Development of multi-functional optical coherence tomography

DISSERTATION

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Abstract

Optical coherence tomography (OCT) has been introduced as a non-contact non-invasive cross-sectional imaging for biological sample. The purpose of the research is to show the potential improvements in the resolution and in the functionalities of OCT and to demonstrate the application of OCT. An ultra-high resolution spectral domain OCT (SD-OCT) was developed by using a supercontinuum laser. A spectrometer was designed for the OCT. High resolution imaging of the sweat ducts of a finger was demonstrated by the system. Spectroscopic analysis on a scattering media was attempted using the SD-OCT. A multi-functional swept source OCT which combined polarization sensitive OCT and Doppler OCT was developed with fiber-based system to visualize birefringence and blood flow in tissue in addition to the internal structure of sample.
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1. Introduction

Optical coherence tomography (OCT) has been introduced as a non-contact non-invasive cross-sectional imaging for biological sample [1]. Since then in vivo imaging by OCT was demonstrated in a wide range of medical fields, such as dermatology, dentistry and cardiology, and especially has become an indispensable imaging tool in ophthalmology [2-3]. Besides, applications of OCT in industrial non-destructive testing and evaluation were also attempted.

The OCT is based on interferometry technique to measure the internal structure of sample. Frequency domain OCT (FD-OCT) acquires the spectrum of the interference signal and Fourier analysis of the spectrum provides the depth profile of the sample. FD-OCT can be implemented by either using a broad bandwidth light source with a spectrometer, which is known as spectral domain OCT (SD-OCT), or a wavelength sweeping light source with a photodetector, which is known as swept-source OCT (SS-OCT). The axial resolution of OCT are known to be dominated by the center wavelength and the bandwidth of the light sources. The development of broad bandwidth light sources brings resolution improvement to SD-OCT. The availability of wavelength sweeping light sources which have narrow linewidth enable swept-source OCT to image over large depth range with low signal decay. The feature of low signal decay made possible in the implementation of polarization sensitive SS-OCT, a functional extension OCT to visualize depth-resolved birefringence of sample, without the requirement of any active modulation optical component.

As a functional extension of OCT, polarization sensitive OCT (PS-OCT) has been developed to measure depth-resolved birefringence of sample [4]. Based on birefringent properties of ocular tissues, PS-OCT has been used to study the properties of retinal nerve fiber [5,6] and scar tissue in age-related macular degeneration [9,10]. Measurement of polarization scrambling by PS-OCT enables discrimination of the retinal pigment epithelium [9,10].

Doppler OCT has been used to image depth-resolved flow, particularly the flow in blood vessels of the retina. Visualization of vasculature using Doppler OCT shows its importance in the diagnosis of diseases such as glaucoma [14] and age related macular degeneration [15] which can be characterized by abnormalities in the blood flow and/or the vascularization.
Recently, several functionally extended OCT have been demonstrated based on swept source OCT (SS-OCT) technology [9,10]. This is partially because of the advantages of SS-OCT over spectral domain OCT, including improved sensitivity with greater imaging depth, k-linear sampling [16], and balanced detection.

Considering the advantages of optical fibers, such as system alignment and handling, many functional OCT systems have been developed with the use of fiber optics. However, for PS-OCT, at least two polarization states of the incident light are required to measure the Jones matrix of a sample [8] because of the dynamic birefringence in the single-mode fiber of the sample arm. For this purpose, Jones matrix PS-OCT systems with polarization modulators have been demonstrated [8, 9].
2. Purpose

Each functional imaging of OCT visualize unique depth resolved characteristic of sample. The purpose of the research is to enhance the potential application of OCT by improving the resolution and functional imaging of multiple physical properties of sample. An ultra-high resolution SD-OCT was developed by using a supercontinuum laser. High resolution imaging of sweat ducts of the finger was demonstrated using the system. A multi-functional SS-OCT which combined polarization sensitive OCT (PS-OCT), and Doppler OCT was developed to visualize the properties beyond the resolution of OCT. The research showed a novel implementation of PS-OCT with only passive polarization components. The PS-OCT visualizes the birefringence of sample which corresponds to the organization of fibrous tissue in biological sample that cannot be observed with a conventional OCT. The PS-OCT was used for the ocular imaging to discriminate fibrous tissue and to identify the abnormalities in tissue. The Doppler OCT enhances contrast of the moving scatters in sample. It was used to measure the flow of blood which visualize the vasculature in retina. The multi-functional SS-OCT simultaneously measured birefringence and blood flow in tissue in addition to the internal structure of sample.
3. Principle of Optical Coherence Tomography

3.1 Fourier Domain Optical Coherence Tomography

Optical coherence tomography is an imaging technique based on low-coherence interferometry [17]. Figure 3.1 shows the scheme of Michelson interferometer, which consists of a light source, a detector, a beam splitter, a reference mirror and a mirror as sample. The beam splitter splits the light source into reference arm and sample arm. Reflections from both arms return and pass through the beam splitter, then interfere at the detector. Fields of the two arms at the detector can be given by

\[
\begin{align*}
E_R(k) &= \sqrt{R_R} S(k) e^{i(kz_R+\omega t)} \\
E_S(k) &= \sqrt{R_S} S(k) e^{i(kz_S+\omega t)}
\end{align*}
\]

Here, \( S, k, \) and \( \omega \) are power spectrum, wavenumber, angular frequency of the light source. \( R \) and \( Z \) are intensity reflectivity and optical path length. The interference at the
detector which is proportional the photocurrent of the detector can be expressed by the following. Here, \( \langle \rangle \) denotes integration over the response time of the detector.

\[
I = \langle \left| E_R + E_S \right|^2 \rangle = \langle (E_R + E_S)(E_R^* + E_S^*) \rangle
= \langle E_R E_R^* \rangle + \langle E_S E_S^* \rangle + \langle E_R E_S^* \rangle + \langle E_R^* E_S \rangle .
\]  

3.1.2

Defining double-pass optical path length as \( \Delta z = \frac{1}{2}(z_R - z_S) \), the equation can be expressed in two DC terms and an interference term, as shown below and demonstrated in Fig. 3.2(a).

\[
I(k) = S(k) \left( R_R + R_S + 2\sqrt{R_R R_S} \cos(2k\Delta z) \right).
\]  

3.1.3

The time dependent term \( \omega t \) disappeared and the interference term is a function of the optical path difference between the reference mirror and the sample. In the case of multiple reflectors, for instance multilayer sample, the interference occurs between layers within sample, than the detected signal can be expressed as

\[
I(k) = S(k) \left( R_R + \sum_n R_{S_n} \right)
+ 2S(k) \sum_n \sqrt{R_R R_{S_n}} \cos(2k\Delta z_{S_n})
+ 2S(k) \sum_{n \neq m} \sqrt{R_{S_n} R_{S_m}} \cos(2k\Delta z_{S_{nm}}) .
\]  

3.1.4

The first term is DC term which is total intensity reflected from the reference mirror and the sample, the second term is cross-correlation which is the interference between the reference mirror and the sample, and the third term is auto-correlation term which is the interference between the layers of sample. The DC term appears at the zero delay line. The cross-correlation term is particularly important for the reconstruction of the depth profile of sample. The auto-correlation term is very small and close to the zero delay line in most of the applications. The cross-correlation term can be separated from the auto-correlation term by moving the reference mirror. Two implementation are available to acquire the spectrum of the interference signal given by Eq. 3.1.4. One implementation which is known as Spectral Domain OCT (SD-OCT) uses a broadband light source with a spectrometer as the detector. Another implementation is Swept-source OCT (SS-OCT). SS-OCT is implemented with a wavelength sweeping light source and a photodetector. Both SD-OCT and SS-OCT share the same data processing
algorithm in general. Inverse Fourier transform of the interference Eq. 3.1.4, which gives depth profile of the sample can expressed as,

\[
i(z) = s(z) \left( R_r + \sum_n R_{S_n} \right) + s(z) * \sum_n \sqrt{R_r R_{S_n}} \left[ \delta(z + 2\Delta z_{S_n}) + \delta(z - 2\Delta z_{S_n}) \right] + s(z) * \sum_{n \neq m} \sqrt{R_{S_n} R_{S_m}} \left[ \delta(z + 2\Delta z_{S_m}) + \delta(z - 2\Delta z_{S_m}) \right],
\]

3.1.5

where \( I(k) \xleftarrow{\beta} i(z) \), and \( * \) denotes convolution, \( \delta \) is the Dirac delta function and

\[
S(k) \xleftarrow{\beta} s(z)
\]

3.1.6

is the Fourier transform relation of the power spectrum of the light source. A well-known Fourier transform relation

\[
\cos(kz_0) \xleftarrow{\beta} \frac{1}{2} \left[ \delta(z + z_0) + \delta(z - z_0) \right]
\]

3.1.7

is used to derive the Eq. 3.1.5. Given that

\[
\begin{align*}
X(k) & \xleftarrow{\beta} x(z) \\
Y(k) & \xleftarrow{\beta} y(z)
\end{align*}
\]

3.1.8

the convolution theorem shows that

\[
X(k) \cdot Y(k) \xleftarrow{\beta} x(k) * y(k).
\]

3.1.9

In numerical data processing, Discrete Fourier transform (DFT) is used to obtain the depth profile. Figure 3.2(b) demonstrates the DFT of an interference between reference mirror and a single reflector. The peak at zero delay line is DC term. Two peaks are the signal of the single reflector. Symmetrical appearance of the peak in the positive and negative depth direction is the property of DFT. In the case of multilayer sample, multiple peaks which correspond to cross-correlation terms and auto-correlation terms will appear. The cross-correlation terms provide the depth profile of the measured sample.
3.2 Sensitivity and Resolution

One advantage of OCT is that the axial resolution is dominated by the bandwidth the light source. The depth profile OCT in Eq. 3.1.5 shows that point spread function \( s(z) \) is the inverse Fourier transform of power spectrum. By assuming Gaussian shaped power spectrum

\[
S(k) \propto \exp\left(-\frac{(k-k_o)^2}{\Delta k}\right)
\]

with a spectral bandwidth of \( \Delta k \), which is defined by the half-width of the spectrum at \( 1/e \) of its maximum, centered at wavenumber \( k_o \), the resulting inverse Fourier transform is

\[
s(z) \propto \exp\left(-z^2\Delta k^2\right).
\]

Then the full width at half maximum (FWHM) of \( s(z) \) which is defined as the axial resolution is given by [18,19]

\[
\Delta z = \frac{2 \ln 2}{\pi} \frac{\lambda^2}{\Delta \lambda}.
\]
Here, the relations between wavelength and wavenumber, $\lambda_0 = 2\pi/k_0$ and $\Delta k = 2\pi\Delta\lambda/\lambda_0^2$ are used. Besides the axial resolution, maximum measurable imaging depth range and sensitivity are important quantities to evaluate the performance of an OCT system. The maximum measurable imaging depth range $\Delta z_{\text{max}}$ is given by [20]

$$\Delta z_{\text{max}} = \frac{\lambda_0^2}{4\delta\lambda}.$$  \hspace{1cm} 3.2.4

Here, $\delta\lambda$ is the wavelength resolution. Sensitivity at shot noise limited performance can be calculated from [20]

$$-10\log\frac{\eta P_0}{h\nu f}[dB],$$  \hspace{1cm} 3.2.5

where $h$ is the Planck constant, $P_0$ and $\nu$ are the optical power and frequency of the probe beam, respectively. The $\eta$ is the quantum efficiency of detector or spectrometer. $f$ is wavelength-sweeping frequency in SS-OCT. In SD-OCT $f$ in replaced by $t/N$, where $t$ is the exposure time and $N$ is the number of sample [21]. $N$ appears in the equation because the optical power is divided into each pixel in the case of SD-OCT.

### 3.3 Spectral Calibration

#### 3.3.1 Phase Method

The Discrete Fourier Transform (DFT) which is defined by

$$X_k = \sum_{n=0}^{N-1} x_n e^{-2\pi i kn/N}$$  \hspace{1cm} 3.3.1

is used obtain the depth profiles from the OCT interference spectra. The DFT implied the requirement of evenly spaced sampling. Due to the spectrometer design of SD-OCT or non-linear wavelength sweeping of the light source of SS-OCT, usually the measured spectrum are not sampled in evenly spaced wavenumbers. The unevenly spaced sampling results in broaden of OCT signal. Hence, spectral calibration to map the measured spectra into evenly spaced spectra is required prior to DFT calculation. However, latest commercial wavelength sweeping light sources implement a specific hardware to generate frequency clock signals to eliminate the necessity of spectral calibration.
The wavenumber of the measured spectrum can be expressed as a sum of wavenumber and a polynomial [22],

\[ k' = k + p(m), \]

where \( m \) is the index of the sample,

\[ p(m) = a_0 + a_1m + a_2m^2 + \cdots + a_n m^n. \]

To perform spectral calibration, interference signal of a single reflector is required. Common path interference signal in air, such as interference between a mirror and a slide glass is preferable to eliminate the influence of dispersion in the calibration data. Ignoring the DC term, the interference signal of a single reflector can be expressed with the unevenly spaced wavenumber \( k' \) as

\[ I(k) = 2\sqrt{R_\mu R_s} \, S(k') \cos(2k' \Delta z). \]

The DFT of this signal yields,

\[ i(z) = 2\sqrt{R_\mu R_s} \, s(z) \ast \left\{ \begin{array}{l} e^{2j(m) \Delta z} \ast \delta(z - 2\Delta z) \\ e^{-2j(m) \Delta z} \ast \delta(z + 2\Delta z) \end{array} \right\}. \]

A window function \( w(z) \) is used to filter out the signal of the peak at one side of DFT, for instance, filter out the first term in Eq. 3.3.5. The center of peak is shifted to zero delay line and then inverse DFT of this peak gives

\[ I(k') = 2\sqrt{R_\mu R_s} \, \delta \left[ w(z) \ast \left\{ \begin{array}{l} e^{2j(m) \Delta z} \ast \delta(z) \\ e^{-2j(m) \Delta z} \ast \delta(z) \end{array} \right\} \right] e^{2j k' \Delta z}, \]

where the term \( \exp(2jk\Delta z) \) comes from the shift theorem when the peak is shifted to zero delay line. Finally, the equation becomes

\[ I(k') = 2\sqrt{R_\mu R_s} \, \delta \left[ w(z) \ast S(k') e^{2j(m) \Delta z} e^{2j k' \Delta z} \right]. \]

Here, the phase of the complex valued \( i(k') \) is

\[ \phi(m) = 2j k' \Delta z. \]

The wavenumber of the measured data \( k' = k + p(m) \) can be obtained from the unwrapped phase \( \phi(m) \) to map the measured data into interference signals with evenly spaced
wavenumber, then the depth profiles are the DFT of the spectral calibrated interference signals.

3.3.2 Zero-crossing Fringe Matching Method

A zero-crossing fringe matching method is proposed to determine the wavenumbers of the OCT spectrum. A fringe pattern is required to be generated and acquired by the OCT system and a commercial spectrometer. The acquired fringes, expressed by Eq. 3.1.3, are band-pass filtered by Fourier method to remove the DC term and sinusoidal modulation of the fringes as given by Eq. 3.3.4 will remain. Zero-crossing points are obtained from the fringes and matched. Polynomial fitting is used to obtain the relation between the wavelengths and the pixel indices of the OCT. The OCT spectrum can be resampled into evenly spaced wavenumber using the fit polynomial.

3.4 Dispersion Compensation

A potential factor that degrades the axial resolution of OCT is the dispersion mismatch between reference and sample arms, because of the frequency dependence in the propagation function for materials in the both arms. The effect of dispersion mismatch can be described by including a propagation function $\beta(k)$ in the interference signal as

$$I(k) = S(k) \left[ R_R + R_S + 2 \sqrt{R_R R_S} \cos(2k\Delta z + \beta(k)) \right].$$

Taylor series expansion of the propagation function around the central wavenumber $k_0$ is given by

$$\beta(k) = \beta(k_0) + \frac{\partial \beta}{\partial k} \bigg|_{k_0} (k - k_0) + \frac{1}{2} \frac{\partial^2 \beta}{\partial k^2} \bigg|_{k_0} (k - k_0)^2 + \cdots + \frac{1}{n!} \frac{\partial^n \beta}{\partial k^n} \bigg|_{k_0} (k - k_0)^n. \quad 3.4.2$$

The first term $\beta(k_0)$ is the propagation constant at the central wavenumber, which introduces additional phase in OCT signal. The second term $\partial \beta / \partial k$ corresponds to the first-order dispersion and the inverse group velocity. The third term is related to the group velocity dispersion. The second, third and high order terms cause distortion of OCT signal and hence degrade axial resolution. Dispersion compensation techniques based on dispersive optical material and numerical methods were proposed. Numerical methods are preferred because of its robustness in variation of the thickness between different subjects.

To compensate the mismatched dispersion numerically, a polynomial phase
\[ \theta(k) = a_0(k - k_0)^2 + a_1(k - k_0)^3 + \cdots + a_{n-2}(k - k_0)^n \]

is subtracted from the spectral calibrated interference \( I(k) \) by \( I(k)e^{-j\theta(k)} \). Then the DFT of \( I(k)e^{-j\theta(k)} \) gives dispersion compensated OCT depth profile. The coefficients \( a_0, a_1, \ldots, a_n \), can be obtained automatically by using an iterative procedure to measure and optimize the sharpness of the OCT image.
4. Principle of Functional Imaging

4.1 Doppler Optical Coherence Tomography

![Diagram of Doppler detection of a flow at velocity \( v \) with an angle of \( \theta \) relative to the probe beam.](image)

The detected interference signal of a moving reflector in OCT can be expressed as [24]

\[
I(k) = S(k)(R_R + R_S + 2\sqrt{R_R R_S} \cos(2k(\Delta z + v_j t))).
\]

4.1.1

Here, \( \Delta z \) is the optical path difference between the reference mirror and the moving reflector and \( t \) is the time variable. Given \( v \) as the velocity of the reflector, the velocity paralleled to the incident beam is \( v_j = v \cos \theta \), where \( \theta \) is the angle between the incident beam and the moving direction of the reflector, as shown in Fig. 4.1. The moving reflector induces additional phase shift in the interference signal. Fourier transformation of the interference signal is given by

\[
i(z) = s(z)(R_R + R_S) + 2\sqrt{R_R R_S} s(z) \star \left( \delta(z - 2\Delta z)e^{2j\alpha v_j t} + \delta(z + 2\Delta z)e^{-2j\alpha v_j t} \right).
\]

4.1.2

The phase of the signal at one side of the Fourier transformation is

\[
\Phi = 2k_0 v_j t.
\]

4.1.3
The phase difference between two measurements of a moving reflector in a time separation of $\Delta t$ gives

$$\Delta \Phi = 2k_0v_j \Delta t.$$  \hspace{1cm} 4.1.4

The velocity of the moving reflector paralleled to the incident beam can be expressed as

$$v_j = \frac{\Delta \Phi}{2k_0 \Delta t} = \frac{\lambda_0}{4\Delta t} \frac{\Delta \Phi}{\pi}.$$  \hspace{1cm} 4.1.5

Where central wavenumber and central wavelength is related by $k_0 = 2\pi/\lambda_0$. From Eq. 4.1.5, the velocity of the moving reflector can be obtained by

$$v = \frac{\lambda_0}{4\Delta t \cos \theta} \frac{\Delta \Phi}{\pi}.$$  \hspace{1cm} 4.1.6

In the measurement of high velocity flow, square of the phase $\Delta \Phi^2$ is use for the visualization. The measureable phase is known to be limited by phase wrapping $|\Delta \Phi| \leq \pi$ and signal-to-noise ratio (SNR). As a result, the measureable velocity range can be expressed as [25]

$$\frac{\lambda_0}{4\Delta t \sqrt{SNR}} \leq |v_j| \leq \frac{\lambda_0}{4\Delta t}.$$  \hspace{1cm} 4.1.7
4.1.1 Bulk Motion Compensation

The motion of the whole sample has to be compensated to obtain the velocity of the flow inside the sample, especially when performing in vivo measurement. As shown in Fig. 4.2, the measured velocity by Doppler OCT becomes $v_{\parallel}(z) + v_b$, given the motion of the whole sample is $v_b$, which is known as bulk motion. The bulk motion $v_b$ is a constant within the whole sample and $v_{\parallel}(z)$ is the velocity of the flow at depth $z$. In addition to bulk motion, in the case of swept-source OCT, usually the start sweeping of wavelength jitters. This jittering induces additional depth dependent phase shift, the last term in Eq. 4.1.9, which can be described by the shift theorem of the Fourier transform,

$$F(k + \Delta k_s) \xrightarrow{\beta} e^{-j\beta k_s} f(z).$$

Considering the bulk motion and the jittering of the light source, the phase difference between two measurements can be expressed as [26].

$$\Delta \Phi(z) = 2k_0\left(v_{\parallel}(z) + v_b\right)\Delta t - z\Delta k_s$$
$$= 2k_0v_{\parallel}(z)\Delta t + 2k_0v_b\Delta t - z\Delta k_s.$$

Here, $\Delta k_s$ is the difference of the wavenumber due to the jittering of the light source. Assuming that the flows occupy negligible small portion of sample, the last two terms of Eq. 4.1.9 can be obtained by OCT signal amplitude weighted linear fitting of the
unwrapped phase as proposed in [26]. The algorithm minimizes the weighted residual sum of squares $\varepsilon$,

$$
\varepsilon = \sum_z W(z) |\Delta \Phi(z) - (az + b)|^2 .
$$

Here, $z$ is the depth of the measurement and $W(z)$ is the amplitude of the OCT signal which is thresholded with an empirically determined noise level. The linear fitting with coefficients of $a$ and $b$ which correspond to the bulk motion and the jittering of the light source is subtracted from the raw phase difference,

$$
2k_0V(z) \Delta t = \Delta \Phi(z) - (az + b).
$$

The resulting velocity of flows is free from the bulk motion of the whole sample and the wavelength shifting of the jittering.
4.2 Polarization Sensitive Optical Coherence Tomography

4.2.1 Polarization in Material

Polarized light can be decomposed as two orthogonally oscillating electric fields $E_x$ and $E_y$, as shown in Fig. 4.3. The oscillation of the electric fields can be described by the notation of phasor, namely

$$
\begin{align*}
E_x &= E_{x0} e^{i(2\pi n z_x / \lambda + \omega t)} \\
E_y &= E_{y0} e^{i(2\pi n z_y / \lambda + \omega t)}
\end{align*}
$$

where $E_{x0}$ and $E_{y0}$ are the amplitude, $z_x$ and $z_y$ are the optical path length which is a multiplication of refractive index $n$ and propagated physical length $l$, $z = nl$, then $k$ and $\omega$ are the wavenumber and the angular frequency of the light. Birefringent materials are anisotropic materials which exhibit different refractive indices. Figure 4.3 shows the simplest type of birefringent material which has two refractive indices $n_x$ and $n_y$ in orthogonal directions. The difference of the refractive indices,

$$
\Delta n = n_x - n_y
$$
is used to quantify birefringence. Because of different refractive indices, when light passed through the material, the electric fields $E_x$ and $E_y$ of the light propagate in different optical path lengths at the two directions. This results in phase retardation between the two electric fields, which can be obtained from \[ \eta = (\angle E_{x\text{out}} - \angle E_{y\text{out}}) - (\angle E_{x\text{in}} - \angle E_{y\text{in}}) = 2\pi n L/\lambda , \] \[ 4.2.3 \]

where $\angle$ denotes the phase of a complex number, $E_{x\text{in}}$ and $E_{y\text{in}}$ are the electric fields before entering the material, $E_{x\text{out}}$ and $E_{y\text{out}}$ are the electric fields after passing the material and $L$ is the thickness of the material.

### 4.2.2 Jones Calculus

The oscillating electric fields in two orthogonal directions can be expressed in complex vector form

\[ E = \begin{bmatrix} E_x \\ E_y \end{bmatrix} = \begin{bmatrix} E_x e^{i2\varphi_x/\lambda} \\ E_y e^{i2\varphi_y/\lambda} \end{bmatrix} e^{i\phi} = \begin{bmatrix} E_x e^{i\phi_x} \\ E_y e^{i\phi_y} \end{bmatrix} e^{i(z \lambda/\lambda)}. \] \[ 4.2.4 \]

Here, $\phi$ and $z$ are the phase offset of the electric fields and the common optical path length, respectively. The polarization of a material can be expressed by a $4 \times 4$ complex matrix,

\[ J = \begin{bmatrix} J_{11} & J_{12} \\ J_{21} & J_{22} \end{bmatrix}, \] \[ 4.2.5 \]

which is known as Jones matrix [29]. The Jones matrix of birefringent materials with attenuation can be expressed by a diagonal complex matrix,

\[ D = \begin{bmatrix} P_x \exp(j \eta/2) & 0 \\ 0 & P_y \exp(-j \eta/2) \end{bmatrix}, \] \[ 4.2.6 \]

where $P_x$ and $P_y$ are amplitude attenuations at two axes. These attenuations are quantified by diattenuation, which is defined by [27,28]

\[ d = \frac{P_x^2 - P_y^2}{P_x^2 + P_y^2}. \] \[ 4.2.7 \]
\[ J = U^\dagger DU, \tag{4.2.8} \]

where \( \dagger \) denotes conjugate transpose. The rotational matrix \( U \) is a complex matrix formed by the orientation of the optic axis, \( \Phi \) and \( \theta \), as defined by.

\[
U = \begin{pmatrix}
\exp(j\Phi/2) & 0 & \cos(\theta) & \sin(\theta) \\
0 & \exp(-j\Phi/2) & -\sin(\theta) & \cos(\theta)
\end{pmatrix}. \tag{4.2.9}
\]

Because the matrix \( U \) is a unitary matrix, \( J \) is a diagonalizable matrix in Eq. 4.2.8, and the diagonal elements of the diagonal matrix \( D \) are the eigenvalues of \( J \) \[27\], namely

\[
\begin{cases}
\lambda_1 = P e^{j\eta/2} \\
\lambda_2 = P e^{-j\eta/2}
\end{cases}. \tag{4.2.10}
\]

The phase retardation Eq. 4.2.3 and diattenuation Eq. 4.2.4 can be obtained from the eigenvalues as shown by the following, where \( ^- \) denotes complex conjugate

\[
\begin{cases}
\eta = \angle\lambda_1\lambda_2^- \\
d = \frac{|\lambda_1|^2 - |\lambda_2|^2}{|\lambda_1|^2 + |\lambda_2|^2}.
\end{cases} \tag{4.2.11}
\]
4.2.3 Jones-Matrix Based Polarization Sensitive OCT

The implementation of the polarization detection can be described using Jones calculus. In the sample arm, the light is separated into two orthogonal polarization states and propagated over different optical path lengths, of which the polarization states can be expressed as

\[
\begin{align*}
E_\uparrow & = \begin{bmatrix} 1 \\ 0 \end{bmatrix}, \\
E_\downarrow & = \begin{bmatrix} 0 \\ e^{i\delta} \end{bmatrix},
\end{align*}
\]

where \( k \) and \( \delta \) are the wavenumber of the light source and the relative optical path difference, respectively. The two polarization states of the light are combined before illuminated to sample as

\[
E_0 = E_\uparrow + E_\downarrow = \begin{bmatrix} 1 \\ e^{i\delta} \end{bmatrix}.
\]

Considering the birefringence of the fiber, the Jones matrix at the detection arm \( J_{\text{measured}} \) can be expressed by using the Jones matrix from the point where the optical path of the orthogonal polarization states is separated into the sample surface \( J_{\text{in}} \), a round-trip Jones matrix of the sample \( J_{\text{sample}} \), and the Jones matrix from the sample surface to the detection arm \( J_{\text{out}} \) as

\[
J_{\text{measured}} = J_{\text{out}}J_{\text{sample}}J_{\text{in}} = \begin{bmatrix} J_{11} & J_{12} \\ J_{21} & J_{22} \end{bmatrix}.
\]

In the detection arm, the respective electric fields of the sample arm and the reference arm are expressed as

\[
\begin{align*}
E_S &= A_SJ_{\text{measured}}E_0e^{i\delta z_S}, \\
E_R &= A_R \begin{bmatrix} 1 \\ 1 \end{bmatrix}e^{i\delta z_S},
\end{align*}
\]

where \( z_S \) and \( z_R \) are the path lengths of the sample and reference arms, and \( E_S \) and \( E_R \) are scalar constants that define the amplitudes of the sample and reference beams.
In the detection arm, the two beams were split into two orthogonal polarization states. The interferogram of the polarization state, labeled the horizontal state, is the square of the summation of the horizontal elements of \( \mathbf{E}_S \) and \( \mathbf{E}_R \) from Eq. 4.2.15:

\[
\begin{align*}
\mathbf{E}_{S,H} & = A_S \left( J_{11} + J_{12} e^{i \zeta} \right) e^{i z_S} \\
\mathbf{E}_{S,V} & = A_S \left( J_{21} + J_{22} e^{i \zeta} \right) e^{i z_S} \\
\mathbf{E}_{R,H} & = A_R e^{i z_R} \\
\mathbf{E}_{R,V} & = A_R e^{i z_R} \tag{4.2.16}
\end{align*}
\]

The detected current from the balanced photodetector at the corresponding state is given by

\[
\begin{align*}
I_H \propto |\mathbf{E}_{S,H} + \mathbf{E}_{R,H}|^2 & = A_R^2 + A_S^2 \left( J_{11}^2 + J_{12}^2 \right) + 2A_R^2 J_{11} J_{12} \cos(k \zeta) \\
& + A_R A_S \left( J_{11} e^{i 2 \Delta z} + J_{12} e^{i (2 \Delta z + \zeta)} \right) + \text{c.c.} \tag{4.2.17}
\end{align*}
\]

\[
\begin{align*}
I_V \propto |\mathbf{E}_{S,V} + \mathbf{E}_{R,V}|^2 & = A_R^2 + A_S^2 \left( J_{21}^2 + J_{22}^2 \right) + 2A_R^2 J_{21} J_{22} \cos(k \zeta) \\
& + A_R A_S \left( J_{21} e^{i 2 \Delta z} + J_{22} e^{i (2 \Delta z + \zeta)} \right) + \text{c.c.}
\end{align*}
\]

where c.c. denotes the complex conjugate of the terms which are associated with \( J_{11} \) and \( J_{12} \), and optical path length difference \( \Delta z = z_S - z_R \). A discrete Fourier transform of this signal yields an OCT signal.

\[
\begin{align*}
\tilde{I}_H & \propto A_R A_S \left[ J_{11} \cdot \delta(z - \Delta z) + J_{12} \cdot \delta(z - (\Delta z + \zeta)) \right] \\
\tilde{I}_V & \propto A_R A_S \left[ J_{21} \cdot \delta(z - \Delta z) + J_{22} \cdot \delta(z - (\Delta z + \zeta)) \right] \tag{4.2.18}
\end{align*}
\]

As illustrated in Fig. 4.4, \( J_{11} \) and \( J_{12} \), and \( J_{21} \) and \( J_{22} \) are separated by a distance \( \zeta \), the process above has provided us with two of the elements of \( J_{\text{measured}} \) in a depth-resolved manner.

Because the measured Jones matrix consists of the birefringence of the fiber, it must be compensated. The Jones matrix at the surface of sample is extracted as a matrix of the fiber,

\[
J_{\text{Surf}} = J_{\text{Out}} J_{\text{In}} \tag{4.2.19}
\]

The matrix is inverted, and then multiplied by the measured Jones matrix to cancel the birefringence of the fiber as

\[
J_c = J_{\text{surf}}^{-1} J_{\text{measured}} \]

\[
= \left( J_{\text{in}}^{-1} J_{\text{out}}^{-1} \right) \left( J_{\text{out}} J_{\text{sample}} J_{\text{in}} \right) \]

\[
= J_{\text{in}}^{-1} J_{\text{sample}} J_{\text{in}} \tag{4.2.20}
\]
where the fiber is assumed to be lossless, and whose Jones matrix is expressed as a unitary matrix. This assumption allows us to obtain the birefringence of the sample by decomposing the Jones matrix obtained into a diagonal matrix $D$ and a unitary matrix $U$, as shown in Eq. 4.2.11. Then again $J_c$ is also a diagonalizable matrix [8],

$$J_c = J_{in}^r U^r DU_{in} = (UJ_{in})^r D(UJ_{in}).$$ \hspace{1cm} 4.2.21

The phase retardation and diattenuation of the sample can be obtained from the eigenvalues of $J_c$, as shown in Eq. 4.2.8.

---

Fig. 4.4 OCT profile of a sample obtained from a single balanced photodetector.
4.3 Spectroscopic Analysis

Spectral interferogram of a single reflector as sample is given by Eq. 4.3.1, where $k$ is wavenumber, $S(k)$ is the spectrum of the filtered light from the laser source, $\Delta z$ is the optical path difference between the sample and the reference mirror, and $r_s$ and $r_r$ are amplitude reflectivity of the sample and reference mirror, respectively,

$$i(k) = S(k) \times (r_s^2(k) + r_r^2(k) + r_r r_s \cos(k\Delta z)) .$$  \hspace{1cm} 4.3.1

Fourier analysis of the spectral interferograms provides the depth profile of the sample, illustrated in Fig. 4.5(a), as

$$I(\Delta z) = |F[i(k)]| .$$  \hspace{1cm} 4.3.2

Fig. 4.5 Illustrations of data processing in spectroscopic analysis. (a) Illustration of interferogram acquired by the line scan camera, Gaussian window and windowed interferogram. (b) Depth profiles of a single reflector obtained by Fourier transform of original interferogram and windowed interferogram.
Spectroscopic analysis can be obtained by short time Fourier transform (STFT) [3, 4], Fig. 4.5(b). The spectral interferogram is numerically multiplied by a sliding window function $w(k_0)$ before Fourier transformed to obtain depth profile as

$$I(k_0, \Delta z) = |F[w(k_0) \times i(k)]|, \quad 4.3.3$$

where $k_0$ is the center wavenumber of the window function.

The depth profiles obtained by the windowed Fourier transformed are associated with the center wavenumber $k_0$. Thus spectroscopic information of the sample can be extracted from the resulting depth profiles. Spectroscopic OCT has been demonstrated to measure the concentration of hemoglobin [33], which is important in the diagnosis of various diseases including cancer.
5. Implementation of Spectral Domain OCT

The multifunctional OCT is a spectral-domain OCT system based on the micro-optical coherence tomography [31], as shown in Fig. 5.1. The system is mainly built with bulk optical components due to the lack of availability of broadband wavelength components of its fiber based alternatives. A fiber is used to connect the probe unit and spectrometer unit for the easiness of alignment and the flexibility of probe unit.

A commercial supercontinuum white light laser (SuperK COMPACT, NKT Photonics, Denmark) which emits a broad spectrum ranged from 450 nm to 2400 nm is used as the broadband light source. The spectrum of the source is filtered and shaped by a custom made spectral filter which is designed to output the wavelength in a range from 680 to 940 nm. By using this wavelength range the OCT theoretically provides approximately 1 micrometer axial resolution. The filtered light is collimated, reflected by a beam splitter and then coupled into a fiber. The light is again collimated and its wavefront is split by a 45° rod mirror into the reference arm, which consists of a focusing lens and a reference mirror, and the sample arm. At the sample arm, a dual axis scanning galvanometer mirror is used to scan the probe beam over the sample. The backscattered lights from the reference mirror and sample are coupled back into the fiber, transmitted through the beam splitter and interfered. The interferogram is detected by a spectrometer made of an achromatic collimation lens, a transmission grating, a homemade multi-element focusing lens, and a 4096 × 2 pixels line scan camera (Basler sprint spL4096-140k, Basler AG, German). Prior to the construction, the OCT was assembled in a 3D computer-aided design (SolidWorks, Dassault Systèmes SolidWorks Corp., MA), shown in Fig. 5.2, to prevent mechanical conflation between components.

5.1 Spectral Filter Design

As shown in Fig. 5.1, the custom made spectral filter consists of two prisms and a cylindrical lens. The light source with equipped with a collimator emits collimated beam. The beam is dispersed by a dispersive prism and collimated with a cylindrical lens. An aperture stop is used to block undesired wavelengths completely. The light which passes through the aperture stop is reflected by a right-angle prism back toward the dispersive prism, shown in Fig. 5.1 (side view). The light is combined by the dispersive prism and coupled into a fiber. The output of the fiber is used as the light source of the OCT.
5.2 Spectrometer Design

The design of spectrometer is important to the performance of an OCT. It characterizes the sensitivity, the signal roll-off, and the maximum measurable depth range of an OCT. The sensitivity of the OCT is affected by the efficiency of the spectrometer. The maximum measurable depth range is determined by the wavenumber spacing between pixels shown by the Eq. 3.2.4. The Figure 5.3 show the spectrometer design which consists of a grating, a focusing lens and a line scan camera. The spectral resolution $\delta_r\lambda$ of a grating at a wavelength $\lambda$ according to the Rayleigh’s criterion is given by $\lambda/\delta_r\lambda = mN$ where $m$ is the order of diffraction and $N$ is the illuminated number of groove. High groove density grating was used to achieve high spectral resolution. A volume phase grating (Wasatch Photonics, Utah) with a groove density $G$ of 1200 line per mm (lpmm) was used in the design. The incident angle of the grating was calculated using the grating equation,

$$mg\lambda = \sin \theta_i + \sin \theta_d$$  \hspace{1cm} 5.2.1

where $m$ is the order of diffraction. Since incident angle $\theta_i$ and diffraction angle $\theta_d$ are equal at the Bragg wavelength $\lambda_B$, which was as 840 nm, and the first order of diffraction $m=1$ was used in the spectrometer, the incident angle at the Bragg wavelength for its maximum diffraction efficiency given by $\theta_i = \sin^{-1}(mG\lambda_B/2)$ was calculated to by 30.3°. The grating equation Eq. 5.2.1 expressed by Bragg wavelength of the grating is given by

$$mG(\lambda - \lambda_B/2) = \sin \theta_d.$$ \hspace{1cm} 5.2.2

By normalizing the diffraction angle to the center wavelength $\theta_c = (\theta_{max} + \theta_{min})/2$ by using the diffraction angle of the maximum $\theta_{max}$ and the minimum wavelengths $\theta_{min}$ as $\theta = \theta_d - \theta_c$, the focused location $L$ on the camera by a focusing lens with a focal length of $f$ for each wavelength is $L = f \tan \theta$. The focal length must be chosen to focus the whole wavelength range onto the length of the sensor $\Delta L = f (\tan(\theta_{max} - \theta_c) - \tan(\theta_{min} - \theta_c))$. The designed wavelength range of the OCT was 680 – 940 nm. As a result, the focal length of the lens must be about 111 mm.

5.2.1 Focusing Lens Design

Figure 5.4 shows the multi-element focusing lens of the spectrometer designed with the optical design software Code V (Synopsys, CA). The software was used to minimize
focused spot size on each pixel of sensor and to fit the whole range of wavelengths into the length of sensor. Only stock lenses were used in the design to lower the cost. The starting point of the design was to use achromatic doublets to correct the chromatic aberration and two meniscus lenses were placed at both sides of the doublets to correct aberrations. The achromatic doublets were chosen from a list of stock lenses. Two or three same doublets were stacked together as a single doublets to obtain the desired focal lengths which were not in the list of stock lenses. Two glass blocks were placed at the both sides of the doublets. The curvatures of the surfaces and thickness of the glass blocks and the separation between elements were optimized by the optical design software. These glass blocks were evolved into meniscus lenses. The meniscus lenses were then split into combinations of plano-convex and plano-concave. After splitting the curvature and thickness of these lenses and the spacing between all lenses were further optimized. By adding constraints of the curvature and thickness in the optimization, the plano-convex and plano-concave lenses which matched the stock lenses were partially replaced. Repeating the optimization and replacement of stock lenses, eventually all lenses could be replaced by stock lenses. The optimization was repeated from choosing the achromatic lenses when the performance was not satisfied. Figure 5.4 shows the result of the designed lens. A line scan camera with two rows of 4096-pixel sensors was used in the design. Each pixel of which is $10 \times 10 \mu m$. The camera is operated as a single row sensor in vertical binning mode so the area of each pixel would be $10 \times 20 \mu m$. The spectrometer was optimized for its efficiency using the spot diagram as the optimization metric. Figure 5.5 shows the optimization result of the airy disc diameters, the root-mean-square (RMS) spot diameters and the geometric (GEO) spot diameters of the spot diagrams of the designed wavelengths.

The airy disc diameters are limited by the focal length of the focusing lens which is determined by the length of the spectrum on the camera. The calculated airy disc diameters of the spectrum were in the range 11 - 15 $\mu m$. The RMS and GEO spot diameters were smaller than the airy disc diameters and pixel size (20 $\mu m$) in the vertical direction (X). No optical power loss is expected in in the vertical direction of the optimized design. However, in the spectral dispersion direction (Y), the Y GEO spot diameter is larger than pixel size (10 $\mu m$) in some wavelength range. This would result in decrease of the wavelength separation between adjacent pixels and degrade signal roll-off in axial direction [8].
Fig. 5.1. Scheme of spectral domain optical coherence tomography
Fig. 5.2. 3D CAD of the OCT system
Fig. 5.3. Spectrometer Design
Fig. 5.4 Lens design of the multi-element focusing lens of the spectrometer.
Fig. 5.5 Airy disc diameters, root-mean-square (RMS) spot diameters and geometric (GEO) spot diameters of the spot diagrams of designed wavelengths. GEO and RMS were also calculated in X and Y directions, which are the vertical direction of the sensors and the spectral dispersion direction, respectively.
5.3 Fluctuation Optical Power in the Supercontinuum Light Source

The supercontinuum light source is generated by launching a pulsed laser into a nonlinear photonic crystal fiber. The pulsed laser was operated at a range of repetition rate, 21 ~ 22 kHz. As shown in Fig. 5.6, the optical power vary at the operated repetition rate.

To evaluate to fluctuation in the light source, the line-scan camera of the spectrometer in the SD-OCT system was used to measure the optical power of the light source. The exposure time of the camera was set to 300 μs in the measurement. Figure 5.7 and 5.8 are measurement without and with synchronization of the camera and the light source. In the case of synchronization, the camera was triggered using the clock signal generated by the supercontinuum light source. Figure 5.7(a) and 5.8(a) show the measurement in time and 5.7(b) and 5.8(b) are the average of each measurement. Figure 5.7(c) and 5.8(c) are the overlap of all measurements, each individual measurement is plotted with different color. The thicker plot in different colors indicates larger fluctuation in the light source. Strips or vertical lines are obviously seen in the Fig. 5.7(a), while Fig. 5.8(a) has periodic variation with small magnitude. Although the mean of the averaged pixel values are different in the two measurement, 483 in Fig. 5.7(b) and 713 in Fig. 5.8 (b), because the beam direction of the light source are slightly different on camera, the standard deviation of the averaged pixel values are 36 in Fig. 5.7(b) and 10 in Fig. 5.8(b), respectively. The low standard deviation indicates that the synchronized measurement has low fluctuation of the optical power in time, although it fluctuates periodically.

![Fig. 5.6. Variation of output power of the supercontinuum light source measured by a photodetector.](image)
Fig. 5.7. Optical power measurement of the light source by a line-scan camera without synchronization.

Fig. 5.8. Optical power measurement of the light source by a line-scan camera without synchronization.
5.4 Wavelength Calibration of Spectrometer

Figure 5.9 shows the zero-crossing points, the band-pass filtered fringes acquired by OCT and commercial spectrometer. The envelopes of the fringes were completely different due to the different optical paths and the absorption of the fiber of the commercial spectrometer. Hence, the shapes of the envelopes were not reliable for the fringe matching. The landmark indicated by an asterisk (*) in Fig. 5.9 was used for the fringe matching. Figure 5.11 shows the original and resampled fringes of the OCT spectrometer. The fringe was resampled evenly, Fourier Transformed and shown in Fig. 5.12. The broadened point spread function was improved and became sharp after the fringe was resampled using this method.

Fig. 5.9. Matching of the band-pass filtered fringes acquired by OCT and commercial spectrometers and zero-crossing points. Asterisk (*) indicates the landmark used for the fringe matching.
Fig. 5.10. Wavenumber of each pixel before and after wavelength calibration.

Fig. 5.11. Original and resampled fringes of the OCT spectrometer

Fig. 5.12. Point spread functions obtained from the Fast Fourier Transform of the original and resampled OCT fringes.
5.5 Sensitivity and Axial Resolution Measurement

The optical power on the sample was 46 μW. The exposure time of the line scan camera was set to 990 μs. A mirror was used as a single reflector for the measurement of the OCT performance. The mirror was placed at the focal point and the reference mirror was fixed at several optical path lengths to quantify the OCT performance at different depth positions. Figure 5.13 shows the measured axial resolution and sensitivity at each mirror position. The mean resolution within 1 mm depth range was measured to be 1.9 μm in air which is close to theoretical value, estimated as 1.1 μm. The sensitivity was estimated to be 91.5 dB at the zero delay point by extrapolating the measured sensitivity. The sensitivity was lower than the theoretical value, estimated as 109.2 dB without the consideration of the efficiency of spectrometer and power loss in the system. The signal decayed at a rate of 16.4 dB/mm within 1 mm depth range.

![Graph showing sensitivity and resolution](image)

Fig. 5.13. Measured sensitivity and resolution

5.6 Measurement of Sweat Ducts in A Middle Finger

Sweat ducts of the tip of a middle finger were measured as a biological sample and a static sample by the ultra-high resolution. The tip of finger was scanned with a 1.0 mm line pattern, because currently the volumetric data was able to be obtained due to motion of the finger. As shown in results of sensitivity in Fig. 5.13, the OCT has stronger signal as the sample approaches the zero delay line, so the zero delay line was placed close to the skin surface for the sweat duct measurement to achieve high sensitivity in the shallow region.

Figure 5.14 shows the measured cross-sectional OCT image of the sweat ducts and the double sided foam tape. Dermis was weakly perceived at the bottom of the OCT
images in Fig. 5.14 (a-b). The sweat ducts inside the epidermis layer was clearly observed beneath the skin surface in Fig. 5.14 (a) and above dermis in Fig. 5.14(b). To the best of our knowledge, this is the first demonstration of ultra-high resolution visualization of sweat duct by OCT.

Fig. 5.14 OCT cross-sectional images of (a-b) sweat ducts and (c) a double sided foam tape.
5.7 Measurement of A Double Sided Foam Tape

An area of 1.0 mm×1.0 mm on the double sided taped was scanned by the ultra-high OCT for the volumetric visualization. A drop of water was dropped onto the tape to prevent strong surface specular reflection. However, the water altered the optical property of the adhesive from transparent to opaque. The zero delay line was placed at the deeper region of the tape.

The cross-sectional OCT image of the double sided foam tape is shown in Fig. 5.14(c). The volumetric visualization of the internal structure of the double side foam tape is shown in Fig. 5.15. The surface of the water droplet was visible on top of the OCT image, in Fig. 5.14(c). The adhesive of the tape appeared as a hyper-reflective layer on top of the foam. Void spaces were observed inside the foam.

Fig. 5.15 Volumetric data of a double sided foam tape.
6. Implementation of Multi-functional OCT

6.1 Doppler and Polarization Measurement by Swept Source OCT

As shown in Fig. 6.1, the MF-OCT is based on a fiber-based Mach-Zehnder interferometer. A 100-kHz wavelength sweeping laser (Axsun Technologies Inc, MA), which sweeps over a spectral range of 109 nm around a center wavelength of 1.06 μm, was utilized. The light is split by a fiber coupler by a power ratio of 90:10 into two arms: a sample arm and a reference arm.

In the sample arm, a polarizer (GL5-B, Thorlabs, NJ) and a polarization beam splitter (NT49-870, Edmund Optics Inc., NJ) are used to split the light into two mutually orthogonal linear polarization states. Both states are reflected by Dove prisms (PS992, Thorlabs) and propagate in two different optical paths. By translating one of the Dove prisms, the delay between the two incident polarization states can be arbitrarily controlled. The two polarization states with different delays are recombined by a second polarization beam splitter (NT49-870, Edmund Optics Inc.) and coupled into an 80:20 fiber coupler. 80% of the light which traveled into the fiber coupler is then focused on the sample. The power of the probe beam was adjusted to be 1.6 mW at the cornea, which is lower than the ANSI standard safe exposure limit. The backscattered light from the sample is collected and directed into a polarization diversity detection arm.

Using two fiber collimators, the reference arm collimates the light and couples the light into a fiber which is connected to the detection arm. The optical path length of the reference arm is delayed by translation of the fiber collimator which is used to couple the beam. In the measurement, the optical path length of the reference arm is adjusted in order to make position of sample to be within the imaging range.

The light beams from the sample arm and the reference arm are recombined using a non-polarization beam splitter (BS; NT47-124, Edmund Optics Inc.) in the detection arm. This combined light is then split into two orthogonal states of polarization by two polarization beam splitters (PBS; NT49-871, Edmund Optics Inc.). The interferograms of the two polarization states are detected using two balanced photodetectors (PDB430C, Thorlabs) for the data processing. Prior to the BS, a polarizer is used to ensure equal reference power after the PBSs, as described by Eq. 4.2.15.
Fig. 6.1 Scheme of the MF-OCT. A/D: Analog-to-digital converter, BS: Beam splitter, FC: Fiber collimator, PBS: Polarization beam splitter, PC: Polarization controller, P: Polarizer.
6.1.1 Sensitivity and Axial Resolution Measurement

A mirror was used as a reflector sample in the measurement of the system performance. The mirror was used to reflect the probe beam and to couple the beam back into the collimator of the probe. The optical power of the coupled light was maximized and a neutral density filter was inserted between the mirror and the collimator to attenuate the reflection power. Without scanning the probe beam, a series of measurements were performed at several optical path lengths of the reference arm. The optical path lengths of the reference arm were recorded to calculate the physical length of each pixel in depth, and the FWHM of the mirror signals were measured as axial resolution. Sensitivities at several depths were calculated from signal-to-noise ratio (SNR) and the attenuation of the neutral density filter.

Figure 6.2 shows the measurement of the axial resolution and depth in pixel at several physical depths. Averaged axial resolution defined by the width at -3 dB of the point spread function was measured to be 8.3 µm in air. Physical length of each sampling point in was measured to be 5.17 µm/pixel, which is the slope of depth in pixel plot. As shown in Fig. 6.3, the sensitivity was measured at several depths. The sensitivity was measured to be 84.6 dB at zero depth, and the roll-off of the sensitivity was -1.45 dB/mm.

In the measurement, each spectrum was sampled with 1440 points. Because of the Nyquist-frequency of the sampling theorem, half of the sampling point can be used for a conventional OCT. As shown in Fig. 4.4 the PS-OCT uses only half of the conventional OCT imaging depth which is 360 pixels in the current setup. Using the measured physical size of each pixel, imaging depth of the PS-OCT is 1.86 mm. Half of the conventional OCT imaging depth range in current setup is expected to be 1.86 mm. In the data processing, Hanning window was applied to each spectrum prior to discrete Fourier transform. The resultant spectrum became narrower, and the effective FWHM of the spectrum was expected to be 55 nm. According to Eq. 3.2.3, the axial resolution is expected to be 9.0 µm. Theoretically the sensitivity is 96.4 dB with consideration of optical power loss in the system.

The measured axial resolution and imaging depth agree well with the expected value, a small discrepancy could be accounted by the departure of the spectral shape from Gaussian spectrum. However, the measured sensitivity was 11.80 dB lower than expected value. Although the measured sensitivity is not very high, it is possible to perform sample measurement with the current setup. One possible reason for the low
sensitivity is that the OCT detection is not shot noise limited, and this problem can be solved by using optical components which have less optical power loss.

6.2 Polarization Measurement of Polarizer and Eighth-wave Plate

A polarizer and an eighth-wave plate were measured as standard samples to quantify the capability Jones matrix measurement. Each standard samples was measured and rotated from $0^\circ$ to $180^\circ$. 2048 measurements without scanning the probe beam was obtained for each orientation.

As shown in Fig. 6.4 and 6.5, the diattenuation of the polarizer was measured to be $1.00 \pm 0.00$ and the double pass phase retardation of the wave plate was measured to be $88.5 \pm 3.7^\circ$. The measured relative axis orientation of each sample was fitted to a linear regression line. The slope of the fitted lines became 0.99 with $R^2 = 1.00$, and 0.97 with $R^2 = 1.00$ for the measurement of polarizer and eighth-wave plate, respectively.

The expected results for the diattenuation of the polarizer and the double pass phase retardation of wave plate are 1 and $90^\circ$, respectively. As a conclusion, the measured results showed that the PSOCT is capable of high accuracy measurement of diattenuation, double pass phase retardation and relative optic axis orientation.
Fig. 6.2 Axial resolution and depth at several physical depths.

Fig. 6.3 Sensitivity measurement.
Fig. 6.4 Diattenuation and relative axis orientation measurement of a polarizer

Fig. 6.5 Double pass phase retardation and relative axis orientation measurement of a eighth-wave plate
6.3 In vivo Polarization Measurement

A retina without marked posterior disorder was scanned in vivo by the swept-source OCT, where each volume consists of 1024 B-scans and each B-scan consists of 512 depth-scans. Four B-scans were acquired at a single location to for Jones matrix averaging [30]. Figure 6.6 shows the en face images of intensity (a), phase retardation measured at a retinal pigment epithelium (b) obtained from a single measurement of the optic nerve head (ONH). Figure 6.7 (a) is the intensity average of 4 elements of a Jones matrix, and Fig. 6.7 (b) has been obtained from averaged Jones matrix. Figures 6.7 (a) and (b) are the cross sections of the intensity and the phase retardation at the horizontal line in Fig. 6.6 (a).

The structure of the ONH was visualized by the intensity images, Figs. 6.6 (a) and 6.7 (a). As shown in Fig. 6.6 (b), bow-tie pattern of moderate high phase retardation was observed in the en face image of phase retardation. The bow-tie pattern is an appearance of thick retinal fiber layer (RNFL), and this pattern can also be observed by the previously developed PS-OCT system and other polarization imaging apparatus [6]. As indicated by arrows in the cross section image, high phase retardation was observed in the sclera and the lamina cribrosa which are birefringent tissues consist of collagen fibers.

In comparison to conventional OCT which generally utilizes single polarization state for the imaging, fully measuring polarization in tissue, PS-OCT provides polarization insensitive intensity images and free of birefringent artifact. As shown in the phase retardation images, contrast was enhanced in birefringent tissues, and this additional contrast provides useful information for the tissue discrimination. In summary, PS-OCT simultaneously visualizes structure of sample and enhances contrast on birefringent tissues.
6.4 *In vivo* Doppler Measurement

Doppler flow was calculated from the same volumetric data measured in the *in vivo* polarization measurement. Figure 6.6 (c) and (d) show the vasculature of the retina visualize using $\Delta \varphi^2$. Figure 6.6 (d) shows a magnified image of the area indicated by a black box in Fig. 6.6 (c). In $\Delta \varphi^2$ images, Figs. 6.6 (c) and (d), detailed vasculature can be seen. Small vessels, indicated by arrows, were clearly observed. In the cross-section intensity image, structures such as retina and choroid can be clearly identified. Asterisk marks (*) in both intensity images indicate the position of a blood vessel. Absorption of blood vessel shows low signal in the intensity image. Because of the absorption, large vessels were observed in the *en face* image, Fig. 6.6 (a).
Fig. 6.6 En face images of (a) intensity, (b) phase retardation, and (c) power of Doppler shift, of an optic nerve head. (d) Magnified image of the square area in (c). [12]
Fig. 6.7 Cross sectional images of (a) Intensity and (b) phase retardation taken from the dashed line in Fig. 6.6(a)
6.5 Spectroscopic Measurement of Intralipid

Intralipid is lipid emulsion which has been used as a material to mimic biological tissue. Hence, the optical scattering properties of Intralipid was widely studied. Intralipid-20% solution was diluted with purified water in 10 v/v % and 50 v/v % for the spectroscopic measurement. The attenuation coefficient \( \mu \) was calculated from the slope of the intensity depth profile in logarithm scale as \( \log(I(z))=-\mu z+\log(I_0) \) in different wavelength. Figure 6.8 shows the measured result. The spectrum of the 50% solution is approximately 5 times of the spectrum of the 10% solution, which indicates that the spectroscopic analysis has potential in the measurement of the concentration of a solution.

![Fig. 6.8 Spectroscopic analysis of diluted Intralipid.](image-url)
7. Conclusion

A spectral domain optical coherence tomography (SD-OCT) with 1.9 μm high resolution was constructed using supercontinuum light source. The spectrometer of the OCT consists of a homemade multi-element focusing lenses. The lens was designed using only stock lenses to reduce to cost. The whole OCT system was assembled in a 3D computer-aided design prior to its construction. The tip of a middle finger and the internal structure of a double sided foam tape were successfully imaged using the SD-OCT. The volumetric internal structure of the double sided foam tape was visualized. Sweat ducts were observed in the OCT images at very high resolution.

Polarization measurement requires modification of OCT hardware, so a PS-OCT has been implement by a swept-source OCT system. The PS-OCT system was implemented with only passive components and requires no polarization modulation, so advantages such as easy to maintain, high stability are expected. Because of these advantages, the improvement might enable clinical routine use of the PS-OCT in large scale study of eye diseases. Ultimately, by combining additional function, such as Doppler OCT with the PS-OCT to develop multi-functional OCT, wider potential clinical use of OCT is expected.

In conclusion, spectroscopy, polarization and Doppler imaging of OCT were performed. The potential applications of OCT were shown in the measurement of adhesive tapes, skin and retina. Particularly only polarization imaging require hardware modification. In the further improvement, combined functional imaging into a single OCT can be realized using polarization sensitive OCT to enhance the potential application of OCT.
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Publications


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