, the magnitude of light scattering and absorption within dense, biological tissues is inversely proportional to wavelength of light used in imaging system [60, 102]. Hence, using infrared in OCT has become a mean to enhance imaging penetration depth within biological samples. First confirmation of this theory was done in [102], where authors demonstrated better probing depth in highly scattering tissue for 1300 µm wavelength compared to 800 nm, used in the same optical setup. Successful approaches to use FF-OCT in the near-infrared region were made in [100, 101] and the authors confirmed an increasing in their systems’ probing depth.

The 1300 nm wavelength range has been successfully employed in OCT a number of times [91, 100, 102, 143], but light from telecom bands, like 1550 nm range, has been far less popular due to greater absorption by water, that dominates absorption in tissues, for this band. Dental tissues contain significantly less water than other biological tissues (by volume: enamel 8-12%, dentin ~20%), so using 1550 nm OCT in dentistry might be justified. Authors of [138] calculated the mean total attenuation coefficient of 1550 nm light in sound enamel to be ~3.8 cm\(^{-1}\) (compared to 3.1 cm\(^{-1}\) for 1310 nm wavelength, see section 5.1.1). It gives the mean free path equal to approximately 3 mm for 1550 nm.

The mean free path value along with high optical power available from an AFC-EDFA light source, give hope to reach enamel-dentin interfaces 1-2 mm beneath the sample surface. On the other hand, high surface scattering and quality of the anti-Stokes camera detection capabilities put that goal into question.
5.2. Teeth imaging.

In this work, the main goals of teeth imaging are to obtain images of the samples’ internal microstructure, reveal enamel-dentin interfaces and perhaps discover other interfaces or material discontinuities inside samples. Identification of carious lesions and other dental diseases, investigation of various states of enamel (hypercalcified, stained, etc.) to perform comparison with a sound tissue are not subjects of this project. Demineralization in tooth structure alters its refractive index, so care has been put to select sound samples, or at least image only sound-looking parts of teeth [139].

This subchapter will continue with describing the imaging methodology followed by the presentation and interpretation of the acquired data. The performance of the tested OCT setup will be compared to a commercially available 1310 nm swept-source system.

5.2.1. Experiment.

To perform stable imaging conditions while operating a free-space FF-OCT system, tooth samples were sat in a plastic discs. In a dedicated apparatus, samples were assailed with plastic powder and isolated from air. Then, under increased pressure and temperature conditions, powder melted, embraced the sample and after cooling formed a solid-state disc. The front facets of the discs were polished to obtain flat surfaces and to uncover enamel tissue. A flat specimen surface is necessary to minimize undesired reflections of incident light and maximize its injection inside sample. Samples could be placed in the apparatus in various orientations, so different
parts of teeth are revealed during polishing. Created discs (examples in figure 5.4) are solid-state objects ready to attach to the OCT system.

After the preparation, certain areas of tooth crown samples were investigated using 1550 nm FF-OCT system. For comparison, the same areas of samples were imaged using a commercially available, ThorLabs 1310 nm, swept-source OCT (depth resolution 6 µm, lateral resolution 16 µm). Availability of other, proven OCT, allows better evaluation of acquired images of highly scattering samples and helps recognizing interfaces detected by the 1550 nm FF-OCT. A direct comparison of functional parameters of both OCT systems is not a purpose of this work. For simplification, in this chapter from now on, the tested 1550 nm full-field OCT and the commercial 1310 nm swept-source OCT systems will be referred to as FF-OCT and SS-OCT, respectively.

The architecture of SS-OCT enables producing images of large volumes. For this work, the scanning area was 2.5 x 5 mm and 3 mm in depth direction. To obtain images of such a large area using FF-OCT, a number of adjacent scans have to be recorded and merged using MatLab or other mathematical application.

Figure 5.4. Example of tooth samples sat in plastic discs and polished.
5.2.2. Tooth sample no.1.

5.2.2.1. Sample description and system settings.

As seen in figure 5.5, sample no.1 is a molar tooth with its top of the crown cut and polished to approximately 1 µm roughness. The marked area roughly indicates the examined area of the sample. It was chosen to scan side of the tooth, due to structural aspects of dental enamel. As explained in section 5.1.1, this tissue is mostly formed by ordered array of rods oriented normally to the tooth surface. Light scattering distributions in enamel are mostly anisotropic and depend on tissue orientation relative to incident light beam direction [140]. To investigate the latter fact, the sample is scanned from its side towards the middle. This allows FF-OCT and SS-OCT scans to cover rods creating enamel structure to be oriented in parallel, slantwise and orthogonally to the incident light beam.

![Figure 5.5. Tooth sample no.1. (a) presents a sketch showing how the sample was prepared for examination. (b) shows front facet of the sample disc with scanned area marked red.](image-url)
The reference scan produced by SS-OCT is 5 x 2.5 x 3 mm (X, Y, Z – depth) large, while FF-OCT scan is 1.83 x 0.68 x 2.40 mm. A-scans in FF-OCT were performed with 10 µm steps, while SS-OCT has a fixed number of pixels per A-scan (512). The acquired data is shown in next subchapter.

5.2.2.2. Data visualization.

Here, visualization of sample no.1 internal microstructure made by both systems are presented as 8-bit grey scale images. The FF-OCT setup scanned a much smaller volume, but the FF-OCT scanning area is a part of the larger, reference scan, acquired by SS-OCT. Figure 5.6 presents volumetric images of the sample from both systems. Figures 5.7, 5.8, 5.9 show corresponding B-scans, made by both systems, separated from each other by approximately 100 µm.
Figure 5.6. Volumetric image of a marked area in tooth sample no.1. Upper image was made by SS-OCT (dimensions are 5 x 2.5 x 3 mm). Below, image made by FF-OCT (1.83 x 0.66 x 2.40 mm). Volume imaged by FF-OCT is contained within volume scanned by SS-OCT.
Figure 5.7. Corresponding B-scans of tooth sample no.1 made by both SS-OCT (top image) and FF-OCT (bottom image). The red frame indicates corresponding region shown in both images.
Figure 5.8. Corresponding B-scans of tooth sample no.1 made by both SS-OCT (top image) and FF-OCT (bottom image). The red frame indicates corresponding region shown in both images.
Figure 5.9. Corresponding B-scans of tooth sample no.1 made by both SS-OCT (top image) and FF-OCT (bottom image). The red frame indicates corresponding region shown in both images.
5.2.2.3. Observations.

There is a significant difference in quality between images produced by the SS-OCT and tested FF-OCT setups. Axial resolution of SS-OCT equal to 6 µm, compared to over 30 µm resolution of FF-OCT using EDFA light source, is a major factor governing this gap in quality and readability of images. Volumetric images from figure 5.6 are the best representation of that differences. The 3D image from FF-OCT is blurred and any identified internal interfaces are fading in the background. All FF-OCT produced scans have a large, bright peak indicating the air-sample interface. It is a result of a flat surface of the specimen disc facet (~1 µm roughness) and quadrature light intensity response of anti-Stokes imaging device, employed in FF-OCT (see section 3.4 for more details). The latter fact means, high intensity signals detected by the anti-Stokes camera are very well recognizable, while low intensity ones are barely detectable. As seen on figures 5.7 – 5.9, SS-OCT scans have air-enamel and enamel-dentin interfaces sharp and well defined. Their corresponding FF-OCT images, have air-enamel boundary excessively pronounced, but enamel-dentin interface is barely visible and incomplete.

It is known, the enamel is made of organized rod-like structures perpendicular to the tooth surface. All B-scans from figures 5.7 – 5.9 are characterized by brighter left side, where rods are supposed to be positioned slantwise orientation to the incident light. This could suggest better reflectivity of enamel rods in that sort of orientation than rods parallel to the direction of the incident light, and confirm an anisotropic distribution of scattered light in enamel tissue.
5.2.3. Tooth sample no.2.

5.2.3.1. Sample description and system settings.

Sample no.2 is a molar tooth cut and polished from its side to approximately 10 µm (Fig. 5.10). The marked area roughly indicates the examined area of the sample. The author’s intention was to obtain a thin, wedge-like layer of enamel and to uncover dentin tissue for direct examination, as depicted in figure 5.10a. Unfortunately, during polishing the thinnest enamel layer had been ripped off by abrasive and further polishing had been stopped to avoid further damage to the sample. As it was the only tooth specimen sat in plastic in such orientation, it was decided to proceed with its use.

The 10 µm roughness of sample no.2 disc facet will cause greater surface scattering, comparing to sample no.1, and will limit amount of photons reaching specimen interior. Hence, the maximum depth of penetration of both OCT systems is

![Figure 5.10. Tooth sample no.2. (a) presents a sketch showing how sample was prepared for examination. (b) shows front facet of the sample disc with examined area marked red.](image)
expected to be smaller than in case of sample no.1. No visible anisotropy of back-reflected light is expected to be revealed on OCT images, as enamel rods are more or less parallel to the direction of the incident light beam in sample no.2.

The reference scan produced by SS-OCT is 5 x 2.5 x 3 mm (X, Y, Z – depth) large, while FF-OCT scan is 2.5 x 0.68 x 3 mm. A-scans in FF-OCT were performed with 10 µm steps, while SS-OCT has a fixed number of pixels per A-scan (512). The acquired data is shown in next subchapter.

5.2.3.2. Data visualization.

Here, visualization of sample no.2 internal microstructure made by both systems are presented as 8-bit grey scale images. FF-OCT setup scanned much smaller volume, but FF-OCT scanning area is a part of the larger, reference scan, acquired by SS-OCT. Sample reflectivity data acquired by the latter system is shown in figure 5.11 as a partially transparent, volumetric image. Figures 5.12, 5.13, 5.14 show corresponding B-scans, made by both systems, separated from each other by approximately 100 µm.
Figure 5.11. Volumetric image of a marked area in tooth sample no.2 made by SS-OCT (dimensions are 5 x 2.5 x 3 mm). Volume imaged by FF-OCT is contained within volume scanned by SS-OCT.
Figure 5.12. Corresponding B-scans of tooth sample no.2 made by both SS-OCT (top image) and FF-OCT (bottom image). The red frame indicates corresponding region shown in both images.
Figure 5.13. Corresponding B-scans of tooth sample no.2 made by both SS-OCT (top image) and FF-OCT (bottom image). The red frame indicates corresponding region shown in both images.
Figure 5.14. Corresponding B-scans of tooth sample no.2 made by both SS-OCT (top image) and FF-OCT (bottom image). The red frame indicates corresponding region shown in both images.
5.2.3.3. Observations.

A significant difference in quality between images acquired by both OCT systems remained. Figure 5.11 contains a half transparent, volumetric presentation of reference data recorded by SS-OCT. A hole created in the disc facet by accidental removal of a thin enamel layer is visible. Volumetric image from FF-OCT setup is not attached, because interfaces discovered within specimen are fading away in the rest of the image background, like in 3D image of sample no.1 shown in figure 5.6. Hence, only B-scans (X-Z planes) are presented (Fig. 5.12, 5.13, 5.14).

Greater roughness of the sample no.2 disc surface is noticeable mainly in SS-OCT made images. Air-sample boundary peaks are wider comparing to sample no.1 scans. Sample penetration by both systems is much shallower due to greater surface roughness and because of presence of more than one discovered interfaces in the same X-Y coordinates within the specimen.

FF-OCT made images are characterized by thick peaks recognized as an air-enamel boundary, blurred and weakly pronounced internal discontinuities of refractive indexes and very weak detection of air-dentin interface (far left side of images from figures 5.12 – 5.14). Dentin tissue had been uncovered while polishing and accidental removal of thin enamel layer and its surface is probably rougher than 10 µm and highly irregular. Despite greater surface scattering and greater light absorption by dentin, due to absorption by water, it would be expected to obtain a sharper image of this particular interface. The enamel-dentin internal interfaces are more pronounced, despite smaller difference in refractive indexes between both tissues. Refractive indexes of enamel and dentin are $n_e = 1.631 \pm 0.007$ and $n_d = 1.540 \pm 0.013$ respectively and their difference is equal to ~0.091 (compared to 0.54 difference between refractive indexes of dentin and air).
Images from FF-OCT still suffer from the presence of high intensity artifacts, described in chapter 4.4. They are especially visible in figure 5.13 on both sides of the air-enamel peak and have destructive impact on image contrast and visibility of sample internal structure.

As indicated by red frames, the reference data do not entirely match FF-OCT made scans. However, it was possible to find corresponding areas, due to presence of characteristic internal interfaces and a hole in the sample no.2 disc facet.

5.3. Suitability for dental imaging – advantages and limitations.

This subchapter will summarize testing assembled FF-OCT system in terms of dental imaging. To have full insight in the subject, teeth samples should be examined without additional polishing. However, basing on already conducted work, several key advantages and limitation will be pointed out.

- High optical power from AFC-EDFA light source in 1550 nm wavelength band offers probing depth up to 2 mm in enamel. On the other hand, AFC-EDFA is the likely source for parasitic artifacts distorting OCT images.

- 1550 nm wavelength band and bandwidth of this particular EDFA light source governs axial resolution to be 36.7 µm. It is much worse than commercially available OCT systems.

- A nonlinear light intensity response of the system detector, the anti-Stokes imaging device, is a reason of aberrations in produced OCT images. High reflection signals, like air-sample boundaries, are excessively pronounced. Low intensity signals, mostly originated from photons reflections from interfaces within sample, are barely detectable, however visible.
• In case of sample no.2, FF-OCT setup barely records an air-dentin interface. It is most likely due to issues in recording rough surfaces that are steep, which means not perpendicular to the direction of the incident light.

• All above points contribute to images contrast, which is worse comparing to the commercial ThorLabs SS-OCT system. Some recognized discontinuities of refractive indexes of examined samples tend to fade away in background due to parasitic artifacts.

• The full-field OCT images are created using phase stepping algorithm and images averaging for quality improvement. This approach requires a fast detector array to minimize data acquisition time. With anti-Stokes imaging device operating up to 100 Hz (assuming no data is lost during operation – see section 3.4 for details), recording 300 en-face images with $N=100$ frames averaged lasts more than ten minutes. For comparison, recording a $2.5 \times 2.5 \times 3$ mm volume with SS-OCT lasts approximately two minutes.
6. Discussion and conclusion.

The aim of this research was to build a functional, cost effective full-field OCT system operating in 1550 nm wavelength band. The two main components chosen to be parts of this setup were: a 1550 nm superluminescent diode as a light source, and the anti-Stokes imaging device as a detector. The SLD output power was found insufficient for full-field imaging and was replaced by a combination of SLD and EDFA amplifier operating in similar wavelength band.

This section will summarize all previous chapters in terms of presentation and evaluation of assembled OCT system properties and its capabilities of imaging dental tissues. Eventually, an analysis of possible improvements to the system will be outlined.


A comprehensive characterization of the FF-OCT setup is presented in chapters 4 and 5. Chapter 4 covers detailed description of the system, presentation of its functional parameters and tests, like glass plate thickness and profilometric measurements. Chapter 5 takes on examining dense, highly scattering dental tissues.

The main functional parameters are outlined in Table 7. The AFC-EDFA light source offers trade-off between moderate axial resolution high optical power. Lateral resolution can be adjusted using proper imaging lenses. However, anti-Stokes camera array has spatial resolution equal to 50 µm and resolution adjustments brings another trade-off. Better resolution and larger region of interest in counter to optical power density across detector area and limited probing depth of the system. FF-OCT dynamic
range is limited only by camera detection capabilities. Data acquisition speed is dependent on camera frame rate. Theoretical operation speed of the camera is approximately 100 Hz, assuming there are no discarded data. More insight into this subject can be found in chapter 3.

Capability of successful retrieving data about samples internal microstructure has been confirmed as well. Figure 6.1 contains example images presenting stack of two sapphire windows and a scan of a human tooth with enamel-dentin interface revealed.

![Figure 6.1](image)

Figure 6.1. Example B-scans acquired by the FF-OCT system. (a) is an image of a stack of two sapphire optical windows. (b) is an image of a crown of a sound, human tooth.
Table 7. Main functional parameters of the proposed FF-OCT.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Axial resolution</td>
<td>36.7 µm</td>
</tr>
<tr>
<td>Lateral resolution</td>
<td>12.5 µm</td>
</tr>
<tr>
<td>Dynamic range</td>
<td>~61 dB</td>
</tr>
<tr>
<td>Source optical power</td>
<td>20.85 mW</td>
</tr>
<tr>
<td>Center wavelength</td>
<td>1584.1 nm</td>
</tr>
<tr>
<td>Camera frame rate</td>
<td>varying, up to 100 Hz</td>
</tr>
</tbody>
</table>

6.2. System limitations.

The component that is the limiting factor in many key fields in this work is the anti-Stokes imaging device. This phosphor coated CCD array was designed as a cost-effective, but comprehensive, measurement system for optical fibers output evaluation in telecommunication. When this particular research started, there were not many options among cameras working in 1550 nm band, hence first attempts to incorporate this device in interferometry appeared [122]. As a part of the FF-OCT system, the camera performance is poor for several reasons.

- Nonlinear wavelength response of the phosphor coating is a source of losses of useful illumination.
- Quadrature light intensity response of the phosphor coating is a reason why OCT images, which are light reflectivity maps, cannot be properly evaluated. High reflection signals are excessively pronounced and low intensity signals hardly detectable.
• Phosphor coating causes light diffusion onto CCD array and limits camera spatial resolution to 50 µm. It is not possible to match OCT lateral resolution to CCD pixels size (CCD pixels are 7.4 x 7.4 µm large).

• Camera operates in software event-driven environment and number of collected frames within a unit of time varies, so perfect synchronization with other devices is impossible. This forces using simplified phase-stepping algorithm instead of much faster continuous phase modulation algorithms [73, 74].

• Theoretical maximum frame rate of the anti-Stokes camera is 100 Hz. FF-OCT requires two or more phase shifted images to compose an OCT image and image averaging or summing to improve system dynamic range. Hence much faster cameras are desired in FF-OCT applications.

Axial resolution of presented FF-OCT setup is at least row of magnitude worse than modern high resolution OCT systems. Axial resolution value is proportional to the light source central wavelength $\lambda_C$ squared and is inversely proportional to source FWHM. For example, to reach 2 micron axial resolution with $\lambda_C = 1550$ nm, light source bandwidth should be wider than 500 nm. It is difficult to obtain high axial resolution OCT when operating in NIR band, comparing to visible light region.

6.3. System improvement analysis.

Any improvement of existing FF-OCT requires replacing one or two key components of this setup, light source and detector array. More suitable detector would be cameras employing InGaAs arrays operating in NIR region. They offer high quantum efficiency (>80%) between 900 nm and 1700 nm, decent spatial resolution
(12.5 – 25 µm), dynamic range up to 80-90 dB and linear wavelength and intensity response [144, 145]. Frame rates of those detectors usually do not exceed 60-100 Hz [145], so cost-efficient, real time imaging with full-field OCT would still be impossible. The fastest reported InGaAs detector has 346 Hz [154], however in this case, the price becomes an issue. InGaAs cameras have successfully been used in the past in full-field OCT imaging, especially in 1200 and 1300 nm range [91, 92, 101, 143]. On the other hand, if the anti-Stokes camera device is to be utilised furtherly, it is possible to use microscope objectives with smaller NA, than 0.25, without degradation of effective lateral resolution. This would reduce optical aberrations resulting from microscope objectives and improve transmission of light. Furthermore, the depth of focus would be larger, so dynamic focusing adjustments could have better tolerance or even be entirely avoided.

In case of a light source, there are wide selection of SLDs and LEDs available in 1550 nm region [7, 107]. To obtain high optical power and high axial resolution OCT, it is possible to couple several emitters with adjacent bandwidths into one broadband source [53, 109]. The depth resolution could be furtherly improved either by shaping the source effective spectrum, like authors of [148] performed on the combination of LEDs, or by adjusting the spectral response of the image sensor [149].

Authors of [150] presented a FF-OCT system employing a Mirau interferometer, instead of Linnik interferometer. As Mirau interferometer has only one objective and the almost-common path layout, it is less sensitive to environmental disturbances and easier to assemble and adjust, compared to Linnik configuration. Although the system presented in [150] is water immersed and is operating in visible light range, it is believed adapting it into infrared region could be possible by using glass plates suitable for NIR illumination. Then the immersing medium could be
omitted and dynamic focusing feature implemented instead, exactly as shown in section 2.3.5.2.

A very interesting idea would be extending existing setup and assemble a swept-source full-field OCT (SS-FF-OCT) operating in 1550 nm range to improve dynamic range and data acquisition time [151, 152]. Transition into the frequency domain would simplify the optical setup, as no axial scanning would be necessary. Employing SS technology into en-face scanning, would require destroying the lateral spatial coherence of the source to reduce cross-talk. The light sources used in the SS-FF-OCT systems in the past were SLDs with an acousto-optic tunable filter (AOTF) as frequency tuning device [98, 99, 147], a tunable laser [96, 146] or a dedicated tunable light source [97]. Authors of [153] presented a comparison of state-of-the-art swept source technology, including devices operating in 1550 nm wavelength range. Those include:

- Insight akinetic swept source – up to 13 mW optical power, up to 80 nm sweep width and 200 kHz frequency sweep speed at 1550 nm region,
- Exalos external cavity laser with resonant MEMS-based one-dimensional scanning mirror – 20 mW optical power, 120 nm sweep width and 150 kHz frequency sweep speed at 1550 nm region,
- FDML laser – more than 60 mW optical power, 115 nm sweep width, 115 kHz frequency sweep speed at 1550 nm region.

As modern NIR light sources emerge, there is still a need of cost-effective, high speed array detectors operating in NIR. Possibly, with future technological development, new detection cameras will make the 1550 nm FF-OCT a viable option for imaging in the future, both in time domain and in frequency domain.
Glossary

AOTF – Acusto-Optic Tunable Filter
BS – Beam Splitter
CCD – Charge Coupled Device
CMOS – Complementary Metal-Oxide Semiconductor
EDFA – Erbium-Doped Fiber Amplifier
DOF – Depth Of Focus
FD-OCT – Fourier Domain Optical Coherence Tomography
FF-OCM - Full-Field Optical Coherence Microscopy
FF-OCT – Full Field Optical Coherence Tomography
FOV – Field Of View
FT – Fourier Transform
FWHM – Full Width at Half Maximum
HT – Hilbert Transform
InGaAs – Indium Gallium Arsenide
LCI – Low Coherence Interferometry
LED – Light Emitting Diode
MMF – Multimode Optical Fiber
MO – Microscope Objective
MQW – Multiple Quantum Well
MTS – Motorized Translation Stage
NA – Numerical Aperture
NIR – Near Infrared
OCDR - Optical Coherence-Domain Reflectometry
OCM – Optical Coherence Microscopy
OCT – Optical Coherence Tomography
OPD – Optical Path Difference
GLOSSARY

PMT – Photomultiplier Tube
PSF – Point Spread Function
PSI – Phase Shifting Interferometry
PS-FF-OCT – Polarization Sensitive Full Field Optical Coherence Tomography
PS-OCT – Polarization Sensitive Optical Coherence Tomography
PZT – Piezoelectric Transducer
QW – Quantum Well
RIN – Relative Intensity Noise
SD-OCT – Spectrum Domain Optical Coherence Tomography
SLD – Superluminescent Diode
SMF – Singlemode Optical Fiber
SNR – Signal to Noise Ratio
SS-FF-OCT – Swept Source Full Field Optical Coherence Tomography
SS-OCT – Swept Source Optical Coherence Tomography
TD-OCT – Time Domain Optical Coherence Tomography
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BIBLIOGRAPHY


Appendix A – Publications

The list of published work related to the described project is consisted of two international SPIE conference proceedings. Both publications are presented.
1550 nm Superluminescent Diode and Anti-Stokes Effect CCD camera based Optical Coherence Tomography for Full-Field Optical Metrology

Lukasz Kreuzinski and Michael J. Connelly

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ABSTRACT

Optical Coherence Tomography (OCT) is a promising non-invasive imaging technology capable of carrying out 3D high-resolution cross-sectional images of the internal microstructure of examined material. However, almost all of these systems are expensive, requiring the use of complex optical setups, expensive light sources and complicated scanning of the sample under test. In addition most of these systems have not taken advantage of the competitively priced optical components available at wavelength within the main optical communications band located in the 1550 nm region. A comparatively simple and inexpensive full-field OCT system (FF-OCT), based on a superluminescent diode (SLD) light source and anti-stokes imaging device was constructed, to perform 3D cross-sectional imaging. This kind of inexpensive setup with moderate resolution could be easily applicable in low-level biomedical and industrial diagnostics. This paper involves calibration of the system and determines its suitability for imaging structures of biological tissues such as teeth, which has low absorption at 1550 nm.

Keywords: full field optical coherence tomography, Anti-Stokes imaging device, biomedical diagnostics, dentistry

1. THE OCT SYSTEM

1.1. Experimental setup

Full-field OCT is a type of OCT that enables the capture of two-dimensional en-face images in a single exposure. Depth scanning is facilitated by scanning the reference mirror or moving the sample axially in the fasion of conventional time domain system (TD-OCT)[4]. The lateral resolution is limited by the scanning optics and camera pixel size. The axial scan resolution is limited by the source bandwidth. The schematic of our full-field OCT system is shown in Fig. 1. It is based on a free-space Michelson interferometer with low numerical aperture (0.2 NA) focusing lenses in both arms. The light source is a superluminescent diode (Superlum SLD-761, centre wavelength $\lambda_c = 1563.5$ nm, optical power $P = 4.5$ mW) with a roughly Gaussian spectrum with a 44 nm bandwidth, leading to an OCT resolution of approximately 24 $\mu$m. The interferometer output is projected onto a phosphor coated 2D CCD array camera (Ophir-Spiricon, model L070-1550). The phosphor coating is an anti-Stokes material that up-converts radiation in the 1460 nm to 1625 nm range to visible light measurable by the CCD. As shown in Fig. 2 the conversion efficiency of the camera is highly non-linear. As a result, some wavelengths are just severely suppressed by the camera phosphor coating. This leads to large loss of optical power striking on CCD array. The camera array size is 640 x 480 pixels, has an external trigger and allows image data to be processed in real-time by LabView and MatLab by the use of a USB
connection to a Personal Computer (PC). 2x2 binning is applied to increase the effective dynamic range of the camera.

The reference arm contains a neutral density filter to reduce the power of the reference signal in order to avoid camera saturation and improve the interferometer visibility. A translation stage (ThorLabs MTS50/M) is used to provide depth scanning. Its mirror is mounted on a piezoelectric stage actuator (PZT; PI model S-303), which is used to shift the reference beam phase four times (0, \(\lambda_c/4\), \(2\lambda_c/4\), and \(3\lambda_c/4\)) for every depth scan step, so four stacks of \(N\) camera images are acquired for each depth scan step, a technique similar to that described in previous papers \(^2\)-\(^4\). Increasing the number \(N\) of frames acquired to calculate every depth scan, improves visibility and contrast of the final image. The phase-shifting, depth scanning and image acquisition are controlled in real-time by the PC.

Fig. 1. Full-field OCT setup.

![Fig. 1. Full-field OCT setup.](image)

Fig. 2. Signal required versus wavelength to achieve camera full signal illumination by Anti-Stokes up conversion material \(^6\).

The grey area indicates the spectral range of the SLD.

1.2. Image restoration method
The signal $I_p(x, y)$ detected by each camera pixel, with coordinates $(x, y)$, corresponding to each phase shift can be expressed as:

$$I_{pn}(x, y) = I_0(x, y) + I_{coh}(x, y) \cos \left( \phi(x, y) + (p - 1) \frac{\pi}{4} \right), \quad p = 1, 2, 3, ..., n = 1, ..., N$$  \hspace{1cm} (1)

Where $I_0(x, y)$ denotes the average signal intensity and $\phi(x, y)$ denotes the optical phase. $I_{coh}(x, y)$ is the intensity of the coherent signal, which is proportional to the time-averaged cross-correlation of the sample and reference optical fields. For every phase shift step, a $N$ number of frames is collected and summed up.

$$I_p = \frac{1}{N} \sum_{n=1}^{N} I_{pn}(x, y)$$  \hspace{1cm} (2)

Standard number of frames used to build single phase step image is 30. The four equations in (2) can be combined to obtain $I_{coh}(x, y)$ as:

$$I_{coh} = [(I_1 - I_2)^2 + (I_4 - I_3)^2]/4$$  \hspace{1cm} (3)

This method allows reasonably faster data computations than Fourier or Hilbert transforms based algorithms. Another important role of the detection scheme is to eliminate the dc component $I_0(x, y)$. Full-field OCT systems using similar algorithms [5-8] rely on direct synchronization between the detector and PZT, which yields fast data acquisition speed but introduces additional modulation of the optical phase. This approach cannot be realized here, due to limitation in the camera design, the CCD camera frame rate can vary significantly even over timescales of a few seconds.

1.3. System calibration

As a first demonstration of the OCT system, it was used to detect the interfaces between two dielectric layers. The dielectrics are silicon and sapphire windows characterized by ~55% and ~15% reflectivity and 3.7 and 1.7 refractive indexes respectively at 1564 nm. Scans were made across 5 mm and 3.2 mm optical paths with a 10 µm depth scan step. The acquired data was post processed in MatLab. The experimental results are shown in Fig. 3. The bright lines in Fig. 3 are the detected window-window and air-window interfaces. The detected interface between the two windows is somewhat blurred. This is because in this region there are actually two reflections, one from each mirror boundary. Because the axial resolution is of the order of 10s of microns it is not possible to resolve these two surfaces precisely and measure the point spread function of the system. The air-window interfaces are indicated more precisely. Because the sample targets have smooth and very reflective surfaces the contrast is very close to 1.
Fig. 3. Top: 5 mm depth scan of stacked 0.5 mm thick sapphire and 1 mm thick silicon windows surrounded by air in 2D (left) and 3D (center) mode and single-point depth scan of highest detected intensity spot (right). The mirror-mirror interfaces (labelled) and mirror-air interfaces are clearly visible.

2. TISSUE IMAGING

The principle of this study was to acquire high-quality tomographic scans of a tooth using the described setup. A human tooth, as illustrated in Fig. 4, consists of a crown and a root. The junction between them is called the cervical margin. The tooth's crown is covered by enamel, which is an acellular, highly mineralized tissue. Enamel is an ordered array of inorganic crystals surrounded by a matrix made of protein, lipid, and water. The crystals are approximately 30–40 nm in diameter and can be as long as 10 μm. The crystals are clustered together in 4-μm-diameter prisms-like structures, which are roughly perpendicular to the tooth surface. The bulk of the tooth is made of dentin, which is a hard, elastic tissue that primarily consists of collagen, fibrils, and inorganic components and water (~18-20% by volume). The main dentin structural components are micrometer-sized tubules. The interior of the tooth is called the pulp, a soft connective tissue that is innervated and highly vascular. Due to the complex nature of these structures, the scattering distributions are anisotropic and depend on tissue orientation relative to the irradiating light source. In [10], the total extinction of sound enamel has been estimated as ~3.8 cm⁻¹ and it is dominated by absorption of water [10] (for enamel it is 2-12% by volume). 

This yields approximately 3 mm mean free path for a photon travelling in this tissue before getting absorbed or scattered. Optical imaging of dentin is limited by scattering. The major role in this process is played by tubules, which are the most important scatterers in dentin [11]. In [12], the averaged refractive indices of enamel and dentin for near-IR light have been estimated as 1.63±0.007 and 1.54±0.013, respectively. Light scattering is primarily determined by a combination of Mie and Rayleigh scattering and has been shown to be high in the visible range [12]. Since absorption in near-IR wavelength range is much lower, these wavelengths are more suitable for diagnostics in dentistry.
Tooth samples were set in a plastic disc and polished to 1μm roughness. The aim of the polishing was to achieve a flat enamel surface with an enamel-dentin interface a few 100s of microns below. Then, approximately same area of sample has been scanned using the experimental OCT and commercially available ThorLabs swept-source system. The experimental results are shown in Fig. 5. As shown in Fig. 6a, the experimental OCT system has poor penetration depth and the rather poor feature resolution and in particular it is difficult to resolve the enamel-dentin interface. Blurred and non-uniform air-enamel boundary signs poor depth resolution of the order of 10s of microns (theoretical 24 μm). A-scan or XY images were mostly uniform and deprived of information about the depth scan or air-enamel surface profile. Image contrast is also poor.

It was decided to pick a different type of sample, less absorbing and with different structure, so a piece of garlic skin sample was also used. This time it was difficult to retrieve air-sample interfaces, but some internal structure information was recorded and shown in Fig. 7.

3. CONCLUSION

We have proposed a simple and inexpensive FF-OCT system working in 1550 nm range. Its performance is currently poor and requires further improvements. Its penetration depth and resolution are limited. An Erbium-doped amplifier has been coupled with SLD to produce more optical power, but resulting 14 mW output power (SLD output power is 4.5 mW) did not increase the penetration depth significantly. Furthermore, the amplifier causes a narrowing of the source spectrum and therefore increasing the coherence length. It also brought some false targets into depth scans due to internal reflections inside the device cavity. Test results with the amplifier have not been included to this paper. A problem of nonlinear response of CCD camera phosphor coating has not been yet resolved. It causes severe power losses for longer wavelengths in the source spectrum. Surprisingly, it was not possible to retrieve any details about air-enamel surface or enamel internal profile. However it succeeded with examining a cellular sample, but lateral resolution did not allow recording much details and resulting A-scan is somewhat blurred.

Next step will be optimizing OCT current performance by means of improving lateral resolution and penetration depth. A first step will be to replace the scanning optics and sample and reference arm of the interferometer. The latter could be solved by extending the setup with polarization sensitive elements to improve system settings.
Fig. 6. Enamel-dentin interface XZ images obtained by experimental FF-OCT setup (a) and Thorlabs swept-source system (b). Images dimensions are 400 µm x 1 mm. Image (a) have been acquired by combining several en-face scans. Depth scan made with 10 microns steps. The scale is linear.

Fig. 7. – 50 x 200 µm A-scan of a garlic skin, approximately 50-60 µm below the surface. The image is a combination of 4 en-face scans. The scale represents back-reflected light intensity, it is linear.

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Anti-Stokes effect CCD camera and SLD based optical coherence tomography for full-field imaging in the 1550 nm region

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ABSTRACT

Full-field Optical coherence tomography is an en-face interferometric imaging technology capable of carrying out high resolution cross-sectional imaging of the internal microstructure of an examined specimen in a non-invasive manner. The presented system is based on competitively priced optical components available at the main optical communications band located in the 1550 nm region. It consists of a superluminescent diode and an anti-stokes imaging device. The single mode fibre coupled SLD was connected to a multi-mode fibre inserted into a mode scrambler to obtain spatially incoherent illumination, suitable for OCT wide-field modality in terms of cross-talk suppression and image enhancement. This relatively inexpensive system with moderate resolution of approximately 24μm x 12μm (axial x lateral) was constructed to perform a 3D cross sectional imaging of a human tooth. To our knowledge this is the first 1550 nm full-field OCT system reported.

Keywords: full-field optical coherence tomography, spatially incoherent illumination, infrared imaging, mode scrambling

1. INTRODUCTION

Full-field optical coherence tomography (FF-OCT) is an OCT modality that enables the capture of two-dimensional images in transverse orientation of reflected light within specimens in a single exposure. It is built of a Limnêk-type interferometer illuminated by broad bandwidth light source and an array-based detector, usually a CCD\cite{1} or InGaAs\cite{2} camera, to enable parallel detection and avoid the necessity of lateral scanning. Depth scan is performed by scanning the reference mirror or moving the sample parallel to the scanning lens optical axis in the fashion of conventional time-domain OCT. The great advantage of FF-OCT is decoupling lateral resolution from axial resolution. Transverse resolution is limited only by the microscope objectives NA, thus it is similar to that of conventional microscopy and can reach the order of ~1 μm. On the other hand, axial resolution is limited by source central wavelength value and its bandwidth and can vary from fraction of μm to few tens of μm.

FF-OCT usually uses thermal light sources with its extremely broad and smooth spectrum and low spatial coherence properties\cite{3,4}. Nevertheless, efforts have been made to try longer wavelengths for possible penetration depth extension in highly scattering media\cite{5}. We propose a FF-OCT that uses competitively priced optical components available at the main optical communications band located in the 1550 nm region, a superluminescent diode (SLD) and an anti-stokes imaging device. SLDs do not seem as an appropriate source for FF-OCT due to its high spatial coherence\cite{6}, which creates cross-talk effect in en-face images and therefore lateral resolution and signal-to-noise ratio (SNR) degradation\cite{7,8}. To overcome this issue, the single mode fibre
coupled SLD was connected to a large core diameter multi-mode fibre inserted into a mode scrambler to obtain spatially incoherent illumination. Imaging within tissues creates signal losses generally due to scattering and absorption\(^1\). Light scattering becomes weaker as the wavelength increases, but absorption of water, which dominates total absorption in tissues, increases\(^6, 9\). Also the coherence length of the source, which defines system axial resolution, is proportional to the source wavelength value squared. Hence the 1550 nm region is not useful for OCT examination of many biological tissues. However, it is applicable for examining low water content samples, such as teeth.

2. THE OCT SYSTEM

2.1. Experimental setup

The schematic of the FF-OCT system is shown in Fig. 1. It is based on a free-space Michelson interferometer with identical microscope objectives (10x, NA = 0.25, no immersion medium) in both arms. The light source is a superluminescent diode of centre wavelength \( \lambda_C = 1563.4 \) nm, output optical power \( P = 4.5 \) mW and an approximate Gaussian spectrum of 44 nm bandwidth, leading to an OCT axial resolution\(^1\) of approximately 24 \( \mu \)m in air. The interferometer output beam is projected onto an Anti-stokes device, which is a phosphor coated 2D array CCD camera (Ophir-Spiricon), used in full-field metrology for the first time in \(^{16}\). The phosphor coating is an Anti-stokes material that up-converts radiation in the 1460 nm to 1625 nm range to visible light measurable by the CCD array. Unfortunately, the phosphor coating conversion efficiency is highly non-linear, as shown in Fig. 2. This leads to a reduction in the useful SLD illumination, because for some longer wavelengths that part in building SLD spectrum, a phosphor coating response is an order of magnitude different than for other wavelengths.

The camera array size is 640x480 pixels with a 12 bit resolution. A 2x2 binning is applied to increase the effective dynamic range of the camera. A motorized translation stage provides depth scanning of specimen under test. Another translation stage provides dynamic focusing\(^2\) adjustments of the microscope objective position in the sample arm. It is necessary to keep the objective focal plane coinciding with coherence plane within the sample. Otherwise interferometric fringes may become blurred, which leads to possible reduction of final image contrast, and eventually vanish, when both planes recede from each other too much\(^5\). The reference mirror is mounted on a piezoelectric stage actuator (PZT), which introduces discrete phase-shifts in the interferometer by mechanical displacement of the reference mirror. The purpose and operation of a multimode fibre and a mode scrambler will be explained in next sub-chapter. Lateral resolution of the system was estimated to be approximately 12 \( \mu \)m using a USAF 1951 test chart. There is a possibility of improving that feature by replacing L2 lens with a longer focal length lens. The calculated theoretical transverse resolution is 4.5\( \mu \)m. The image acquisitions, calculations and display are realized by means of customized LabView and Matlab software. The phase-shifting, depth scanning and dynamic focusing are controlled in real-time by the PC.
Fig. 1. Full-field OCT setup: MO – microscope objective, PZT – piezoelectric phase shifter, MTS – motorized translation stage, BS – beam splitter, C – collimator, L1, L2 – achromat doublet lenses, RM – reference mirror, MS – mode scrambler, MMF – multimode fiber

Fig. 2. Signal required versus wavelength to achieve camera full signal illumination by Anti-Stokes up conversion material. The grey area indicates the spectral range of the SLD.

2.2. Spatially incoherent illumination

According to previous studies, a typical SLD produces highly spatially coherent light, which is undesirable for full-field imaging due to speckle and cross-talk effects. A cross-talk phenomenon arises in scattering samples when multiply scattered photons traveled the distance that matches the path length of the sample depth within source coherence length. In point-scanning OCT, the narrow illumination and confocal aperture of the single mode fiber eliminates this effect. An efficient way to reject cross-talk was to use a spatially incoherent illumination from a thermal source. Another author proposed coupling the input light into few tens of meters of a large core diameter and large NA multimode fiber. Instead of propagating light through long lengths of fiber, we use a multimode patch cord and a mode scrambler to obtain a stable and repeatable modal distribution independent of the initial modal distribution caused by the light launch conditions. Mode scrambler employs specially designed surfaces that create bending to a short length of a
multimode waveguide. The bending force is controlled by a precision translation stage. These perturbations increase light scattering and redistribution of the power among the guided modes of the fiber to induce mode coupling greatly. However, mode scrambling needs to be very precisely controlled. Excessive bendings can create large optical power losses, distortions to the guided light spectrum and damages to the waveguide eventually. To generate spatially incoherent illumination useful for a particular FF-OCT system, a selected fiber lateral coherence area must be smaller than the system lateral resolution\textsuperscript{15}.

We use a 200 μm core diameter and 0.39 NA step-index fibre able to guide up to 50000 modes. Images of the USAF test chart made with spatially coherent and spatially incoherent illumination are presented in figure 3. Figure 3a is less blurred than corresponding figure 3b, which indicates that cross-talk has been reduced, but not entirely rejected. The speckle, visible as a background granulation pattern on both figures, has been reduced by a spatially incoherent illumination.

![Fig.3 Images of a USAF test chart created with a spatially incoherent illumination (a) and a spatially coherent illumination (b). Image (b) is more blurred and exhibits a granulation pattern due to speckle.](image)

### 2.3. Image restoration method

The interferometric images are obtained using a simple phase-stepping technique. The signal $I_p(x, y)$ detected by each camera pixel, with coordinates $(x, y)$, corresponding each phase shift can be expressed as:

$$I(x, y) = I_0(x, y) + I_{coh}(x, y) \cos \phi(x, y)$$  \hspace{1cm} (1)

Where $I_0(x, y)$ denotes the average signal intensity and $\phi(x, y)$ denotes the optical phase. $I_{coh}(x, y)$ is the intensity of the coherent signal, which is proportional to the time-averaged cross-correlation of the sample and reference optical fields. To retrieve the information carrier and to eliminate the dc component $I_0(x, y)$, a number of discrete phase shifts is performed until optical phase reaches the multiplication of the $\lambda/8$ (or $\pi/4$) value. All frames recorded during phase shifting are summed up which is denoted by the integral in eq. 2. This procedure is repeated four times for every depth scan to acquire four stacks of frames.

$$I_{pn}(x, y) = \int_{\phi_0}^{\phi_0+\pi/4} n[I_0(x, y) + I_{coh}(x, y) \cos[\phi(x, y) + \phi]] d\phi$$

$$p = 1, 2, 3, 4, \ n = 1, ..., N$$  \hspace{1cm} (2)
For every discrete phase shift step, a N number of frames can be collected and averaged.

$$I_p = \frac{1}{N} \sum_{n=1}^{N} I_{p,n}(x, y)$$  \hspace{1cm} (3)

The four equations in (3) can be combined to obtain $I_{coh}(x, y)$ as:

$$I_{coh} = \sqrt{(I_1 - I_2)^2 + (I_2 - I_4)^2}$$  \hspace{1cm} (4)

Full-field OCT systems using similar algorithms1-5 rely on direct synchronization between the detector and PZT, which yields fast data acquisition speed but introduces additional modulation of the optical phase. This approach cannot be realized here, due to limitation in the camera design, the CCD camera frame rate can vary significantly even over timescales of a few seconds.

Previously, a slightly different algorithm was used. Its principle was to shift the reference beam phase four times ($0$, $\lambda/8$, $\lambda/4$ and $3\lambda/8$) for every depth scan step, so four stacks of N camera images are acquired for each step. The new approach exhibits improvement in SNR of single dBs and is more resistant to time-varying “signal floating” effects due to external conditions, such as movement of particles in air in free-space interferometer.

It is worth mentioning that increasing the number N of frames acquired to calculate every depth scan, as well as performing more discrete phase shifts to build a single $I_p$ stack (in terms of decreasing the $d\phi$ variable), improves the SNR of the final image, but decreases the data acquisition speed of the system.

3. TOOTH IMAGING

The principle of this study was to acquire investigate a human molar tooth crown using described setup. The tooth’s crown is covered by enamel, which is an acellular, highly mineralized tissue containing 2-12% of water6. The bulk of the tooth is made of dentin, a hard, elastic tissue that primarily consists of collagen, fibrils and inorganic components and approximately 24% water content8. The interior of the tooth is called the pulp, a soft connective tissue that is innervated and highly vascular. Due to the complex nature of these structures, the scattering distributions are anisotropic and depend on tissue orientation relative to the irradiating light source. In 8 the total extinction of sound enamel has been estimated as $\sim 3.8 \text{ cm}^{-1}$ and it is dominated by absorption of water. This yield approximately 3 mm mean free path for a photon travelling in this tissue before getting absorbed or scattered. Optical imaging of dentin is limited by scattering7,9,11. In 10 the averaged refractive indices of enamel and dentin for near-IR light have been estimated as 1.631±0.007 and 1.540±0.013 respectively. Light scattering is primarily determined by a combination of Mie and Rayleigh scattering and has been shown to be high in the visible range9.

Since absorption in near-IR wavelength range is much lower, these wavelengths are more suitable for diagnostics in dentistry.

The tooth specimen has been cut from the side to leave a thin enamel layer. Then the cut plane was polished to approximately 1 μm roughness and examined. Five separate scans were made to construct a vertical XZ profile of the sample with total 700 μm in depth and 1 mm in width (Fig. 5). En-face scans were performed with a 10 μm steps. On Fig. 5 there is a well visible, thin enamel layer and an enamel-dentin interface. The signal backscattered from dentin is very weak, but it was able to discover another well reflective interface deep inside dentin. In another tooth sample, a crown was polished from the top to uncover the internal structure of the specimen. Then several en-face scans were performed and results were saved. Figure 6 presents an enamel-dentin interface of that sample. System was adjusted to examine the tooth 50 μm below the polished surface. Scanned sample had many major cracks and other faults and it is seen on the picture, as a discontinuities in backscattered and
backreflected signal. Dentin is clearly a weakly reflective tissue and a higher power light source would be useful to retrieve more details about it.

4. CONCLUSION

We have proposed a simple, inexpensive and easy to operate FF-OCT system working in 1550 nm range. Its performance is currently limited and requires further improvements. Penetration depth and axial resolution are mainly the issue. A balance needs to be found between focused spot size and optical power density on the sample side. On the other hand, SLD illumination was successfully converted into spatially incoherent, which reduced lateral cross-talk and speckle pattern and therefore prevented the associated loss of resolution and improved images signal-to-noise ratio.

ACKNOWLEDGEMENTS

This work was supported by Science Foundation Ireland Research Frontiers Programme grant 08/RFP/ENE1491.

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Appendix B – Data sheets

The available data sheets and relevant manufacturer catalogue pages for the two key components in the project, the Sumerlum SLD-761-HP1 diode and the Ophir-Spiricon L070-1550 camera, can be found here. The last two pages from Ophir-Spiricon describing the phosphor coating technology originate from the data sheet of a different camera model, but its content fully apply to the L070-1550 camera.
**Features:**
- two power categories, up to 10 mW ex SM fiber
- wide spectrum with small Fabry-Perot modulation depth

**Packages:** fiber coupled: Butterfly

**Additional & customized:**
- PD monitors (for selected models)
- PM fiber pigtails (slow axis alignment; 45 degree orientation upon request)
- FC/APC terminated pigtails

**Specifications**
*(Nominal Emitter Stabilization Temperature +20 °C)*

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<td>HP2</td>
<td>-</td>
<td>-</td>
<td>600</td>
</tr>
<tr>
<td>Forward voltage, V</td>
<td>All</td>
<td>-</td>
<td>1.9</td>
<td>2.5</td>
</tr>
<tr>
<td>Central wavelength, nm</td>
<td>All</td>
<td>1480, 1530, 1550</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Spectrum width, nm</td>
<td>All</td>
<td>30</td>
<td>45</td>
<td>-</td>
</tr>
<tr>
<td>Residual spectral modulation depth, %</td>
<td>All</td>
<td>-</td>
<td>-</td>
<td>5.0</td>
</tr>
<tr>
<td>Secondary coherence subpeaks (10 log), dB</td>
<td>All</td>
<td>-</td>
<td>-</td>
<td>-20</td>
</tr>
<tr>
<td>Slow / fast polarization ratio (PM modules)*, dB</td>
<td>All</td>
<td>5</td>
<td>10</td>
<td>-</td>
</tr>
<tr>
<td>Operating temperature (case) at full power, °C</td>
<td>HP1</td>
<td>-55</td>
<td>-</td>
<td>+70</td>
</tr>
<tr>
<td></td>
<td>HP2</td>
<td>-</td>
<td>-</td>
<td>+60</td>
</tr>
<tr>
<td>Cooler current, A**</td>
<td>All</td>
<td>-</td>
<td>-</td>
<td>1.2</td>
</tr>
<tr>
<td>Cooler voltage, V**</td>
<td>All</td>
<td>-</td>
<td>-</td>
<td>3.5</td>
</tr>
</tbody>
</table>

* Pseudo-depolarized version (light is launched into the fiber with its polarization oriented at 45° to the birefringent axes) is available upon request
** 2.5 A / 4 V TEC cooler may be used to extend the operating temperature range

The following part numbers should be used when **ordering:**

SLD-761-(b)-(c)-(d)-XXXX,
where:
- b – power category (HP1 or HP2),
- c – package type,
- d – SM (isotropic) or PM (polarization maintaining),
- XXXX – required wavelength (in nanometers).

Example: SLD-761-HP2-DBUT-1550.

A maximum feedback of $\approx 30$ dB ($10^{-3}$) is allowed to run HP-series SLDs safely at full power.

All specifications are subject to change without notice.

**Applications:**
- optical sensing
- optical coherence tomography
- optical measurements

**PERFORMANCE EXAMPLES**

**SLD-761-HP2-SM. Light-current curve**

**SLD-761-HP2-SM-1550. Spectrum**

**Detailed spectrum trace**

**Extended displacement**

Mirror displacement = Optical path difference / 2
LIGHT-CURRENT and CURRENT-VOLTAGE CHARACTERISTICS @ T=20 C

MAXIMUM SLD CURRENT IS 315 mA

PERFORMANCE PARAMETERS @ 315 mA PUMPING CURRENT and T=20 C

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Spectral center, nm</td>
<td>1563.2</td>
</tr>
<tr>
<td>Spectral bandwidth, FWHM, nm</td>
<td>41.4</td>
</tr>
<tr>
<td>Spectral Ripple, %</td>
<td>4.3</td>
</tr>
</tbody>
</table>
USB Silicon CCD Cameras

L-series

Features

- USB 2.0 compatible
- 12-bit A/D, 66dB true system dynamic range
- Choice of software for sophisticated measurements and ease of use
- Spectral range: 350 - 1100nm
- Optional optical trigger module senses and synchronizes camera with laser pulse

USB L070, L130, L230

USB L070, L130, L230
# Camera Selection Chart

## USB2 CCD Cameras

<table>
<thead>
<tr>
<th>MODEL (1)</th>
<th>Spiricon L230</th>
<th>Spiricon L130</th>
<th>Spiricon L070</th>
<th>UNITS</th>
<th>CONDITIONS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of Elements</td>
<td>1616 x 1216</td>
<td>1392 x 1040</td>
<td>640 x 480</td>
<td>Pixels</td>
<td>H X V</td>
</tr>
<tr>
<td>LBA-PC Maximum Digitized Resolution</td>
<td>1616 x 1216</td>
<td>1392 x 1040</td>
<td>640 x 480</td>
<td>Pixels</td>
<td>H X V</td>
</tr>
<tr>
<td>Element Pitch ((\mu m))</td>
<td>4.4 x 4.4</td>
<td>4.65 x 4.65</td>
<td>7.4 x 7.4</td>
<td>(\mu m)</td>
<td>H X V</td>
</tr>
<tr>
<td>Area</td>
<td>7.1 x 5.4</td>
<td>6.5 x 4.8</td>
<td>4.7 x 3.5</td>
<td>(\mu m)</td>
<td>H X V</td>
</tr>
<tr>
<td>Max Viewable Beam, mm (2)</td>
<td>5.4</td>
<td>4.6</td>
<td>3.6</td>
<td>(\mu m)</td>
<td>100% of Beam</td>
</tr>
<tr>
<td>Material</td>
<td>Silicon CCG</td>
<td>Silicon CCG</td>
<td>Silicon CCG</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Spectral Response</td>
<td>190 - 1100 nm</td>
<td>190 - 1100 nm</td>
<td>190 - 1100 nm</td>
<td>nm</td>
<td></td>
</tr>
<tr>
<td>Readout</td>
<td>Progressive Scan Interline Transfer</td>
<td>Progressive Scan Interline Transfer</td>
<td>Progressive Scan Interline Transfer</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

## CW OPERATION

<table>
<thead>
<tr>
<th>Full Video</th>
<th>(\mu W/cm^2)</th>
<th>0.6</th>
<th>1.0</th>
<th>1.1</th>
<th>(\mu W/cm^2)</th>
</tr>
</thead>
<tbody>
<tr>
<td>S/N (5) (13)</td>
<td>dB</td>
<td>59</td>
<td>59</td>
<td>61</td>
<td>dB</td>
</tr>
<tr>
<td>At gamma = 1, (\gamma) set to 1.95 (13)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

## PULSED OPERATION

<table>
<thead>
<tr>
<th>Full Video</th>
<th>nJ/cm²</th>
<th>42</th>
<th>66</th>
<th>18</th>
<th>nJ/cm²</th>
</tr>
</thead>
<tbody>
<tr>
<td>Max. Pulse Rate (3)</td>
<td>Hz</td>
<td>12</td>
<td>15</td>
<td>60</td>
<td>Hz</td>
</tr>
</tbody>
</table>

## MECHANICAL SPECIFICATIONS

<table>
<thead>
<tr>
<th>Dimension</th>
<th>Inch</th>
<th>3.5 x 3.5 x 1.1</th>
<th>3.5 x 3.5 x 1.1</th>
<th>3.5 x 3.5 x 1.1</th>
<th>Inch</th>
<th>H X W X D</th>
</tr>
</thead>
<tbody>
<tr>
<td>Weight</td>
<td>Oz</td>
<td>10.2</td>
<td>10.2</td>
<td>10.2</td>
<td>Oz</td>
<td></td>
</tr>
</tbody>
</table>

## APPLICATIONS (R=Recommended/Function Available)

<table>
<thead>
<tr>
<th>Electronic Shutter (4)</th>
<th>R</th>
<th>R</th>
<th>R</th>
</tr>
</thead>
<tbody>
<tr>
<td>Full Resolution</td>
<td>R</td>
<td>R</td>
<td>R</td>
</tr>
<tr>
<td>Low Noise</td>
<td>R (8 or 12 bits)</td>
<td>R (8 or 12 bits)</td>
<td>R (8 or 12 bits)</td>
</tr>
<tr>
<td>Interface Type</td>
<td>USB2</td>
<td>USB2</td>
<td>USB2</td>
</tr>
<tr>
<td>Integration</td>
<td>R</td>
<td>R</td>
<td>R</td>
</tr>
<tr>
<td>Remote Head</td>
<td>R (11)</td>
<td>R (11)</td>
<td>R (11)</td>
</tr>
<tr>
<td>Long Wavelength (&gt;1100 nm)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Short Pulse YAG at 1.06 (\mu m)</td>
<td>R (27)</td>
<td>R (27)</td>
<td>R (27)</td>
</tr>
<tr>
<td>CE Mark</td>
<td>R</td>
<td>R</td>
<td>R</td>
</tr>
</tbody>
</table>
The following are observations relative to the various cameras available to enable you to make the best possible selection for your application. The notes below are coupled to the numbers in parenthesis at each point of the selection chart.

Please note that cameras purchased from Spiricon for use with the Laser Beam Analyzer include special alignment and option features. These features add increased cost over the list price of a basic camera. For the majority of laser beam diagnostic applications these options are essential for accurate and reliable operation. These include such options as: windowless detector to eliminate fringe patterns, low noise electronics, extended frame integration, special synchronization timing, etc.

Notes To Camera Selection Guide

(1) The cameras described in this comparison are available with many factory options which are too numerous to detail on this chart. The information contained herein is general in nature and is not an assurance that any camera not purchased through Spiricon will operate as recommended.

(2) Refers to the maximum beam dimension that can be displayed on the Laser Beam Analyzer in lowest magnification.

(3) In normal (non-shuttered) camera operation, this is the fastest rate at which the laser may pulse and the camera can still separate one pulse from the next. It is not the rate at which the Beam Analyzer can acquire pulsed data. With electronic shutter operation, (see note 4) higher rate laser pulses can be split out by matching the laser repetition rate to the shutter speed.

(4) Various shutter speeds and options are available with different makes and models. Consult manufacturer specifications.

(5) Signal to Noise Ratio measured with the LBA-PC using the Histogram of background noise.

(6) With pulsed lasers, the Interlaced Frame Transfer type camera will capture single laser pulses in only one field thus having the effect of reducing the stated camera resolution by 1/2.

(7) CCD cameras function at wavelengths as short as 190 nm. However wavelengths shorter than 245 nm will damage the camera sensor and lead to camera failure.

(8) This is the value specified when supplied with a low noise factory option.

(9) These Intellitron Transfer cameras offer the highest resolution with pulsed lasers because they output the pulse in both fields.

(10) The TM-6/7 cameras have been observed to create a ghost image when operating with pulsed YAG lasers at 1064 nm. Therefore, we recommend the COHU 4812 and the SMD-1M15 for this wavelength and pulse condition.

(11) These cameras are not available with remote heads, but are already so small that they are approximately equivalent in size to other cameras which have separate remote heads.

(12) The FireWire® camera maximum rate depends on the ROI (Region of interest) size, the bits readout, and the number of cameras on the same bus. The 220HR20 operates at 7.5 kHz@12 bits and 15 Hz@8 bits. It operates up to at least 25 Hz with a smaller ROI, it slows down to 3.75 Hz with 2 cameras on the same FireWire® bus. However, up to 3 FireWire® to PCI cards have been installed in one desktop computer and have operated 3 cameras simultaneously without slowing down the rate. Performance also depends on the speed of the computer.

(13) S/N given is with camera in natural state of γ = 1. With γ changed to 1.95 in the LBA-PC to compensate for the phosphor non-linearity, the noise is raised and the S/N becomes about 30 dB.

(14) For CW Uses chopper. Camera model PV-11-C.

(15) The LBA-PC digitizes analog cameras at the vertical pixel pitch in both horizontal and vertical axes. A video low pass filter in the camera enables digitizing the horizontal pixels at a pitch different from the actual pixel pitch. This creates a square digitized pixel.

(16) These cameras require an optional synchronous chopper to operate in CW mode.

(17) These cameras are supplied with AR coated windows. Contact factory for details of coatings.

(18) The SU-320M now has a baseline above zero so it is now possible to UltraCal the camera and obtain accurate beam width measurements.

(19) The COHU 4812 is the one CCD camera that has response to 1310 nm without blooming, even though the responsivity is greatly reduced.

(20) The FireWire and USB2 cameras work well with short pulse YAG lasers when the external trigger module is used to trigger the camera synchronous with the laser.

### DAMAGE THRESHOLD POWER/ENERGY DENSITIES FOR VARIOUS CAMERA DETECTOR TYPES

<table>
<thead>
<tr>
<th>Detector</th>
<th>CW power in mW/sq.cm.</th>
<th>Pulse energy in mJ/sq.cm.</th>
<th>Multiples of video saturation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pyrocam III</td>
<td>8000</td>
<td>20 (1us pulse)</td>
<td>2X</td>
</tr>
<tr>
<td></td>
<td></td>
<td>600 (1us pulse)</td>
<td>60X</td>
</tr>
<tr>
<td>PbS Vidicon</td>
<td>10</td>
<td>10</td>
<td>5X</td>
</tr>
<tr>
<td>CCD</td>
<td>.15</td>
<td>.1</td>
<td>&gt;1000X</td>
</tr>
</tbody>
</table>

*Information obtained from camera manufacturers. Spiricon is not responsible for the accuracy of this information.*
3.1.2.5 1440-1605nm Phosphor Coated CCD Cameras For NIR Response

Features
- 1440-1605 nm Wavelengths
- NIR Telecom mode field analysis
- NIR Laser beam analysis

Available Models
- USB models: SP503U-1550
  - SP620U-1550
- Firewire models: GRAS20-1550
- Analog Camera™: SP-155OM

*When used with beam profiling software, requires LBA-FC-10 fiber grabber system (sold separately)

Phosphor Coating Technology

The up-conversion from NIR to visible light in the 1550 series cameras is nonlinear. The anti-Stokes phosphor coating produces visible photons at a rate roughly the square of the input signal. This is shown dramatically where the camera total output increases dramatically faster than a linear output shown in the bottom line. The CCD camera saturation in the center of a beam, the up-converted visible signal drops as the square of the input signal. Thus the lower signal wings of a beam are suppressed, resulting in the appearance and measurement of a beam width much smaller than actual.

This illustration a comparison of the cross-section of a beam with and without correction. (As seen, the real width of the beam is much greater than would be observed without correction.)

![Non-linearity of SP-155OM Camera at 1550nm](image1)

![Comparison of Beam Shape with and without Correction Factor](image2)

Non-linear output of the 1550 series cameras.

Cross-section of a fiber beam with and without non-linearity correction.
Wavelength Response

The anti-Stokes up-conversion efficiency is very wavelength dependent. This graph shows the typical spectral response curve of a new high response coating. As seen, we have calibrated the response from 1527nm to 1605nm. We have extrapolated the shorter wavelength region by comparing our measured response to data published over the entire range.

Phosphor Coated Cameras with Spiricon’s BeamGage software

Spiricon engineers have carefully measured the non-linearity of the signal generated by the Phosphor Coated series cameras. The software in the BeamGage incorporates an algorithm to correct for the non-linearity. This illustration shows the linearity obtained, showing in the top line that the low level signals drop linearly, rather than at the square of the input, seen in the lower line.

The two photos show the uncorrected and corrected camera beam shape in 3D. See the BeamGage section for additional information on the beam analyzer.

Beam profile of a fiber beam with non-linearity correction.

Beam profile of a fiber beam without non-linearity correction.

SP-1550M: RS-170 monitor display when used without a frame grabber.