Optical Coherence Tomography Systems

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1. INTRODUCTION

1.1 Background

Tomographic imaging techniques, such as x-ray computed tomography, magnetic resonance imaging, and ultrasound imaging, have found widespread applications in medicine. These techniques measure different physical properties of the tissue and have resolution and penetration depth that prove advantageous for specific applications. [1] Biomedical research and clinical medicine are always in search of fast, portable, inexpensive imaging techniques that do not require ionizing radiation, radioactive isotopes, or removal of tissue to detect early signs of illness. One intriguing possibility would be to convert safe, cheap optical technology into a means of getting those images. But most of the existing optical techniques were not easy to be used for imaging the body, since most tissues are at least partially opaque to most wavelengths of light.

Passing light through a tissue, however, is only one way of obtaining an image. Another is to exploit a branch of optics ---- coherence interferometry to measure weak reflections of light off structures within the body. [2] A new imaging modality based on low-coherence interferometry, Optical Coherence Tomography (OCT), is a noninvasive, non-contact method using infrared radiation to provide cross-sectional images of biological tissues with high spatial resolution on the order of ~10 μm. This technique is analogous to ultrasound, in which acoustic waves bounce off tissue and are translated into images with a resolution of a few hundred microns. Optical coherence tomography, however, depends on the interaction of light waves and tissue to create images, and it does not require a conducting medium between the probe and the tissue. It can image through air or water. [3]
1.2 Motivation

Periodontal diseases are plaque-induced disorders that result in loss of connective tissue attachment and alveolar bone. An important aspect of periodontal disease assessment is determining the location of the soft tissue attachment to the tooth surface. Currently, mechanical or pressure sensitive probes are used to assess periodontal conditions. These probes can be painful for the patient and have several sources of error resulting from variations in insertion force, inflammatory status of tissue, and anatomical tooth contours. OCT is not sensitive to these errors and thus should be a more reproducible and reliable method for determining attachment level. Moreover, directly imaging tooth and soft tissue structures and contour in vivo may provide information that would allow diagnosis of periodontal diseases before attachment loss occurs. The periodontal applications of OCT could be expected in following aspects:

(A). Detection of caries: Dental caries are a common disease that can be easily treated if detected early enough. If undetected and untreated, caries may progress through the outer enamel layer of a tooth into the softer dentin, requiring extraction of the tooth or causing inflammation of the periodontal tissue surrounding the tooth. The standard methods for detecting caries in teeth are by visual inspection or by the use of dental x-rays. Both methods are unreliable for the detection of small caries. In addition, dental x-rays subject the patient to ionizing radiation, a known mutagen. OCT imaging offers a safe, noninvasive alternative for locating potential and actual sites of caries incursion and therefore improves early disease detection and treatment.

(B). Restoration placement/evaluation: Dental restorations are used to provide a barrier restricting oral fluids and bacteria from entering through the tooth into the
systemic system as a result of dental decay or trauma. An inadequate seal can result in a loss of tooth structure, infection, and dissemination of bacteria. The most commonly used methods for evaluating the seal and structural integrity of restorations is visual and tactile examination. OCT has the advantage over these methods of visualizing structural and marginal restoration defects before significant leakage occurs, minimizing tooth loss and decreasing the number of unnecessary replacement restorations. [4]

This thesis research is part of a 4-year research project funded by NIH for periodontal applications of optical coherence tomography technique. The task of this thesis research is to develop prototype OCT systems, which could be used to study the principles of low-coherence interferometry, and could be finalized to be clinically applicable, consisting of a small, compact, self-contained unit that will image the entire tooth-bearing regions within the oral cavity.

1.3 Fundamental theory of interferometry [5]

1.3.1 Interference by plane waves

Interference occurs when radiation follows more than one path from its source to the point of detection. The simplest example of interference is that between plane waves (Figure 1.1). For two waves that have complex amplitude of the form

\[ U(\vec{x}) = a e^{jkr \cdot \vec{x}} , \]

(1.1)

(where \( a \) is a constant, \( \vec{r} \) is a unit ray vector, and \( \vec{x} \) is a position vector of some point in the wavefront), the total amplitude at a position \( \vec{u} \) in the plane of observation is

\[ U(\vec{u}) = a_1 e^{jk\vec{r}_1 \cdot \vec{u}} + a_2 e^{jk\vec{r}_2 \cdot \vec{u}} , \]

(1.2)
where the position vector $\vec{x}$ has been replaced by the vector $\vec{u} = \vec{x} - \vec{x}_0$. The intensity at $\vec{u}$ is

$$I(\vec{u}) = a_1^2 + a_2^2 + 2a_1a_2 \cos k[(\vec{r}_2 - \vec{r}_1) \cdot \vec{u}]. \quad (1.3)$$

The intensity is made up of a uniform background $a_1^2 + a_2^2$ on which is superimposed a cosinusoidal variation

$$G(\vec{u}) = 2a_1a_2 \cos k[(\vec{r}_2 - \vec{r}_1) \cdot \vec{u}] \quad (1.4)$$
This variation $G(\mathbf{u})$ is the *interferogram*.

The vector $\mathbf{r}_2 - \mathbf{r}_1$ is a measure of the angle between the two interfering waves.

Denote $\mathbf{\theta}$ as the angular *tilt* between the interfering beams, that is

$$
\mathbf{\theta} = \mathbf{r}_2 - \mathbf{r}_1. \quad (1.5)
$$

In terms of this vector,

$$
G(\mathbf{u}) = 2a_1 a_2 \cos k \mathbf{\theta} \cdot \mathbf{u} = 2a_1 a_2 \cos \left(2\pi \mathbf{\theta} \cdot \mathbf{u} / \lambda \right)
$$

(1.6)

The contrast of interferogram is given by the *visibility*, defined as

$$
\xi = (I_{\text{max}} - I_{\text{min}})/(I_{\text{max}} + I_{\text{min}})
$$

(1.7)

For the interference described by (1.3),

$$
\xi = \frac{2a_1 a_2}{a_1^2 + a_2^2} = \frac{2I_1^{1/2} I_2^{1/2}}{I_1 + I_2}, \quad (1.8)
$$

where $I_1$ and $I_2$ are the intensity due to each beam alone. The visibility is a maximum with a value of unity, when $I_1 = I_2$.

When the two interfering waves are traveling in the same direction, the observed intensity can be expressed as

$$
I(R) = a_1^2 + a_2^2 + 2a_1 a_2 \cos 2\pi R / \lambda, \quad (1.9)
$$
where $R = R_2 - R_1$ is the difference between the two distances $R_1$ and $R_2$ traveled by the wave. The part that varies with the path difference is

$$G(R) = 2a_1a_2 \cos 2\pi R / \lambda = 2a_1a_2 \cos 2\pi \nu \tau$$

(1.10)

where $\tau$ is the delay, the optical path difference $p = nR$ divided by the speed of light,

$$\tau = p / c$$

(1.11)

The interferogram $G$ is now a function of the path difference, or the delay, rather than the position $\vec{u}$ in the plane.

1.3.2 General waves

When two waves represented by the general equation

$$U(\vec{u}) = a(\vec{u}) \exp \left[ k\vec{r} \cdot \vec{u} - \psi(\vec{u}) \right]$$

(1.12)

interfere, the resulting intensity is

$$I(\vec{u}) = a_1^2(\vec{u}) + a_2^2(\vec{u}) + 2a_1(\vec{u})a_2(\vec{u}) \cos \left[ k\vec{\theta} \cdot \vec{u} + \psi_2(\vec{u}) - \psi_1(\vec{u}) \right]$$

(1.13)

where $\psi_2 - \psi_1$ is the phase difference between two waves.

When the interference is between an unknown wave of amplitude $U_2(\vec{u})$ and a uniform plane reference wave of complex amplitude $U_1$, the interferogram has the form

$$G(\vec{u}) = 2U_1|U_2(\vec{u})| \cos \left[ \arg U_2(\vec{u}) + \text{constant} \right]$$

(1.14)
This is, to a constant factor, the real part of $U_2$. The imaginary part is obtained if the phase of the reference wave is changed by $\frac{1}{2}\pi$. Thus the complex amplitude of a wave, which is otherwise unobservable, is made observable by the addition of a reference wave from the same source. Once the distribution of complex amplitude across a wave has been found, it can be related back to whatever caused it: irregularities on a reflecting surface, variations of refractive index in the light path, etc.

### 1.3.3 Heterodyne interferometry

If the two interfering waves have different frequencies $\nu$ and $\nu + \Delta \nu$, the time dependent term $2\pi \nu t$ must be retained in the complex amplitude, which is then

$$U(\vec{r}, t) = a \exp j(k\vec{r} \cdot \vec{x} - 2\pi \nu t)$$

for a uniform plane wave. When two waves of different frequency interfere, the interferogram is

$$G(\vec{u}) = 2a_1a_2 \cos 2\pi \left[\vec{\Theta} \cdot \vec{u} / \lambda - n \Delta \nu (r_1 + r_2) \cdot \vec{u} / c - t \Delta \nu\right]$$

This represents a traveling intensity wave with a frequency $\Delta \nu$, and a detector will convert the interferogram at any point $\vec{u}$ into an alternating electrical signal. In order to maintain constant frequency difference between two waves, practical heterodyne interferometry requires the use of two beams from the same source and a certain method of producing a constant frequency difference between them.
1.3.4 Example: Michelson interferometer

The Michelson interferometer shown in Figure 1.2 is a typical two-beam interferometer with one of the mirrors movable. In Michelson interferometer, one beam splitter is used; the two beams being reflected back to recombine where they started.

![Figure 1.2 Schematic of Michelson interferometer](image)

1.3.5 Coherence time

The Coherence is a statistical description of radiation, expressed in terms of correlation functions. Interference is the simplest example of correlations between light beams.
For the Michelson interferometer described in §1.2.4, the interference appears as a cosinusoidal variation of intensity when expressed as a function of the path difference $p$ or the delay $\tau = p/c$ for the two arms. For radiation of a single frequency $\nu$, this variation has the form $1 + \cos 2\pi \nu \tau$. When the radiation has a range of frequencies, the total intensity is proportional to the integral of this variation over all frequencies. Thus if the source has a spectral line of profile $g(\nu - \nu_0)$, with a bandwidth $\Delta \nu$ and mean frequency $\nu_0$, the intensity can be expressed in terms of the cosine transform $G_c$,

$$I(\tau) = G_c(0) + G_c(\tau) \cos 2\pi \nu_0 \tau$$

(1.17)

As for a single frequency the modulation is still cosinusoidal, but the amplitude now decreases with increasing $\tau$. The delay for which interference fringes may be observed is limited: it is the width of the function $G_c(\tau)$ and this is related to the bandwidth of the source spectral line by the uncertainty relation

$$\Delta \tau \Delta \nu \geq 1.$$  \hspace{1cm} (1.18)

This time $\Delta \tau$ is the *coherence time* of the radiation. A *coherence length* can also be defined as

$$\Delta p = c \Delta \tau,$$

(1.19)

and this can be related to the bandwidth of the radiation, now expressed as a wave number, by

$$\Delta p \Delta \sigma \geq 1.$$  \hspace{1cm} (1.20)
These results show the *temporal coherence* of the radiation, which increases as the bandwidth is decreased.

### 1.3.6 Cross-correlation function

For a two-beam interferometer such as the Michelson interferometer, on one side of the interferometer there is a source of radiation, on the other side there is some plane at which the interference is observed. A point \( P' \) in this latter plane has two images as seen from the source, one from each beam, and these appear at \( P_1 \) and \( P_2 \). In addition, the time taken for radiation from the source to reach \( P' \) by the two paths may differ by an amount \( \tau \). Then, as far as the source is concerned, the radiation arriving at \( P' \) through the interferometer is the sum of amounts that would, in free space, arrive at \( P_1 \) and \( P_2 \) at time \( \tau \) apart through filters of the same transmission as the interferometer. If \( P_1 \) and \( P_2 \) have position vectors \( \vec{u}_1 \) and \( \vec{u}_2 \), the analytic signal at \( P' \) can be written as

\[
V(P', t) = k_1 V(\vec{u}_1, t) + k_2 V(\vec{u}_2, t + \tau), \tag{1.21}
\]

where the factors \( k_1 \) and \( k_2 \) are the amplitude transmission factors for the two beams of the interferometer.

The quantity measured in optics is the intensity, the time average of the square of the real signal. To a factor of 2,

\[
I(\vec{u}) = \langle V^* (\vec{u}, t) \ V(\vec{u}, t) \rangle, \tag{1.22}
\]

so that
\[ I(P') = \left| k_1 \right|^2 I(\tilde{u}_1) + \left| k_2 \right|^2 I(\tilde{u}_2) + 2\Re \left[ k_1^* k_2 \Gamma(\tilde{u}_1, \tilde{u}_2, \tau) \right], \]

(1.23)

Where \( \Re \) denotes the real part, and \( \Gamma \) is defined by

\[ \Gamma(\tilde{u}_1, \tilde{u}_2, \tau) = \langle V^*(\tilde{u}_1, t) \ V(\tilde{u}_2, t + \tau) \rangle. \]

(1.24)

In this definition, \( P_1 \) is regarded as the reference point to which \( P_2 \) is referred.

The function \( \Gamma(\tilde{u}_1, \tilde{u}_2, \tau) \) is the cross-correlation function. It is the correlation function for the essentially random signals \( V(\tilde{u}, t) \) at two points \( \tilde{u}_1 \) and \( \tilde{u}_2 \) at times \( \tau \) apart and is denoted as \( \Gamma_{12}(\tau) \) and its real part as \( G_{12}(\tau) \).

The cross-correlation function is an observable quantity of optics, and has the dimensions of intensity. From it can be defined as the dimensionless degree of coherence

\[ \gamma(\tilde{u}_1, \tilde{u}_2, \tau) = \left\{ I(\tilde{u}_1, \tilde{u}_2, \tau) \right\}^{-1/2} \Gamma(\tilde{u}_1, \tilde{u}_2, \tau). \]

(1.25)

The transmission factors in (1.21) can be eliminated by using the intensities \( I_1(P') \) and \( I_2(P') \) that are produced at \( P' \) by each beam separately; \( I_1(P') = \left| k_1 \right|^2 I(\tilde{u}_1), \) etc.

Then

\[ I(P') = I_1(P') + I_2(P') + 2[I_1(P')I_2(P')]^{1/2} \Re \left[ \gamma(\tilde{u}_1, \tilde{u}_2, \tau) \right] \]

(1.26)

The expression has the same form as (1.3), a uniform field on which an interferogram is superimposed.
1.3.7 Temporal coherence and quasi-monochromatic radiation

When $P_1$ and $P_2$ coincide, the cross-correlation function reduces to the autocorrelation function $\Gamma(\vec{u}_1, \vec{u}_1, \tau)$ or $\Gamma_{11}(\tau)$, which contains only temporal effects. Its real part represents the interferogram. As it is an autocorrelation function, the Wiener-Khinchin theorem states that its Fourier transform is the power spectrum $g(\vec{u}_1, \vec{u}_1, \tau) \equiv g_{11}(\nu)$ of the radiation. Since $\Gamma$ as well as $V$ is an analytic signal and has no components of negative frequency,

$$g_{11}(\nu) = \int_{-\infty}^{+\infty} \Gamma_{11}(\tau) e^{2\pi j \nu \tau} d\tau \quad (\nu \geq 0)$$  \hspace{1cm} (1.27)

and

$$\Gamma_{11}(\tau) = \int_{0}^{\infty} g_{11}(\nu) e^{-2\pi j \nu \tau} d\nu$$  \hspace{1cm} (1.28)

$g_{11}(\nu)$ is the transform of $G_{11}(\tau)$, the real part of $\Gamma_{11}(\tau)$ that describes the intensity modulation, that is, the interferogram. The interferogram and the power spectrum are thus Fourier transforms of each other, and bandwidth and coherence time are related by the uncertainty relation (1.18).

In addition to the power spectrum $g_{11}(\nu)$, the cross-power spectrum can be defined as the transform of $\Gamma_{12}$,

$$g_{12}(\nu) \equiv g(\vec{u}_1, \vec{u}_2, \nu) = \int_{-\infty}^{+\infty} \Gamma(\vec{u}_1, \vec{u}_2, \tau) e^{2\pi j \nu \tau} d\tau,$$  \hspace{1cm} (1.29)

so that
\[ \Gamma_{12}(\tau) = \int_0^{\infty} g_{12}(\nu) e^{-2\pi j \nu \tau} d\nu. \]  

(1.30)

For quasi-monochromatic radiation \((\Delta \nu \ll \nu_0)\), the cross-power spectrum has an appreciable magnitude only for frequencies close to the mean frequency \(\nu_0\). With a change of origin to \(\nu_0\), we can write

\[ \Gamma_{12}(\tau) = e^{-2\pi j \nu_0 \tau} \int_{-\nu_0}^{\infty} \tilde{J}_{12}(\nu) e^{-2\pi j \nu \tau} d\nu, \]  

(1.31)

where \(\nu = \nu - \nu_0\) and \(\tilde{J}_{12}(\nu) = g_{12}(\nu)\). Then \(\tilde{J}_{12}\) will have appreciable values only around \(\nu = 0\), and its transform \(J_{12}(\tau)\), the interferogram, or the cross-correlation intensity, will have only low-frequency Fourier components, and also is a slowly varying function of \(\tau\), changing little for

\[ |\tau| \ll 1/\Delta \nu. \]  

(1.32)

When the interferometer contains dispersive media, the time difference is no longer constant, but is a function of frequency, \(\tau(\nu)\), and equation (1.30) is not a simple Fourier transform. However, \(\nu \tau\) can be expanded as

\[ \nu \tau(\nu) \approx \nu_0 \tau(\nu_0) + (\nu - \nu_0) \tau_g, \]  

(1.33)

where

\[ \tau_g = \left[ \frac{d}{d\nu} \nu \tau \right]_{\nu = \nu_0} = \left[ \tau + \nu \frac{d \tau}{d\nu} \right]_{\nu = \nu_0} \]  

(1.34)
Hence \( \tau_g \) is a group delay, and \( \tau \) being the phase delay. Equation (1.29) can be rewritten for dispersive media with \( \tau_g \) in the exponent inside the integral, and \( J_{12} \) becomes a function of \( \tau_g \). The center of the interferogram, where the visibility is maximum, is thus given by \( \tau_g = 0 \) rather than \( \tau = 0 \).

The cross-correlation function for quasi-monochromatic radiation can then be expressed as

\[
\Gamma_{12}(\tau) = J_{12}(\tau_g)e^{-2\pi jv_0\tau},
\]

(1.35)

and the degree of coherence as

\[
\gamma_{12}(\tau) = \mu_{12}(\tau_g)e^{-2\pi jv_0\tau},
\]

(1.36)

where

\[
\mu_{12}(\tau) = (I_1I_2)^{-1/2}J_{12}(\tau_g).
\]

(1.37)

Both \( J_{12} \) and \( \mu_{12} \) are functions that vary only slowly with \( \tau \) in comparison with the rapid fluctuations of \( \exp(-2\pi jv_0\tau) \). The modulus of \( \mu_{12} \) is the same as that of \( \gamma_{12} \), namely \( m_{12}(\tau_g) \). If the phase of \( \mu_{12} \) is written as \( \beta_{12}(\tau_g) \), the degree of coherence can be expressed as

\[
\gamma_{12}(\tau) = m_{12}(\tau_g)\exp j[\beta_{12}(\tau_g) - 2\pi v_0\tau].
\]

(1.38)
This expression shows the significance of $\beta_{12}$ as an effective phase difference between the radiation at $P_1$ and $P_2$ over and above the phase due to the time-difference $\tau$, namely $-2\pi \nu_0 \tau$.

If the real part of equation (1.38), which describes the interferogram, is compared with that for monochromatic waves, (1.10), it is seen that the change to an extended quasi-monochromatic source has introduced this phase term $\beta_{12}$, but the cosinusoidal variation, $\cos 2\pi \nu_0 \tau$, is unaltered.

### 1.4 Principle of Optical Coherence Tomography

While interferometry, the heart of optical coherence tomography, has been around since the 1800s, the ramp up from basic principles to high-performance instruments began with the publication of a key paper in *Science* in 1991.\textsuperscript{[1,3]} In its simplest form, as shown in Figure 1.3,\textsuperscript{[6]} OCT uses a Michelson interferometer along with a low-coherence source to detect reflected light from a volume scatterer with a photodiode. To acquire a one-dimensional data (A-scan profile), an axial (longitudinal) depth scan is performed. The two-dimensional OCT image (B-scan) is build up as a series of adjacent axial depth scans (A-scans). Depth is gated by low-coherence interferometry, where the sample is placed in one arm of the Michelson interferometer (sample arm) and a scanning optical delay line, either mechanically or optically, is located in the other arm (reference arm). The reflected light from the sample only coherently interferes with the light from the
reference arm if the path lengths are equal to within the source coherence length. With signal-processing algorithms and related hardware, the interferogram can be translated into images. The longitudinal (depth) resolution of the image is limited by the coherence length of the source, and the lateral (transverse) resolution of the image is limited by the beam spot diameter inside the sample. The two-dimensional scanning measurements map optical scattering as a function of depth and transverse position in an "optical section" of the tissue sample, and the resulting data array can be viewed directly by gray scale or false-color map.

The basic measurement performed in OCT imaging is an interferometric cross-correlation, $J_{rs}(\Delta l)$, (where $r$ denotes reference and $s$ denotes sample), of light returning from the reference and sample arms as a function of the pathlength difference $\Delta l$ between the arms. The interferometric part of the photodetector current, $I_d(\Delta l)$,
(where $d$ denotes photodetector), is proportional to the interferometric cross-correlation as:

$$I_d(\Delta l) = \rho J_{rs}(\Delta l),$$  

where $\rho$ is the detector responsivity given by $\rho = \eta \hat{\lambda}_0 e / hc$. Here $\eta$ is the detector quantum efficiency, $\hat{\lambda}_0$ is the optical source center wavelength, $e$ is the electronic charge, $h$ is Plank’s constant, and $c$ is the free space speed of light.

The interferometric cross-correlation can be expressed as the product of the cross-correlation function $J_{rs}$, which is defined as the complex envelope of the interferometric cross-correlation, and a complex exponential carrier:

$$J_{rs}(\Delta l_g, \Delta l_\phi) = J_{rs}(\Delta l_g) e^{-jk_0 \Delta l_\phi}$$  

Here $\Delta l_g$ and $\Delta l_\phi$ are the group delay and phase delay, respectively, expressed as pathlength differences, and $k_0$ is the center wavenumber of the optical source. The cross-correlation function, $J_{rs}(\Delta l_g)$, is a function of group delay, while the carrier is a function of phase delay. The cross-correlation function can be expressed as the convolution of the autocorrelation function of the optical source, $J_{rr}(\Delta l_g)$, and the amplitude backscatter profile of the sample, $r_s(\Delta l_g)$, which can be thought of as a train of impulses with various amplitudes responding discrete reflections or scattering locations in the sample:
An OCT system with a perfect mirror in the sample arm measures the interferometric autocorrelation of the source, $J_{rr}$, which can be expressed similarly to equation (1.40):

$$J_{rr} (\Delta l_g , \Delta l_\phi) = J_{rr} (\Delta l_g) \otimes r_s (\Delta l_g) e^{-j k_0 \Delta l_\phi}$$  \hspace{1cm} (1.42)

When the path length difference is scanned by a scanning delay line in the reference arm, the photodector response is a time domain signal related to the interferometric autocorrelation by the scan velocity of the delay. The carrier of the detector response signal is related to the carrier of the autocorrelation by the phase delay scan speed, and hence the center heterodyne frequency $f_0$ of the detector response signal can be written in terms of the center wavelength of the optical source spectrum:

$$f_0 = \frac{V_\phi k_0}{2\pi} = \nu_0 \frac{V_\phi}{c} = \frac{V_\phi}{\lambda_0}$$  \hspace{1cm} (1.43)

Here, $V_\phi$ is the scan speed of the phase delay, defined as the time derivative of the phase delay: $V_\phi = \frac{d\Delta l_\phi(t)}{dt}$, and $\nu_0$ and $\lambda_0$ are the center frequency and the center wavelength, respectively, of the optical source. The carrier frequency corresponds to the Doppler shift frequency of the center wavelength component of the reference arm light, and equivalently to the beat frequency of the optical heterodyne detector response. The frequency components of the detected signal, expressed as offset from the carrier frequency $f' = (f - f_0)$, are related to the complex envelope of the autocorrelation.
by the scan speed of the group delay. They can thus be written in terms of the offset frequency $v' = (v - v_0)$, or wavelength components of the optical source:

$$f' = v' \frac{V_g}{c} = \left( \frac{1}{\lambda} - \frac{1}{\lambda_0} \right) V_g$$  

(1.44)

Differentiating equation (1.44) gives the expressions for the required bandwidth of the detector response in terms of the optical source frequency bandwidth $\Delta v$, or wavelength bandwidth $\Delta \lambda$:

$$\Delta f = \Delta v \frac{V_g}{c} = \frac{\Delta \lambda V_g}{\lambda_0^2}$$  

(1.45)

The group delay scan speed $V_g$ is defined as the time derivative of the group delay:

$$V_g = \frac{d\Delta l_g(t)}{dt}.$$  

In the case of a simple scanning retroreflecting mirror, $V_\phi = V_g = 2s$, where $s$ is the velocity of the mirror. When $V_\phi = V_g$, the expression $\Delta f / f = \Delta \lambda / \lambda$ holds true. Note that if the scan is linear, then $V_\phi$ and $V_g$ will be constants, while in the case of a nonlinear scanning delay line, $V_\phi$ and $V_g$ are time-varying functions.

Assuming a Gaussian optical source spectrum (and therefore a Gaussian autocorrelation function), the ideal detector bandwidth corresponds to approximately two times the signal bandwidth. In OCT, as in any optical heterodyne detector, the detected signal-to-noise ratio (SNR) in the shot noise limit is proportional to the optical power illuminating the sample and inversely proportional to the detection bandwidth:
\[ SNR = \frac{\rho P_s R_s}{2eB}, \]  

(1.46)

Where \( P_s \) is the power incident on the sample, \( R_s \) is the reflectivity of the sample, \( B \) is the detection bandwidth. From equation (1.45) and (1.46), the signal-to-noise ratio of OCT in the shot noise limit, using the ideal detection bandwidth, becomes:

\[ SNR = \frac{\rho P_s R_s \lambda_0^2}{4e\Delta \lambda V_g} \]  

(1.47)
2. DESIGN OF OCT SYSTEMS

2.1 Prototype low-coherence interferometry system for OCT study

In order to study the basic principles and specifications of optical coherence tomography, a prototype low-coherence interferometry [8-14] system has been constructed. The schematic of the low-coherence interferometry system is shown in Figure 2.1. The light with a centered wavelength 1550nm from the erbium-doped fiber amplifier source was coupled to a single-mode fiber connected to one arm of a $2 \times 2$ fiber coupler. The

![Schematic of the prototype low-coherence interferometry system for OCT study](image)

Figure 2.1 Schematic of the prototype low-coherence interferometry system for OCT study. O1, O2, microscope objectives.
coupler split the light into beams of equal intensity, which were propagated to the reference arm and the sample arm, respectively. The fiber lengths to the reference arm and the sample arm are initially chosen to be equal. The light propagated to the reference arm was collimated by a microscope objective O1 and reflected by an oscillating mirror which was driven by a 1KHz PZT oscillator and produced 20KHz Doppler frequency to modulate the received signals. At the sample arm, a microscope objective O2 was used to focus the light on the object to be studied and to collect the scattered signal from the object. The sample was placed on an X-Y-Z stage that could be manually moved to allow both lateral and depth scanning of the object. A polarization controller was placed at the reference arm to match the polarizations of lights from both the reference and sample arms to generate maximum interference intensity between reference and sample scattered signals.

An interference signal at the fiber coupler is produced only from scattering sources inside the object whose path length difference from the reference arm is within the coherence length of the source. The interference signal was detected, amplified, and band-pass filtered at the center heterodyne frequency 20KHz. The filtered signal is then displayed and measured with a digital oscilloscope.

2.1.1 Erbium-doped fiber amplifier source

The erbium-doped fiber amplifier (EDFA) is used as a source for the low-coherence interferometry system. EDFA is an optical fiber that can be used to amplify an optical input. Erbium rare earth ions are added to the fiber core material as a dopant in typical levels of a few hundred parts per million. The fiber is highly transparent at the
erbium lasing wavelength of two to nine microns. When pumped by a laser diode, optical gain is created, and amplification occurs. EDFA is also used as a low-coherence source by taking advantage of its broadband spontaneous emission when no optical input is applied. The schematic diagram of EDFA is shown in Figure 2.2. It consists of a 23 m long erbium-doped fiber, which is pumped by a 980 nm pump laser. Erbium atoms embedded in the glass matrix absorb the laser light, and the excited erbium atoms radiate spontaneously at wavelengths near 1550 nm. For sufficiently high pump power, the erbium-doped glass exhibits optical gain at wavelengths near 1550 nm. Since the pump laser is absorbed over the entire length of the erbium fiber, the spontaneous emission having short coherence length generated by the erbium atoms is amplified as it propagates through the fiber. Since the gain of the amplifier can be easily changed by varying the pump laser power, the power output and the coherence length of the source can be easily varied. Output powers in excess of 10 mW has been obtained with ~ 100 mA of pump laser current. The optical isolator at the output improves the source stability from unwanted reflections.

Figure 2.2 Schematic of erbium-doped fiber amplifier (EDFA)
2.1.2 Detector and Amplifier

The Optiphase V500 fiber optic analog receiver is used for detecting and amplifying the heterodyned interference signal.

2.1.3 Band-pass filter

A homemade band-pass filter \cite{15} with center pass-band frequency 20KHz and 40dB gain is used in this system. It is obtained by cascading two identical second-order multi-feedback band-pass filters. The design of circuit which realizes the second-order band-pass filter is shown in Appendix A.1.

2.2 Low-coherence interferometry system with motor scanning

reference arm

Since the prototype low-coherence interferometry system described in §2.1 has fixed reference mirror, which provides only the fundamental phase modulation, continuous depth scanning into the sample can only be performed by manually moving the sample axially with a translation stage. A computer-controlled linear motor (Newport MFN25CC) was added to the prototype low-coherence interferometry system shown in Figure 2.1 to achieve uniform depth scanning by linearly moving the reference arm mirror, as shown in Figure 2.3. With this system, A-scan profiles were obtained for a depth scanning range of several millimeters.
Figure 2.3 Schematic of the low-coherence interferometry system with motor scanning reference arm. O1, O2, microscope objectives. Compared with Figure 2.1, the display is replaced by computer, and also a linear motor is added. The linear motor is controlled by computer with LabVIEW™ program.

The 20KHz phase modulation of the reference arm is produced by the PZT oscillation as in the prototype low-coherence interferometry system shown in Figure 2.1. The linear moving of the linear motor at a maximum speed of 0.3mm/s produces another Doppler shift of maximum 387Hz, which is filtered out by the band-pass filter centered at 20KHz in this system.
2.3 OCT system with motor-galvanometer configuration

OCT imaging requires two-dimensional scanning, one is in the depth direction, the other is in the transverse direction. However, for the low-coherence interferometry system described in §2.2, only A-scan profiles can be measured automatically by linear motor moving. To achieve optical scanning in two-dimension, a transverse scanning is added to sample arm of the low-coherence interferometry system described in §2.2. The schematic diagram of this OCT system with motor-galvanometer configuration is shown.

![Schematic diagram of the OCT system with motor-galvanometer configuration. O1, O2, microscope objectives. S, louder speaker. G, galvanometer. L, lens. The galvanometer and linear motor are controlled by PC.](image)

Figure 2.4  Schematic of the OCT system with motor-galvanometer configuration. O1, O2, microscope objectives. S, louder speaker. G, galvanometer. L, lens. The galvanometer and linear motor are controlled by PC.
in Figure 2.4. The same erbium-doped fiber amplifier (EPDA) at a center wavelength of 1550 nm is used as the low-coherence source. The optical power output and 3dB coherence length of this EDFA at 40mA current are 0.43mW and 25μm, respectively. An optical circulator is introduced into the source arm of the interferometer to recover light lost in previous low-coherence interferometer design that used 50/50 coupler in which half of the light returning from the sample and the reference arm is lost. The three-port optical circulator is a nonreciprocal device that couples light that is incident upon port 1 to port 2 and light that is incident upon port 2 to port 3. The optical circulator reduces light power loss by coupling the light returning toward the source arm to a detector. The optical circulator enables the power-efficient fiber-optic interferometer, together with balanced heterodyne detection, to be constructed.

The light from the source enters the optical circulator OC, and it splits into a reference for the differential receiver input and a signal, which is passed through a 3 dB fiber coupler FC, where the outputs form a Michelson interferometer. The reference arm of interferometer consists of a microscope objective O1, a plane mirror, a loud speaker S, and a linear DC motor. The mirror is glued on the bowl of the loud speaker, which is mounted on the linear motor. The loud speaker is used to provide Doppler shifting frequency and is driven by a sinusoidal wave of 100Hz frequency and 6.2 volts peak-to-peak amplitude. The measured maximum Doppler shifting frequency is 17KHz. The linear moving of the linear motor at a speed of 0.3mm/s produces another Doppler shift of 387Hz, which is much less than the 17KHz peak phase modulation signal generated by the mirror mounted on the speaker bowl, and is filtered out by the 1KHz cutoff frequency high-pass filter used in this system. The depth (Z direction) scan is achieved
by translating the louder speaker and motor assembly. The sample arm consists of a microscope objective $O_2$, a galvanometer scanner $G$, a lens $L$, and the sample. The alignment requirement for the sample arm is that not only the incident light is perpendicular to the flipping axis of the galvanometer mirror, but also the incident light spot is located at the focal point of the lens. With this precise alignment, the transverse (X direction) scanning is achieved by translating the flipping of the galvanometer mirror to a parallel scanning upon the sample without introducing a change in optical path, for the reason that the focus of the lens at the galvanometer mirror side and the infinity at the sample side of the lens are conjugate with respect to the lens.

The data acquisition consists of a dual balanced optical receiver, a homemade high-pass filter, and a PC. A dual balanced differential receiver is used to combine the interference signals while simultaneously rejecting common mode noise that is due to source-intensity fluctuations. Since the modulation of the interference signal generated by cheap loudspeaker is broadband due to nonlinearity of speaker film movement, a high-pass filter is used to take full advantage of the modulated signal. We have found that the high-pass filter tracks the peak Doppler frequency better than a band-pass filter, which is regularly used in OCT. The cut-off frequency of the high-pass filter is 1KHz, and the gain is 40dB. The filtered signals are sampled by an A/D converter of 400KHz sampling frequency, and the sampled data are Hilbert transformed to obtain the envelope of the Doppler signals. An A-scan line is obtained from this envelope signal.

The galvanometer and the linear motor are controlled by a galvanometer driver and a DC motor driver, respectively. The flipping of the galvanometer and the translation of the motor are synchronized by the PC control software. A 10Hz sine wave...
generated by the PC is used to drive the galvanometer controller, which in turn drives the scanning mirror. One set of B-scan data is acquired for 128 cycles of galvanometer flipping and 3.84 mm translation of the reference arm in depth, resulting in a period of 12.8 seconds for acquiring one set raw data of B-scan image. For the images obtained, the transversal scanning range is set to 6 mm, and a total of 400 pixels are used. Thus the pixel size at transversal direction (X) is 15μm, which is less than 3dB spot size (~30μm) of the sample beam. The pixel number at longitudinal direction is 256, twice the scanning frequency ratio 128, and the pixel size at depth direction (Z) is 15 μm, which is less than 3dB coherence length (25μm) of EDFA source at 40mA driving current.

2.3.1 Dual balanced receiver

A Newfocus 2017 dual balanced receiver is used in detecting heterodyned interference signal. The auto-balancing circuit of this receiver uses a low-frequency feedback loop to maintain automatic DC balance between signal and reference arms. In effect, the circuit behaves as a variable-gain beamsplitter. This, in conjunction with the subtraction node, cancels common-mode laser noise with greater than 50dB rejection at frequencies less than 125 KHz.

2.3.2 High-pass filter

A homemade fourth-order high-pass Butterworth filter \[^{[15]}\] with cutoff frequency of 1KHz is used in this system. It is obtained by cascading two second-order Butterworth high-pass filters. The design of circuit which realizes the fourth-order high-pass filter is shown in Appendix A.2.
2.4 OCT system with double-galvanometer configuration

The OCT system shown in §2.3 has been used to image various samples including teeth in vitro. Being acceptable in prototypical study for OCT principles and image evaluation, the OCT system shown in §2.3 is suffered from slow imaging speed, e.g. for a B-scan image with pixel number 256*400, the data acquisition takes 12.8 seconds, not even including the post-processing time for image display. A new OCT setup toward faster data acquisition (1.28 seconds of data acquisition time for a B-scan image with

![Diagram of OCT system with double-galvanometer configuration](image)

Figure 2.5  Schematic of the OCT system with double-galvanometer configuration. O1, O2, microscope objectives. G1, G2, galvanometers. L1, L2, lenses. Both galvanometers are controlled by PC.
pixel number 256*400) by using double-galvanometer configuration has been setup. The schematic of this OCT system is shown in Figure 2.5. The only difference between this double-galvanometer configuration and the motor-galvanometer configuration shown in Figure 2.4 is the reference arm. The reference arm consists of a microscope objective O₁, a galvanometer scanner G₁, a lens L₁, and a mirror. In the previous motor-galvanometer configuration, the phase modulation of the reference signal is performed by mechanically vibrating reference mirror, and the longitudinal ranging is achieved by translating the reference mirror assembly. In this double-galvanometer configuration, both of the phase modulation and the longitudinal ranging of reference arm are obtained by off-center alignment of incident light with respect to the rotation axis of the scanner, as shown in Figure 2.8.

The phase modulation generated by the galvanometer scanner is centered at several KHz when G₁ is driven by a sinusoidal wave of 1/1.28Hz frequency and 0.34v peak-to-peak amplitude. The center frequency and the bandwidth are not stable at the whole range of scanning due to the reason that will be described later; therefore, the same high-pass filter as in the motor-galvanometer configuration is used.

The reference arm and sample arm galvanometer scanners are controlled by two galvanometer drivers separately. The flippings of the galvanometers are synchronized by the PC control software. A 100 Hz sine wave generated by the PC is used to drive the sample arm galvanometer scanner. The other 1/1.28 Hz sine wave generated by the PC is used to drive the reference arm galvanometer scanner. One set of B-scan data is acquired for 128 cycles of sample arm galvanometer flipping and 1 cycles of reference arm galvanometer flipping. The transversal scanning range can be easily changed by
changing amplitude of galvanometer driving signal. However, the longitudinal scanning range is set to 3.0mm due to the limitation described later. For the images obtained, regularly the transversal scanning range is set to 6 mm, and a total of 400 pixels are used. Thus the pixel size at transversal direction (X) is 15μm, which is less than 3dB spot size (~30μm) of the sample beam. The pixel number at longitudinal direction is 128, and the pixel size for depth direction (Z) is 23 μm, which is less than 3dB coherence length (25μm) of EDFA source at 40mA driving current.

2.4.1 Group delay and phase modulation by galvanometer scanning

Consider the reference arm with shifted galvanometer scanner configuration as shown in Figure 2.6. The incident light upon the galvanometer scanner is shifted a distance $\delta$ with respect to the axis of galvanometer. The flipping angle of the galvanometer is $\pm \beta$, and the focal length of the lens is $f$. When the galvanometer mirror is flipping, the light reflected by the galvanometer mirror could be approximately considered as emitting from an equivalent source at point $A$. The point $A$ is set at the focus of the lens; hence, the conjugate image point of source $A$ is at infinity. Therefore, in terms of optical path, $\overline{ABM_0} = \overline{ACM}$, as governed by the principle of equal-pathlength.

For the setup shown in Figure 2.6, the optical path length difference between the edge beam and center beam for a single trip is
Figure 2.6  Schematic of shifted galvanometer scanner setup of reference arm for group delay and phase modulation.

\[
\Delta p = \overline{ICM} - \overline{IBM}_0 \\
= (\overline{ICM} - \overline{ACM}) - (\overline{IBM}_0 - \overline{ABM}_0) \\
= -\overline{BC} + (\overline{AB} - \overline{AC})
\]  \hspace{1cm} (2.1)

Since

\[
\overline{OC} = \frac{\delta}{\sin(\pi / 4 + \beta)}.
\]  \hspace{1cm} (2.2)
\[
\overline{CD} = \overline{OC} \cdot \sin(\beta) = \frac{\delta \cdot \sin(\beta)}{\sin(\pi / 4 + \beta)}
\]  \hspace{1cm} (2.3)

\[
\overline{BC} = \sqrt{2} \cdot \overline{CD} = \frac{\sqrt{2} \delta \cdot \sin(\beta)}{\sin(\pi / 4 + \beta)} \approx 2\delta \beta
\]  \hspace{1cm} (2.4)

\[
\overline{AB} = \frac{\overline{BC}}{\tan(2\beta)} \approx 2\delta \beta \frac{\cos(2\beta)}{\sin(2\beta)}
\]  \hspace{1cm} (2.5)

\[
\overline{AC} = \frac{\overline{BC}}{\sin(2\beta)} \approx 2\delta \beta \frac{1}{\sin(2\beta)}
\]  \hspace{1cm} (2.6)

The Equation (2.1) can be solved as

\[
\Delta p = -2\delta \beta + 2\delta \beta \frac{\cos(2\beta) - 1}{\sin(2\beta)}
\]

\[
= -2\delta \beta (1 + \tan(\beta))
\]  \hspace{1cm} (2.7)

Denote \( \alpha = 2\beta \), which is the total flipping angle of the galvanometer, the

Equation (2.14) can also be expressed as

\[
\Delta p = -\delta \alpha (1 + \tan(\alpha / 2))
\]  \hspace{1cm} (2.8)

and \( l_g = 2\Delta p \) gives the group delay that the mirror flipping can produce. For a

flipping angle of \( \alpha = 10^\circ \), the comparison between theoretical group delay values

calculated by Equation (2.8) and actual values versus \( \delta \), the incident light shift, is

shown in Figure 2.7. Although group delay can be obtained up to about 3.8mm for a
Figure 2.7 Group delay generated by galvanometer scanning with shifted incident light. Horizontal axis, offset (mm) of incident light upon galvanometer mirror with respect to the mirror flipping axis. Vertical axis, group delay range (mm) generated by the galvanometer mirror flipping at a total flipping angle of 10°. Stars, experimental values; line, calculated values.

δ of 10mm, regularly 3mm range (δ is 8mm) is used in image acquisition to avoid possible spot loss due to bigger offset from the galvanometer axis.

The phase delay at the reference arm is also generated by the shifted galvanometer flipping, and is given by $l_\phi = 2\pi \frac{\Delta p}{\lambda_0}$. For a linear change in angle $\beta$ as a function of time $\beta = \Omega t$, where $\Omega$ is a coefficient, the phase modulation frequency $f = \frac{dl_\phi}{dt}$ could be centered at $4\pi \delta \Omega / \lambda_0$. 
From Equation (2.7), it can be seen that $\Delta P$ consists of one term of $\beta$ and one term of $\beta^2$. For a linear change in angle $\beta$ as a function of time $\beta = \Omega t$, where $\Omega$ is a coefficient, both the derivatives of group delay and phase delay have the time-variance term, therefore, the group delay speed and the phase modulation frequency are not uniform through one cycle of galvanometer flipping. Furthermore, there is error considering the equivalent light source at point $A$, this results in a limited group delay range of about 3 mm for a $\delta$ of 8mm.

2.4.2 Flipping ratios of galvanometer scanners

The advantage of this system is that the imaging speed is much higher (1 order) than the motor-galvanometer system described in §2.3. Since both the group delay and the phase modulation are provided by the flipping of reference mirror, at a frequency ratio of 128, one galvanometer scanner can be driven at 100Hz (The load free resonant frequency of the galvanometer scanner is 135. For a mirror load, up to 85% of free resonant frequency sine wave can be used as the driving signal), the other one can be driven at 1/1.28Hz. With these driving signals, one frame of B-scan image can be acquired in 1.28 seconds excluding later data analysis and display time, which is 10 times faster than the motor-galvanometer system described in §2.3.

One frame of B-scan image can be achieved in two options, fast depth (Z) scan slow transverse (X) scan, or slow depth (Z) scan fast transverse (X) scan. For fast Z, due to the relatively fast flipping, the signal tends to be more non-uniform along the entire path ranging (large at edge and small at center) because of higher time derivative gradient
of group delay and phase delay, and also the signal strength is about 10dB less at 100Hz than at 10Hz and lower. Therefore, slow Z fast X option is preferred to get better reference ranging uniformity, better phase modulation in terms of bandwidth, and higher signal-to-noise ratio.

### 2.5 Software programming and data acquisition

For all the 4 systems described above, National Instrument LabVIEW™ has been used for programming control, data acquisition and imaging display softwares. The softwares control the data acquisition from an A/D converter at maximum sampling frequency of 1.25 MHz. The LabVIEW™ program and embedded Matlab™ program process image data and display A-scan profile or gray-scaled B-scan images. The Matlab™ programs for windowing the raw image data after Hilbert transformation done by LabVIEW™ and displaying the image are attached in Appendix A.3.

### 2.6 Summary

Four experimental, low-coherence interferometry and optical coherence tomography systems with gradually improved specifications have been presented. The first one, prototype low-coherence interferometry system, provides basic concepts of low-coherence interferometry; the second one, low-coherence interferometry system with motor scanning reference arm, is capable of doing single linear A-scan; the third one, motor-galvanometer configured OCT system, has been used intensely to perform multiple scanning objectives; and the fourth one, double-galvanometer configured OCT system, images with a relatively high speed. Nonetheless, clinical OCT imaging requires
fast speed, high resolution, linear scanning, narrow band modulation, etc., to facilitate
diagnosis and potential functional imaging. [16-19] Unfortunately, none of the above
mentioned systems could strictly achieve all of the clinical OCT imaging criteria. System
specification improvement stemming from a better source, different reference arm
scanning strategy, and compact probe for \textit{in vivo} scanning will be accomplished later.
3. RESULTS

3.1 Coherence properties of the source

With the prototype low-coherence interferometry system shown in Figure 2.1, the trade off between the coherence length, side lobes \[^{20-22}\], and source power had been studied. These three parameters are very important to characterize OCT imaging quality.

The measured coherence length and sidelobes as a function of pump laser current are shown in Figure 3.1 and 3.2, respectively. Clearly, there is a trade-off between high image quality (shorter coherence length and low sidelobe level) and high source power needed for better signal-to-noise ratio. Since the study on source coherent properties, as well as the relationship between these properties and source power, has not been

![Figure 3.1](image)

Figure 3.1 3dB coherence length as a function of source current. Horizontal axis, source current (mA). Vertical axis, 3dB coherence length (μm) of the source. Stars, experimental values; line, linear fit.
Figure 3.2  Sidelobe level as a function of source current. Horizontal axis, source current (mA). Vertical axis, sidelobe level (V) of the source. Stars, experimental values; line, linear fit.

documented well in the literature, the results obtained in this study have been reported at the Optical Society of America Biomedical Topical Meeting, April 2000, in Miami Beach, Florida.\textsuperscript{[23]}

Since superluminescent diode (SLD) sources have high output power and are commonly used in OCT imaging systems, the coherence properties of the SLD source, which is going to be used in the clinical application along with this OCT system, has been compared with that of erbium-doped fiber amplifier source.

Figure 3.3 shows the measured 3dB coherence length of SLD source vs. source power. SLD source is expected to have constant coherence length determined by its spectrum properties. The deviations from constant value (line) at lower and higher source powers were due to measurement errors. The coherence length of SLD does not change
Figure 3.3 3dB coherence length of Superluminescent Diode source as a function of source power. Horizontal axis, source power, which is represented by the bumping current (mA) of EDFA source at the same output power. Vertical axis, 3dB coherence length (μm) of the source. Stars, experimental values; line, ideal performance per specification.

with the source current, and SLD source has shorter coherence length (12μm) than that of erbium-doped fiber amplifier source, which justifies the use of SLD source for our OCT system.

The effects of source coherence properties on imaging can be seen in Figure 3.4. The A-scan profiles scanning at same location with different EDFA source currents are obtained using the low-coherence interferometry system with motor configuration shown in Figure 2.4. The sample is an extracted tooth, and the depth scan direction is left to right, as shown by the arrow at the bottom of the figure. The higher peak at the left corresponds to the air-enamel interface, and the lower peak at the right is the sidelobe
artifact of the air-enamel interface signal. As the source current goes up, the coherence length increases, resulting in worse resolution, which can be seen by the gradually lost details just underneath the air-enamel interface. However, as the source current goes up, another signal peak is appearing at about 1mm depth, which demonstrates that increasing source power will result in penetration and signal-to-noise ratio improvements. There is clearly a trade-off between resolution and signal-to-noise ratio.

### 3.2 Detection configuration study

Detection circuitry’s signal-to-noise ratio (SNR) has been investigated. High SNR ensures high system sensitivity which is essential for weaker target detection.
Firstly, a single end detection circuitry is implemented in which the detection fiber in Figure 2.1 was directly connected to a receiving amplifier. The receiving amplifier was carefully designed to band-pass the desired 20 KHz Doppler signals. However, it was found that this circuitry was very sensitive to environmental noise and the dynamic range was not high enough to show details of extracted teeth sample. The detection circuitry has been improved by using a dual balanced detection (shown in Figure 2.4) in which two fibers one from detection and one splitted from source were connected to a dual balanced receiving amplifier. The common mode noise from the environment was rejected and the system dynamic range was improved significantly. By using a mirror as a reflecting target, the SNR at the 30 mA source current improved from 40 dB to 55 dB after dual balanced receiving circuitry. With this SNR, small details inside extracted teeth could be seen. Optical circulator was used to avoid unwanted reflection to the detection circuitry from unused arm.

The effect of Doppler frequency has also been investigated. In principle, high Doppler frequency is essential for improving system SNR by avoiding 1/f noise. In the low-coherence interferometry system with phase modulation generated by mirror vibration, the Doppler frequency was about 20 KHz that was determined by the PZT oscillation. This frequency was not high enough. Different types of oscillators have been investigated, for example, regular louder speaker, acoustic PZT transducers, and high-frequency PZT speakers. With a high frequency speaker driven at 7 KHz, 100 KHz Doppler frequency was obtained, which provided factor of 2 increment in modulation signals compared with 20 KHz Doppler frequency signal.
3.3 OCT images, sidelobe artifacts and CLEAN algorithm

The results in §3.1 show that there is sidelobe accompanied with the interference signal for a mirror as the sample. When the imaging medium is changed to an extracted tooth, the sidelobe artifacts appear as two complicated curves on both sides of the signals representing physical refraction index changes inside the tooth, as shown in Figure 3.5 (b).

From Figure 3.5 (b), it can be seen that coherent sidelobes of a source can severely degrade OCT image quality by introducing false targets if no targets are present at the sidelobe locations. Sidelobes can also add constructively or destructively to the targets

![Figure 3.5](image)

(a) Sketch of a sagittal section of human tooth. (b) Typical image of extracted tooth sample by OCT system with motor-galvanometer configuration.
that are present at the sidelobe locations. This constructive or destructive interference will result in cancellation of the true targets or display of incorrect echo amplitudes of the targets. CLEAN, a nonlinear deconvolution algorithm, has been introduced to cancel coherent sidelobes in OCT images of extracted human teeth. The results show that CLEAN can reduce the coherent artifacts to the noise background, sharpen the air-enamel and enamel-dentin interfaces and improve the image contrast. The detailed description of this CLEAN algorithm and the results are described at Appendix A.4, “Cancellation of coherent artifacts in optical coherence tomography imaging,” which is a paper submitted to Applied Optics in December 2000.

3.4 Image comparison between EDFA and SLD sources

The coherence length property has been compared in §3.1. The difference in coherence properties should be reflected by real images.

3.4.1 Point spread function

With the motor-gavanometer configuration OCT system described in Figure 2.4, point spread functions of system for EDFA and SLD sources with a mirror as the sample have been measured, as shown in Figure 3.6 (a) and (b).

The point spread functions of EDFA and SLD showed similar sidelobe positions and sidelobe levels, but different coherence length, which has been verified in §3.1, and different noise level. Since the point spread function of the OCT system from a planar mirror is the convolution of the Fourier transform of the source spectrum with the transfer function of the detection system, the measured sidelobes positions are the combined effects of OCT system and source coherence properties. Hence, the similarity of sidelobe
positions and sidelobe levels shown here could be controlled by the OCT system transfer function, and the coherence length could be controlled mainly by source spectrum.

Figure 3.6  Point spread functions of OCT system for different sources. (a) EDFA source; (b) SLD source
3.4.2 Image comparison of EDFA and SLD sources

The images taken for the same sample, for the same light power illuminating upon sample with only source difference (EDFA 1550nm, SLD 1310nm) are shown in Figure 3.7. There are sidelobes in both sets of images, however, the resolution of SLD images are much better. CLEAN algorithm can be applied to both sets of images.

Figure 3.7 Comparison between images taken with EDFA and SLD sources. Upper row, EDFA images; lower row, SLD images.
3.5 Images obtained with double-galvanometer OCT

An extracted tooth image taken with OCT system with double-galvanometer configuration described in §2.4 is shown in Figure 2.8. The air-enamel interface and the enamel-air interface can be delineated well. However, the background is not uniform as can be seen from the wide darkened strip at the right half the image. This is resulted from the non-uniformity of group delay discussed in §2.4.1.

In the images shown in §2.4.1, sidelobe artifacts appear at both sides of the signal at a special distance of about 1.3mm. However, in the images taken with double-galvanometer configuration OCT, the sidelobe artifacts are not observed through the depth scan range of 3mm. Sidelobe artifacts are the combined effects of source coherence properties and the transfer function of the OCT system, and the disappearance of the sidelobes in Figure 3.8 can only be explained by the change of OCT system transfer function for the non-linear group delay, which is generated by shifted galvanometer scanning, compared with the linear group delay generated by uniform motor moving.

![Figure 3.8 Image obtained with double-galvanometer OCT system](image)
4. CONCLUSION

4.1 Summary

Two low-coherence interferometry systems and two optical coherence tomography systems have been constructed. The coherence properties of sources and the effects of which upon imaging specifications have been studied with the two low-coherence interferometry systems. It has been shown that the two OCT systems are capable of imaging extracted teeth \textit{in vitro}. A deconvolution algorithm, CLEAN, has been employed to cancel sidelobe artifacts resulting from source coherence properties, and a detailed study of which has been presented as a contribution paper to a journal.

4.2 Future work

The optical coherence tomography systems described in this study are constructed toward clinical application, which is \textit{in vivo} periodontal imaging. Several modifications will be employed to achieve this objective. Firstly, a handpiece that can be clinically used in oral cavity will be employed into the sample arm. In the handpiece design, a small linear motor will be used to mechanically move a pigtailed GRIN lens to obtain lateral scanning with spot size less than 10\(\mu\)m. Secondly, since the motor scanning reference arm has the disadvantage of low scanning speed, and the galvanometer scanning reference arm has the limitation of non-uniformity on group delay and phase modulation, a new rapid scanning optical delay line based on diffraction grating \cite{25-26} will be used to achieve fast, uniform group delay, and an electro-optic phase modulator will be used to provide narrow bandwidth, uniform phase modulation.
Besides the application to dentistry, applications of OCT to blood flow measurements and 3-dimensional imaging to dermatology will also be studied later.
REFERENCES


A.1 Design of a band-pass filter

A homemade band-pass filter is used in this system. It is obtained by cascading two identical second-order multi-feedback band-pass filters. The circuit which realizes the second-order band-pass filter is shown in Figure A.1.1.

![Multiple-feedback second-order band-pass filter circuit](image)

Figure A.1.1  A multiple-feedback second-order band-pass filter

A second-order band-pass filter can be expressed, for appropriate values of bandwidth $B$ (rad/sec) and center frequency $\omega_0$, by the transfer function

$$H(s) = \frac{V_{out}(s)}{V_{in}(s)} = \frac{Ks}{s^2 + Bs + \omega_0^2}$$

(A.1.1)
The quality factor $Q$ is defined by $Q = \omega_0 / B$, or if $B$ is in Hz, $Q = f_0 / B$, where $f_0 = \omega_0 / 2\pi$. The gain of the filter is defined as the amplitude of $H(s)$ at the center frequency, which is $\text{gain} = K / B$.

For the multiple-feedback network shown in Figure A.1.1,

\[ B = \frac{2}{R_3 C} \]  \hspace{1cm} (A.1.2)

\[ \omega_0^2 = \frac{1}{R_3 C^2} \left( \frac{1}{R_1} + \frac{2}{R_2} \right) \]  \hspace{1cm} (A.1.3)

The constant $K$ in Equation (A.1.1) is given by $-1 / R_1 C$, and hence the circuit yields an inverting gain (negative) with magnitude $R_3 / 2R_1$.

By cascading two or more identical band-pass second-order filters, a much sharper band-pass filter may be obtained. If $Q_1$ is the quality factor of a single stage and there are $n$ stages, then the $Q$ of the filter is $Q / \sqrt{n} - 1$. If $\text{gain}_1$ is the gain of a single stage, then the $\text{gain}$ of the filter is $\text{gain}_1^n$; for an even $n$, the inverting gain of single stage may be inverted to a noninverting gain.

Designed for $f_0 = 20\text{ KHz}$, $Q = 10$, and $\text{gain} = 10$, the one stage multiple-feedback second-order band-pass filter shown in Figure A.1.1 has a $K$ parameter of 5, and $R_1 = 8K\Omega$, $R_2 = 420\Omega$, $R_3 = 157K\Omega$, for a $C$ of
$0.001 \mu F$. Using a TL082 op-amp and resistances of $3.3K \Omega + 4.7K \Omega$, 
$470 \Omega + 47 \Omega$, $100K \Omega + 47K \Omega$ respectively for $R_1$, $R_2$, $R_3$, the obtained
$f_0 = 19.9 KHz$. By cascading two identical stages, the overall $Q = 15.5$
$(B = 1.29 KHz)$, and gain = 100 (40dB).
A.2 Design of a high-pass filter

A homemade fourth-order high-pass Butterworth filter is used in this system. It is obtained by cascading two second-order Butterworth high-pass filters. The circuit which realizes the second-order high-pass filter is shown in Figure A.2.1.

![Second-order high-pass filter circuit](image)

**Figure A.2.1** A second-order high-pass filter

A second-order high-pass filter can be expressed, for appropriate values of bandwidth $a$ and center frequency $b$, by the transfer function

$$H(s) = \frac{V_{out}(s)}{V_{in}(s)} = \frac{Ks^2}{s^2 + as + b} \quad (A.2.1)$$

An $n$th-order transfer function has numerator $Ks^n$ and an $n$th degree polynomial denominator. For example, a fourth-order function is the product of two second-order functions such as Equation (A.2.1).
The cutoff frequency is $\omega_c$ in radians/sec or $f_c = \omega_c / 2\pi$ Hz. The gain of the filter is $K$. The high-pass Butterworth filter amplitude response is monotonic in both the stop and passband.

For the second-order high-pass filter shown in Figure A.2.1,

$$K = \mu = 1 + \frac{R_4}{R_3} \quad \text{(A.2.2)}$$

$$a = \frac{1}{R_1C} (1 - \mu) + \frac{2}{R_2C} \quad \text{(A.2.3)}$$

$$b = \frac{1}{R_1R_2C^2} \quad \text{(A.2.4)}$$

Figure A.2.2  A fourth-order high-pass filter
By cascading two second-order high-pass filters, a fourth-order high-pass filter may be obtained as shown in Figure A.2.2. Designed for $f_c = 1 \text{kHz}$, and gain = 100 (40dB), the fourth-order high-pass filter shown in Figure 2.6 has a $K$ parameter of 10, and $R_1 = 37K\Omega$, $R_2 = 6.8K\Omega$, $R_3 = 7.8K\Omega$, $R_4 = 68K\Omega$, $R_5 = 42K\Omega$, $R_6 = 6K\Omega$, $R_7 = 6.8K\Omega$, $R_8 = 62K\Omega$ for a $C$ of $0.01\mu F$. For the real circuit, a TL082 op-amp and resistances of $22K\Omega + 15K\Omega$, $5.6k\Omega + 1.2k\Omega$, $5.6K\Omega + 2.2K\Omega$, $47K\Omega + 20K\Omega$, $22K\Omega + 20K\Omega$, $2.7K\Omega + 3.3K\Omega$, $5.6K\Omega + 1.2K\Omega$, $47K\Omega + 15K\Omega$ respectively for $R_1$ to $R_8$ are used.
A.3 Matlab\textsuperscript{TM} programs

A.3.1 Matlab\textsuperscript{TM} program for OCT system with motor-galvanometer configuration

%load raw data
cd d:\daqing\OCT\motor_galvanometer;
load raw_data_motor.txt

%set number of frequency ratio and pixel size
n_ratio=128;%the number of lateral scans within one longitudinal scan, twice of
step_length=400;%pixel number at lateral direction

%Windowing data and convert to 2 dimensional array
for i=1:2*n_ratio;
    if mod(i,2)~=0;
        for j=1:step_length;
            image_data(i,j)=raw_data_motor((i-1)*step_length+j);
        end
    else
        for j=1:step_length;
            image_data(i,j)=raw_data_motor(i*step_length+1-j);
        end
    end
end

%Interpolation is needed to smooth the image
for i=2:2:2*n_ratio-2;
    for j=1:step_length;
        image_data(i,j)=(image_data(i-1,j)+image_data(i+1,j))/2;
    end
end

%Rearrange image array
image_data=rot90(image_data);
image_data=flipud(image_data);
image_data=fliplr(image_data);

%Display image
imagesc(log10(image_data));
colormap(gray);
A.3.2 Matlab™ program for OCT system with double-galvanometer configuration

```matlab
% load raw data
cd d:\daqing\OCT\double_galvanometer;
load raw_data_galva.txt

% set number of frequency ratio and pixel size
n_ratio=128;% the number of lateral scans within one longitudinal scan, % which is the pixel number at longitudinal direction
step_length=400; % pixel number at lateral direction

% Windowing data and convert to 2 dimensional array
for i=1:n_ratio;
    if mod(i,2)~=0;
        for j=1:step_length;
            image_data(i,j)=raw_data_motor((i-1)*step_length+j);
        end
    else
        for j=1:step_length;
            image_data(i,j)=raw_data_motor((2*n_ratio-i)*step_length+j);
        end
    end
end

% Interpolation is needed to smooth the image
for i=2:2:n_ratio-2;
    for j=1:step_length;
        image_data(i,j)=(image_data(i-1,j)+image_data(i+1,j))/2;
    end
end

% Rearrange image array
image_data=rot90(image_data);
image_data=flipud(image_data);
image_data=fliplr(image_data);

% Display image
imagesc(log10(image_data));
colormap(gray);
```
A.3.3 Matlab™ program for CLEAN algorithm

%Set path
cd c:\daqing\data\oct_clean\applied_optics;

%load point spread function data;
load psf_40ma_0831_largewindow_01.txt;

n_ratio=128;%scanning ratio of x/z, equal to 1/2 of pixel number in z
direction
step_length=400;%pixel number in x direction

n=2*n_ratio;

%Normalization of point spread function data
psf_odd=psf_40ma_0831_largewindow_01(1:n)/max(psf_40ma_0831_largewindow_01(1:n));

%zero-padding point_spread_function to 3*n long for shifting subtraction
for m=1:n;
    psf(1:m)=0;
    psf(n+m)=psf_odd(m);
    psf(2*n+m)=0;
end

%load image
load tooth_0830_04.txt; % sample 1
load tooth_0918_04.txt; %metalic filling--sample 2

>windowing measured data to make 2-dimensional image array
for i=1:2*n_ratio;
    if mod(i,2)~=0;
        for j=1:step_length;
            tooth_dirty(i,j)=tooth_0918_04((i-1)*step_length+j);
        end
    else
        for j=1:step_length;
            tooth_dirty(i,j)=tooth_0918_04(i*step_length+1-j);
        end
    end
end

%smooth interpolating
for i=2:2:2*n_ratio-2;
    for j=1:step_length;
        tooth_dirty(i,j)=(tooth_dirty(i-1,j)+tooth_dirty(i+1,j))/2;
    end
end
for mm=1:2*n_ratio;
    for nn=1:step_length;
        image_dirty(nn,mm)=tooth_dirty(mm,nn);
    end
end

image_cleaned(1:step_length,1:2*n_ratio)=0;
for k=1:step_length;
    iloop(k)=0;
    temp(k)=1;
end

alfa=0.1;%loop gain of CLEAN iteration
stop=min(min(image_dirty));%stopping criterion:noise level

%find position of maximum strength in point spread function
[mx_psf,mi_psf]=max(psf);

%iteration of PSF subtraction
for k=1:step_length;
    while temp(k)==1
        iloop(k)=iloop(k)+1;
        [mx_signal,mi_signal]=max(image_dirty(k,1:n));
        image_dirty(k,1:n)=image_dirty(k,1:n)-alfa*mx_signal*psf(mi_psf-mi_signal+1:mi_psf-mi_signal+n);
        if max(image_dirty(k,1:n))<stop
            temp(k)=0;
        end
    end
image_cleaned(k,mi_signal)=image_cleaned(k,mi_signal)+alfa*mx_signal;
end

%function needed for convolve back
smooth=psf(mi_psf-3:mi_psf+3);

image_cleaned_temp(k,1:n+6)=conv(smooth,image_cleaned(k,1:n));
image_last(k,1:n)=image_cleaned_temp(k,4:n+3);
end

%Add back noise to cleaned image
for mm=1:2*n_ratio;
    loop=mm;
    for nn=1:step_length;
        if image_last(nn,mm)==0
            if image_dirty>0;
                tooth_clean(mm,nn)=image_dirty(nn,mm)+stop;
            else
                tooth_clean(mm,nn)=stop*rand+stop;
            end
        else
            tooth_clean(mm,nn)=image_last(nn,mm)+stop;
        end
    end
end
%maximum pixel strength match
shift=mean(mean(tooth_dirty))-mean(mean(tooth_clean));

for mm=1:2*n_ratio;
    for nn=1:step_length;
        tooth_clean(mm,nn)=tooth_clean(mm,nn)+shift;
    end
end

%rearrange array to get conventional display
for pp=1:2*n_ratio;
    for qq=1:step_length;
        dirty(qq,pp)=tooth_dirty(2*n_ratio+1-pp,qq);
    end
end
for pp=1:2*n_ratio;
    for qq=1:step_length;
        clean(qq,pp)=tooth_clean(2*n_ratio+1-pp,qq);
    end
end

figure(1);
%imagesc(log10(tooth_dirty))
imagesc(log10(dirty))
colormap(gray);

figure(2);
%imagesc(log10(tooth_clean+stop))
imagesc(log10(clean))
colormap(gray);
Cancellation of coherent artifacts in optical coherence tomography imaging

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ABSTRACT

Coherent sidelobes of a source can severely degrade OCT image quality by introducing false targets if no targets are present at the sidelobe locations. Sidelobes can also add constructively or destructively to the targets that are present at the sidelobe locations. This constructive or destructive interference will result in cancellation of the true targets or display of incorrect echo amplitudes of the targets. We introduce the use of CLEAN, a nonlinear deconvolution algorithm, to cancel coherent sidelobes in OCT images of extracted human teeth. The results show that CLEAN can reduce the coherent artifacts to the noise background, sharpen the air-enamel and enamel-dentin interfaces and improve the image contrast.

Keywords: Optical coherence tomography, coherent sidelobe, CLEAN, deconvolution, dentistry
OCIS codes: 170.4500, 170.3880, 170.3890, 170.1850

1. INTRODUCTION

Optical Coherence Tomography (OCT) is a new imaging technique for various medical applications. OCT exploits the short temporal coherence of broadband light sources to achieve optical scanning of scattering tissue in depth dimension. Spatial scanning is accomplished by mechanically or optically scanning the beam. At each spatial location, the OCT scanner output is the Fourier transform of the source spectrum convolved with the tissue reflectivity over a narrow scattering angle and with the transfer function of the detection system. Due to the imperfect Gaussian shape of a source spectrum, coherent sidelobes of the source appear in OCT images, and their locations vary with the source and the source current level. In Ref. [9], we have reported that the sidelobe level increases with the source current when an erbium-doped fiber amplifier is used as a source. Coherent sidelobes can severely degrade OCT image quality by introducing false targets if no targets are present at the sidelobe locations. Sidelobes can also add constructively or destructively to the targets that are present at the sidelobe locations. This constructive or destructive interference will result in cancellation of the true targets or display of incorrect echo amplitudes of the targets. In this paper we show that these
sidelobe artifacts in OCT images of extracted teeth can be effectively reduced to background noise level by using an iterative nonlinear deconvolution procedure known as CLEAN.

CLEAN was invented in the middle 1970s in radio astronomy and was modified later for use in microwave imaging. In ultrasound, similar deconvolution procedures were used to reduce the refractive artifacts. In OCT, the use of CLEAN to reduce speckle noise caused by interference of nonresolvable scatterers of highly scattering tissue has been reported by Schmitt. In this reference, CLEAN has been implemented using a theoretical point spread function with a Gaussian envelope. Therefore, the sidelobes due to imperfect Gaussian source spectrum have not been considered. In addition, the author has used a non-conventional OCT system with an array of sources and detectors, which enables CLEAN to be implemented in two dimensions.

A human tooth as illustrated in Figure 1 consists of a crown and root. The junction between the crown and the root is called the cervical margin. The tooth crown covered by an acellular, highly mineralized tissue, enamel. Enamel is translucent and varies in thickness from a maximum of 2.5 mm to a feather edge at the cervical margin. The bulk of the tooth is comprised of dentin. Dentin is a hard, elastic, avascular tissue that is approximately 70% mineralized. The interior of the tooth is called the pulp. The pulp consists of soft connective tissue that is innervated and highly vascular.

In this paper, we report the use of CLEAN, implemented in a conventional OCT system, to cancel coherent sidelobe artifacts in OCT images of extracted teeth caused by the imperfect source spectrum. We show that CLEAN reduces the artifact level to the noise floor, sharpens the air-enamel and enamel-dentin interferences and improves the contrast between the interfaces and the surrounding medium.

2. BASIC PRINCIPLE

2.1 Point spread function
The point spread function (PSF) of an OCT scanner depends on the coherence length of the source and the effective aperture defined by the pupil functions of the source and the detection optics. Therefore, it is a two dimensional (2-D) function and is denoted as \(h(x,z)\), where \(z\) and \(x\) are the propagation and lateral dimensions, respectively. In conventional OCT scanners, a single beam is scanned across the sample, therefore the spatial distribution of \(h(x,z)\) can not be measured. Figure 2 illustrates the pertinent issue. A point scatterer is located in the medium and it scatters the incident light. The reflected light within a narrow angle is received by the detection optics and the off-axis scattered light falls outside of the receiving aperture. If more detectors were located at the off-axis positions, the scattered light could have been received over a larger angle and the \(h(x,z)\) in \(x\) dimension could have been measured. However, since the incident beam is highly focused and the energy received over the beam spot area far exceeds the energy of off-axis scattered light, it is therefore possible to use one-dimensional point spread function \(h(z) \approx \int_{\text{beam-spot-area}} h(x,z) dx\) to estimate the \(h(x,z)\). We have found that, by using
this approximated PSF in CLEAN, a large portion of the sidelobe energy can be removed effectively.

2.2 Conventional deconvolution
Assume that the medium contains N point scatterers each has complex strength C_j. The original image that we wish to reconstruct is

\[ O_{\text{image}}(x,z) = \sum_{j=1}^{N} C_j \delta(x-x_j, z-z_j) \]  

(1)

where \( x_j \) and \( z_j \) are the coordinates of the \( j \)'th scatterer. The "dirty" image is

\[ D_{\text{image}}(x,z) = O_{\text{image}}(x,z) \ast h(x,z) + N(x,z) \]  

(2)

where \( \ast \) denotes convolution and \( N(x,z) \) represents system noise. Conventional deconvolution procedures perform Fourier transform on "dirty" image \( D_{\text{image}}(x,z) \) and PSF \( h(x,z) \) to obtain their spatial frequency spectrum \( D_{\text{image}}(u,v) \) and \( h(u,v) \). The original image can be recovered from the inverse Fourier transform of

\[ \frac{D_{\text{image}}(u,v)}{h(u,v)} = O_{\text{image}}(u,v) + \frac{N(u,v)}{h(u,v)}, \]

where \( O_{\text{image}}(u,v) \) and \( N(u,v) \) are spatial frequency spectrum of \( O_{\text{image}}(x,z) \) and \( N(x,z) \), respectively. Unfortunately, this procedure is very sensitive to \( h(x,z) \) and the system noise. In a practical OCT system, \( h(u,v) \) approaches zero rapidly due to limited bandwidth in axial direction and the effective aperture in the lateral direction. Therefore, this procedure is not robust and the deconvolution results highly depend on system noise.  

2.3 CLEAN procedure
CLEAN provides more robust means to reconstruct the original image. CLEAN is performed in the spatial image domain and it iteratively recovers the original image. The basic steps of CLEAN algorithm are described as follows: First, find the brightest pixel \((x^{(1)}, z^{(1)})\) in the "dirty" image and subtract a fraction of the deconvolution kernel \( a h(x-x^{(1)}, z-z^{(1)}) D_{\text{image}}(x^{(1)}, z^{(1)}) / h_{\text{max}} \) from the "dirty" image. The parameter \( a \) is called loop gain which is less than unity, and it represents the fraction that is subtracted out by each iteration of CLEAN. Second, find the brightest pixel \((x^{(2)}, z^{(2)})\) from the residual of \( D_{\text{image}} - a h(x-x^{(1)}, z-z^{(1)}) D_{\text{image}}(x^{(1)}, z^{(1)}) / h_{\text{max}} \) and repeat the first step. The iteration continues until the maximum value in the "dirty" image reaches the noise floor. Assume that a total of M targets of strengths \( a D_{\text{image}}(x^{(i)}, z^{(i)}) \) are found before iteration is stopped. The final CLEANed image is obtained by convolving the set of delta functions of strengths \( a D_{\text{image}}(x^{(i)}, z^{(i)}) \) at locations \((x^{(i)}, z^{(i)})\) with the clean beam of the point spread function containing only the mainlobe. Generally, the residuals after CLEAN are added back to the CLEANed images to produce a realistic noise level in the final image.

2.4 1-D CLEAN procedure for conventional OCT imaging
As discussed early, a 2-D PSF cannot be obtained from a conventional OCT scanner. Therefore, we have used the 1-D PSF measured from a mirror to approximate the 2-D
In this modified procedure, we first find the brightest pixel \((x^{(1)}, z^{(1)})\) in the “dirty” B-scan image and subtract a fraction of the deconvolution kernel \(c h(z - z^{(1)})D_{\text{image}}(x^{(1)}, z^{(1)})/h_{\text{max}}\) from the “dirty” A-scan line. We then, find the brightest pixel \((x^{(2)}, z^{(2)})\) from the residual of \(D_{\text{image}} - c h(z - z^{(1)})D_{\text{image}}(x^{(1)}, z^{(1)})/h_{\text{max}}\) and repeat the first step. The final CLEANed image is obtained by convolving the set of delta functions of strengths \(aD_{\text{image}}(x^{(i)}, z^{(i)})\) at locations \((x^{(i)}, z^{(i)})\) with the clean beam of the 1-D PSF containing only the mainlobe. The disadvantage of using 1-D CLEAN is that the off-axis sidelobe energy is not removed. However, we have found that the off-axis sidelobe energy is significantly smaller compared with the on-axis sidelobe energy and 1-D CLEAN effectively reduce the sidelobe energy to the noise floor in the images.

3. METHODS

The schematic diagram of our OCT system is shown in Figure 3. The erbium-doped fiber amplifier (EDFA) at a center wavelength of 1550 nm is used as the low coherent source. The optical power output and 3dB coherence length of this EDFA at 40mA current are 0.43mW and 25\(\mu\)m, respectively. The light from the source enters the optical circulator (OC), and it split into a reference for the differential receiver input and a signal, which is passed through a 3 dB fiber coupler (FC), where the outputs form a Michelson interferometer. The reference arm of interferometer consists of a microscope objective (O1), a plane mirror, a speaker and a linear DC motor. The mirror is glued on the bowl of the speaker, which is mounted on the linear motor. The speaker is used to provide Doppler shifting frequency and is driven by a sinusoidal wave of 100 Hz frequency and 6.2 volts peak-to-peak amplitude. The measured maximum Doppler shifting frequency is 17KHz. The depth (Z direction) scan is achieved by translating the speaker and motor assembly. The sample arm consists of a microscope objective (O2), a galvanometer scanner, a lens, and a sample. The alignment requirement for the sample arm is that not only the incident light is perpendicular to the rotation axis of the galvanometer mirror, but also the incident light spot is located at the focal point of the lens. With this precise alignment, the spatial (X direction) scanning is achieved by translating the rotation of the galvanometer mirror to a linear scanning upon the sample without introducing a change in optical path.

The data acquisition consists of a dual balanced optical receiver, a high-pass filter, and a PC. The interferometer output is received by the dual balanced receiver, which is used to reduce the common-mode noise. The resulting signal is passed through a high-pass filter, which is used to preserve the wide frequency range of signals. We have found that a high-pass filter tracks the peak Doppler frequency better than a band pass filter. The filtered signals are sampled by an A/D converter of 400KHz sampling frequency, and the sampled data are Hilbert transformed to obtain the envelope of the Doppler signals. An A-scan line is obtained from this envelope signal.

The galvanometer and the linear motor are controlled by a galvanometer driver and a DC motor driver, respectively. The rotation of the galvanometer and the translation of the motor are synchronized by the PC control software. A 10 Hz sine wave generated by the
PC is used to drive the galvanometer controller, which in turn drives the scanning mirror. One set of B-scan data is acquired for 128 cycles of galvanometer flipping and 3.84 mm translation of the reference arm in depth. For the images presented, the transversal scanning range is set to 6 mm, and a total of 400 pixels are used. Thus the pixel size at transversal direction (X) is 15\(\mu\)m, which is less than 3dB spot size (~30\(\mu\)m) of the sample beam. The pixel size for depth direction (Z) is 15 \(\mu\)m, which is less than 3dB coherence length (25\(\mu\)m) of EDFA source at 40mA driving current.

4. RESULTS

The measured point spread function of the system is shown in Figure 4(a). The distance between the mainlobe and the sidelobe positions is 1.335mm, which is within the scan range of 3.84 mm. When a planar mirror is imaged, the sidelobe artifacts create two parallel lines on both sides of the central image line (see Figure 4(b)). Under this ideal imaging scenario, the sidelobe artifacts can be easily identified. However, when the imaging medium is changed to an extracted tooth, the sidelobe artifacts appear as two complicated curves on both sides of the air-enamel interface (see Figure 4(c)). These artifact curves can be confused with real interfaces in diagnosis. Furthermore, if the enamel-dentin interface were located at the neighborhood of the sidelobe curve, it would not be imaged with correct echo amplitude. The relative amplitude (logarithmic scale) of the measured point spread function is shown in Figure 5. The strengths of the first sidelobe and second sidelobe are -14 dB and -28 dB, respectively. Although the second sidelobe is within the 3.84 mm z-scan range, its strength is too weak to cause any visible image degradation.

Figure 6 shows the “dirty” image of the extracted tooth again (part a) and the CLEANed image (part b). Figure 6(c) is the microscopic picture of the sectioned tooth imaged. The scanned region contained a fissure defect in the enamel (pointed by the black arrow in (c)). Diagnostically, it is important to determine whether or not such a defect extends beyond the enamel layer. Comparing images before and after CLEAN, we can observe significant improvement in image detail in the fissure region (pointed by the arrow in the upper part of (a)). In addition, the two artifact curves on both side of the air-enamel interface shown in Figure 6(a) are reduced to background level after CLEAN. The contrast between the air and the enamel has been improved and the interface is delineated well after CLEAN. Furthermore, the sidelobe artifacts associated with the enamel-dentin interface (pointed by the right arrow at the bottom of (a)) are removed and this interface (pointed by the left arrow at the bottom of (a)) is enhanced after CLEAN. To quantitatively assess the sidelobe artifacts reduction, we show an A-scan line in Figure 7(a), which is obtained at the neighborhood including the air-enamel and enamel-dentin interfaces as well as their sidelobes. The measured peak sidelobes associated with these interfaces are ~15dB and are reduced to the background level after CLEAN (Figure 7(b)), and both interfaces are enhanced.

Another example is shown in Figure 8, wherein a tooth that contained a metallic restoration is imaged. Diagnostically, it is important to verify smooth adaptation of the restoration to the tooth surface and to detect structural defects of the tooth at the interface. A noticeable improvement after CLEAN is the contrast between the restoration and the
enamel. Since the metallic restoration has higher reflectivity than the tooth, the incident light is reflected at the air-restoration interface and the interior of the restoration should appear dark. However, the contrast between the restoration and the enamel is poor in the ‘dirty’ image because of significant sidelobe energy in the restoration. After CLEAN, the restoration tooth interface is better visualized both at the surface and along the internal aspects of the restoration. In addition, the portion of the air-enamel interface, pointed by the bigger arrow at the top of the CLEANed image, is blurred and diffused in the background level in the ‘dirty’ image and is clearly visible after CLEAN. Another noticeable improvement is the boundary of a cavity at the tooth surface, pointed by the smaller arrow at the bottom of the CLEANed image, and it is delineated well after CLEAN.

The residual sidelobe after CLEAN is visible at several spatial locations. Since the reflectivity of the metallic restoration is strong, the signals returned from air-metal interface are saturated at several spatial positions, and the extent of the saturation depends on the angle of incident light with the metal surface. As a result, the ratios of the peak sidelobe to mainlobe strength at these spatial positions are several decibels higher than the ratio obtained from the mirror. Therefore, the sidelobe subtraction at these A-scan positions is not as complete as that obtained at the non-saturation positions, and it leaves larger residual sidelobe energy.

5. DISCUSSION

Coherent sidelobes of a source are caused by the imperfect source spectrum. The strengths and positions of the sidelobes are related to the particular source and the source current level used. As the injection current increases, the gain of the diode increases, and multiple internal reflection occur. Figure 9 is the measured source spectrum of the EDFA used in our system. As shown in the figure, the source spectrum has periodical ripple overlapped on top of the main spectrum waveform. The interval between two ripple peaks is approximately 1nm, which in spatial domain corresponds to 1.060 mm. This distance is roughly close to the measured sidelobe position of 1.335 mm. Since the measured PSF from a planar mirror (Figure 4) is the convolution of the Fourier transform of the source spectrum with the transfer function of the detection system, the measured sidelobes positions are modified by the detection optics.

In this paper, CLEAN is applied to A-scan data obtained from a conventional OCT scanner. Therefore, the CLEANed images still contain a small residual sidelobe energy caused by off-axis scattered light. Another factor that could affect the elimination of sidelobe artifacts would be the nonlinearity introduced by the scanning optics. Since the beam is scanned across the lens at the sample arm (see Figure 3), the center beam alignment and the beam spot size vary with the galvanometer flipping angle. Therefore, a certain spatial nonlinearity is introduced in the image. Since this non-linearity also exists in the mirror image, we have used the corresponding PSF measured at every spatial location to clean the extracted teeth images. However, we did not observe any visible changes in CLEANed images by using a single PSF and the corresponding PSF obtained at each spatial location.
The final image quality is affected by the loop gain and the stopping criteria. Since the signals cannot be distinguished if their levels are below the noise floor in the image, it is straightforward to choose the background noise level in the image as a stopping criteria when CLEAN is used. The loop gain determines the final image quality. Tested with different loop gains of 1, 0.5, 0.25, 0.1, 0.05, 0.025 and 0.01, it is shown that there is distinguishable image quality change for loop gains of 1, 0.5, 0.25 and 0.1, but if loop gain is further reduced, no distinguishable image quality change is observed. In terms of processing time, using Pentium III 800MHz CPU, the computing times for loop gains of 1, 0.5, 0.25, 0.1, 0.05, 0.025 and 0.01 are all about 11 minutes. In our regular CLEAN, therefore, we choose 0.1 as the loop gain and the background noise level as the stopping criteria.

The power of the EDFA at 40 mA pumping current is 0.43 mW, which is lower than that of superluminescent light sources used by many research groups. We could increase the pumping current to deliver more power to the medium, but the source coherence length also increases. In Ref [9], we have reported that the coherence length as well as the sidelobe level increases with the source current. When the pumping current was increased from 40 mA to 100 mA, the coherence length was increased from 25 μm to 48 μm. As a compromise between the coherence length and the source power, we have chosen 40 mA pumping current in our experiments. We have demonstrated that sidelobe artifacts can be removed and image contrast is significantly improved using CLEAN and our source.

6. SUMMARY

Coherent sidelobes of a source can severely degrade OCT image quality by introducing false targets if no targets are present at the sidelobe locations. Sidelobes can also add constructively or destructively to the targets that are present at the sidelobe locations. This constructive or destructive interference will result in cancellation of the true targets or display of incorrect echo amplitudes of the targets. In this paper, we have demonstrated that the non-linear deconvolution algorithm CLEAN can be used to reduce sidelobe artifacts caused by the imperfect Gaussian source spectrum. We have modified CLEAN and adapted it to a conventional OCT system, and have shown that the sidelobe artifacts can be effectively reduced to background noise level. As a result of sidelobe reduction, the image contrast of the extracted tooth has been improved significantly. CLEAN also sharpens the air-enamel and enamel-dentin interfaces, and improves the visibility of these interfaces which will be beneficial to diagnosis.

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Figure 2. A diagram of the conventional OCT scanning. The on-axis reflected light within a narrow angle is received by the detection optics, while the off-axis scattered light falls outside of the receiving aperture. If more detector fibers were located at off-axis positions (broken fiber lines), the scattered light over a wider angle could have been received.
Fig. 3. Schematic diagram of the OCT system. $O_1, O_2$, microscope objectives; $L$, lens; $G$, galvanometer scanner. The galvanometer and linear motor are controlled by PC.

Fig. 4. The point spread function of the system and the resulting image artifacts. (a) Normalized point spread function of the system or A-scan line measured from a planar mirror. The horizontal axis is the propagation depth, and the distance between the main lobe and the sidelobe is 1.335 mm. (b) Image of a mirror. The propagation direction is from left to right. Vertical axis is the spatial scan dimension. The central line is the mirror, and the two lines at both sides of the central line are sidelobe artifacts. (c) Image of an extracted tooth. The central curve is the air-enamel interface. The sidelobe artifacts are clearly shown by two complicated curves, pointed out by the vertical arrows, on both sides of the central curve.
Fig. 5. The relative strength of mainlobe, first order sidelobe and second order sidelobe of the point spread function. The display is in logarithmic scale.

Fig. 6. (a) Original tooth image. (b) CLEANed image. (c) Microscopic picture of the sectioned tooth imaged. The OCT image dimension is 6 mm (vertical, X) by 3.84 mm (horizontal, Z), the pixel size is 15 μm². The sidelobe artifact curves are reduced to the background level. The air-enamel interface is enhanced after CLEAN, and the sidelobe energy of the enamel-dentin inference, pointed by the smaller arrow at the bottom right, is removed after CLEAN.
Figure 7. Normalized A-scan lines (dB) from the “dirty” image (a) and the CLEANed image (b). The A-scan position is shown in Figure 6 (a). After CLEAN, the sidelobes associated with air-enamel and enamel-dentin interfaces are reduced to background noise.
Fig. 8. (a) Image of tooth with a metal filling at the surface. (b) CLEANed image. (c) Microscopic picture of the sectioned tooth. The spatial scan range is indicated by the two arrows. The portion of the air-enamel interface that is delineared well after CLEAN is pointed by the arrow at the top of (b). The indented portion of the air-enamel interface corresponds to a cavity and is pointed by the longer arrow at the bottom of (b). The boundary and the internal aspects of the metallic restoration (within the frame) is visualized better after CLEAN.

Fig. 9. Optical spectrum of EDFA. (a) wavelength range is 1500-1600nm; (b) wavelength range is 1545-1565nm. The period of the ripples on top of the main spectrum waveform is about 1 nm.