High-Speed Wavelength Swept Laser Source for Optical Coherence Tomography

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*To my lovely wife, Ayumu, two daughters, Emma and Sally,
and my son, Siwon.*

*Everything should be made as simple as possible, but not
one bit simpler*

Albert Einstein

Abstract

Optical Coherence Tomography (OCT) is one of promising technologies in medical diagnostic imaging system for the next generation, and which has been gaining great attentions these years. Swept-Source OCT (SS-OCT) that utilizes a tunable laser has advantages of higher sensitivity of 20dB or greater, three times higher resolution, and faster imaging speed with several tens frame per second in comparison with conventional Time Domain OCT (TD-OCT). There are following correlations between image properties in SS-OCT and performance/parameters of swept source.

- 1) Coherence length of the source is proportional to the measurement depth range.
- 2) Wavelength swept range is counter-proportional to the axial resolution of the image.
- 3) Swept rate is proportional to the imaging speed because one sweep corresponds to an A-line (vertical line) of the OCT image.

So high-speed imaging of high-resolution, large depth OCT requires high-speed sweep, wide range, and high coherence (or narrower spectral linewidth) during sweep. However, most of previously introduced technologies have two significant drawbacks that make us difficult to implement it into actual medical equipment for real clinical application. One is the fact that these sources are not fully meeting the simultaneous requirement of three parameters. Particularly, coherence length degrades when the source is swept at very high speed over 20 kHz or higher. In Ophthalmology or Cardiovascular applications, large depth range is required with 40-50 kHz, very high swept rate. Ambiguity removal process by having phase shifter device was proposed to countermeasure this issue, but it also complicates the system configuration. Secondly, the transient response of the sweep is intrinsically non-linear so that it necessitates very complex post-process to calculate the OCT signal, and which makes real-time imaging virtually impossible as a result.

The major purpose of this study is to propose and validate the novel configurations of swept source that overcome these drawbacks and improve the performance of swept source, thus improving the image quality of SS-OCCT.

Followings are major proposals and the work focused in this study;

1. Highly k-linear sweep for high-speed or real-time imaging
2. Coherence length improvement by Quasi-Phase Continuous Tuning
3. Wide range tuning for high resolution OCT imaging
4. Clinical applications

First, a base configuration of a fiber extended cavity wavelength swept laser source was proposed which utilizes a polygon-scanner grating filter in Littrow arrangement that linearizes the frequency sweep. Geometrical compensation technique with using prism expander in the tunable filter was introduced. Theoretical analysis was carried out to estimate the linearity, and the prototype source was fabricated to validate its performance and to evaluate the OCT image quality. Linearity of less than 0.25% is achieved with extremely high repeatability between scans. And actual OCT image measured with the source proved that highly k-linear sweep eliminates the need of wavelength rescaling process and increase the imaging speed.

Second, we introduced the novel idea of Quasi-Phase Continuous Tuning. We have proven that without implementing mirror canceling technique or the other costly mitigation methods, coherence length can be drastically improved by using the proposed technique. Theoretical estimation of linewidth is also carried out.

Third, several schemes for wide range tuning with special filter geometries are proposed. These techniques overcome the physical limitation of tuning range with rather simple configurations.

In this thesis, these studies are divided in the chapters as in following. In Chapter 1, background of OCT and SS-OCT explains general discussion on the requirements for swept source in various specific applications. And it is followed by Chapter 2, 3, 4 describing the above-proposed innovations. SS-OCT algorithm and clinical study examples based on proposed swept source are introduced in Chapter 5 including novel MEMS fiber probe for endoscopic applications that was combined with SS-OCT system. In Chapter 6, feasibility studies on single mode hop-free swept lasers are discussed for the future improvement.

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Chapter 1 Introduction

1.1 Research background

Continuous and repetitively tunable or “swept” lasers have become common in both research laboratories and production environments for measuring optical parameters. The lasers have a narrow spectral line width that is swept rapidly over a wide wavelength range. The light is used as a “probe” for measuring spectral response, for example absorption, light scattering or to measure surface topography using the interference effects of the coherent light. Compared to systems using broadband light sources with optical spectrum analyzers these lasers provide high sensitivity and a rapid measurement time; but nevertheless are still limited by a scan time of typically one second or more for a 100nm scan.

This study proposes a high speed-scanning laser that continuously scans over a 110nm wavelength range at up to 20kHz. This scan speed, which is some 20,000 times faster than conventional swept lasers, opens up a wealth of new application areas some of which have only just started to be researched. Spectroscopic measurements that have been performed with conventional swept lasers can now be performed in real time and, further, transient dynamics can also be monitored. This is critical for applications such as gas analysis in internal combustion engines or industrial processes [1-5], or monitoring vibrations with fiber Bragg gratings in fiber optic sensor systems [6]. The most significant application, however, is real time imaging using a technique called Optical Coherence Tomography (OCT) [7-12]. For this application the rapid scan is indispensable and the technology is actively being researched for medical applications in particular [13-15].

In this section, we describe first, the research background of Swept-Source OCT followed by general requirements as well as application-specific requirements of the swept source. We, then, explain what technology is best fit to achieve the required performance while describing the drawbacks of previous works. Finally, we explain the scope of this thesis and contents of this study.

1.2 SS-OCT and swept source

OCT is non-invasive imaging technique that was originally introduced in early 90's, and diagnosis systems based on which are now widely practiced in ophthalmology application. Compared to conventional medical imaging technologies such as Magnetic Resonance Imaging (MRI), Positron Emission Tomography (PET), X-Ray Computerized Tomography (CT) and Ultrasonography, OCT provides a safe, high resolution solution at a cost point that will enable widespread use in hospitals and clinics. Most of these OCT systems use Time-Domain optical interferometry in which the optical path length difference between reference mirror and sample in the Michelson or Mach-Zehnder interferometer is modulated in time. Time-Domain or TD-OCT had opened up the potential of optical biopsy but there is performance limitation for further extension of the applications. First, imaging speed is relatively slow because of mechanical delay modulation. Second, even when higher frequency scanning is possible, detection sensitivity drops because of detection bandwidth in return. Fourier-Domain OCT [16,17] is a break-through technology to achieve high sensitivity and high speed imaging at the same time. FD-OCT relies on analyzing the individual frequency components of backscattered light from the sample or tissue. There are two ways within FD-OCT. One is Spectral-Domain OCT (SD-OCT) that uses low coherence light source and spectrometer where frequency components are spatially analyzed on the CCD array. Fast readout speed of CCD benefits to the imaging speed and high signal-to-noise ratio (SNR) gives 20-30dB advantage over conventional TD-OCT [18,19]. However, there are also disadvantages. Image gets blurred and degraded when the sample arm motion washes out interference fringe on the CCD during pixel integration time [20,21]. Unavailability of InGaAs CCD with higher pixel resolution also limits the application of FD-OCT in in-vivo endoscopic applications.

The other approach is Swept-Source OCT that uses continuous and repetitively tunable or "swept" light source where frequency components are analyzed in time with a single photodetector. Each wavelength scan generates depth information; lateral scanning of the laser beam then enables a cross section image to be constructed. This technique has a theoretical sensitivity benefit equal to that of SD-OCT, while overcoming the disadvantage of SD-OCT such as fringe washout and allowing the use of longer wavelength over 1 μm to 1.5 μm range for endoscopy.

Here the question is how we can realize the "swept-source" that meets requirements for the use in OCT while meeting the same cost-performance as "CCD".

Next section explains correlation of source performance to that of SS-OCT and other general requirements for the source.

1.3 General requirements for the swept source

Output power

The higher the output power on the sample, the higher the SNR improves in principle [23]. However, maximum permissible exposure on the human tissue is regulated by ANSI(American National Standards Institute) standard. 0.7mW in average at 800nm range and about 1mW at 1060nm range at the probe arm are allowable exposure for ophthalmic applications. The other tissues can be exposed over >10mW in endoscopic applications. The use of further longer wavelength such as 1300nm makes the max power less critical in terms of ionization of tissue by high power exposure.

Sweep linearity

OCT signal is processed by time-sampling the backscattered light as the swept source sweep the wavelength followed by Fourier transform (FFT). Ideally, the sweep should be linear in k space ($k=2\pi/\lambda$). But actual sweep curve of the most of proposed swept sources are non-linear in time because of its intrinsic tuning mechanism. For examples, the use of Galvano mirror [24], or fiber Fabry-Perot filter imposes sinusoidal sweep due to its driving characteristics [25]. If simply applying FFT on time-sampled interferogram when this non-linearity is present, it blurs resolution of the signal and signal power also decays. So, in general, most of SS-OCT systems implement either nonlinear sampling with the use of optical clock having another set of interferometer and detector, or post processing approach; so called “wavelength rescaling process”.

Wavelength range

The choice of wavelength band in OCT is dependent on the water absorption and scattering property of the sample or tissue of interest [26-29]. Figure 1.1 shows the absorption and scattering coefficient over visible to infrared wavelength bands. In general, 800nm range is used for retinal imaging because of low absorption in vitreous humor [30,31], and recently 1060nm range gains attentions because of large penetration in retinal tissue as well as low dispersion property in tissue[32-34]. In endoscopic application 1310nm range or longer range is commonly used because of low scattering resulting in large depth penetration [36,37]. The OCT in 1310nm range has another benefit; abundant available optical components in this range that were developed for optical fiber telecommunication applications. Spectral width of OCT signal, i.e. axial resolution is inversely related to the envelope of temporal power profile via Fourier transforms, and which is given by equation;

$$\delta z = \frac{2 \ln 2}{\pi} \frac{\lambda_0^2}{n \Delta \lambda} \quad (1.1)$$

For examples, 10um resolution can be achieved by 30nm at 820nm and 75nm at 1310nm, respectively.

Figure 1.1 Absorption and scattering coefficients of light

Swept rate (scanning speed)

Wavelength swept rate, or scanning speed of the swept source is directly reflected on the imaging speed like readout speed or refresh rate of CCD in SD-OCT. Swept rate corresponds to A-line rate in OCT. Increasing A-line rate makes it possible to accommodate more A-lines per frame or increase the frame rate. In practical applications the ability to produce video rate images is of critical importance. This not only removes imaging artifacts that is creased by undesired movement, but also enables a large area/volume measurement without compromising resolution in a short amount of time. Depending on applications swept rate of 10 kHz to 100 kHz range is required.

Instantaneous linewidth / Coherence length

Instantaneous linewidth; this terminology is very specific to the high-speed swept-source. Sometimes it is called, dynamic linewidth. It doesn't mean linewidth of a single longitudinal

mode as usually referred in the laser oscillation. The most of swept sources for SS-OCT are partially coherent light sources, and spectral width is the width of envelope that contains group of cavity modes. Since instantaneous linewidth is finite and the OCT signal is convolution of its spectrum and interferogram, its fringe visibility drops for the higher frequency components, i.e. at deeper range. Coherence length is defined as the optical round trip delay or twice of depth range where fringe visibility drops half or Fourier-transformed OCT signal drops 6dB compared to the signal power at zero delay. The relation of instantaneous linewidth, $\delta\lambda$ and coherence length, C_l is given in equation;

$$C_l = \frac{2 \ln 2}{\pi} \frac{\lambda_0^2}{\delta\lambda} \quad (1.2)$$

Coherence length of most of medical applications ranges between 2 and 10mm except for intraocular length measurement as discussed in Chapter 5.

Relative Intensity Noise

Intensity noise of the source at the signal detection band directly impacts on the SNR of the OCT system. Source of the noise attributes to the injection current from the driver circuitry and the cavity mode beating and its harmonics at the frequency equal to the reciprocal of cavity life time. Ripple on the spectral profile of the sweep also influences on the OCT signal as an alias noise. In OCT system the source RIN can be reduced if balanced detection is used for common mode noise reduction. RIN can be measured with a photodiode and an electrical spectrum analyzer.

Polarization

Single polarization output or output with high degree of polarization (DOP) is desired to suppress the effect of PMD in the interferometry system. Single polarization output may be useful when polarization sensitive OCT can be realized by adapting the Muller matrix method.

Reliability

Long term reliability and durability of the source is desired for harsh environmental conditions in clinical use. Meeting EMC/EMI standards is also important so that the instrument doesn't interfere with the other surrounding equipments. In most cases, 5 years product life should apply, corresponding to over average 40,000hrs of continuous usage.

1.4 Requirements for specific OCT applications

The above explained three parameters; Swept rate and wavelength range and coherence length are interrelated and have been selected to provide optimum imaging resolution, imaging depth and imaging speed for OCT. These parameters can be reconfigured for different imaging requirements or applications.

Figure 1.2 shows typical examples of requirements for different applications. Here, swept range and instantaneous linewidth are enclosed as a parameter of Finesse; the ratio of instantaneous linewidth to swept range.

Figure 1.2 Requirements of swept source for different clinical application

Cardiovascular applications: Three dimensional structure and properties of plaque formed inside the coronary is the interest of imaging. Saline flush process to image the sidewall of the vessel allows only several seconds for the image acquisition of a few centimeter long volume with push-pull method [40]. A-line rate required here is 40 to 50 kHz. Since this is not a molecular level biopsy like cancer detection, axial resolution of 10 μ m is sufficient for most of diagnosis that corresponds to about 80 to 100nm swept range in 1310nm wavelength range.

Gastro-Intestinal endoscopic applications: Cancer detection is everybody's ultimate goal as optical biopsy. In GI-endoscopy, axial resolution is the most critical parameter in this regard.

Requirement of swept rate is somewhat compromised in this case even though 10 to 20 kHz is necessary to mitigate the motion of sample and endoscope's probe.

Ophthalmic applications:

(1)*Retinal imaging:* Retinal tissue consists of multiple thin layers. High resolution is important factor here again. Eye ball is constantly vibrating at the frequency of several Hz. So, major glaucoma assessment requires whole three-dimensional tomograph to be captured within a few seconds in order to register the mapping of different nerve layers. Because thickness of retina to the bottom (Sclera) is relatively small in the order of 1mm, coherence length can be smaller in this application. 840nm or 1060nm band is chosen for the wavelength band because of low absorption in vitreous humor.

(2)Anterior chamber

A diversity of diagnosis in this area includes early diagnosis of glaucoma, post-LAISK, and iris tumor. Shape analysis is important aspect of anterior segment pathologies. So the depth range needs to reach down to the lens surface that is about 7mm in physical depth. However, since cornea and vitreous gel is transparent and less scattering compared to non-ophthalmic human tissues, ex. skin, the decay of OCT signal due to finite coherence length does not fade the image contrast as abruptly as it does for skin. Hence, 5 mm coherence length is sufficient for most of cases. The use of 1.3 μm range is advantageous in terms of getting clear contour in less scattering surfaces and deeper penetration near iridocorneal angle opening [41].

(3)Intraocular distance measurement:

Axial length of eye, or so-called intraocular distance [15] is the important parameter that needs to be accurately measured prior to intraocular lens implantation. The axial length ranges 20 to 35mm for the normal human eye. Since this is only one-dimensional length measurement, swept rate can be around 1 kHz or higher. In order for the light to reach retina, 1060nm range or lower wavelength range should be chosen to minimize the absorption of water in vitreous humor.

Hard and soft oral tissues or dental applications:

Oral cancer and caries detection are interest of diagnosis for OCT [42]. The requirement is similar to that for GI-endoscopy, except for dental hard tissues, i.e. tooth. Since a tooth is made of enamel and dentin that have less water presence, less than 10% of volume, 1.3 μm and 1.55 μm ranges are useful especially for caries detection.

1.5 Extended cavity lasers

Tunable lasers

So what type of tunable laser meets these requirements? In tunable lasers, there are many variations of technology. But no particular technologies other than extended cavity

Table 1.1 Comparison of different tunable laser technologies

Type of tunable lasers	Tuning mechanism	Tuning range [nm]	Continuous tuning	Speed	Comments	Ref
Grating	200	○	Slow	Pivot tuning	
	AO filter	160	×	Fast		43
	Ring resonator	132	×	-	Vernier effect	44
DFB array	47	△	Fast	Need a booster SOA	45
SSG-DBR	114	×	Fast	Discrete tuning	46
VCSEL	50	○	Fast	Low power	103

lasers suffice the requirements of swept source for wide range tunability in the first place. (see table 1.1). For a monolithic device like DFB array, tuning range per device is limited by principle and cascading in arrays necessitates the need of a booster amplifier. SSG DBR, on the other hand, can tune over 100nm but its discrete tuning nature requires a quite complex control of multiple nodes that leads to rather slower tuning despite the fast response time of device itself. Tunable vertical cavity surface emitting laser (VCSEL) is the only one approach that can realize true continuous single mode tuning but output power and tuning range is limited to a few milliwatt and several tens of nanometers, respectively at this moment.

Extended cavity lasers

Extended cavity lasers or external cavity lasers (ECL) are common sources for transmitter for the long-haul transmission in telecommunication as well as tunable sources for measurement [43,44]. The most of these lasers operate in a single longitudinal mode oscillation. It is difficult to achieve wide range and high-speed sweep operation simultaneously in these lasers because mode-hop free operation during the rapid sweep is

complex, and the control of which is time-limited process.

On the other hand, most of high-speed swept source lasers for OCT that have been introduced since early 2000's have ECL structure with a relatively long cavity [47-49]. The cavity is typically extended meters long with using a single mode fiber so that oscillation occurs with multiple of cavity modes within the Gaussian envelope of the tunable filter, and which can be gradually shifted like a tidal wave on the cavity modes as shown in Fig.1.3. Hence, the continuity of tuning, or "sweep" is ensured by this mode of operation. However, in strictly saying, this mode of operation is no longer "high-coherence" laser, it should rather called, "partially coherent" or "coherent amplification of ASE". This fact imposes us to compromise the coherence length, but achieving several millimeters is in fact sufficient for the most of OCT application. In this thesis, we also adapt the long extended cavity structure for realizing a swept-source. In the following discussion, we explain three key elements/devices suitable for configuring a swept source for OCT applications.

Figure 1.3 Spectral dynamics of a swept-source

Semiconductor Optical Amplifier (SOA) as gain medium

Semiconductor Optical Amplifier (SOA) is considered most suitable for a gain medium of swept source. It is sometimes just called "gain chip" when it is integrated into the device. SOA offers a wide range of selection of wavelength bands, ranging from 0.8 μ m to 1.6 μ m with use of either GaAs or InP based III-V compound semiconductors. Gain bandwidth can be typically over 100nm at 1.3 μ m range that is optimized by carefully engineering the bandgap structure of quantum well. Response time of SOA as short as sub nanoseconds is the most of interest in the use for swept source because over kHz rapid sweep requires intensity modulation in the order of nanoseconds on the longitudinal mode. Gain medium such as rare earth doped fiber amplifier has similar performance in gain level and bandwidth but response time of milliseconds is too slow for rapid sweep greater than kHz swept rate.

Grating based filter

Grating based filter either in Littrow or Littman-Metcalf configuration is considered best-fit as a tunable filter for the swept-source. Several reasons for this; one is filter shape.

Filter shape is in Gaussian profile with box-like shape while Fabry-Perot filter has very peaky and not enough suppression over FSR(Free Spectral Range), and also monotonically decreasing at its slope while AO(acousto-optic) filter has undesired side-lobes that lead to insufficient suppression of ASE during lasing operation. FSR of grating filter is essentially large with appropriate geometry and grating constant. The other reason is that it can be designed into a single polarization selective filter so that oscillation of swept source can be locked in a single polarization within a cavity. Fluctuation of polarization because of undetermined polarization state with polarization insensitive devices leads to polarization phase noise during swept operation that affects on imaging quality as a result.

Wavelength tuning can be realized by changing the incident angle on the grating and which can be repetitively changed by beam deflector if using a resonant mirror or a polygon scanner.

Polygon scanner

Polygon scanner is industrially matured beam deflection device that is used for a broad range of applications. Motor can rotate up to the speed of 50,000 revolutions per minute when using air-bearing axis. Repeatability of beam deflection by using the polygon mirror is quite high; timing jitter less than 0.1%, constancy of speed less than 0.025%, facet-to-facet angle variation less than 0.2mrad, which can realize highly constant and repeatable wavelength sweep over time. High repeatability also enables us to take advantage of wavelength rescaling process in a post-processing method with a pre-calibrated table in the case that non-linearity of sweep should be compensated. However, there is a design rule that restrains FSR of the tunable filter and swept rate. This is discussed in Chapter 4.

Previous works

There are two major approaches that have been studied in MIT and Harvard, and frequently cited by many other works in SS-OCT area. The most of them are all based on extended cavity laser structure as previously mentioned.

Figure 1.4 Fourier Domain Mode Locked laser source (top), ECL with polygon scanner-based filter with Telescopic lens configuration (figures taken from ref. [50], and [51] respectively).

One is Fourier Domain mode locked laser (FDML) [50]. In this laser, a dispersion managed large delay with long rounds of fiber is incorporated into the laser cavity and the narrowband optical filter is tuned periodically at the cavity round-trip time, or a harmonic of the round-trip time (Fig. 1.4 (top)). This produces a quasi-stationary mode of operation. Light from one frequency sweep propagates through the cavity and returns to the filter at the exact time when the transmission window of the optical bandpass filter is tuned to the same optical frequency. This laser has advantage of high-swept rate over 100 kHz maintaining several millimeter coherence length. However, its sinusoidal sweep necessitates the need of real time wavelength rescaling process. And manufacturability is still questionable because of long rounds of fiber that needs to be precisely controlled in length with centimeter order and also polarization should be carefully managed over time.

The other notable approach is similar to the configuration proposed in this thesis; polygon scanner based filter with telescopic lens system [51]. Geometry is in Littman configuration but beam deflection is converted to converging angle via telescopic lens system so that the beam is effectively scanned for wavelength tuning. This configuration has limitation in size

and Finesse. Bulky configuration is also problematic when ensuring durability of the source and its long-term reliability.

Table 1.2 shows performance of a variation of swept-sources proposed from leading research groups including MIT and Harvard.

Table 1.2 List of references of swept sources

Date	Author	Institution	Technology	Main Characteristics Swept rate/range/ coherence length	Finesse	Ref
October 2003	SH.Yun	Harvard(MGH)	Telescopic lens +Polygon scan	15.7kHz/73nm/6mm(0.127nm)	575	51
MARCH 2005	WY. Oh	Harvard(MGH)	2x(Telescopic lens +Polygon scan)	20kHz/ 145nm/3mm(0.25nm)	580	52
December 2005	R.Huber JY. Jiang	MIT Thorlabs	Galvano, Free-space cavity	16kHz/ 133nm/ 4mm(0.19nm)	700	53
December 2005	SH.Yun	Harvard(MGH)	Dual telescopic lens+polygon	115kHz/ 80nm/3.3mm(0.23nm)	348	54
April 2006	R.Huber	MIT	FDML	59kHz/74nm(FWHM)/12mm (0.063nm)	1670	50
August 2008	SMR.Motag hian Nezam	Caltec	Dual reflection with Littman arrangement	50kHz/91.6nm/6mm (0.12nm)	760	55

Although tunable laser technologies have been studied over 30years, there had been no introduction of practically usable source meeting three important parameters since the concept of SS-OCT was first introduced by Fercher (Wien University) in 1995 [15]. Since around 2004, several proposals of swept sources have been reported. However, most of these have two significant drawbacks that make us difficult to implement it into actual medical equipment for real clinical application. One is the fact that these sources are not fully meeting the simultaneous requirement of three parameters. Particularly, coherence length degrades when the source is swept at very high speed over 20 kHz or higher. In Ophthalmology or Cardiovascular applications, large depth range is required with 40-50 kHz very high swept rate. Ambiguity removal process by having phase shifter device was proposed to countermeasure this issue [56-59], but it also complicates the system configuration. Secondly, the transient response of the sweep is intrinsically non-linear so that it necessitates very complex post-process to calculate the OCT signal, and which makes real-time imaging impossible as a result.

1.6 Scope of this study

1.3 μm and 1.06 μm range swept sources are very attractive light sources for SS-OCT in many different clinical applications. Adaptation of SS-OCT in actual diagnostic system has just started.

The major objective of this study is to propose the high-speed wavelength swept source that achieves simultaneous realization of high-speed, high-coherence (narrow linewidth), wideband sweep. Experimental and theoretical study was also carried out for several specific applications.

What we are targeting in terms of specifications in comparison with previous works is better balanced laser. If the swept source design allows sufficient design margin, it means that it is capable of either achieving equally superior on three key parameters (i.e. swept rate, swept range, coherence length), or emphasizing extremely high performance on one specific parameter for a specific OCT application. Target swept rate ranges from 2 to 50 kHz and swept range up to 160nm without compromising on coherence length with any cases.

We propose and demonstrate three major technical innovations in this thesis.

1. Highly k-linear sweep for high-speed or real-time imaging

Accurate information about the wavelength throughout the scan is critical in generating an image with accurate proportions. Wavelength meters or monitoring devices are too slow in response to monitor the instantaneous wavelength at any point during a scan. We have addressed this issue optically with compensating optics inside the tunable filter. Linearity, defined as the frequency deviation/error from a linear extrapolation in time, of less than 0.2% is achieved with extremely good repeatability between scans.

2. Coherence length improvement by Quasi-Phase Continuous Tuning

Deeper penetration is always desired for most of OCT applications. We have proven that without implementing mirror canceling technique or the other costly mitigation method, coherence length can be improved by the proposed technique.

3. Wide range tuning for high resolution OCT imaging

Wide range tuning techniques and special filter geometries based on polygon scanner are proposed. These techniques overcome the physical limitation of tuning range with rather simple configuration.

In this thesis, Chapter 1 is devoted to introductory for the background of OCT and SS-OCT discussing general requirements for the swept source. Review of previous works is introduced to highlight the important parameters that should be overcome.

Chapter 2 describes basic configuration of the proposed swept source, followed by the design for highly k-linear scan/sweep. Other basic performance of the source is also described. OCT image quality with and without wavelength rescaling process is compared for the validation of the fabricated k-linear source.

In Chapter 3, coherence length improvement is demonstrated by Quasi-Phase Continuous Tuning (QPCT) technique. Concept of this technique is discussed and performance in 1.3 μ m and 1.06 μ m source is demonstrated. Measurement of intraocular distance of 20-30mm of pig eye and human eye in 2D tomograph was demonstrated to prove the effectiveness of this method. Estimate of spectral linewidth by introducing multimode rate equations are briefly discussed in the end.

In Chapter 4, various configurations to effectively achieve wide range sweep at higher swept range are disclosed. Two techniques are introduced to effectively maximize the gain bandwidth of SOA. The other two approaches are variations to polygon scanner based Littrow grating filter to broaden the Free Spectral Range (FSR) of the tunable filter or higher the Finesse. Swept rate of 50 kHz is demonstrated using this technique.

In Chapter 5, SS-OCT system is described. Configuration and signal process step is briefly disclosed. OCT image samples with various tissues are shown with a brief clinical point of view.

In Chapter 6, feasibility study on single mode hop-free swept lasers is discussed for the future improvement.

In Chapter 7, concluding remarks are addressed.

The benefits of OCT using a high-speed scanning laser are clear. OCT will significantly reduce the need for surgical biopsy for cancer diagnosis. OCT will transform the diagnosis of retinal diseases such as glaucoma. OCT will be an invaluable tool in treating the occurrence of heart disease and reducing the need for surgery. These are just a few of the many new technologies that will be supported by this new type of high speed scanning laser. Other applications including fiber sensing and spectroscopy will surely follow a similar path taking advantage of this new high speed scanning laser technology.

Organization of thesis

Chapter 2 General design of high-speed wavelength swept source and k-linear sweep

2.1 Introduction

Recent studies advocate the sensitivity advantage of frequency domain or swept source OCT(SS-OCT) over conventional time-domain OCT(TD-OCT)[1,2]. In addition the SS-OCT technique can reduce undesirable imaging artifacts. Motion artifacts caused by the presence of pulsation, internal organs motion, or endoscopic motion significantly affect the quality of in vivo images. Utilizing a high-speed imaging or rapid image capturing technique can reduce detrimental effects of motion. This feature has been discussed previously by Yun, et al. [21]. A key component of the SS-OCT technique is the light source; this must be stable, widely tunable and operate at high sweep speeds. Lasers have been identified as the best approach to realize ideal SS-OCT light source, however development of OCT lasers has been slow since first proposed over a decade ago [15]. In this time much of tunable laser development was focused on single mode oscillation and not designed to sweep repetitively, continuously, over wide range at high repetition rate. For example, semiconductor tunable lasers for telecommunication transmitters are too narrow in tuning range and bulky solid-state lasers are unstable in operation.

Several types of swept source specifically designed for OCT have been reported, many are ring cavity-based lasers integrated with a fast scanning filters utilizing either fiber Fabry-Perot tunable filters or diffraction grating with various beam deflection mechanisms [47-55]. At the same time as these swept sources advances there have been many developments of the OCT technique to deal with specific issues such as mirror image cancellation and phase-resolved measurement. Most swept sources are based on quasi-continuous tuning. Lasing and tuning are realized by multi-mode oscillation in extended fiber ring cavities with an integrated scanning filter. As the filter is tuned continuously, a multiple of longitudinal modes in the envelope of the filter gradually shift to the neighboring modes over the gain bandwidth, building up the gain from the ASE noise with the incoming modes and suppressing the passing modes at the same time. Gain medium

in this case is a semiconductor optical amplifier (SOA). Previous works discussed mechanism of lasing and impact on system performance. As indicated in previous work [49], the faster the speed of wavelength tuning, the wider the instantaneous laser linewidth. This is because the cavity longitudinal modes are stationary with respect to the moving filter window, and the window of filter is moving to next adjacent group of modes at a rate faster than the average cavity lifetime when the sweep rate is over kHz order. This results in insufficient built-up of the gain and suppression of the passing modes. Numerical analysis to explain the detailed dynamics including nonlinear effects of SOA was previously reported by Bilenca, et al. [60].

Another important aspect in swept lasers is the sweep performance including repeatability, swept range and linearity of sweep. Swept sources that use a scanning Fabry-Perot filter or a diffraction grating with a resonant mirror have a non-linear, sinusoidal temporal change of frequency. In OCT systems, non-linear temporal change of scan frequency necessitates the calibration process in k-space after data acquisition of interference signal. Since data acquisition (DAQ) boards samples data in constant time intervals, this calibration process burden the imaging speed because of the buffering of data for processing, thus hinders real-time display of images. This aspect is drawbacks for high-quality, high-speed imaging in OCT systems.

In this Chapter, we present and discuss a high speed, wide range wavelength-swept laser source that has highly linear sweep. The swept laser is designed in a simple and robust configuration with using a diffraction grating and polygon scanner combined with novel compensation optics. The swept source also incorporates a built-in start photodiode that provides external trigger signal to be fed to DAQ board for each axial scan for synchronization of axial scan and frame. The source has a linear k scan capability taking advantage of high linearity and repeatability of sweep of this laser configuration. The laser generates 15mW peak output power with 100nm tuning range FWHM and 117nm in total range at 20KHz scanning rate. The performance of the laser is investigated, including the coherence length that restricts the axial measurement range for OCT application. An OCT image is obtained from a system that utilizes this laser source with and without wavelength calibration.

2.2 Principle and design

2.2.1 Laser Configuration

Figure 2.1 shows the configuration of the wavelength-swept laser source. The laser is configured with an SOA (COVEGA: BOA1017) as the gain medium and a grating-based scanning filter in the extended cavity. Wavelength tuning is achieved by using a diffraction grating (1312line/mm) and a polygon scanner in Littrow arrangement. A 50% partial reflector at the other end of the cavity is attached to the GRIN lens collimator from which the output light is extracted. Amplified spontaneous emission within the window of this tunable filter is oscillated in the cavity. Insertion loss of the filter was less than 1.5dB, and bandwidth of filter is $\sim 0.15\text{nm}$ with the beam width ($1/e^2$) of collimator of 0.8mm. The polygon scanner has 30 facets mirrors to achieve 20 kHz scanning rate at 40,000rpm and is

Figure 2.1 Swept laser configuration

Figure 2.2 Temporal evolution of beam deflection inside the filter block

sealed inside the protective package with an AR-coated glass window; a dust-absorbing filter is implemented to prevent facet contamination inside the package. The rotational direction of the polygon scanner is chosen to have positive saw-tooth scanning starting from shorter wavelength to longer wavelength since the reverse sweep, high to low wavelength, induces a self-frequency shift due to four-wave mixing by carrier density modulation in the SOA [61,62] lowering laser output power as a results. A scan trigger signal derived from an integrated detector monitoring light in the 0th order diffraction light path of the grating is provided to enable the start of each scan to be synchronized. Only when the diffracted 0th order light beam at shorter wavelength intersects the photo detector, does a voltage output arise as a Gaussian response in time. Adding a narrow bandpass filter in front of the detector enhances sharp PD temporal response. The voltage at preset threshold is converted to the TTL output by conversion circuitry integrated in the unit.

2.2.2 Scanning filter design

In the scanning filter the polygon scanner continuously deflects the beam to the diffraction grating with a range of angle determined by the number of its facet. This change of angle corresponds to the change of selected wavelength inside the cavity. Longitudinal mode separation is determined by the round trip path length of the cavity, in this case a 4m length corresponds to approximately 70 MHz mode spacing. A collimated beam from the SOA is expanded by a series of prism expanders in the plane parallel to the deflection in order to narrow the filter bandwidth, thus narrowing the linewidth of laser output. The total magnification of three prism expanders M^3 before the polygon scanner is adjusted to have about a beam diameter as small as 1/3 of polygon facet width. This ensures the effective duty cycle of scanning greater than 2/3 or 65%. Incident angle to the prism expander is set to near the Brewster angle. Setting the prism at Brewster angle minimizes insertion loss because of the grating and prism combination, the diffractive grating having higher diffraction efficiency in S polarization. The other sides of the prisms that have nearly normal incidence is AR coated for additional reduction of insertion loss. The four beam expanding prisms placed in a complementary direction with respect to one after another in terms of refraction of the beam so that the resultant chromatic dispersion of the prism pair is effectively zero. The magnification of single expander M for first three is set to x1.5 by arranging the prism angle $\alpha=30\text{deg}$, and beam incidence θ around 60deg which is near the Brewster angle. The filter bandwidth of diffractive grating is expressed in Eq. (2.1):

$$\delta\nu = \frac{c}{2\pi\omega_0 \tan(\beta - \nu)} = \frac{\lambda a \cos(\beta - \nu)}{\pi w_0} \quad (2.1)$$

where β is the relative angle of the prism with respect to the grating, ν is the exit angle of

the beam from the normal to the prism second facet, and a is the pitch of the grooves of the diffraction grating, or the inverse of grating period Λ . Provided that the beam width after the polygon mirror is w_i , then the magnification of the beam after the prism expander w_o is given by:

$$w_o = M_4 \cdot w_i \quad (2.2)$$

$$M_4 = \frac{\cos \phi \cdot \cos \nu}{\cos \theta \cdot \cos \mu} \quad (2.3)$$

where ϕ and μ are the refracted angle at first and second facet inside the prism. On the other hand, filter wavelength of the diffraction grating in Littrow configuration can be simply expressed as:

$$\lambda = 2a \sin(\beta - \nu) \quad (2.4)$$

2.2.3 Sweep linearity

High linearity of frequency sweep is an important design goal for a scanning laser to be used in OCT investigations. By minimizing the deviation of the laser frequency from a linear extrapolation in time, high-resolution images are obtainable using simple time-constant FFT techniques. Many reported swept lasers have non-linear frequency change over the period of scan. This can be because of mechanical limitations in the tuning element of lasers, such as the sinusoidal drive of a galvanometer. In addition intrinsically non-linear tuning elements, such as diffraction gratings, will result in non-linear scan. For such lasers the instantaneous frequency can only be known by continuous monitoring with an external wavelength or frequency reference. A trigger signal derived from the frequency monitor is used to recalibrate measured OCT interference signals into frequency or k-space before they are sequentially processed to generate images. This process introduces the implementation of an appropriate algorithm for rescaling which may lead to a large task on the data processor and may interrupt the real time imaging capability due to the need of data buffering. Also, this error, the mismatch or inaccuracy of reference to the actual instantaneous frequency results in degradation of signal to noise ratio and resolution. In the case of linear frequency sweep in time, no complicated data rescaling is required, nor the external frequency reference necessary.

The linearity of the laser described in this study is measured to be less than 0.25% with extremely good repeatability between scans. The high degree of linearity in the scan of the laser is a result of using a prism to compensate for the intrinsic non-linear wavelength selection of the diffraction grating. The polygon scanner used in these investigations has a

linear rate of beam deflection since the velocity of revolution is precisely locked to a clock-circuit, the deviation of which is less than 0.025%. If the beam is directly incident upon the grating, then diffracted wavelength or frequency is a sinusoidal function of incident angle as in Eq. (2.4). The frequency change is positive parabolic slope over change of angle. On the contrary, the refracted angle of exit beam from prism expander (that equals the incident angle to the grating) has a negative curve of non-linearity in terms of the angle variation. So, in the case that there is a prism arranged in between polygon scanner and grating as shown in Fig.2.2, these two factors may be cancelled out to yield linear variation of frequency over time as a result.

Here, frequency change in terms of deflection angle θ at polygon scanner is given by:

$$f(\theta) = c/\lambda = \frac{c}{2\alpha \sin\left[\beta + \sin^{-1}\left\{n \cdot \sin\left[\alpha - \sin^{-1}(\sin \theta / n)\right]\right\}\right]} \quad (2.5)$$

Linearity, here, is defined as the deviation of optical frequency from linear extrapolation of a single trace of frequency scan over time. A certain combination of parameters (i.e. α and β) in Eq. (2.5), if appropriately optimized, can reduce the non-linearity of sinusoidal change of wavelength. It is more intuitive to understand the mathematical solution of high linearity though numerically study rather than by analytical derivation. Figure 2.3 shows the linearity with various angles β when $\Lambda = 1312$ lines/mm, provided that the required wavelength scan range is fixed at 1260 to 1360nm. Chromatic dispersion of refractive index of the prism glass (BK7) is also taken into account for the beam refraction in which frequency change as the beam deflects.

As seen in Fig. 2.3, when the deviation from linear extrapolation traces the third order polynomial curve, and the middle flexion point coincides with the center of the frequency range, the linearity reaches near minimum. This is when β is approaching to 68.5 degree in the case that $L=1312$ lines/mm and $\alpha=30$ degree. This translates that high linearity solution exists when $d^3 f(\theta)/d\theta^3 = 0$ around at the center frequency. Figure 2.4 shows the high linearity solutions with different diffraction constants, implicating that smaller constant would yield better compensation thus higher linearity. For an example, when $\Lambda = 1000$ lines/mm linearity can be minimized to less than 0.05% while 1312 lines/mm yields 0.25% linearity.

Figure 2.3 Linearity over frequency range with different β

Figure 2.4 Linearity over frequency range with different Λ

In addition to the linearity performance, when the angle β is provided in the range close to the calculated value in the above, the filter bandwidth tends to be uniform over the range of incident angle corresponding to the wavelength scanning range. This is because the magnification of the prism expander decreases toward the longer wavelength due to steeper incident angle so that it cancels the narrowing of the filter bandwidth according to Eq. (2.1). This balances the bandwidth over the wavelength range as a result. This uniformity is advantageous performance especially when the source is used for OCT system combined with spectroscopic application or for the other applications that need the spectral parameter encoding such as the fiber sensing system. Figure 2.5 shows the calculated filter bandwidth of diffractive grating with respect to the wavelength scanning range, where parameters are

Figure 2.5 Filter bandwidth over wavelength range

set as $\alpha = 30^\circ$, $\beta = 68.5^\circ$ and $\Lambda = 1312$ lines/mm. The filter bandwidth in this arrangement is uniform having a bandwidth of 20GHz (0.13nm) over the entire 1260 – 1360nm-tuning range.

Figure 2.6 shows a simulation result of point-spread functions in the case of the ideal linear sampling with no error (IDEAL), and k-linear scan with an error of 0.1%, 0.25% and 0.8%, respectively. Frequency traces are derived from Eq. (2.5) for each case. Simple FFT is performed with Gaussian window on the sinusoidal fringes equivalent to the data sets that can be obtained when the mirror is positioned at 1mm depth position in the OCT system. K-linear scan with about 0.25% linearity exhibits moderate but tolerable axial resolution, and no S/N degradation, while 0.8% shows significant degradation in axial resolution, which is not tolerable in the OCT imaging.

Figure 2.7 shows depth dependence of axial resolution with different linearity values. 0.25% linearity gives tolerable resolution up to about 1mm, and 0.1% linearity gives better

resolution up to 2mm depth range. Choice of other window functions may additionally improve the tolerance against depth range. [63]

Figure 2.6 Simulated point spread functions (signal spectrum)

Figure 2.7 Depth dependence of axial resolution

2.2.4 Other parameters

Other important design parameters for the source are discussed in the followings. When the faster scanning rate is desired, the number of polygon facets needs to be increased because the upper rotational speed of polygon scanner is limited. However, this change decreases the FSR of the scanning range because of reduction of the deflection angle as well as so called walk-off issue as depicted in Fig.2.2. Scanning range and rate of the scanning filter considering the above issues are determined by the following equations:

$$F_s = Nf_p \quad (2.6)$$

$$\Delta\lambda_{Littrow} = 4a \cos \theta_0 \cdot \Delta\theta_m / M \quad (2.7)$$

$$\Delta\theta_e = 720(1 - \frac{w_o}{L}) / N \quad (2.8)$$

where θ_0 is the incident angle at center wavelength and $\Delta\theta_e$ is effective deflection angle range and L is the facet width of polygon scanner with a diameter, D , $L = \pi D / N$. The term $1 - w_o / L$ in Eq. (2.8) represents the effective duty cycle of the scanning filter. Polygon scanner is industry-proven technology and suitable adaptation in order to realize repetitive beam scanning at high-speed rate. The polygon scanner of air suspended axis type has a rotation speed ranging from 5,000rpm to 40,000rpm. Scanning rate of a swept source is simply the product of the number of facet N of the polygon scanner and rotation rate f_p that is the inverse of rotation speed as in Eq. (2.6). For an example, the rotation rate f_p of the polygon scanner used in this laser is 666.7Hz (40,000rpm), yielding a scan rate of 20kHz when the number of facet N is 30. Eq. (2.7) yields 125nm range where parameters are set as $w_i = 2.7\text{mm}$, $D = 60\text{mm}$

Trade-off relations of scanning rate and range is shown in the Fig.2.8 according to the Eq.s (2.6), (2.7), (2.8).

When the scanning rate requirement ranges around 5 to 40kHz, diffraction gratings with a grating constant in between 1000 to 1300lines/mm suffice the scanning range of practical 80 to 100nm over which the gain bandwidth of SOA can cover.

Figure 2.8 Scan rate and scan range vs. the number of polygon facet

2.3 Laser performance

Figure 2.9 shows the temporal laser output power spectrum and a start trigger signal of a fabricated swept source measured with the oscilloscope. The peak and average power were 15 mW and 9 mW, respectively over the 1267 to 1384 nm range. Duty cycle is about 75%, which is greater than the expected value of 65% calculated from the term $1-w_o/L$ in Eq. (2.8). This estimation supposes the lasing is on duty when full beam width is contained on the single facet of polygon. 15% increase of duty cycle indicates that a small portion of beam despite the walk-off transition on the polygon is partially contributed to the lasing. Total scanning range is 117 nm and FWHM is 100 nm.

In the fiber extended external cavity lasers like our example, laser stability is mainly dependent on two factors; polarization stability and the filter block alignment. In the case of 1.3 μm range laser, polarization alignment is relatively insensitive to the stability because of small polarization dependent gain of about 1dB or less of the SOA device in 1.3 μm range. Figure 2.10 shows the photograph of the fabricated scanning filter part and mechanical drawing of laser source. All the optics is aligned on the metal base plate allowing simple but precise positioning of optical arrangement in a single degree of adjustment axis. Total optical path length of the filter part is only 10cm in single path owing to the compact size. This configuration provides not only production robustness but also high reliability of the operation. Measured output power fluctuation were less than 1% including scan-to-scan variation over a temperature range of 10 to 40 degC over 8hours. In order to further verify the rigidity of the proposed structure, we have carried out the mechanical integrity test according to Telcordia GR-63 and confirmed the fabricated unit passed the tests including mechanical shock and vibration tests (table2.1). Temporal change of wavelength was measured by counting the fringes from the output of a Michelson interferometer with FSR of 4nm.

Wavelength jitter of the start trigger was measured to be less than 0.1nm, and the linearity of tuning in frequency was found less than 0.25% as shown in Fig.2.11, close to the predicted linearity. Further optimization is possible by refining the combination of prism angle and grating pitch. Figure 2.12 shows the relative intensity noise (RIN) of the light source during operation. It had a peak value of -100dB/Hz at near DC to 4MHz range, and -130dB/Hz over 20MHz in between other peaks of about -100dB/Hz at harmonic frequency of 70MHz that is resulted from beating of the laser longitudinal mode.

Figure 2.9 Output spectrum of the laser source per scan at 20 kHz scan rate

Table 2.1 Reliability test results

Test items	Conditions	Criteria	Pass/Fail
Vibration Test	0.3mmp-p amplitude, frequency 10 to 55Hz	Max output power change <1dB p-p, and the rest of spec in spec.	Pass
Shock test 1	side fall: side of the unit is lifted in 30deg from the floor and dropped in free fall(4 sides)		Pass
Shock test 2	corner fall: side is lifted 20mm and opposite side of corner is lifted 25mm and dropped in free fall(4 corners)		Pass
Drop test	When pakced in cardoboard box and fropped from 80cm height(6 faces)		Pass
Temperature range	10 to 40 degC		Pass
Temperature cycle	-20 degC 30min. <-> 60degC 30min. (1degC/min slope) 100cycles		Pass
Damp heat test	60degC 80% RH, 100hours		Pass
Operational Vibration test	Office vibration (Telcordia GR-63CORE) 0.1G, 5 to 100Hz, 0.1 octave/min.	In spec before/after	Pass
Operational Vibration test	Earthquake1 (Telcordia GR-63CORE) 1.6G, 15 to 50Hz, 2 octave/min.		Pass

Figure 2.10 Photograph of the scanning filter block (upper) and 3D CAD schematic of fabricated laser source (bottom)

Figure 2.11 Measured linearity of the scan in dots and theoretical line

Figure 2.12 Relative intensity noise at 20KHz scanning rate

2.4 OCT System evaluation

An SS-OCT system with a fabricated swept laser source was configured as shown in Fig.2.13, in order to demonstrate the capability of the swept source in actual OCT images and the effectiveness of its linear sweep. Output beam from the source is introduced into a fiber-based Mach-Zehnder interferometer with a 1:99 1x2 fiber coupler, two circulators, and a 50:50 2x2 fiber coupler. The through port with 90% coupling ratio of the first coupler is connected to a probe arm through a three-port circulator. Probe arm comprises a collimator lens with a focal length of 10mm (Thorlabs.Inc.) and a galvano scanner (Cambridge Technology 6210H), achromatic lens with focal length of 100mm yielding the lateral resolution of about 25 μ m. The 1% port of the first coupler is connected to the reference arm which is configured symmetrically with a same circulator, collimator and achromatic lens as those used in the probe arm. Probe and reference beams are recoupled by 2x2 coupler and two outputs are fed into the balanced photoreceiver (newfocus 1544). A variable optical attenuator is used to adjust the dynamic range of interference signal and optical delay line combined with the objective lens and a reflective mirror is used to adjust the zero delay to match the focus area of the probe arm. An additional visible laser light at 633nm is

Figure 2.13 SS-OCT system configuration

CP1: 1x2 1:99 fiber coupler, CP2: 2x2 50:50 fiber coupler, CP3: 1x2 30:70 fiber coupler, OC: Optical Circulator, PC: Polarization Controller, ATT: Variable Optical Attenuator, OL: Objective lens, G: Galvano scanner, ODL: Optical Delay Line, BR: Balanced Receiver, LPF: Low Pass Filter

introduced from the reference arm through the fiber coupler and is used for a guide maker on the sample surface in order to make it easy to aim the same target position to be measured. The spectral interference signal is converted to voltage, and digitized by a 14-bit DAQ board (PDA14, Signatec.Inc.) with sampling rate of 100 Msample/sec that is mounted inside a PC unit.

The start trigger for A-scan synchronization is fed from the integrated start trigger output from the source into the A/D converter. Each start trigger initiates the segmented sampling of about 3500 counts that are transported to the PC, followed by FFT calculation. The start trigger is set to lag about 10-15% of total scan range behind the actual rise of the lasing start in order to maintain stable detection. Because of this, pre-trigger mode is used in order to make use of 100% sampling over the effective scan range. Pre-trigger mode is a function in the DAQ board to recall the sampled data prior to the input start trigger using internal buffer on the board. Wavelength rescaling of interference signal is processed in the following steps. First, we obtain an interferogram with using a reflective mirror positioned at 1.5mm depth position in order to extract the calibration data, i.e. the time array of peaks and valleys of the interferogram. This time array is then polynomially approximated up to 4th order. By using this data set to rescale the sampled data in actual OCT measurement, the nonlinearity of the temporal frequency change can be properly compensated. Similar rescaling process is discussed in ref. [41]. Signal processing is performed with using LabVIEW program. 512x1024 pixel image can be output within 0.3sec including process time for wavelength rescaling. The process time without wavelength rescaling was less than 0.1sec, however, with LabVIEW-based program we cannot accurately estimate the duration of process time because the other programs are running simultaneously consuming PC's memory and run time. We expect the reduction of process time is significant when this is adapted to a programmed FPGA board.

The sensitivity of this SS-OCT is over 100 dB while the shot-noise limited sensitivity is 119 dB. Figure 2.14 shows the power spectrum at different depth (a) with and (b) without wavelength rescaling process using about -40dB reflector. Signal power drops 6dB at around 2.5mm in both cases, and no significant degradation of S/N is observed even for the case with no rescaling process. The arrangement of the intracavity prism in the source to achieve 0.25% linearity has a slightly larger beam magnification in comparison with the case of 0.8% linearity, which results in a narrower linewidth thus a larger coherence length. This may account for the reason why the signal with rescaling process appears to drop slightly faster against depth position than that without rescaling process. The calculated linewidth using the measured coherence length L_c from Eq. (2.9) is 22GHz, which is almost equal to the filter bandwidth.

$$Lc = \frac{2 \ln 2}{\pi} \frac{c}{\delta \nu} \quad (2.9)$$

In the case when wavelength is rescaled by calibration data as shown in Fig.2.15, axial

(a) With wavelength rescaling process

(b) Without wavelength rescaling process

Figure 2.14 Power spectrums

resolution is about $12.5\mu\text{m}$ ($9\mu\text{m}$ in tissue) at FWHM in air at the depth of 1mm, while theoretical value is calculated to $7.7\mu\text{m}$ in air. Large discrepancy between the theoretical value according to Eq. (2.9) and the measured value may be attributed to the wavelength dependent loss of the components such as optical circulators and collimator lenses used in

the interferometer. Accumulated loss at the edges of 100nm wavelength range was measured about 2dB with respect to the center of the range, which narrows the spectral profile of the output power of or the envelope of the interferogram by about 25%. Other calculation errors and noise factors as well as polarization mismatch in the interferometer due to the miss-adjustment of polarization controller may account for additional degradation of axial resolution to some degree.

Figure 2.15 Power spectrums at 1mm

Figure 2.16 Axial resolutions as a function of image depth

Dashed line shows a spectrum at 1mm depth without rescaling process. Axial resolution of this case is about $13\mu\text{m}$ while the value by previous simulation (in Fig.2.7) was $11\mu\text{m}$. The wavelength dependent loss of the system and the chromatic dispersion of the interferometer of the system may account for the error in this case. Here, the magnitude of the axial resolution is mostly dominated by the factor of nonlinearity, but not as much by the wavelength dependence as in the case of rescaling process. Spectrum width below -20dB tends to be wider as the depth gets deeper beyond 1mm point.

The results indicate that a linearity of frequency sweep as high as 0.1 to 0.3% can maintain moderate axial resolution as expected and practically eliminate the need of an auxiliary optical k-triggering to be implemented into the OCT system as long as the depth range is relatively shallow about 1-2mm range which is practical range of use in the most of OCT applications as shown in Fig.2.16. The previous calculation as in Fig.2.4 suggests that the use of a lower density grating of less than 1100lines/mm is effective to improve the linearity to 0.1% or less, and which can further improve the axial resolution and S/N at deeper range. As proposed in the previous report, the double pass filter configuration may overcome the disadvantage of shorter coherence length when using lower modulation grating that will be discussed in chapter 4.

In addition, we found that the uniformity of wavelength dependent loss of the interferometer system is another important factor to improve in order to reduce the axial resolution below $10\mu\text{m}$ order or to reach closer to the theoretical value.

Figure 2.17 shows two-dimensional tomograph of human skin on the finger. Sweat ducts and epidermis are apparent in the images (b), (c). (a) is the image with no wavelength rescaling process with 0.8% linearity scans. In this case, the broadening of the FFT spectrum blurs the spiral appearance of the sweat ducts and boundary structures such as surfaces of dermis and epidermis. On the other hand, no obvious degradation in the image quality is observed in the case of 0.25% high linearity scan without rescaling process (b) with a quality comparable to the one with rescaling process with 0.8% linearity (c).

(a) Finger skin image with no wavelength rescaling process (linearity 0.8%)

(b) with no wavelength rescaling process(linearity 0.25%) (c) with wavelength rescaling process (linearity 0.8%)

Figure 2.17 OCT image of human finger (Sweat ducts and epidermis)

2.5 Summary

In this chapter, we presented and discussed a high speed, wide range wavelength-swept laser source that has highly linear scan. The laser exhibits 15 mW peak output power with 100 nm tuning range in FWHM and 117nm in total at 20 kHz scanning rate. The source has k-linear scan capability taking advantages of the high linearity and repeatability of sweep of this laser configuration. Impact of sweep linearity of the source on OCT images was also discussed. In addition to the basic performance of the source, OCT images are also obtained from a system using the fabricated swept laser source, and the effectiveness of linear scan is verified with actual OCT images for the first time.

The simple and robust configuration of the demonstrated laser source should expand the field of OCT in clinical applications in the near future.

Chapter 3 Coherence length improvement by Quasi-Phase Continuous Tuning (QPCT) technique

3.1 Introduction

Fast wavelength-swept lasers are increasingly being used as light sources for applications that require dynamic or real time measurements with high sensitivity and wide dynamic range. Examples include interferometric measurements such as swept-source optical coherence tomography (SS-OCT) and surface profiling, as well as spectroscopic measurements including fiber sensor systems [6], and gas sensing [1-5]. In the case of SS-OCT and surface profiling, coherence length, or instantaneous linewidth of swept source is a key parameter that determines the measurement depth range in cross-sectioned images. In the fiber sensor systems or gas sensing systems, instantaneous linewidth determines the measurement resolution of the system. To maximize the data acquisition rate of such measurement systems it is desirable to increase the scan rate of the swept laser. In general as lasers move to higher sweep speeds there is an increase in instantaneous linewidth, and decrease in coherence length resulting in reduced measurement depth. Typical coherence length previously reported for SS-OCT was limited to 5 to 6 mm [52-54] corresponding to spectral linewidth of about 0.13nm at swept rates ranging from several kHz to a few tens of kHz. A 5 to 6mm coherence length corresponds to the single side measurement depth of 2.5 to 3 mm in OCT, which has been sufficient for many clinical OCT applications where the depth of interest is limited to the superficial layer and fundamental absorption and scattering in the tissue limits light penetration. However, increase of coherence length is now desired because of following three reasons; 1) demand of increase of the swept rate while maintaining fair amount of depth range, 2) demand of increase of mechanical clearance to the sample surface especially in the case of endoscopic applications so that it can give certain tolerance to the position of the catheter [11,38,40], and 3) the need of larger scale 3D profiling in industrial applications. Previous work has reported on fiber-extended cavity tunable lasers [24,25,47-49]. In these lasers a tunable filter sweeps over the gain bandwidth of a semiconductor optical amplifier (SOA) and accumulated gain on the cavity longitudinal modes within the envelope of the filter will result in lasing output from the fiber cavity. In

most cases, cavity longitudinal modes are virtually stationary during the sweep as shown in Fig.3.1. If the tunable filter is stationary or slowly tuned, there is strong cavity mode selection within the filter window resulting in narrow spectral envelope of laser output. On the other hand, if the tunable filter is swept at over a few kHz rates, cavity mode selection is reduced resulting in an increase output linewidth. The increase in mode competition is due to the mismatch between the modes preferentially amplified in the SOA and the center of the scanning filter over each roundtrip. This is mainly because the most of swept sources are designed to have a fixed cavity length where cavity modes are relatively stationary with respect to the high-speed swept tunable filter.

In order to overcome the increase in output linewidth at higher scan rates, several ideas have been introduced such as the phase matching technique using accousto-optic filter, that matches the wavelength shift over a round trip to the phase shift generated by the filter itself [64]. Another technique is Fourier domain mode locking (FDML), whereby the tunable filter scan frequency is matched to the optical round trip time resulting in a higher Q factor of the cavity in frequency domain [50, 65-68]. These two approaches, however, require both to operate at a preset resonant condition, i.e. at fixed swept rate, and the latter case needs long fiber length to accommodate several tens of kHz swept rate or slower. Other than using these techniques, adding the ambiguity or complex conjugate removal by adding external phase shifter in the OCT system is known as an alternative way [56-59], but it is not preferable when the system design is cost-sensitive.

Figure 3.1 Diagram of multi-mode lasing with Gaussian filter envelope

In this study, on the contrary to the above approaches in frequency domain, we demonstrate a rather simple but effective configuration to achieve narrow instantaneous linewidth in time

domain. We here call this approach Quasi-Phase Continuous Tuning (QPCT) technique, which is analogous to the classical pivot tuning mechanism in external cavity single longitudinal mode laser [69] which we modified and adapted to the fiber-extended cavity.

3.2 Concept of Quasi-Phase Continuous Tuning (QPCT)

Figure 3.2 shows the concept of a QPCT laser in comparison with the single mode external cavity laser with perfectly phase-continuous tuning. In a conventional single mode-hop free tuning as shown in Fig.3.2(a), in a simplified case of Littrow arrangement, if the grating is rotated around the so called pivot, the cavity mode and the filter center is completely synchronized over large tuning range without having a mode-hop [69]. This is a novel tuning

Figure 3.2 (a) Configuration for phase continuous tuning (b) Concept of quasi-phase continuous tuning for mode-hop free tuning

Figure 3.3(a) Phase diagram of Lasing in Fig 2(a)

(b) Phase diagram of Lasing in Fig 2(b)

method because only one rotary motion is required for continuous tuning, and it has been widely applied to the tunable laser instruments. However, the tuning speed or slope is limited to a few seconds per sweep over 100nm at 1.3 or 1.55 μm wavelength range. This is because the pivot of the rotation is largely off from the optical axis or the point of diffraction, the faster modulation of large momentum is impossible to achieve with today's mechanical actuation technologies. In the case of multi-mode envelope tuning as previously explained, wavelength is tuned continuously over a group of longitudinal modes, gradually shifting to next adjacent modes ensuring continuity of tuning. Figure 3.2 (b) shows the schematic of wavelength swept laser based on QPCT with an SOA and a tunable filter that comprises a diffraction grating and deflection mirror in a Littrow arrangement. Deflection of the collimated beam from the fiber output of the SOA by the scanner mirror changes the incident angle to the diffraction grating thus changes the wavelength of diffracted light couple back to the cavity. At the same time during wavelength sweep, the deflection of the beam modulates the phase by the change of diffraction point on the diffraction grating, which corresponds to the cavity length change. If we set an appropriate geometry so that the gradient of wavelength change and phase are nearly equal, the wavelength and the phase can be synchronously modulated to a certain degree. Figure 3.3 (b) shows the evolution of phase in step by step. Even though the filter's peak wavelength is not perfectly matched to each one of longitudinal mode over entire wavelength range, a majority of longitudinal modes remains in a filter window over the small increment of time allowing a few round trip of light during a scan. As a result, it is expected to enhance the coherent build-up of the laser modes and increase the Q factor of the cavity during the sweep, which was confirmed in the experiment described in section 4.

3.3 Conditions for QPCT

As previously mentioned, the relation of gradient of wavelength and the phase change simply determines the condition for quasi-phase continuous tuning. Here, total cavity length L is a function of grating incident angle θ , which is expressed in the sum of the optical length from the deflection point to the grating, $H/\cos\theta$ and the length to the other end of cavity, L_1 , thus $L=L_1 + H/\cos\theta$

Wavelength of the filter, λ_g and the wavelength of the cavity longitudinal mode, λ_L at the center of wavelength range are expressed in:

$$\lambda_g = 2a \sin \theta \quad (3.1)$$

$$\lambda_L = \frac{L(\theta)}{L_0} \lambda_L(\theta_c) \quad (3.2)$$

L_0 , θ_c , a are the length and the diffraction angle at the center of wavelength range, and grating pitch, respectively. Then the gradients of these Eqs.s are expressed in (3.3), (3.4)

$$\lambda'_g = d\lambda_g / d\theta = 2a \cos \theta_c \quad (3.3)$$

$$d\lambda_L / d\theta = \lambda_c H \tan \theta_c / (L_0 \cos \theta_c + H) \quad (3.4)$$

, and are set to equal, then following relation of geometry is derived.

$$H = \frac{\cos \theta_c}{\tan^2 \theta_c - 1} \cdot L_1 \quad (3.5)$$

This is a basic condition for QPCT design, where $\theta_c = \arcsin(\lambda_c/2a)$.

Figure 3.4 shows the ratio of H/L_1 with various diffraction grating constants ($A=(1/a)[\text{g/mm}]$) and with respect to the center wavelength of the tuning range. This graph indicates that depending on the selection of the grating constants A , appropriate ratio of lengths around the point of deflection, H/L_1 has a large variation with respect to the desired wavelength range, and also shows that the lower the wavelength range is such as 1050nm range, the higher the grating constant should be chosen in order to fulfill the condition. Figure 3.5(a) shows an example of wavelength and phase trace when the above condition is fulfilled. Figure 3.5(b) shows the cavity mode order n that coincides with filter's peak wavelength over tuning range. In terms of value R as in reference [72], which is defined as the relative frequency change of the cavity modes with respect to the free spectral range during a single roundtrip, R is substantially small. This mode of operation is completely opposite approach of modeless operation as described in ref. [72].

We can quantify the degree of phase synchronization in terms of temporal phase lag between longitudinal modes and the filter's peak wavelength over entire wavelength range as the difference of gradients;

$$P = \frac{d\lambda_g - d\lambda_L}{d\theta} \quad (3.6)$$

Figure 3.4 QPCT conditions with different wavelength ranges

Figure 3.5 (a). Phase variation over diffraction angle range

(b). Cavity mode order over wavelength

Figure 3.6 Out-of-phase ratio P over wavelength range

Figure 3.6 shows the factor of phase lag, P over wavelength range with grating constant of 1312 g/mm. As shown in Fig. 3.6, the calculated phase lag over small increment of wavelength change is at worst 40 % at the edges of tuning range, in the other words, 60 % are synchronous resulting in a high cavity resonance leading to narrow linewidth of the source. For an example, for an optical cavity length of 3 m corresponding to a cavity round trip time of 10nsec, and the filter tuning 100 nm at 20 kHz, the tuning slope is 2 nm/ μ sec. As described in previous work [49], when using an SOA with ASE power of a few milliwatts and over 20 dB gain and a tunable filter with 2-3 dB insertion loss, it takes about three round trips to build up to the saturated optical output. During three round trips, i.e. 30 nsec, filter wavelength shifts by 0.06 nm, which is nearly a half of filter window. In the other words, if the filter window shifts about a single increment of window in the amount of filter bandwidth; 0.12nm, there will be no contribution of any preceding cavity modes from the previous window of the filter to a next increment of window. On the other hand, in the case of QPCT, there is always over 70 % of phase or cavity modes carried over from previous round or window of oscillation as it travels the cavity contributing to the build up of the gain, resulting in narrower linewidth. It is equivalent to the slowing the tuning speed of the tunable filter with respect to the cavity modes.

Cavity length is another important factor in order to improve the resonance of the cavity. The shorter the cavity length is, the higher the gain builds up on lasing cavity modes during the high-speed sweep, by allowing many roundtrips within the filter's moving window. However the shortest cavity length is limited by the continuity of sweep. If it is too short, unstable mode competition or bistability within a small number of cavity modes within the window results in unstable or noisy temporal optical power profile. In order to have a continuous and stable sweep of wavelength, a few tens of centimeters to a few meter long cavity lengths is considered appropriate. In this experiment, since the use of single mode fiber instead of free space makes it much easier to configure the laser cavity in terms of the alignment and handling, a pigtailed SOA is used as a gain element which is connected to the tunable filter via a collimator lens and terminated with a partial reflection mirror on the other side of fiber pigtail to form an extended cavity structure as shown in Fig. 3.7. The shortest fiber length that we can prepare for the pigtail part was 0.6m from end to end corresponding to optical length L_1 of 0.85 m. To have the length of H in practical range of a couple of tens of centimeters for loss-less free-space coupling by the collimator, we selected a diffraction grating constant of 1312 g/mm according to the Eq. (3.5) that gives sufficient synchronization over the tunable range considering the practical length that can be achieved by fiber splicing. From the Fig. 3.4 or Eq. (3.5), the condition for the quasi-phase continuous tuning is found to be when $H=100$ mm. Total cavity length at center wavelength at 1310 nm is calculated to 2.5 m per round trip optical path that corresponds to the photon

lifetime in the cavity of 8.3 nsec. In this case, the number of longitudinal modes is about 150 within the window of 0.1 nm bandwidth of the filter.

Additionally, Doppler shift may arise as a concern when the cavity length modulation is as large as 100 mm in this case. When swept rate is at 20 kHz, the rate of cavity modulation v_f is 100 mm/ 50 μ sec =2,000 m/sec, and the Doppler shift calculated by $\Delta f_{Doppler} = f / (c + v_f)$ is about 1.5 GHz in optical frequency or 9pm in wavelength. This is negligible magnitude because the spectral linewidth of interest in this study is in the order of 0.05 to 0.1 nm. However, either when further linewidth narrowing is necessary to the order equal to the Doppler shift, or when increasing the swept rate resulting in larger Doppler shift, the Doppler shift becomes non-negligible order to the linewidth, and this needs to be accounted for the design. One way to countermeasure this problem is either scaling down the cavity length or choosing the condition of $H \ll L_1$.

3.4 Experimental setup

Figure 3.7 shows the schematic of the QPCT laser that was used to verify the proof-of-concept. The laser comprises an SOA (COVEGA BOA1137) and a polygon scanner based grating filter in a Littrow arrangement. Polygon mirror has an inscribed diameter of 60 mm with 30 facets, and its rotation speed is set at 40,000 rpm for 20 kHz swept rate and 5,000 rpm for 2.5 kHz rate. The rotational direction of the polygon scanner is chosen to have positive saw-tooth scanning starting from shorter to longer wavelength since the reverse sweep induces a self-frequency shift due to four-wave mixing by carrier density modulation in the SOA [61,62] lowering laser output power and also widening the spectral linewidth as a results. Collimated beam with a beam diameter ($2w$ at $1/e^2$) of 800 μ m is expanded to 2 mm in lateral direction before the polygon mirror by three prism expanders. Polygon facet width of 6.3 mm is wide enough so that there is no clipping of the beam within effective FSR of the tunable filter. Polarization controllers at the two arms of fiber pigtail are used to match polarization states of the light travelling back and forwards with respect to the diffraction grating and SOA so that the collimate beam is aligned in S-polarization in free-space before the grating, and input and output beam going into the SOA is aligned in TE mode. Modulation constant of the 1312 g/mm is chosen to tune over 100 nm wavelength range around 1320 nm. FSR of tunable filter is 180 nm. The bandwidth of tunable filter is calculated to 0.13 nm. Tunable range is limited by the physical size of the diffraction grating that should cover the effective deflection angle of the beam at the distance H . The prism expander between polygon scanner and the diffraction grating is arranged in a way such that the linearity of sweep slope becomes higher, eliminating the need of so called k-triggering and wavelength rescaling process [41].

Figure 3.7 Diagram of polygon scanner -based swept laser source

PR: Partial reflector, Col: Collimator lens, PC: Polarization controller,
S: SOA, PR: Prism, G: Grating, POL: Polygon scanner

3.5 Results

Instantaneous linewidth of the source with QPCT was measured at various conditions. The bandwidth (FWHM) of the grating filter of 0.126 nm and 0.05 nm was chosen for validation of spectral narrowing effect by QPCT at two different speeds, 2.5kHz and 20kHz. Filter bandwidth is adjusted by the number of prism expanders inserted in the free-space, thus changing the magnification of the beam. Figure 3.8 shows the oscilloscope trace of the three consecutive scans. Scan range measured by optical spectral analyzer was 55 nm, from 1271 nm to 1326 nm. Peak and average power are 12 mW and 8m W, respectively. Relative intensity noise (RIN) of the source was measured less than -110dB/Hz over all frequency range as shown in Fig. 3.9. The cavity modes spacing of 120-130 MHz, or 0.68 pm to 0.74 pm in wavelength, accounted for the peak at 120-130 MHz that is inverse of the cavity length, and the splitting of which is due to the modulation of cavity length.

In order to measure the instantaneous linewidth, the swept output is fed into a simple Mach-Zehnder interferometer as shown in Fig. 3.10, and a high-speed receiver with 2.5 GHz bandwidth is used to detect the fringe signal. When the amplitude of fringe drops in half compared to that at zero delay, the amount of delay equals to the coherence length by definition. Relation of instantaneous linewidth and coherence length is given by Eq. (3.7). If

the dual side of the range around zero delay is included, the range doubles as long as mirror imaging can be cancelled [56-59].

$$Lc = \frac{2 \ln 2}{\pi} \frac{c}{\delta \nu} \quad (3.7)$$

Figure 3.8. Temporal optical output power

Figure 3.9. Relative Intensity Noise

Figure 3.10. Balanced Mach-Zehnder interferometer

Figure 3.11 shows the signal power spectrum at several depth positions, with and without QPCT. The one without QPCT has a ratio of H/L_1 about less than 0.05 ($H=60$ mm, $L_1=1.15$ m), while the source with QPCT has a ratio of 0.25 ($H=210$ mm, $L_1=0.85$ m). Here we kept the same cavity length for both cases, about 2.5 meters round trip, at the center wavelength. As shown in Fig. 3.11(a), signal power drops 6 dB at the depth of about 2 to 2.5 mm if it is without this technique. The corresponding linewidth nearly equals to the filter bandwidth of 0.13 nm. While, it is extended to over 5 mm if this quasi-phase continuous condition is satisfied, and which corresponds to 0.075 nm of spectral linewidth. If we slow down the swept rate, we get further extension of single side depth up to around 8 mm at 2.5 kHz as shown in Fig.3.12. Axial resolution at 2 mm depth as shown in Fig. 3.11(b) is found about 24 μ m, which corresponds to about FWHM of 55 nm of swept range. The obtained swept range was such narrower than FSR of tunable filter 180 nm (duty ratio of about 30 %) mainly because the length of grating we used is not long enough to cover whole deflection angle of the beam. We believe that by using longer grating in size or shortening the total cavity length can improve the swept range over the gain bandwidth of SOA. Figure 3.13 shows the comparison of spectral linewidth with and without QPCT over different swept rates.

Coherence length improvement or linewidth narrowing effect was in the factor of about X1.8 to X2 from the case without using QPCT over 1-30 kHz swept rate.

(a)

(b)

Figure 3.11.(a) OCT signal at different depth positions with and without QPCT condition

(b) Close-up of spectrum (20 kHz swept rate)

Figure 3.12. OCT signal at different depth positions with and without QPCT condition (2.5 kHz swept rate)

Figure 3.13. Spectral linewidth vs. Scan rate

Replacing the grating to the one with higher modulation constant of 1500 g/mm resulted in a further increase of the coherence length. The calculated filter width in the same configuration except the grating was 0.051 nm. Figure 3.14 shows the measured OCT signal spectrum at different positions at two swept rates, 2.5, 20 kHz. Coherence length was 30 mm, and 17 mm, corresponding to 0.025 nm, 0.043 nm of linewidth respectively at center wavelength of 1289 nm. It was a factor of X1.2 to X2.0 improvement or narrowing with respect to the filter bandwidth depending on the swept rate. Improvement factor becomes lower than the previous case. This is because the phase lag becomes larger when using modulation constant of 1500 g/mm than the one with 1312 g/mm and also because the phase shift per round trip becomes non-negligible order with respect to the order of linewidth. Table 3.1 summarizes the results with different setting of parameters.

Figure 3.14. OCT signal at different depth positions with and without QPCT condition

Table 3.1. List of summary results

Parameter	Without QPCT		With QPCT (1)		With QPCT (2)	
Scan rate	20 kHz	2.5 kHz	20 kHz	2.5 kHz	20 kHz	2.5 kHz
Center wavelength	1310 nm		1310 nm		1290 nm	
Filter bandwidth	0.126 nm		0.126 nm		0.05 nm	
Instantaneous linewidth	0.149 nm	0.09 nm	0.075 nm	0.048 nm	0.043 nm	0.025 nm
Coherence length	5.1 mm	8.4 mm	10 mm	16 mm	17 mm	30 mm
Narrowing factor from filter bandwidth	X0.85	X1.4	X1.7	X2.1	X1.2	X2.0

3.6 Intraocular distance measurement

3.6.1 Introduction

Most standard treatment for Cataract is the implantation of intraocular lens (IOL). Prior to the surgery, IOL power should be obtained by measuring the axial length of the eye that is a distance from retina to cornea. The accuracy of measurement is important for conducting precise implantation of IOL. This ranges from 25mm to 35mm in normal human eye. Conventionally, it was measured by Ultrasonography. Measurement accuracy in this case is limited to 0.05 to 0.1mm range, and measurement itself is an invasive measurement in the sense that the measurement probe should be in contact with the cornea through matching gel.

Figure 3.15 Schematic cross-section of a healthy human eye (left: courtesy of the National Eye Institute, National Institute of Health), and intraocular lens (right)

On the other hand, recently introduced optical methods such as time-domain optical coherence reflectometry [15] improve the accuracy, and which is a non-invasive method by its principle. However, because the measurement time is relatively slow due to slow scanning of mechanical delay line, if the movement of the eye during the measurement may degrade the measurement accuracy. On the other hand, Fourier domain methods such as the OCT for the retinal imaging are known as a popular technique for making high-resolution, high sensitivity imaging possible. Recent work [73] based on Fourier Domain low coherence interferometry similar to SD-OCT at 840nm implements two separate reference arms with different optical length so that you can image two depth ranges simultaneously and calculate the axial length just by adding the offset length between two arms. This approach countermeasures the shortcoming of shallow depth range of spectral domain based measurement that is intrinsically restricted by the pixel resolution of spectrometer. However, this approach still forces the user to search an approximate position to find the signal within the measurable range. This will necessitate the mechanical scanning of the

delay line and also the processing to distinguish the overlapped signals from anterior and posterior area.

Recently, 1060nm range wavelength that has gained good attentions for the retinal OCT imaging because of larger penetration compared to that at 840nm range [74]. This range combined with Fourier domain methods, either in spectral domain, or in swept source OCT promises a next generation OCT system for the retinal imaging. Our further expectation is that OCT can be extended to the large geometry measurement such as for measuring axial length for IOL calculation. However, in a similar way as explained for the SD-OCT, the most of previously proposed swept sources for OCT have the depth range defined by coherence length ranging around 2mm, and which is far below the measurement range required for the axial length of the eye, or swept rates of tunable lasers were too slow [75].. In this report, we propose a swept source that implements quasi-phase continuous tuning (QPCT) [76] and multiple beam expanders in order to improve the coherence length up to 28mm range at swept rate of 2.5 kHz, which is about 5 to 10 times larger than what can be achieved in present systems [77,78]. This swept source enables us a very simple operation of the measurement of the axial length of eye where you only need to divide the obtained distance by the known refractive index. We demonstrate the actual OCT measurement of axial length with using a pig eye. We have confirmed the positions and image of retinal and cornea in 1D and 2D OCT images, and the axial length was measured about 27mm in optical length.

3.6.2 Principle and design

Figure 3.16 shows the actual configuration of the wavelength-swept laser source demonstrated in this study. A fiber pigtailed SOA (COVEGA: BOA1142) and a polygon scanner-based diffraction grating filter are connected together to form an extended ring cavity laser. Effective cavity length is 2.6 meter in round trip. The filter bandwidth is set to about 0.024 nm by using six prism expanders in free-space and a diffraction grating with 1714 lines/mm modulation constant. Output power is extracted from the fiber coupler with 30% coupling ratio. Polygon mirror has 30 facets, 60mm diameter, and rotates at a speed of 5,000 rpm corresponding to 2.5 kHz swept rate. Repetitive deflection toward the diffraction grating in a Littrow arrangement enables the stable sweep of wavelength and the modulation of large optical delay at the same time. The condition of QPCT i.e. virtually equal gradients of wavelength (i.e. $d\lambda/d\theta$) and phase variation (i.e. $d\phi/d\theta$) over the tuning range was realized by having the distance between the deflection point and the diffraction grating set in a certain specific ratio with respect to the total cavity length. This condition is same as in

section 3.3. With given geometry and parameters above, H and L_1 yield 12cm and 1m, respectively for the total cavity length of 2.6 m round trip in optical length.

As seen in the Fig. 3.17, the phase is not perfectly synchronized over whole tuning range between tunable filter and cavity modes. Figure 3.18 shows the degree of phase synchronization in terms of temporal phase lag between longitudinal modes and the filter's peak wavelength over entire wavelength range as the difference of gradients; $(d\lambda_g - d\lambda_L)/d\theta$. There are maximum 40% of phase shift with respect to the tuning wavelength over 40nm of tuning range. Conversely, about at least 60% of cavity longitudinal modes remain in-phase during the sweep over small increment of time, or in the other words, the phase shift per round trip is reduced to 40% compared to the case of non-QPCT conditions, and which is expected to result in the enhancement of cavity Q factor, or large coherent build-up of gain per round trips.

Figure 3.16 Laser configuration

Figure 3.17 Phase change over the range

Figure 3.18 Phase shift ratio over wavelength range

3.6.3 Design trade-off

In the followings, the limitation of the performance is explained. What limits the performance by the geometry is the polygon scanner's rotational speed and diameter, D .

In general, polygon scanner's speed can go up to around 45,000 rpm with sustainable durability and stability with a maximum of 60 to 80mm diameter of the mirror in a higher grade when using the one with air-suspended axis. While, the lower grade scanner with slower speed and shorter diameter allows us to use those off-the-shelf, low-cost polygon scanners from the mass produced industrial product lines like the ones for laser printers.

Swept rate f is the product of rotation rate of the polygon f_p and the number of facets N , $f = f_p N$. The filter width is defined by the Eq. (3.8), where w_i is incoming beam width on the facet of the polygon from the collimator with some magnification prior to the input to the polygon. Effective deflection angle accounting for beam clipping is $\Delta\theta_e = 4\pi (1 - w_i / W) / MN$ where facet width $W = \pi D / N$. This is related to the FSR of tunable filter, $\Delta\lambda$ as given in Eq. (3.9). M is the total beam magnification by cascaded prisms by p after the polygon deflection, $M = m^p$ where individual magnification of each prism expander, m ranges about X1.6 if the angle with respect to the normal is set around 55 degree.

$$\delta\lambda = \frac{2\sqrt{2\ln 2}\lambda a \cos\theta_0}{\pi M w_i} \quad (3.8)$$

$$\Delta\lambda = 4a \cos\theta_0 \cdot \Delta\theta_e = \frac{8\pi a (1 - w_i / W) \cos\theta_0}{MN} \quad (3.9)$$

$$M = m^p \quad (3.10)$$

$$F = \frac{2\pi^2 w_i}{\sqrt{\ln 2} N \lambda_o} \left(1 - \frac{w_i}{W}\right) \quad (3.11)$$

In order to achieve very large coherence length thus very narrow linewidth, original filter bandwidth $\delta\nu$ before QPCT enhancement should also be narrower. However, when we narrow $\delta\nu$ by expanding the beam width w_i , or by reducing grating pitch a , it will also narrow the swept range, $\Delta\lambda$ because effective deflection angle is decreased and the beam clipping will be larger. Finesse is constant as long as w_i and N are fixed as expressed in Eq. (3.11), in this case which is calculated to about 1140 when accounting for beam clipping by the polygon mirror when $w_i = 2.7$ mm. Figure 3.19 shows the relation of FSR and filter bandwidth. Finesse of the filter is counter-proportional to the facet number, and which is reduced when the system requires a faster swept rate with a limited rotational speed of polygon. For the axial length measurement as opposed to the 2D or 3D OCT imaging, the

swept rate of a few kHz is sufficient for the one dimensional measurement of the axial length.

3.6.4 Laser performance

Figure 3.20 shows the temporal power profile of the swept source measured by a detector and an oscilloscope. Output power was 6.2mW at peak, and 3mW in average. Swept wavelength range was 18nm around 1064nm with 40% duty ratio. Coherence length measured by simple Mach-Zehnder interferometer was 28mm corresponding to 0.018nm in instantaneous linewidth at 2.5kHz swept rate. Instantaneous linewidth is narrowed by a factor of X1.37 from the filter bandwidth of 0.024nm corresponding to the coherence length of 20.4 mm. We also measured the coherence length in the case that QPCT is not used by having geometry not meeting QPCT condition but keeping the same cavity length. Without QPCT, instantaneous linewidth was equal to the filter bandwidth. QPCT with factor of X1.37 narrowing is not as effective as in the experiment with a similar configuration using 1.3 μ m range SOA that we have done in section 3.5. We confirmed a factor of X2 improvement from the filter bandwidth of 0.05 nm to instantaneous linewidth of 0.025 nm to achieve a comparable coherence length of 30 mm in 1.3 μ m using a grating with 1500 g/mm constant and a single prism expander. This difference in the results in terms of improvement factor can be accounted for by the difference in the cavity gain and loss factor attributed to the performance of SOA and the loss of tunable filter for 1.06 μ m and 1.3 μ m. Net gain, i.e. the sum of lower gain of 1.06 μ m SOA compared to that of 1.3 μ m and larger loss of circulator in the filter for 1.06 μ m, as well as lower ASE power of SOA enforces more round trips until the gain gets accumulated to the saturated regime than in the case of 1.3 μ m, resulting in a larger phase shift between tunable filter and lasing wavelength. It is also partially because the degree of phase synchronization is not sufficiently high when using a grating with modulation constant as high as 1714 g/mm in this case.

Figure 3.19 Filter bandwidth vs. FSR of tunable filter

Figure 3.20 Temporal profile of output power

However, factor of X1.37 improvement provides about 8mm increase in coherence length in this case compared to the coherence length without using this technique, which is still a significant improvement of measurement range.

Difference in the measured swept range, 18 nm, and theoretical value, 28 nm according to the Eq. (3.9) is due to the additional beam cut-off by the edges of prisms and diffraction grating. The use of optimized polygon scanner with a smaller number of facets as well as prisms and a diffraction grating with larger size will increase the swept range further up to over 80nm, maintaining the same coherence length. As in Fig.3.19, a polygon mirror with ten facets should give over 80nm of FSR while maintaining several kHz swept rate. So lasing with a finesse of over 2,000 is possible in the future by optimizing the polygon mirror and other optics.

Figure 3.21 OCT system setup for IOL measurement

3.6.5 Axial length measurement

The OCT system for the measurement comprises a circulator-less, unbalanced interferometer as shown in the Fig. 3.21, which doesn't use optical circulator to efficiently circulate the signal back to the receiver. We can compromise the sensitivity to a certain degree because we only want to measure a distance of two points. The other process step is similar to the previous work [41]. Optical output from the source is introduced into a fiber-based Mach-Zehnder interferometer with a 70:30 2x2 fiber coupler, and a 50:50 2x2

fiber coupler. The 30% reference port is connected through the collimator link to the second coupler. The through port with 70% coupling ratio of the first coupler is connected to a probe arm and the detected light goes backward to meet at the second coupler. Probe arm comprises a collimator lens with a beam width of 0.8 mm ($1/e^2$ FWHM) and a Galvano scanner (Cambridge Technology 6210H). We didn't use an objective lens after the Galvano scanner to form a telecentric scan, instead we scanned the beam as coming from the collimator in order to gain a long working distance in terms of focusing depth. Lateral resolution of 0.8 mm is not important because the scanning is just to give a range and geometry of imaging area to be captured. Distance between two collimator links is adjusted so that the zero delay matches the focus area of the probe arm. Here, optical lengths in the fiber and in the free space are carefully trimmed and spliced so that ratio of those two almost matches between both paths in order to suppress the wavelength dispersion due to the interferometer itself. Probe and reference beams are recoupled by the second coupler and two outputs are fed into the balanced photoreceiver (newfocus 1544). Pig-eye was used as a sample in this study. The spectral interference signal is converted to voltage, and digitized by a 14-bit DAQ board (PDA14, Signatec.Inc.) with sampling rate of 100 Msample/sec that is mounted inside a PC unit. The start trigger for A-scan synchronization is fed from the integrated start trigger output from the source into the A/D converter. Each start trigger initiates the segmented sampling of about 40000 counts that are transported to the PC, followed by FFT calculation. In the FFT calculation, wavelength-rescaling process in k-space is performed by using non-linear calibration data taken at about 12mm position with a simple reflector. Probe power was kept below about 1.5mW in average.

OCT signals after the above rescaling process with 35 dB anti-reflection plate at different depths (single pass) are shown in the Fig. 3.22. Sensitivity is over 95 dB near zero delay. Signal drops 6 dB about at the 14 mm depth corresponding to the coherence length of 28 mm, and the signal can still be recognized even at the depth of over 30 mm

Figure 3.22. Signal power at different depth

3.6.5 Results

Figure 3.23(a) shows the tomograph image of whole eye over 5mm width. The image that we constructed has 200 A-lines in transverse direction over 5mm. So, pixel resolution is simply $25\mu\text{m}$. Contour of cornea and iris, lens surface and the retina at far side are all recognized, although the detail of each segment is slightly blurred because lateral resolution is only 0.8mm. Figure 3.23(b) shows 1-dimensional signal indicating the positions of different parts. Peaks at cornea, lens and retina are apparent and the distances of each other are 3.5 mm and 27 mm between cornea and lens, lens and retina, respectively. Assuming the refractive index of 1.35, physical length is calculated to be 20 mm as axial length.

The distant side of posterior segment experiences larger loss than anterior due to absorption and scattering through a large distance for the light to travel. This results in additional signal drop on top of that due to a finite coherence length. W Repeatability of measurement was less than $20\mu\text{m}$ that is better than the performance of commercial IOL measurement equipment (e.g. IOL Master, Zeiss Meditec)

(a) 2-dimentional image of whole eye (b) 1-dimentional signal of one A-line(red solid line in (a))

Figure 3.23 IOL measurement of the pig-eye

Results with human eye

Fig.3.24 (a) shows tomography of whole eye (volunteered healthy Asian male, with diopter -7.5) within a window of 5 mm width. Probe power was below 1mW for eye-safety. Structure of anterior segment including cornea, iris, and lens surface is clear, retina is found at a distance of 39mm which can be converted to physical size of 28.9mm using effective refractive index of 1.35. The distant side of posterior segment experiences larger loss than anterior due to absorption through a large distance for the light to travel. This results in additional signal drop on top of that due to a finite coherence length. Fig.3.24 (b) shows the image set in negative side around zero delay where the posterior segment is reversed in the depth scale so that it balances the absorption loss and the signal drop due to coherence length maintain the visibility of signal for the retina.

(a) OCT image in positive delay side

(b) OCT image in negative delay side

Figure 3.24 IOL measurement of human eye

3.7 Estimate of spectral width in multimode lasing in quasi-stationary state

When QPCT is close to the ideal case, *i.e.* perfect phase matching condition, instantaneous linewidth is expected to be as narrow as the stationary linewidth. So estimating the stationary linewidth will provide the quantitative factor of how much narrowing effect can benefit to instantaneous linewidth during the scan in QPCT. When the lasing is stationary in wavelength, the spectrum can be simply measured by optical spectral analyzer (OSA) unless the linewidth is smaller than resolution of OSA.

Theoretical estimation of spectral width in this type of multimode Gaussian envelope lasing can be derived from the analysis of mode suppression ratio that is calculated by expanding rate equations in multimode oscillation [79],

$$\frac{dN}{dt} = \frac{\eta_i I}{qV} - \frac{N}{\tau} - \sum_m v_g g_m N_{phm} \quad (3.12)$$

$$\frac{dN_{phm}}{dt} = \Gamma v_g g_m N_{phm} + \Gamma \beta_{sp} R_{spm} - \frac{N_{phm}}{\tau_{ph}} \quad (3.13)$$

where variables of N and N_{phm} , I are total carrier density, photon density on m th cavity mode, and injection current, respectively, and constants of v_g , V , Γ , τ , n_g , g_m , β_{sp} , R_{spm} are group velocity, volume of active region in the gain medium, confinement factor, cavity life time, group index, spontaneous emission factor, spontaneous electron recombination rate, respectively. In the steady state (*i.e.* $dN/dt=0$, $dN_{ph}/dt=0$), Eq. (3.12), (3.13) yield the followings; (3.14), (3.15) for photon density and injection current.

$$N_{phm} = \frac{\Gamma \beta_{sp} R_{spm}}{1/\tau_{ph} - \Gamma v_{gm} g_m} = \frac{\Gamma \beta_{sp} R_{spm} / v_{gm}}{\alpha_i + \alpha_{mm} - \Gamma g_m} \quad (3.14)$$

$$I = I_{th} + \frac{qV}{\eta_i} \sum_m v_g g_m N_{phm} \quad (3.15)$$

Here, I_{th} , α_i , α_{mm} , η_i are threshold injection current, internal optical loss of gain medium, the sum of mirror loss and filter loss at m th mode, internal efficiency, respectively.

Optical power at a cavity mode order (m) can be expressed as in Eq. (3.16).

$$P_m = F_{lm} v_{gm} \alpha_m N_{phm} h \nu V_p = \frac{\alpha_m \Gamma R'_{spm} h \nu V_p}{\alpha_i + \alpha_{mm} - \Gamma g_m} \quad (3.16)$$

Here, F_{lm} , α_m , is the ratio of power coupled out, and mirror loss that is virtually equal for all modes in this case because of narrow filter window. Mode suppression ratio (MSR) is

simply given as the ratio of output power of primary mode (m) to that of the distant n th mode (n).

$$MSR = \frac{P_n}{P_m} = \frac{g_{thn} - g_n(N)}{g_{thm} - g_m(N)} \quad (3.17)$$

If modal gain and threshold gain at each mode are given, MSR can be calculated from Eq. (3.17). Figure 3.25 shows the optical power per mode with a small gain or loss difference. In the case of multimode laser oscillation with narrow Gaussian envelope as depicted in Fig.3.25 (b), modal gain and carrier densities are localized in degenerate states so that the group of modes experiences simply a cavity loss variation enforced by the tunable filter at different mode positions. Figure 3.26 (b) shows the MSR against carrier density per mode. Since carriers are spread to a number of modes within the window, MSR is expected to be substantially lower than that in the case of single mode lasing. Alternatively, total power can be expressed as in Eq. (3.18), and actual suppression ratio can be simply the ratio of the total output power $P_{total}(\alpha_{mm}(\lambda c))$ to the output power with insertion loss equivalent to the filter transmission loss at n th mode position, $P_{total}(\alpha_{mn}(\lambda c + \delta\lambda))$.

$$P_{total} = \eta_i \frac{\alpha_m}{\alpha_i + \alpha_{mm}} \frac{h\nu}{q} (I - I_{th}) = \frac{\alpha_m}{\alpha_i + \alpha_{mm}} h\nu V \sum_m v_g g_m N_{phm} \quad (3.18)$$

$$MSR = \frac{P_{total}(+\Delta\alpha)}{P_{total}} = \frac{(\alpha_i + \alpha_{mm})(I - I_{thn})}{(\alpha_i + \alpha_{mn})(I - I_{th})} \quad (3.19)$$

where I_{th} , I_{thn} are total threshold current on all modes and threshold current when having the filter loss coinciding on n th mode position, respectively. The output power at each mode position is emulated by changing the cavity loss, for an example by adding the attenuation inside the external cavity. 1.5 dB and 3 dB neutral density attenuation plate are inserted in the optical path. Figure 3.27 shows the output power against injection current to the SOA with 0, 3, 6 dB cavity loss increase. The ratio of output power at two different losses with respect to the 0 dB loss corresponds to MSR stretched from filter's profile as shown in Fig. 3.26(a), which is about 7 dB and 11 dB in this example. This gives an approximate profile of the lasing spectrum as shown in Fig. 3.28. Calculated linewidth from the Gaussian fitted profile using the MSR is 0.075 nm, and the stationary linewidth measured with OSA was 0.052 nm which is close to the instantaneous linewidth at 2.5 kHz swept rate; 0.048 nm. Insufficient data sampling of profile and the error of inserted attenuation value as well as mismatch between actual filter profile and the ideal Gaussian envelope account for discrepancy between the calculated and the measured linewidth. However, it gives an approximate estimate of linewidth especially when the spectral linewidth is smaller than the

measurable resolution of OSA, which is a convenient way to assess the linewidth limit when using QPCT technique.

(a) Single mode lasing

(b) Multi-mode lasing

Figure 3.25 Light vs. carrier density for m -th cavity mode

(a)

(b)

Figure 3.26 Mode suppression ratio

Figure 3.27 Output power vs. Injection current with additional cavity loss

Figure 3.28 Spectrum comparison

In addition, Eq. (3.18) indicates the relation of the modal density contributing to the laser. If cavity mode density within the laser spectrum window is lower, distributed carrier density per mode becomes higher leading to higher modal gain; $g(N)$, thus resulting in higher MSR as can be seen in Fig. 3.26(b). It suggests that the cavity length should be shorter to give lower cavity density for narrowing the linewidth. Shorter limit for cavity length is restricted by the requirement of continuity of tuning, not inducing bistability of lasing or significant mode hopping. Increase of injection current will also increase the MSR up to the saturated regime, while MSR starts to decrease in the saturated regime when the gain is compressed by excessive depletion of carrier.

3.8 Summary

In this chapter, we proposed and demonstrated a high-speed wavelength swept laser with a technique to improve coherence length drastically by novel quasi-phase continuous tuning (QPCT). QPCT configured in the extended cavity tunable laser in Littrow configuration enhances the instantaneous linewidth while maintaining the continuous wavelength sweep at high swept rate. Using this technique, instantaneous linewidth of 0.048nm and 0.075nm was realized at sweep speed of 2.5kHz, 20kHz, respectively, which corresponds to the coherence length of 16mm, 10mm at 1310nm wavelength range. Linewidth narrowing factor by QPCT is about X1.7 to X2.0 with respect to the tunable filter's bandwidth depending on the swept rate. We confirmed this approach is non-resonant with respect to the swept rate unlike frequency-shifted wavelength swept source or FDML laser proposed in the previous works. Further optimization with using a shorter cavity design and smaller grating modulation constant will improve the enhancement factor. The QPCT technique will contribute not only to the large-depth imaging in OCT applications but also to the high-resolution spectroscopy or dynamic fiber sensing applications as these are gaining attentions in the near future.

Chapter 4 Wide range swept-source for high resolution OCT imaging

4.1 Objective

High-resolution imaging is important especially when optical biopsy is necessary such as for cancer diagnosis. Wide tuning range is one of demanding features of the swept source because it improves the axial resolution in the OCT imaging, thus broadening the window of applications for SS-OCT [80]. What limit the tuning range of the swept source are two factors; gain bandwidth of SOA and FSR of tunable filter in the case of extended cavity type laser source. Previous works reported a total tuning range of 133 nm with a single SOA [53] and 145 nm with a dual ring configuration with two SOAs [52]. However, such combination of Littman configuration and polygon scanner necessitates the relatively complicated optics in order to minimize the beam walk-off by polygon scanner. In this chapter, we discuss and propose wide range wavelength-swept laser source with a simple configuration using a diffractive grating and a polygon scanner based on Littrow configuration with a single SOA device. The proposed Littrow configuration with a polygon scanner and prisms realizes a simple arrangement leading to good manufacturability while maintaining solid performance. Comparison of Littrow and Littman arrangement is also discussed in terms of efficiency of tuning.

Further wide tuning is proposed by implementing spatial spectral shaping technique that controls the wavelength dependent loss inside the laser cavity so that SOA's gain is efficiently used over the wavelength range. The laser exhibits 20 mW output power with 170 nm edge to edge scan range, 160 nm FWHM at scanning rate of 20 KHz. Alternative technique we introduce is the method to modulate the injection current over wavelength range synchronizing the timing with A-trigger from the source.

In the last section, further improvement of sweep speed of 50 kHz is proposed which exceed the Finesse of the swept source that uses telescopic lens system [51,52,54].

4.2 Semiconductor Optical Amplifier

The SOA (COVEGA: BOA1132) has very wide gain bandwidth which is almost equal to the ASE bandwidth which is typically 100 nm FWHM with 28 dB gain at 1312 nm as shown in Fig. 4.1. Swept range is limited by finite gain level at the edges of bandwidth. In normal swept source operation, injection current is set to constant over time, so what limits the maximum swept range is maximum rating current and catastrophic optical damages (COD) on the gain chip facet. Further improvement of intrinsic gain bandwidth requires careful engineering of quantum well structure of the chip.

Table 4.1 Typical specification of SOA (COVEGA: BOA1132)

ITEM	SYM.	MIN.	TYP.	MAX.	UNIT
Operating current	I _{op}		600	750	mA
Center Wavelength	λ_c	1290	1300	1315	nm
Optical 3dB bandwidth	BW	70	90		nm
Saturation Output Power	P _{sat}	12	15		dBm
Small Signal Gain	G	17	20		dB
Gain Ripple(p-p)	δG		0.3	0.5	dB
Polarization Dependent Gain	PDG			1	dB
Noise Figure	NF		12	14	dB
Forward Voltage	V _f		1.4	2	V
TEC operation					
TEC current	I _{tec}		tbs	tbs	A
TEC voltage	V _{tec}		tbs	tbs	V
Thermistor Resistance	R _{th}		10K		ohm
Integrated isolator					

4.3 Discussion of tuning range: Comparison with Littman configuration

This section discusses the comparison of Littrow and Littman configuration, explaining the reason why we have selected the Littrow configuration for the combination with polygon scanner in order to achieve the wide scanning range. First, we compare the efficiency of sweep by quantifying the tuning range at a same scanning rate for Littman and Littrow configuration as shown in Fig. 4.2.

The tuning ranges for Littman and Littrow are expressed in following respectively,

$$\Delta\lambda_{Littman} = p \cdot \cos\phi \cdot \Delta\theta_m \quad (4.1)$$

$$\Delta\lambda_{Littrow} = 4a \cos\theta_0 \cdot \Delta\theta_m / M \quad (4.2)$$

Where p and a are the pitch of diffraction grating for the case of Littman and Littrow respectively, and θ_0 is the Littrow angle corresponding to the center wavelength in the range, ϕ is the diffraction angle in Littman configuration as shown in Fig.4.2, M is a magnification of the prism expander in the proposed Littrow configuration, $\Delta\theta_m$ is the deflection angle by scanning mirror.

While, the filter bandwidth for each case is given by:

$$\delta\nu_{Littman} = \frac{\lambda p \cos\phi}{2\pi\omega_i} \quad (4.3)$$

$$\delta\nu_{Littrow} = \frac{\lambda a \cos\theta_0}{\pi M \omega_i} \quad (4.4)$$

where, ϕ is the incident angle to the grating in the case of Littman configuration.

For this comparison we consider the case that both configurations have the same filter bandwidth, both have the same input beam width w_i as well as same deflection angle of scanner mirror. These condition yields the relation of $p \cos\phi/2 = a \cos\theta_0/M$. Together with this, comparing Eq. (4.3) and (4.4) derives to

$$\Delta\lambda_{Littrow} = 2 \frac{\cos\phi}{\cos\theta_0} \Delta\lambda_{Littman} \quad (4.5)$$

So, for an example, at the glazing incidence, angle ϕ for Littman is normally around 1.2rad for conventional diffraction grating. If the p is set at 1/800~1/1300 lines/mm, then $\lambda_{Littman} = p(\sin\phi + \sin\phi)$ yields $\phi=0.0.11\sim0.89$ rad. So, the term $\cos\phi/\cos\theta_0$ becomes in the range of 0.7 ~ 1.1. So the tuning range of Littrow is almost equivalent to that of Littman,

provided that the linewidth is set to equal in both cases.

Furthermore, when combined with a polygon scanner, Littman configuration suffers beam walk-off issue as depicted in Fig. 4.3 unless the complementary optics such as telescopic lens pair is implemented between a diffraction grating and polygon scanner. On the other hand, since the beam width against the polygon facet is relatively small in the case of Littrow, loss of tuning range by walk-off is practically small so that sufficient duty cycle of lasing can be achieved. Considering these facts, the proposed Littrow configuration with a polygon scanner and prisms simplifies the arrangement leading to better manufacturability, reliability while maintaining equivalent performance to the case using Littman configuration. Efficiency of geometry or design trade-off can be correlated with Finesse, the ratio of tuning range to linewidth at a certain speed, which is introduced as a comparison factor in section 4.7.

Figure 4.2 Comparison of Littman configuration (left) and Littrow configuration (right)

Figure 4.3 Beam walk-off issue: As the polygon rotates, a portion of the beam is cutoff during the transition to a next facet of polygon scanner

4.4 Spatial Spectral shaping technique

It is possible to control wavelength dependent loss in a cavity by an aperture with a tailored shape as shown in Fig.4.4. A knife-edge shaped aperture having different edge profile along the position is placed at the wavelength dispersed space in front of diffraction grating, and which attenuates the power level at different wavelength so that arbitrary shaping of output scanning spectrum is realized. Designing the shape of this aperture tailors the desired profile of spectrum.

In general with SOA, the higher the current is injected, the wider the gain bandwidth becomes. On the other hand, in the swept source configuration, the optical power level over the scanning range increases as the injection current increases. In the extremely high power emission regime, the chip facet of the SOA is damaged or degraded by so called catastrophic optical damage (COD). A primary mechanism of COD is known to be a thermally induced material meltdown due to high optical and current densities at the output facet [81-83]. Especially, AlGaAs multiple-quantum well is known to have lower COD threshold. For this reason, just increasing the current cannot realize the wider scanning range.

Figure 4.4 Spatial Spectral Shaping(SSS) Technique

By using spectral shaping technique one can overcome this limitation. COD-limited peak power can be clipped by adding attenuation at this region while increasing the gain around at edges of the scanning range. As a result, a wider and symmetric Gaussian-like shape profile can be provided. ASS-technique is possible in not only the aperture but also ND filter and thin film filter.

Figure 4.5 shows the output spectrum measured with the optical spectral analyzer in peak hold mode. Ripple that is observed around 1360 nm corresponds to the water absorption

peaks. The peak power of the source was 20 mW, tuning range was over 160nm FWHM, 170 nm in total from 1230 nm to 1400 nm as shown in Fig.4.5 and Fig.4.6, and coherence length was 3.6 mm. Fiber ring cavity for this laser consists of single mode fiber with single mode cut-off wavelength at 1260 nm. Therefore, it is notable that the normal SMF fiber excites the secondary lateral mode when the propagating light is below cut-off wavelength of 1260 nm, and which could degrade the coherence of the measurement as a result. In order to avoid this multi-mode propagation at shorter wavelength range, we spliced the output fiber with HI1060 type fiber that have cut-off wavelength at around 960 nm so that it allows only single lateral mode propagation over 1230 nm to 1400 nm at optical output. The wavelength range of subsequent OCT system including an interferometer and a probe should be matched in this aspect.

Figure 4.5 Output spectrum of the laser at 20 kHz scanning rate measured with optical spectrum analyzer in peak hold mode

Figure 4.6 Temporal intensity and wavelength profile

4.5 Wide range sweep by injection current modulation

For conventional instrument type tunable lasers, there is a function to control injection current of gain chip or LD so that the output optical power remains in constant over tuning range. This is usually called “automatic power control-APC” mode. The reason why we cannot do this and need to take a measure like we did in the previous section, i.e. spatial spectral shaping is mainly because the sweep of the source is intrinsically “free running” in a way. Polygon scanner keeps rotation at quite fast speed so there is no simple way to actively monitor the power and feedback to the power level. However, since the temporal power profile is quite repeatable over a sweep because of stable polygon scanner’s rotation, we can take advantage of this repeatability instead. If the temporal power profile is given, we can insert a temporal offset to the injection current with using calibration table over and over for each sweep. In order to realize this, we need to synchronize the timing of the offset by using the integrated start trigger signal as shown in Fig.2.1. We have designed the LD driver circuitry that can initiate temporal offset to the injection current with external trigger input. Temporal injection current profile has a nearly inverse of optical output power profile. In this way gain at band edges can be lifted by increase of the injection current, thus broadening the total swept range in terms of FWHM. This technique is effective way especially when the maximum drive operation is not only limited by COD but also limited by damage threshold of injection current at the peak or center of wavelength range. 1 μ m SOA is typical of this example.

Figure4.7 Peak-hold mode spectrum with OSA

4.6 Double pass configuration

Further improvement of coherence length can be realized by applying a double pass configuration to the same geometrical arrangement. The free space four ports circulator block as proposed in Fig. 4.8 is designed to direct the light beam twice from the collimator to the diffraction grating via polygon scanner in the exactly same incident angle. Subsequent beam in identical polarization state that exits from this block has an offset beam shift of about 2mm in perpendicular direction to the beam deflection plane (details depicted in next page). In results, the filter bandwidth is narrowed by about a factor of two; this is comparable to that of a Littman configuration of grating filter. Figure 4.9 (a) shows narrowing of the spectral bandwidth of tunable filter and Fig. 4.9 (b) shows the improvement of coherence length. The measured coherence length when using double pass configuration was found to be 7mm which is twice as large as that with single pass configuration.

Figure 4.8 Swept source in a double pass Littrow configuration with using a free space circulator block

Figure 4.9 (a) Filter bandwidth

(b) Amplitude of interference fringes vs. depth

Figure 4.10 Mechanism of single polarization recovery module

The structure of the device is explained as follows. The birefringent crystal Rutile is used as a polarization beam splitting device. A faraday rotator and zeroth order half-waveplate of quartz are bonded to the end face of birefringent crystal and a quarter waveplate with gold mirror on the incident facet in between entrance beam and exit beam. S polarized input light travels straight through the birefringent crystal, at the exit end the half waveplate rotates the polarization +45deg with respect to the propagation of the light, then the Faraday rotator rotates 45deg with respect to the direction of magnetization regardless of optical propagation. Thus, input beam via port 1 in S polarization propagates straightforward through the birefringent crystal to the end where polarization rotates +45 deg and -45 deg at Faraday rotator and half waveplate, respectively, hence remains in the initial state of polarization exiting at port2. The light is retro-reflected from the grating upon entering the splitter at port 2 the polarization is rotated -45 deg by the half waveplate and by -45 deg by Faraday rotator, a total of -90 deg change, converting the light from S to P polarization. Then P polarization, extra-ordinary ray, propagation is offset with respects to the input light path, propagating to the quarter waveplate and reflected by gold mirror where it rotates to S polarization again. The subsequent process is exactly the same as the first pass just shifted in the optical axis. The beam exits through port 3 in S polarization, and after reflected by diffraction grating for second time, it is routed back to Port 4 in the same process but only in P polarization state to the output collimator fiber. Polarization controllers within single mode fiber ring cavity are used to adjust the two orthogonal polarization states at port 1 and 4 so that a single polarization state is locked when lasing occurs.

4.7 Double reflection configuration

High-speed wavelength swept-lasers have been widely adapted to applications such as optical coherence tomography (OCT), fiber sensor system, gas sensing, and surface profiling. In a swept-source OCT (SS-OCT) system, requirements for the swept rate and swept range as well as coherence length are slightly different depending on applications. For example, in cardiovascular imaging, swept rate of over 30 kHz or higher is desired because three-dimensional volume acquisitions should be completed within several seconds during the saline flush inside the coronary. Also in ophthalmology applications, both faster and wider sweep is desired to be able to acquire high-resolution image without the influenced of motion artifact due to eye-movement.

Previous works suggest that polygon-scanner based wavelength swept sources in various configurations can provide the high-speed swept rates while maintaining practical swept range and coherence length [51-55]. The use of polygon scanner ensures the stable sweep over long term with high reliability as well as unidirectional saw tooth sweep, which is convenient for OCT system design. In this case, Finesse of the tunable filter inside the swept source determines the swept range and coherence length corresponding to instantaneous linewidth.

Design trade-off between these three parameters; wavelength range, tuning speed and coherence length, are attributed to the mechanical limitation of polygon scanner. Maximum rotation speed of industrially proven polygon scanners ranges about 40,000 to 45,000 rpm with a scanner mirror diameter of 60 mm. The swept rate is simply determined by the facet number of polygon mirror. The deflection angle is proportional to the swept range and widening the illuminating beam width onto the diffraction grating narrows filter bandwidth. So the swept range and filter bandwidth are automatically restrained by deflection angle and facet width, both of these are reduced as the facet number increases. In one polygon-based scheme, a telescopic arrangement is used to convert diverging angular dispersion from the grating into converging angular dispersion so that polygon scanner reflects back the imaged beam on the facet, the rotation of which tunes the wavelength unidirectionally [47,51]. The combination of telescope configuration and additional multiple reflections [54] has been shown to increase finesse at higher swept rates. This approach, however, may have a problem in robustness in terms of manufacturing process because of complex and bulky optics to configure telescopes in the filter. We proposed a swept source using a simplified Littrow arrangement with a prism expander that demonstrates 120 nm swept range at 20 kHz swept rate [48, 84]. However, this approach has limitation in swept range when the swept rate increases up to 50 kHz range. While, a swept source providing 91.6 nm at 50 kHz swept rate has been recently presented which doesn't incorporate telescope configuration such as

in ref.[55] and overcome the limits of Littrow arrangement. High Finesse was expected by utilizing multiple reflections from the polygon facets and grating illuminations. However, in this scheme, there is concern that the beam clipping at the polygon facets broadens the linewidth and excessive loss due to multiple diffractions at the grating reduces the swept range.

In this section, a further simplified configuration without telescopic configuration is proposed that matches the theoretical Finesse demonstrated by ref. [55] with improved performance, minimizing the effect of beam clipping. The schematic diagram of the filter configuration is shown in Fig.4.11.

The beam from the collimator is expanded to the size comparable to the size of polygon facet and deflected by the rotation of polygon scanner. Unlike in a conventional filter where the deflected beam is projected directly onto the diffraction grating, we have used instead an additional mirror placed in front of polygon scanner that returns the beam to the scanner for a second deflection. Wavelength is tuned from shorter to longer wavelength by polygon's rotation, by using a second deflection the total deflection angle of the beam is doubled compared to that of a single deflection. The position of the additional mirror is calculated to return the beam to the center of the second facet over the course of one complete wavelength scan, thereby synchronizing first and second deflections and eliminating beam clipping. In this way we have achieved a twofold increase in FSR compared to a single scanner deflection, without beam clipping at the second facet.

Figure 4.11 Synchronous double-deflection

We derive the optimum geometry for the filter by equating the beam shift;

$dx = 4 \Delta\theta L / \cos\phi \cos(\phi + 2\Delta\theta)$ and polygon facet width $d = 2\pi/N$ as a function of polygon mirror diameter D , where $\Delta\theta$ is the partial rotation angle per facet and ϕ is incident angle to the first facet and L is the distance between external mirror and the facet.

Condition yields the simple relation:

$$L = \cos\phi \cos(\phi + 2\Delta\theta) D / 8 = D / 8 \quad (4.6)$$

Here, the size of external mirror d_2 should be;

$$W + dx / 2 < d_2 < L(\tan\phi + \tan(\phi + 2m\Delta\theta)) \quad (4.7)$$

so that the beam is not clipped by the mirror's edges, where the second projection on the polygon is on the m th facet from the first facet of reflection.

In this demonstration, we use a polygon with diameter of 60 mm, L was calculated to 7 mm when the beam returns second nearest neighbor facet. The mirror is fixed at the window of the polygon scanner casing.

FSR and filter bandwidth of the tunable filter is given in Eqs. (4.8), (4.9)

$$\Delta\lambda = 8a \cos\theta_0 \cdot \Delta\theta \quad (4.8)$$

$$\delta\lambda = \frac{4\sqrt{\ln 2} \lambda a \cos\theta_0}{\pi W} \quad (4.9)$$

Finesse is given by Eq. (4.10) and theoretical value assuming no beam clipping at first facet is 1400 which is identical to that in ref. [55] when facet number is set to 70.

$$F = \frac{4\pi^2 W}{\sqrt{\ln 2} \lambda_0 N} \quad (4.10)$$

The configuration for the swept tunable laser is shown in Fig.4.12. The gain medium used was a semiconductor optical amplifier, SOA (COVEGA BOA1132) with an emission bandwidth of over 160 nm centered at 1310 nm. Polygon scanner rotation was set to 43000 rpm or 717 rev/sec to provide 50 kHz swept rate with 70 facets of polygon mirror. The collimated beam from the collimator was expanded with a series of prism expanders to the $1/e^2$ width of 1.8 mm, set smaller than a facet width of 2.7 mm to reduce the beam clipping. A second beam expander is used between the scanner and diffraction grating with 1312 line/mm constant. Although the placement of two expanders reduces the diffraction angle range at the grating, it contributes to the narrowing of the filter bandwidth by expanding the beam. Thus, overall finesse is the same. The combination of the use of expanders and selection of grating constant gives the flexibility of design to optimize scan range and linewidth as well as enabling k-linear sweep [84] eliminating the need of k-triggering in

SS-OCT system. The designed filter FSR is 124 nm with a bandwidth of 0.134 nm, corresponding to coherence length of 5.6 mm.

Figure 4.12(a) FSR vs. Scan rate

(b) Scan range, Coherence length vs. Scan rate

The fiber ring used in this study is configured using components with polarization maintaining fiber (PMF) pigtails so that polarization controllers for adjusting the polarization states with respect to the tunable filter and an SOA are eliminated. As a result, the fiber ring cavity can be spliced into a cavity as short as 1.3m in optical length. This way suppresses the minimum wavelength shift per round trip. Moreover, the use of PMF will ensure the long-term reliability of the lasing performance while SMF type might have a variation of polarization over long term and under mechanical stress that affects on the lasing spectrum.

The slow axis of the collimator lens is aligned in the polarization for the highest diffraction efficiency, i.e. S-polarization (or TM) with respect to the diffraction grating. Lasing output light is coupled to the slow axis of PMF. Here, since the amplified spontaneous emission from the SOA is also coupled into the orthogonal axis of PMF: i.e. fast axis, it creates beating noise due to cross polarization interference when it is connected to the OCT interferometer with single mode fiber which may result in the ghost image. In order to avoid this beating noise, a fiber polarizer is inserted at the output fiber to filter out the orthogonal polarization, or increasing the polarization extinction ratio over 40dB. The measured degree of polarization at output is over 99% polarized light.

Figure 4.13 Laser configuration

Figure 4.14 (a) Output power spectrum

(b) OCT signal power at different depths

Figure 4.14 shows the laser output temporal spectrum and OCT signal power at different depth positions with wavelength rescaling process. Peak power was 15 mW and the edge-to-edge swept range was 99 nm (1260 nm to 1359 nm) and the duty cycle was 78%, which agrees well with theoretical FSR of the tunable filter. Imperfect alignment of the external mirror and beam clipping by the edges of prism expanders account for difference between theoretical and measured swept range. Signal drops 6 dB at 2.7 mm corresponding to the coherence length of 5.4 mm and instantaneous linewidth of 0.139 nm, which is in good agreement with theoretical value owing to the cavity length as short as 1.8 m in optical

length. Axial resolution of about 12 μm can be confirmed from the OCT signal power spectrum at 1mm.

An actual OCT image was obtained to demonstrate the imaging capability of the source. OCT system consists of a symmetrical Mach-Zehnder interferometer with a balanced detector connected to the DAQ board with 100MS/s. An image was constructed in 250 A-lines/frame at 50,000 Hz swept rate of the proposed source. Lateral resolution set by the probe is 25 μm . Figure 4.15 shows the image of skin at the finger. Sweat ducts and epidermis, dermis are clearly observed in the image of 250 x 1048 pixels.

(a) 3D image (Image size: 250x250x1024 voxel)

(b) Sliced 2D sectioning (250x1024 pixel)

Figure 4.15 OCT image of finger

4.8 Summary

In this chapter, various approaches to overcome the design trade-off between swept range and coherence length at high speed sweep were proposed. We proved that spectral shaping and injection current modulation are effective to take full advantage of wide gain band of SOA. Both techniques can expand the bandwidth without worrying the issue of COD. Maximum of 170 nm tuning range was demonstrated with the above method.

Double pass and reflection configuration were also proposed and found that both configurations are effective to countermeasure the design limit of tunable filter using diffraction grating and polygon scanner. Double reflection configuration enabled 100nm wide range sweep at 50 kHz swept rate while maintaining practical 5mm coherence length during the sweep. 50 kHz fast sweep proved the fast acquisition of three-dimensional tomograph.

Chapter 5 OCT system demonstration and clinical studies

In this chapter, SS-OCT system is disclosed more in detail than in previous chapters, with introductory discussion of the sensitivity advantage over conventional time-domain system that was already given in previous works [18-20]. Secondly, the system configuration and signal calculation algorithm are introduced. Following sections will show some of example OCT images taken with using the fabricated swept source in the previous chapters. Some works are collaborative work with other institutions for clinical studies in oral tissues and anterior chamber. All clinical images shown here are approved by ethical committee of each institution with informed consent from volunteer patients. In the last section, novel optically modulated MEMS scanning probe is introduced for endoscopic OCT applications.

5.1 Principle of SS-OCT

Sensitivity of the Fourier-domain system

Figure 5.1 shows basic schematic of SS-OCT. The output of the swept source is divided into a reference arm and a sample arm. Backscattered and recoupled light from the sample arm creates interference with a light reflected from reference arm. Interferogram is scanned over frequency domain with the wavelength sweep over and detected with a photoreceiver. A-scan or A-line signal along the depth is obtained by fast Fourier transform (FFT) or discrete Fourier transform (DFT) on the time-sampled signals [85].

Raw signal is expressed as

$$i_{\text{det}}(t) = \frac{\eta q}{h\nu} \left(P_r + P_o \int r^2(z) dz + 2\sqrt{P_r P_o} \int r(z) \Gamma(z) \cos(2k(t)z + \phi(z)) dz \right) \quad (5.1)$$

where η , q , $h\nu$, are detector sensitivity, quantum electric charge, single photon energy, respectively, and P_r , P_o are optical power reflected from the sample, optical power illuminating the sample, respectively.

The third term represents the OCT signal and the first and second DC components. G is the coherence function of the instantaneous laser output, which is related to the linewidth and coherence length.

For a swept-laser source with a Gaussian spectral envelope, the axial resolution is given by

$$\delta z = \frac{2 \ln 2}{\pi} \frac{\lambda_0^2}{n \Delta \lambda} \quad (5.2)$$

Where λ_0 is the center wavelength, $\Delta \lambda$ is the FWHM of swept range, and n is the group refractive index of the sample. The depth range Δz is given by

$$\Delta z < \frac{\lambda_0^2}{4n\delta\lambda} \quad (5.3)$$

where $\delta\lambda$ is the sampling interval of wavelength and N_s is the number of samples within FWHM of $\Delta\lambda$. Sampling interval should be smaller than instantaneous linewidth of the source. $\delta\lambda$ is typically less than half of instantaneous linewidth to keep the usable depth range.

$$cl = \Delta L \times 2 = \frac{2 \ln 2}{\pi} \frac{c}{\delta\nu} \quad (5.4)$$

Figure 5.1 OCT interferometer

Detection sensitivity

The OCT signal current at the detector can be expressed as

$$i_s(t) = \frac{\eta q}{h\nu} 2\sqrt{P_r P_s} \cos(2k(t)z_0) \quad (5.5)$$

Where we assume the linewidth is sufficiently narrow so that $\Gamma(z)=1$ within the depth range, and $P_s=\tau_2 P_0$ denotes the optical power reflected from the sample at the photodetector. While, current for noise components can be expressed as [85]

$$\langle i_n^2(t) \rangle = \left(i_{th}^2 + 2 \frac{\eta q}{h \nu} (P_r + P_s) + \left(\frac{\eta q}{h \nu} \right)^2 RIN (P_r + P_s)^2 \right) BW \quad (5.6)$$

where each term represents thermal noise, shot noise, the relative intensity noise (RIN), respectively.

OCT signal is FFT or DFT of signal current $i(k_m)$;

$$F(z) = \sum_{m=0}^{N_s-1} i(k_m) \exp(-2j\pi d m / N_s) \quad (5.7)$$

Note that in Fourier domain, absolute square of the peak value of F_s at z_0 is proportional to the reflectivity. The fact that Parseval's theorem, $\sum F^2 = N_s \sum i^2$ holds for both signal and noise leads to the relation of SNR in comparison with SNR of time-domain ranging [23];

$$(SNR)_{FD} = \frac{|F_s(z)|^2}{\langle F_n^2 \rangle} = \frac{N_s}{2} (SNR)_{TD} = \frac{N_s}{2} \frac{\langle i_s^2(t) \rangle}{\langle i_n^2(t) \rangle} \quad (5.8)$$

N_s is the number of sampling points which should be equal or greater than a half of the number of frequency resolvable points, i.e. $2N_s = \Delta\lambda / \delta\lambda$, which is the ratio of tuning range to the linewidth. For an example, when $\Delta\lambda = 100$ nm and $\delta\lambda = 0.05$ nm, then SNR of Fourier-domain OCT is x1000 or 30 dB better than that of time-domain OCT in ideal or in a simplified estimation. In reality, duty cycle of tuning is less than 100%, and source's spectrum is in Gaussian profile and sampling range spans beyond FWHM of the source spectrum. In this case, N_s is the number of sampling points within FWHM of the source and the optical powers, P_r and P_s is the time-averaged value over on sweep cycle. For a shot noise limited system, eq. (5.6) and (5.8) with $P_s \ll P_r$ leads to

$$(SNR)_{FD} = \frac{\eta P_s}{h \nu f_A} \quad (5.9)$$

where f_A , is the A-line rate or equal to swept rate of the source. Hence, the effective detection bandwidth of the system is equal to the A-line rate instead of detector's bandwidth.

Finally, the sensitivity is expressed as

$$Sensitivity[dB] = -10 \log \left(\frac{\eta P_s}{h \nu f_A} \right) \quad (5.10)$$

In practical system, to reduce the source noise, dual balanced detection is employed [86]. The differential detection gives 3dB benefit in dynamic range as well as suppression of RIN of the source, self-interference noise [19], and fixed-pattern noise or background noise by its common-mode noise rejection property.

However, cross-beating noise such as polarization cross beating or slow ripple on the gain spectrum of SOA results in fixed ripple or noise on the signal spectrum which ends up in fixed alias noise in the OCT image. Such alias can be eliminated by unwrapping the noise component from the signal [39]. But it is desired to have smooth sweep spectrum of the source without such ripple and polarization noise in the first place so that you can avoid unnecessary post-processing.

5.2 OCT system configuration

An SS-OCT system with a swept laser source is shown in Fig.5.4. Output beam from the source is introduced into a fiber-based Mach-Zehnder interferometer with a 1:99 1x2 fiber coupler, two circulators, and a 50:50 2x2 fiber coupler. The through port with 90% coupling ratio of the first coupler is connected to a probe arm through a three-port circulator. Probe arm comprises a collimator lens and a galvano scanner, achromatic lens. The lateral resolution is determined by a diameter of collimator lens D , and focal length of achromatic objective lens f , as given in Eq. (5.11).

$$dx = \frac{4f\lambda}{\pi D} \quad (5.11)$$

The 1% port of the first coupler is connected to the reference arm which is configured symmetrically with a same circulator, collimator and achromatic lens as those used in the probe arm. Probe and reference beams are recoupled by 2x2 coupler and two outputs are fed into the balanced photoreceiver (newfocus 1544) with a band pass of 80 MHz. A variable optical attenuator and polarization controllers are used to adjust the dynamic range of interference signal and optical delay line combined with the objective lens and a reflective mirror is used to adjust the zero delay to match the focus area of the probe arm. An additional visible laser light at 633nm is introduced from the reference arm through the fiber coupler and is used for a guide maker on the sample surface in order to make it easy to aim the same target position to be measured. The spectral interference signal is converted to voltage, and digitized by a 14-bit DAQ board (PDA14, Signatec.Inc.) with sampling rate of 100 Msample/sec that is mounted inside a PC unit. The overview of the integrated SS-OCT system in the special rack is shown in Fig.5.2.

Algorithm

Algorithm of OCT program is shown in Fig. 5.5.

The start trigger or A-trigger for A-scan synchronization is fed from the integrated start trigger output from the source into the A/D converter. Each start trigger initiates the segmented sampling of 2048 counts that are transported to the PC, followed by fast Fourier transform (FFT) calculation. The start trigger is set to lag about 5% of total scan range after the actual rise of the lasing start in order to maintain stable detection. Because of this, pre-trigger mode is used in order to make use of 100% sampling over the effective scan range. Pre-trigger mode is a function in the DAQ board to recall the sampled data prior to the input start trigger using internal buffer on the board. Wavelength rescaling of interference signal is processed in the following steps. First, we obtain an interferogram with using a reflective mirror positioned at 1.5 mm depth position in order to extract the

calibration data, i.e. the time array of peaks and valleys of the interferogram. This time array is then polynomially approximated up to 4th order. By using this data set to rescale the sampled data in actual OCT measurement, the nonlinearity of the temporal frequency change can be properly compensated [41]. Signal processing is performed with using LabVIEW program. 512x1024 pixel image can be output within 0.3sec including process time for wavelength rescaling. The process time without wavelength rescaling was less than 0.1sec, however, with LabVIEW-based program (Fig.5.6) we cannot accurately estimate the duration of process time because the other programs are running simultaneously consuming PC's memory and run time. We expect the reduction of process time is significant when this is adapted to a programmed FPGA board.

The sensitivity of this SS-OCT is over 100 dB while the shot-noise limited sensitivity is 119 dB.

To realize a three-dimensional tomography such as an image shown in Fig.5.3, the galvano mirror for the B-scan is driven by a triangular wave. A B-scan trigger, which is synchronously generated for each B-scan, is used as an update clock of incremental position change in C-scan axis, and the C-scan trigger initiates spectral detection. Complete spectral signals and A-triggers for a single C-scan are stored in the on-board memory of the A/D board, then transferred to the main memory of PC after the measurement, and finally stored on a hard disk. The stored data is loaded during the post-processing process, and arranged into a three-dimensional spectral signal by using the A-scan triggers; this is followed by the application of the above-mentioned signal processing.

Figure 5.2 Rack-mounted SS-OCT system

Figure 5.3 3D OCT image of human skin (finger)

Figure 5.4 Hardware configuration

OCT program sequence per frame

OCT program sequence per 3D volume

Figure5.5 SS-OCT program sequence

Figure 5.6 Example of SS-OCT system display

Figure 5.7 Photograph of a developed handheld probe

5.3 OCT image examples

Unless otherwise noted, 1.3 μm swept source with 120 nm swept range at 20 kHz A-line rate is used for SS-OCT imaging in following examples. Coherence length of the source was 6mm.

A. Oral tissues

This work was done in collaboration with National Gerontology Center in Ohubu,-Shi, Japan. Oral tissue such as the structure of tooth can be imaged in cross-section by SS-OCT system using a compact handheld type probe as shown in Fig. 5.7. Clinical interest includes caries detection, inspection of the filling, and diagnosis of periodontal diseases as well as oral cancers [87-91]. Figure 5.8 shows an OCT image of a bottom-front tooth in comparison with image by X-ray. In OCT measurement, the beam from the probe is projected sideways with using a handheld probe shown in Fig. 5.2.7. Lateral resolution in this case was 15 μm . X-ray image doesn't give three-dimensional information and its resolution is limited to 0.1mm range. While, OCT image in cross-section reveals clear structure of tooth and gum, inside of which you can confirm the boarder of enamel and dentin junction and alveolar bone (yellow arrow). Observation of loss of alveolar bone with OCT can diagnose the periodontitis that may be able to replace the current method of physical probing to check the perio-pocket depth.

Figure 5.8 Comparison of X-ray and OCT image of the tooth

(①photo of front teeth, ②X-ray image from the side ③X ray image from the top ,OCT image (cross-section) in yellowed line in ① of the tooth#3 on the bottom)

Figure 5.9 3D image of tooth with emphasized alveolar bone

Figure 5.9 shows extracted and 3D-rendered OCT image of the tooth and gum with internal alveolar bone emphasized.

Figure 5.10 shows another example of a tooth that has resin filling near the gum. Left image shows very clear boundary of resin and enamel region and right image at tooth discovered a sign of decay inside the resin creating a bubble between the resin and dentin.

Figure 5.10. Resin filling

Figure 5.11 shows another 3-D OCT image of caries with C3 level, which means highly progressed caries, dematerializing the porous structure of the enamel and dentin. 3D image can be manipulated to show the structure in any arbitrary angle, slice, or perspectives when using a commercial 3D viewer program.

Figure 5.11 3D OCT image of caries

Figure 5.12 shows a capillary-venous malformation of the buccal mucosa; a type of oral tumor. The biopsy or pathological diagnosis of oral cancer is impossible because of danger

of spreading the cancerous cells through the mouth to internal organs when extracting the sample. Hence, the OCT makes great advantage when it is used especially for oral tissues. Figure 5.13 shows 3D image of human Labial glands, which is, best to our knowledge, observed in real 3D tomograph for the first time. Observation of Labial glands provides with interesting aspect when diagnosing so called, “dry-mouth” symptom.

Figure 5.12 OCT image of a capillary-venous malformation of the buccal mucosa

Figure 5.13 3D OCT image of human Labial glands

B. Ophthalmology

(1) Anterior chamber imaging

Figure 5.14 Diagram of Anterior chamber of human eye (courtesy of the National Eye Institute, National Institute of Health)

Figure 5.15 SS-OCT image of Anterior chamber of in-vivo human eye

Figures 5.15 (a), (b) show the tomographs of an in vivo human anterior eye segment. Several anatomic features are visible in (a) and (b) including sclera (SL), iris stroma (IS), iris pigment epithelium (IPE), crystalline lens (CL), scleral spur (SS), trabecular meshwork (TM), Schlemm's canal (SC). The corneal epithelium (CE) and corneal stroma (CS) are evident. This is the first time to visualize the full depth of anterior eye chamber, e.g., from the top of the cornea to bottom of the iris, by high-speed SS-OCT. The iridocorneal angle opening (ICA) and the curvature of the cornea are also evident in these figures. The probe power of these measurements was 2.9 mW, which is well below the ANSI safety standard.

Figure 5.16 3D OCT image of anterior segment

Figure 5.16 shows three-dimensional tomographs of in vivo human anterior eye segments. (a) and (b) show the entire structures of the human iris (IR) and cornea, respectively. The measurement ranges are 16.5 mm (horizontal), 16.5 mm (vertical), and 6.0 mm (depth). (c) reveals the iridocorneal angle opening. In (d), the iridocorneal angle opening, crystalline lens, and nucleus lentis (NL) are evident. Trabecular meshwork (TM) and zonules of Zinn (ZZ), which are difficult to recognize in a two-dimensional OCT image, can be observed. The measurement ranges of (c) and (d) are 6.1 mm (horizontal), 6.1 mm (vertical), and 6.0 mm (depth)

5.4 Optically-actuated MEMS Scanning Fiber Probe for Medical Imaging

This section presents a novel configuration of MEMS scanning endoscope that is actuated by external optical modulation. Light at a wavelength in the 1550nm range is used to modulate a scanning MEMS mirror. The instrument also provides for the delivery of a second beam at a wavelength of 1310nm, commonly used for medical diagnostic application such as optical coherence tomography (OCT). The two wavelengths are combined with using a WDM filter and launched into a single mode fiber. This novel approach provides for the operation of a scanning endoscope without the need for directly powering up and wiring to the scanning element, in this way a less hazardous apparatus for in-vivo diagnosis is possible as well as compact size. Design of the module and MEMS scanning mirror is discussed. A fabricated module is demonstrated in which a MEMS mirror scans at the resonant frequency of 300Hz generating an optical scanning angle of 8degrees.

Catheter or fiber probe for endoscopic OCT applications

The scanning fiber-optic probe/catheter technique has increased the potential of endoscopic medical applications. Optical coherence tomography is one of such applications that require the beam scanning inside the organs such as when diagnosing a gastrointestinal tract or monitoring plaque in the vascular vein [11,40,92]. Scanned line or area is visualized as a cross section or a 3-dimensional volume image for diagnosis imaging. Conventional way to scan the beam across the subject tissue area is to simply rotate or translate the fiber mechanically by manipulating from outside of the body as shown in Fig. 5.17. Either pulling the fiber out or rotating the fiber by using the rotary joint makes translation or rotation of the probe possible [40].

Figure 5.17. Rotary type scanning catheter for OCT (ref [40])

However, such technique always involves concerns on accidents of breakage of the fiber due to excessive stress during operation of the fiber probe. Amongst the various technologies that facilitate the scanning of an optical beam for use in endoscopic procedures it is MEMS-base catheters that have many advantages in size, speed, accuracy, and repeatability over the conventional scanning techniques that requires the whole fiber endoscope to be mechanically rotated or translated with respect to the tissue [93,94].

Previously, for endoscopic OCT, MEMS scanning mirrors have typically required a relatively higher voltage supply in order to operate. The high voltage supply is required to deflect the electrostatic MEMS mirror to a sufficiently large angle to cover a wide imaging area. However, high voltage electricity is hazardous for in vivo diagnosis since the wire along the fiber to the endoscope tip can be accidentally exposed to the intralipid or blood that can result in electrocution. In order to avoid this kind of accident, whole fiber probe should be perfectly sealed so that any liquid immerses inside the housing and the wire. This protection may increase the size and the radius of the fiber probe. And modulation of electricity itself may even create electro-magnetic interference (EMI) to the other medical apparatus during the operation.

Figure 5.18 Schematic diagram of optically modulated MEMS probe (Top)
and OCT system configuration with the probe (Bottom)

In this section 5.4, an endoscope with a low voltage-driven MEMS scanner and photovoltaic detector is proposed,. Modulation of the MEMS scanning mirror is achieved using an optical signal that is introduced from outside of the body, thereby negating the need for electrical supply and wiring within the endoscope and eliminating EMI. MEMS scanner is designed to operate at resonant frequency of the device so that relatively higher scanning angle can be achieved. Photovoltaic detector with relatively higher response speed is used for conversion of optical signal into drive voltage for MEMS actuation. Multiplexing with a WDM bandpass filter inside the fiber probe accommodates different wavelengths respectively for OCT signal and MEMS actuation. This application is not just limited to the use in OCT but also suitable for the endoscopic confocal microscopy, or fluorescence spectroscopy as long as the wavelength range of the interest in the measurement is different from the optical modulation signal for MEMS actuation.

Concept of optically modulated MEMS scanning endoscope

The proposed MEMS scanning endoscope consists of a GRIN lens, a WDM bandpass filter, a photovoltaic detector, and a MEMS scanning mirror packaged in a cylindrical tube as shown in Fig. 5.18. The Photovoltaic detector and MEMS mirror are mounted on a ceramic submount and connected electrically. Two light signals at different wavelength ranges are combined with a WDM combiner filter, launched into a single mode fiber and transmitted to the endoscope. The two signals are; incoming optical probe light from an OCT light source at 1310nm and the externally coupled modulation signal at 1550nm. After collimation by a GRIN lens, the modulation signal at 1550nm is reflected by a WDM bandpass filter to the photovoltaic detector and converted to voltage that activates the MEMS scanning mirror. The probe light at 1310nm passes through the WDM filter and propagates through a focusing lens to the MEMS scanning mirror.

The WDM filter in this case is a low pass edge filter type, having 1260-1360 nm passband and 1550nm reflection band with isolation of over 30dB in 20 to 45degree tilt with respect to the optical axis. After transmission through the WDM filter, a lens of 3.26mm focal length focuses the collimated 1310nm signal. The Gaussian beam size before focusing is 400 μ m, the focused spot size is 20 μ m. The working distance of the lens (3.26 mm) was chosen to focus the beam at the endoscope tube circumference (5 mm diameter) after deflection by the MEMS mirror, in this way the endoscope is optimized for inspection of samples in touch with the observation window. Here, Pyrex capillary tube is used for the package. The MEMS mirror is placed right after the lens to reflect the light to the direction perpendicular to the optical axis. When actuated the MEMS mirror scans the 1310 nm beam in the radial direction

with respect to z-axis in order to image the cross sectional of cylindrical organs such as veins or gastro intestines using the OCT technique. If the mirror is place in 45degree tilt, optical scanning angle is 1.4 times the mechanical tilt of the mirror in the case of radial scanning. In this study the target optical scanning angle was set to 10degree, so the required mechanical tilt of MEMS mirror is about 7degree. In order to efficiently convert the photovoltage to a large mechanical scanning angle, MEMS mirror is designed to operate in a resonant mode at a constant frequency corresponding to the image-acquisition rate. Appropriate selection of two lenses can be used to adjust the working distance and focal spot size for the different applications. Outside the body, external laser diode that is directly modulated by the injection current is introduced at the probe arm part of the system, so the application of this fiber probe doesn't limit the type of system as long as different wavelength bands are used for the system and modulation signal.

In the case of OCT application as shown in Fig.5.18(bottom), backscattered light from the sample carrying 1-dimensional depth information is coupled back to the same optical path going through the WDM filter and travels back via single mode fiber toward the OCT system for analysis. Analysis is sequentially carried out over the scanned area to construct the 2-dimensional radial cross section of the sample in results after an appropriate signal processing.

The module design is completely self-contained so that it is easy to make the package sealed at the tip of the fiber probe and the rest is just a single mode fiber along.

Table 5.1 MEMS target specifications

	Specification	Note
Item	1(x)x 2(y) mm	Rectangular
Scanning angle	Single x-axis	
Scanning rate	300Hz	Resonant mode
Scanning angle	10deg (optical)	7deg (mechanical)

MEMS mirror

The MEMS scanning mirror for this module is a single axis scanner and has angular vertical comb (AVC) actuators, that provide a large mechanical tilt angle for low voltage bias [95,96]. Figure 5.19 shows a photograph of the MEMS mirror. The device is made of SOI (30 μ m silicon/ 4 μ m oxide/ 400 μ m silicon substrate).

Figure 5.20 shows the process flow of the MEMS scanner. AVC is formed in a single mask process on the device layer of an SOI wafer. Initial angle offset of movable comb tooth with respect to the stationary comb tooth is realized by so called self-assembly process utilizing in-process stiction in the drying step after the wet sacrificial etching of the membrane [97] as shown in Fig.5.21.

Substrate underneath the mirror is deep dry etched from the bottom so that it liberates the angle of scanning. The stiction pad buckles down to the handle wafer after the sacrificial etching process and is held permanently in place by molecular binding force. This bucking action of the pad introduces the twist of torsion on the hinge that is connected to the stiction pad, and a initial angular offset of the mirror is set as a result. Compared to the other techniques for introducing an offset angle in vertical comb actuators, such as those involving complicated two steps of photolithography and deep dry etching process, our approach only needs one mask for definition of AVC and thus is a robust process through which high device yields are possible. In our design, initial offset angle of about 2-3deg can be created in this design when using 4 μ m BOA layer. The MEMS mirror is designed to operate in resonant mode since the probe allowing it to repeatedly scan over a large scan angle at a constant frequency corresponding to the image-acquisition rate. . The size of comb is 150 μ m long, 10 μ m wide with a gap of 3 μ m in between combs. The number of comb s on the single side of the mirror is 58. The MEMS mirror has dimensions of 1mm wide, 2mm long in the direction parallel to the optical axis. MEMS mirror actuation is driven with a bias voltage between 0 to +5V at the resonant frequency of 300Hz. For applications such as OCT a slower rate, close to video rate, is desirable for lateral scanning of sample. The shape of MEMS chip is elliptically rounded so that it fits well into the cylindrical tubes at a 45 degree tilted angle for radial scanning.

(a)

(b)

(c)

Figure 5.19 MEMS tilt mirror with angular vertical comb drive

Figure 5.20. MEMS fabrication process step with bulk SOI wafer

Figure 5.21 Mechanism of buckling of stiction pad

Photovoltaic detector

The photovoltaic detector used in this study is based on an asymmetric d-doped superlattice structure is used (Fig.5.21 (a))[98,99]. Unlike the other photodetectors, this structure makes the device operate as voltage source and has large bandwidth. The device structure was originally proposed for the direct modulation of field effect transistor (FET) by photovoltage [100].

Figure 5.22 (b) shows the responsivity of the detector. The solid line indicates the voltage generated by 1550nm light focused onto the detector with spot size of 30-40mm, the dotted line shows the voltage generated by the detector when illuminated by a collimated beam of 400μm diameter. We found that although there is a mismatch between the 400μm collimated beam and the effective area of detector, the voltage generated is sufficient for operation of the MEMS mirror. So, in our configuration, the 1550nm beam is directly projected onto the detector after collimation by the GRIN lens and reflection by the bandpass filter. A measured saturated voltage of 5V is the highest open-circuit voltage generated by a single photovoltaic detector illuminated by 1550nm light.

Figure 5.22. (a)Schematic diagram of photovoltaic cell and (b)generated voltage characteristics

Figure 5.23(a) and 5(b) show the response time of the detector at 60Hz, 1kHz modulation, respectively. Illuminating laser is directly modulated by injection current. Input power has a extinction ratio of >90% with 10mW peak power when amplitude-modulated. In the case of 1kHz modulation, the extinction ratio was set to about 6dB above the threshold level of the

Figure 5.23 Response of photovoltaic cell to the optical intensity modulation;

laser. At 60Hz, it can be seen that the generated photovoltage has square waveform almost identical to the input optical signal. As the modulation speed increases up to 1kHz, fall-off tends to slower to the bottom. Rise time is about 50 μ sec, and fall time is about 200 μ sec, which is still sufficiently fast to achieve several hundreds Hz of scan rate for MEMS scanner. Intrinsic response speed of the device in principle is determined by dielectric relaxation time that is on the order of picoseconds.

The size of the chip is about 300x500 μ m and the chip is flip-chip mounted on the ceramic submount. The light is illuminated from the bottom side.

WDM bandpass filter

WDM filter is a thin-film-base bandpass filter (edge filter type) with a wide transmission bandwidth for 1.3 μ m range. A sample of filter was fabricated with ion beam assisted deposition method. Filter is aligned at 20deg to the optical axis so that polarization dependence of input signal and the back-reflection from the photovoltaic detector surface are both negligible. Figure 5.24 (top) shows the simulated filter profile of transmission and reflection at 20 degree incident angle, respectively. 1.3 μ m OCT signal window is flat over a large passband over 1260 nm to 1360 nm and has a high isolation of over 40dB with respect to the 1.5 μ m band. There is an addition of another 40dB that comes from a combiner filter at the system outside. Total of 120dB isolation adds up at the system side between 1310 nm and 1550nm band. Thus, no significant cross talk from modulation signal concerns the OCT systems sensitivity.

As for photovoltage generation in reflected path, 1550 nm has 10dB isolation to the 1310 nm band, which is relatively small compared to transmission path isolation. However, since the OCT probe power level at 1310nm is usually lower than 1550 nm by the factor of over 3dB, total of about 13dB is enough not to be influence by probe power for MEMS operation.

Figure 5.24 (bottom) shows the measured spectrum of WDM filter. Polarization dependent loss is less than 0.1dB for both bands.

Figure 5.24 Bandpass filter characteristics: Reflection spectra (left), Transmission spectra (right)
Measured spectrum with OCT probe signal band and modulation signal band (bottom)

Results

Figure 5.25(right) shows the resonant frequency of MEMS scanning mirror measured by laser Doppler velocimeter. The first order resonance was found at frequency of around 300Hz. Figure 5.25 (left) shows the tilt angle of the MEMS mirror versus applied voltage in DC mode. By operating the mirror in resonance it is possible to amplify the DC tilt angle of about 0.45degree at 5V DC bias by a gain factor of over 15X. Next, basic operation by optical modulation was confirmed. The MEMS scanning mirror is connected to the photovoltaic detector and light at 1550nm is projected onto the detector. The 1550nm laser diode output power used to illuminate the detector is directly modulated between 0mW and 7mW by injection current modulation at 300Hz using a function generator in both TTL square waveform and sinusoidal waveform modes. Scanning angle of the MEMS mirror was measured by projecting a red visible diode laser beam at 633nm on the mirror as shown in Fig. 5.26. By tracing the length of line beam created by scanning the mirror the optical scan angle was found to be 8-10degrees.

Figure 5. 25 MEMS mirror characteristics: DC tilt angle vs. voltage (left), frequency response (right)

The fabricated MEMS scanning endoscope consists of a lens, a WDM bandpass filter, a photovoltaic detector, and a MEMS scanning mirror packaged in a cylindrical tube as shown in Fig.5.27. The Photovoltaic detector and MEMS mirror are mounted on a glass submount and connected electrically. WDM filter is aligned so that input light is divided into transmission toward MEMS mirror and reflection toward the photovoltaic detector. A focus lens with 3.26mm focal length is placed before MEMS mirror so that the probe light focuses right on the circumference of the tube. The assembled optics block on the submount is then inserted into the Pirex glass tube and sealed with the lens housing. This package is completely self-contained so it is easy to realize a hermetically sealed package for reliability and safety.

Figure 5.26 Experimental setup for validating the optical actuation

Figure 5.27 Photograph of fabricated MEMS fiber probe:

Core elements mounted on the strip of glass (Top), The device encapsulated in the Pyrex capillary tube with fiber pigtail

Figure 5.28 SS-OCT image taken with handheld probe with Galvano mirror (a) and an image measured with the fabricated MEMS mirror (b).

Discussion

We have demonstrated an optically modulated MEMS mirror based scanning endoscope. A scanning angle of $\pm 4^\circ$ at a rate of 300Hz was achieved. The scanning angle we achieved is limited by two factors of the present design. First, air-damping introduced by the MEMS mirror acts to reduce the amplitude of the scan. Making the mirror smaller can reduce the air-damping effect. The other reason is the limited electrostatic force. The force applied by the comb drive can be improved by increasing the photovoltage generated by the photovoltaic detector and by increasing the size of the comb drive. There is a room for improvement to realize 45 to 90deg angle scanning by optimization of design.

The same concept applies to the scanning in different direction other than radial. Transverse scanning is possible simply by switching the axis of the MEMS mirror. In this case, factor of X2 accounts for the relation of optical to mechanical scanning angle.

Forward scanning is also possible by implementing a number of reflections inside the module.

Further improvement

A device with improved performance was further investigated. In this prototyping, a MEMS scanning mirror (Fraunhofer Institute (Hiperscan Inc.), MSS1D) [101] with a large scanning angle was used as shown in Fig. 5.29. Typical specification is listed in Table 5.2.

Table 5.2 MEMS mirror (MSM1D) specification

Parameters	Specification	Notes
Device	1D scanner	
Mirror diameter	1.5 mm	
Resonance frequency	250 Hz	Drive frequency 500Hz
Scan range	60 deg (optical)@ 20V	At normal pressure
Reflection coefficient	88%	
Chip size	2.9 mm x 4.3mm	
Shock resistivity	>2000G	When not operating

Figure 5.29 Photograph of MEMS mirror

In order to make a full advantage of scan range of this MEMS mirror, we needed to improve the voltage generated from photovoltaic cell. In this case, we used a photovoltaic cell with new design that has been integrated in parallel with multiple cells around the effective illumination area as shown in Fig. 5.30(a). The structure of each cell is similar to that depicted in Fig.5.22. Four cells are all connected via the electrodes to simply multiply the voltage generation by a factor of four. Figure 5.30 (b) shows improved performance of photovoltaic cell with about 14 V output voltage when 10mW power was focused onto the center of photovoltaic cell with about 100 μm beam diameter. Estimated scan angle at 14V is 45 degree in optical scan with new MEMS mirror.

Instead of Pirex glass capillary plastic tube (PTF) is used as tube material for the probe. PTF of 6mm outer diameter, 5mm inner diameter is used, which is common material for conventional catheter. A diagram of improved design is shown Fig. 5.31. The device is sealed with medical grade UV epoxy (LOCTITE3021) that can pass ETO sterilization process. The rest of basic operation principle is same as initial sample except for the actuation mode of MEMS device. Since this MEMS mirror doesn't have offset angle like the previously fabricated one as seen in Fig.5.29, start of actuation is little bit tricky that you need to sweep the modulation frequency from the value higher than resonant frequency toward resonant frequency to 'trigger' the resonance of the mechanical oscillation.

Figure 5.30 Photograph of Photovoltaic cell and its voltage vs optical power (in courtesy of NEC devices)

Results

To demonstrate the optical modulation, a test similar to the previous mode was carried out, but this time with modulation signal at 1550nm actually propagated via single mode fiber and split and reflected onto the photovoltaic cell by WDM bandpass filter (Fig. 5.32). Visible light in red is projected onto the mirror to scale the scan angle of reflected beam. Measured angle was about 40deg in optical scan with modulation signal with 15mW peak optical power at 500Hz square waveform. 40deg angle corresponds to about 2mm peripheral length in arc provided that focus is just outside the outer diameter of the tube, which gives a relatively wide field of view.

Figure 5.31. Diagram of improved MEMS probe design (Top) and photograph the fabricated sample (bottom)

Figure 5.32 Experimental setup for validation of optical modulation

Figure 5.33 Future design consideration of compact MEMS probe

5.5 Summary

SS-OCT system description was disclosed and a number of clinical data were discussed such as in oral tissues and ophthalmic samples. Even though SS-OCT has advantages over conventional TD-OCT in principle, there should be more rigorous and much larger number of clinical evidence required to support or to prove its advantages [126,127] in the future. One disadvantages of SS-OCT or general OCT system is difficulty in imaging highly or multiple scattering mediums such as tissue or cancer as well as caries of teeth [128-132]. This should be countermeasured in the future by applying further longer wavelength range. In the last section, a new concept of optically modulated MEMS probe was explained. A novel endoscope incorporating a low voltage driven MEMS scanner and photovoltaic detector was proposed. Our approach, that utilizes an optically modulated MEMS scanning mirror, eliminates the need for external electrical voltage source. This concept provides an option of the endoscopes scanning operation to be controlled remotely without introducing high voltage signals and wires that can be detrimental to in-vivo diagnosis. We have demonstrated an optical scan angle of 45 degrees by utilizing 15 mW of 1550 nm light modulated at 500 Hz to actuate a MEMS mirror for 250 Hz scan. Further improvements and optimization of the design for increasing scan angle, and the development of a compact module are necessary for practice in clinical applications. In the future design, a single lens will be used and the band separation is done in focusing path of the light as shown in Fig.5.33. In this way, whole module can be more compact and sensitivity of photovoltaic detector will be further improved. This approach should benefit when OCT is used as combined or complementary modality with the other imaging systems.

Chapter 6 Feasibility study of single mode continuous fast sweep

Introduction

In the previous chapters, basic swept sources all have several millimeter or centimeter order coherence length during high-speed sweep. As explained in Chapter 1 Introduction, integrated devices such as DFB and SSG-DBR or “single mode oscillation lasers” do not meet full requirements of swept source OCT. However, integrated devices are not only advantageous when the market volume is large and cost pressure is high for its mass production capability for OCT applications but also that its single mode oscillation provides coherence length in kilometer that may open up new applications such as dynamic gas analysis, fiber sensing interrogators. So ideally, compact, low-cost, fast sweep single mode continuous source is ultimate goal of this research. In this chapter, potential outlook of three promising technology candidates is introduced and briefly overviewed.

6.1 MEMS VCSEL

Vertical Cavity Surface Emitting Laser diode (VCSEL) is intrinsically single mode oscillation because of such short cavity length between two DBR mirrors built in the vertical direction inside the wafer. Tunable VCSEL [103] has been studied over decades at various wavelength ranges that overlap with the usable range for OCT.

Figure 6.1 Conventional Tunable VCSEL [13]

Figure 6.2 Proposed MEMS hybrid tunable VCSEL

Although the output power of current injection type VCSEL with single lateral mode has relatively lower power, continuous tunability with just one control node (i.e. bias voltage to the deformable membrane) enables easy control of repetitive sweep at high-speed. However, monolithic structure where top DBR mirror is made of 3-5 semiconductor compounds has difficulty in manufacturability [104]. Recently there have been several reports on hybrid structure combined with bulk silicon MEMS membrane on top of half VCSEL [105]. However, control of gap in between top DBR of silicon MEMS and bottom half VCSEL has been difficult in terms of complex actuation mechanism of membrane. In order to make a simple and accurate control of the gap for defining FSR and tuning range, we proposed a vertically movable MEMS mirror that individually has electrostatic electrodes on the MEMS side only as shown in Fig.6.2. In this way, optical cavity and electrostatic gap are separated to have individual and not cross-interference in control. The MEMS vertical mirror is designed in similar process that was used for MEMS mirror for the fiber probe. SOI wafer was simply dry-etched from both sides and sacrificial etching was performed.

Figure 6.3 shows preliminary results of vertical translation against applied voltage measured by laser Doppler interferometer. According to the simple design theory of Fabry-Perot etalon filter, FSR and displacement of MEMS mirror are related to initial air-gap length as in Fig. 6.4.

Figure 6.3 Relation of applied voltage and displacement over time

Figure 6.4 Relation of FSR and Air-gap vs. air-gap length

Further study and results will be reported elsewhere in the near future.

6.2 Integrated ECL

Integration of ECL with the same configuration that was discussed in the previous chapters into single ship device is the most ideal approach if the proportional configuration can be realized in very small footprint. Similar idea was proposed by different group [106,107] where EO beam deflector is built into the LD substrate by using refractive index change induced by injection current. But an effective electro-optic coefficient is not as large as nonlinear optic crystals like LN or PLZT having over several tens of pm/V in coefficient. On the other hand, EO deflector made into slab waveguide by EO polymer materials [108] or by solid-phase epitaxy [109] have practically large EO coefficients over 50 to 100pm/V.

Figure 6.5 Proposal of EO deflection for tunability [106]

Figure 6.6 Monolithic EO tunable ECL [107]

Figure 6.7 Polymer EO deflector with cascaded electrode section [108]

Single mode continuous tuning

EO deflectors based on nonlinear materials can realize nanosecond order response time and a few degree of beam deflection. However, arrangement for pivot tuning cannot be realized in terms of geometry so that single mode continuous tuning is not that simple forward.

We address the novel technique to amend the design limit by taking into account that refractive index change in EO deflector does not only change the angle of deflection but also change the optical path length. We used this modulation to compensate the phase lag between wavelength change by grating filter and longitudinal mode variation by deflection

angle, which is analogous to the QPCT technique. While, this is to find completely synchronous condition taking into account with refractive index change.

Figure 6.8 (a) Pivot tuning

(b) Non-pivot tuning as in QPCT

Figure 6.9 Proposed Hybrid EO deflector ECL type swept source

IS: Isolator, LS: Lens, LD: Gain chip, SS: Spotsizer converter, DG Diffraction grating

Condition can be analytically derived by solving the simple Snell's law on the boundaries of EO deflector and its wavelength of longitudinal mode(λ_m) along the diffraction angle can be expressed in Eqs. (6.1)

$$\lambda_m = \frac{\lambda_o}{L_o} \left[L_1 + b \frac{1}{\cos(\alpha - \theta)} + \frac{L_2 + b \tan \theta_0 \tan(\alpha - \theta)}{\cos \phi - \sin \phi \tan \theta_0} \right] \quad (6.1)$$

$$\theta = \arctan \left\{ \frac{\sin^2 \alpha}{\sin \phi + \sin \alpha \cos \phi} \right\} \quad (6.2)$$

$$L_{eff} = L_1 + b \frac{1}{\cos(\alpha - \theta)} + \frac{L_2 + b \tan \theta_0 \tan(\alpha - \theta)}{\cos \phi - \sin \phi \tan \theta_0} \quad (6.3)$$

where each parameter is defined in Fig. 6.10.

Figure 6.10 Definitions of geometrical parameters

It is difficult to solve the equation analytically, however we found that with an appropriate conditions yield single mode hop free, or continuous tuning over 20nm at 1300nm range.

Here we use the criteria of $|\lambda_g - \lambda_m| < c/4L$ as a mode hop free condition.

Figure 6.11 shows the degree of phase synchronization between longitudinal mode and filter wavelength in the case of non-pivot tuning as in Fig. 6.8(b). Here, at least mode hops should occur about 10 times over a single sweep. While, if we optimize the condition, mode hop will never occur as shown in Fig. 6.12, where $\Lambda=1200\text{g/mm}$, $\alpha=30\text{deg}$, $\lambda_o=1260\text{nm}$, $L_1=8\text{mm}$, $L_2=10\text{mm}$, $b=3.86\text{mm}$.

Figure 6.11 (a) Phase change vs. diffraction angle

(b). Mode hop number

Figure 6.12 (a) Phase change vs. diffraction angle

(b) Mode hop number

6.3 Fiber ECL with saturable absorber

Last candidate is a fiber extended cavity laser with saturable absorber filter which acts as a narrow bandpass filter that automatically tracks the lasing wavelength, thus ensuring the single longitudinal mode operation [110-113] (Fig.6.13). Unpumped erbium doped fiber (EDF) can be used for this purpose at 1550nm band with additionally wavelength tunable filter. Standing wave caused by counter propagating light will self-induce the fiber Bragg grating along the EDF and the center wavelength is tracked with lasing wavelength. If cavity modes are finely packed within the wavelength range and FBG filter is narrower than cavity mode spacing, each adjacent cavity mode is consecutively oscillated, thus quasi-continuous sweep is possible with no mode partition or bistability. Gain medium used in previous work is EDFA so that sweep speed is limited by response time of EDFA resulting in submillisecond or swept rate less than kHz order. If SOA is implemented as a gain medium, much faster swept rate over several kHz can be expected by using same configuration we used for the study in previous chapters as shown in Fig.6.14.

Figure 6.13 Single longitudinal mode tunable fiber laser with a self induced saturable absorber filter ([110],[111])

Figure 6.14 Proposed configuration with saturable absorber filter

Further investigation and experiment is left for future work and will be reported in the future.

6.4 Summary

Single mode continuous tenability is of great importance in many applications. In this chapter potential technologies are discussed for future study. While hybrid tunable MEMS and hybrid ECL are the ultimate or ideal device mode for mass productivity, fiber ECL is the shorted cut to meet the immediate application needs where larger coherence length or narrower linewidth is important to overcome the measurement resolution.

Chapter 7 Conclusion

In this thesis, correlation of characteristics of swept source is investigated in different point of views. Several new techniques are proposed and validated via constructing SS-OCT images using proposed swept source.

Major outcome throughout this study are followings;

1. Littrow based diffraction grating tunable filter combined with a fiber extended cavity laser source is implemented to realize high-speed wideband swept source. Stable operation of over 120 nm swept range and 15 mW output power has been confirmed with fabrication the prototype laser. Compensation optics by using a prism expander in an optimum arrangement enables 0.05 to 0.2% k-linear sweep so that it can eliminate the wavelength rescaling process that was a heavy burden to the system process time. As a result, it opens up a possibility of real time imaging of tomography.
2. Coherence length improvement was realized by novel concept called “Quasi-Phase Continuous Tuning” technique. Theoretical analysis was established to estimate the dynamic spectral linewidth of QPCT operation assuming Quasi-stationary oscillation. Here, multimode rate equations are introduced and extended to carry out the linewidth analysis. Simple rearrangement of Littrow configuration yields coherence length improvement or spectral narrowing effect by a factor of two in comparison with the case not using QPCT. This technique was further carried out for large coherence length swept source for scaling applications such as intraocular distance measurement. Using 1 μ m swept source combined with QPCT successfully performed OCT imaging in 20mm range.
3. Various approaches to overcome the design trade offs between swept range and coherence length at high speed sweep were proposed. We proved that spectral shaping and injection current modulation are effective to take full advantage of wide gain band of SOA. Both techniques can expand the bandwidth without worrying the issue of COD. Double pass and reflection configuration were also proposed and found that both configuration is effective to countermeasure the design limit of tunable filter using

diffraction grating and polygon scanner. Maximum of 170nm tuning range was demonstrated with the above method. Double reflection configuration enabled 100nm wide range sweep at 50 kHz swept rate while maintaining 5mm coherence length during the sweep. 50 kHz fast sweep proved the fast acquisition of three-dimensional tomograph.

4. SS-OCT imaging was carried out in clinical studies in oral tissues and ophthalmic area by developing a system with interferometer and probe that proves the benefits of present inventions with the swept source. Especially, 1.3 μm range source is found out to be suitable for anterior chamber measurement.
5. Novel optically modulated MEMS probe is proposed and fabricated. It has a potential of non-hazardous diagnosis in endoscopic applications.
6. Further possibility of single longitudinal mode continuous tuning is discussed. Three major areas of potential technologies are addressed with initial feasibility study results.

This outcome of this study should accelerate the commercialization of SS-OCT in diagnostic equipment in the near future. And it should also broaden the applications such as industrial inspection system, sensing system as well as basic biological study.

APPENDIX: SS-OCT IMAGE GALLERY

In the Appendix, various image samples are shown with brief explanation. These are works done by OCT research groups all over the world who implemented the developed swept sources based on present invention described in this thesis. Examples of biological samples and industrial samples are measured with our SS-OCT system described in Chapter 5.

(1) Retinal imaging with 1 μm swept source

This work was done at Tsukuba Univ. [141]

The center wavelength is 1060 nm, the scanning range is 72 nm, maximum power of 4 mW at peak. Swept rate of the source is 28 kHz.

Figure A.1 OCT B-scan of *in vivo* human macula captured using (left) 1- μm OCT and (right) 830-nm SD-OCT.

(Image taken from ref [141])

It shows a comparison of the B-scan images of the *in vivo* human macula captured using 1 μm SS-OCT and 830-nm SD-OCT [114,115]. In this figure, each individual layer including the nerve fiber layer (NFL), ganglion cell layer (GCL), inner plexiform layer (IPL), inner nuclear layer (INL), outer plexiform layer (OPL), outer nuclear layer (ONL), the junction between the inner and outer segments of the photoreceptor (IS/OS), RPE complex, and choroids are visible.

The better penetration and higher contrast of the choroid obtained from 1 μm SS-OCT are evident from these images. A strong OCT signal is visible even beneath the choroidal vessels in the 1 μm image because of the wavelength dependent scattering property of the choroid. 1 μm range also has the smallest dispersion in water so depth-dependent axial linewidth broadening effect due to material dispersion in tissue is minimum to keep the uniform resolution along the depth.

Figure A2 (a) shows 3D volume-rendered OCT images of *in vivo* human macula in which the OCT intensity signal is displayed in an inverted-gray color map and the choroidal vessels are displayed with an orange-red color map. (b) shows en face slices of the volumetric

rendering at several different depths, where a semitransparent color map is applied to the OCT intensity volume. (c) A stereoview of the choroidal vessels of the macula.

Figure A.2 3D volume-rendered OCT images of *in vivo* human macula (taken from ref [141])

(2)Anterior segment with PS-OCT using 1.3 mm swept source (20 kHz)

Polarization-sensitive OCT [116-119] that was developed by Tsukuba Univ. group enables the imaging of birefringence of materials. Difference in birefringence of material provides additional layer information to the mere intensity profile.

Fig. A3. Intensity (upper left), phase retardation (upper right) and orientation (lower right). Images of the *in vivo* human anterior eye segment. The size of the images was 5.0 mm $\times$ 3.9 mm in air. (Image taken from ref [144])

(3) Endoscopic applications

Endoscope imaging with using a fiber catheter benefits the imaging of the sidewall of the organs [120-123]. Figure A4 shows in-vivo OCT image of rabbit trachea that has good agreement with histology image (d) (J. Su, ZP.Chen, et al. of Univ. of California, Irvine)[140,142,147]. Figure A5 shows the image of oral cancers that was measured by the group of National Taiwan University [151,152].

Fig. A4 (a) Three-dimensional rebuilt image of rabbit trachea which was 6 mm long; the trachea ring is visible in the middle part. (b) One cross-sectional slice pulled out from (a); the movie shows the whole slices set sequentially. (c) A longitudinal cross-sectional image rebuilt from 3-D image volume; the image size was 6 mm by 1.3 mm. (d) The histology image at corresponding position of (c). (e=epithelium; lp=lamina propria; c=cartilage; sm=submucosa; tm=trachealis muscle.) (Image taken from ref[140])

Fig. A5. SS-OCT scanning images of the cancerous (a) and normal oral mucosal (b) tissues indicated by the red and blue arrows, respectively, in Fig. 2. In the cancerous tissue, the boundary between EP and LP layers disappeared in the whole B-mode scan range. The red and blue vertical lines (labeled as I and II, respectively) and horizontal white arrows indicate the ranges of detailed analyses later. PP: plastic plate. (Image taken from ref[152])

(4) Biological samples

Tadpole is commonly used for developmental biology sample for investigating organogenesis [124]. OCT image reveals inner micro-structure of organs.

Figure A6. OCT image of Tadpole (*Xenopus laevis*, Albino, 2weeks, 2mm in size of head)

Observation of embryo implies the organogenesis, which is the origin of organs from single cell level.

Figure A7 OCT image of Embryo of Tadpole after 24 hrs from first cell division

Figure A8. OCT image of leaf

Figure A9 Orange skin

(5) Industrial/material inspection

A. Semiconductor devices

Silicon wafers and most of dielectric materials are transparent at infra-red range over $1\ \mu\text{m}$. As long as the device is not coated with metal (e.x. gold, or aluminum), OCT imaging is possible. One concern is that multiple reflections from the strong reflective surfaces that cause multiple aliasing on the image that make it difficult to distinguish the actual structure from the image. One way to avoid this problem is simply tilting the sample with respect to the probe beam. Additional adjustment of contrast enhances the image clarity suppressing the aliasing noise. The OCT image of the MEMS pressure sensor was captured. Slice images were taken at different planes through the micromirror device. Microscale features, such as top diaphragm, membranes, and cavity in between these, and bottom electrode are clearly visible in the stereo and the en face images. Scaling of the device is limited by minimum thickness of the layer which is $12\ \mu\text{m}$ divided by refractive index of silicon ($n=3.5$); $3.4\ \mu\text{m}$ which is minimum resolvable scale in silicon, $12\ \mu\text{m}$ in air if scaling the cavity. Scaling accuracy is repeatability of the measurement system which exceeds about submicron level. Furthermore, functional imaging such as monitoring inner stress of material can be realized by implementing phase-resolved approaches [125].

Below MEMS device is imaged with SS-OCT using $1.3\ \mu\text{m}$ source at 20 kHz swept rate.

Figure A10. Three dimensional structure of MEMS pressure sensor

(a) Stereo view, (b) Fly-through from top layer (diaphragm), (b) Membrane, (c) bottom anchor

B. Plastic material

Figure A11 Plastic molding

This is colored plastic. As long as the coloring dye is not absorptive like in black, infrared light can penetrate a certain degree to realize cross sectioning. Left picture shows the bubble trapped during the molding process. OCT can be used as inspection system or monitoring tool for process or quality control.

Figure A12 OCT image of disposable contact lens floating inside the container case

Figure A12 shows image of contact lens. This also promises the surface or contour profiling application in numerous kinds of industrial hardware.

C. Coating material

Figure A13 Paint on the metal

Although it is not possible to image the cross section the metal sample, thin, transmissive material in infra-red band like Alumite (ceramic) can be penetrated so that thickness measurement is possible. Conventional thickness measurement based on white light interferometry is limited to the smooth surface. But OCT can scale and image the boundary of these relatively rough coating to estimate the thickness or coating strength. Other functional OCT such as phase-sensitive OCT also enables accurate measurement of translation or dimension in nanometer order comparable to that of confocal microscopy.

(6) Commercialized medical applications

A. Michelson Diagnostics Incorporated.(UK)

MDL incorporate 1.3 μm source with 160nm wideband sweep range, at 10 kHz swept rate that we designed and manufacture. Application is ex-vivo, in-vivo diagnosis of cancer.

Figure A14. SS-OCT system from MDL

Figure A15. SS-OCT image of human skin with 1 μm source (top) and 1.3 μm source(bottom)
(In courtesy of MDL, <http://www.mdl-ltd.co.uk>)

B. Tomey Corporation(Japan)

- 閉塞隅角緑内障
隅角が閉塞していることがわかります。
また、3次元で画像を取得しているため、360°の隅角情報を得ることが可能です。

- LASIK眼
LASIKのフラップが描出されています。
非接触で撮影するため、術直後でも撮影可能です。

- 濾過泡
(線維柱帯切除術後)
水隙は黒く描出されます。
非接触で撮影するため、術直後でも撮影可能です。

- 円錐角膜
円錐角膜の症状である角膜前面と後面の突出が確認できます。

Figure A16. SS-OCT system for anterior chamber diagnostics (SS-1000) (In courtesy of TOMEY CO.)

(Web site catalogue from http://www.tomey.co.jp/tomey_corp/)

Table A1. Reference of the work that uses our swept-source in SS-OCT for various applications

Author(s)	Institution	Application	Swept source main characteristics	Ref.
Y.Yasuno, M.Miura, et al.	Tsukuba Univ.	Ophthalmology (Retina)	1.0um, 28kHz	139
Jianping Su, et al	UC Irvine	Endoscopy	1.3um 20kHz	140
Yoshiaki Yasuno et al	Tsukuba Univ.	Ophthalmology (Retina)	1.0um, 28kHz	141
Jianping Su, et al	UC Irvine	Endoscopy	1.3um 20kHz	142
Yoshiaki Yasuno et al	Tsukuba Univ.	Ophthalmology(Anterior segment)	1.3um 20kHz	143
Masahiro Yamanari, et al	Tsukuba Univ.	Ophthalmology (Retina)	1.0um 20kHz	144
R K Wang et al	Oregon Health Univ.	General	1.55um 20kHz	145
Masahiro Yamanari et al.	Tsukuba Univ.	Ophthalmology (Retina)	1.3um 20kHz	146
Jun Zhang, et al	UC Irvine	Endoscopy	1.3um 20kHz	147
Jon Holmes, et al	Michelson Diagnostics	Skin, Cancer	1.3um wide 160nm, 10kHz	148
Youxin Mao et al.	National Research Council, Canada	General	1.3um 20kHz	149
Kumar, K, et al	Univ. of Texas	Endoscopy	1.3um 20kHz	150
Meng-Tsan, et al	National Taiwan Univ.	Oral cancer	1.3um 20kHz	151
Meng-Tsan Tsai, et al.	National Taiwan Univ.	Oral cancer	1.3um 20kHz	152

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1. 7,489,713 Tunable fiber laser light source
2. 7,382,809 Tunable fiber laser light source
3. 7,352,783 Tunable fiber laser light source
4. 7,323,680 Optical deflection probe and optical deflection probe device
5. 7,099,358 Tunable laser light source

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