Chapter 5

Integrated-optics-based swept-source optical coherence tomography

We designed, fabricated, and characterized integrated-optics-based swept source optical coherence tomography (SS-OCT) systems in TriPleX™ technology operating in backscattering and off-axis geometries. An external 1300 nm swept-source is coupled to the chip, which contains waveguide structures for interferometric depth ranging and balanced detection. The complete OCT chip has a footprint of 0.4 x 1.8 cm². Light from the chip is focused onto the sample using an aspheric lens with a lateral resolution of 21±1 µm. OCT measurements, performed with a moveable mirror, demonstrate a sensitivity of -80 dB for integrated-optics-based SS-OCT in backscattering geometry and -79 dB in off-axis geometry. In both integrated-optics OCT systems, the maximum imaging depth is 5.09 mm. Corrected for dispersion, the measured OCT axial resolutions are in good agreement with the bandwidth limited resolution. Finally, we demonstrate cross-sectional OCT imaging of a multi-layered tissue phantom with the integrated-optics-based SS-OCT system in both geometries.

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5.1 Introduction

Optical coherence tomography (OCT) is an interferometric imaging technique which can make high resolution images up to a few millimeters deep in scattering tissue [1]. Currently, OCT has its main applications in ophthalmology and intravascular imaging. Still, the widespread use of OCT in medicine, and in other application areas such as forensics, biometrics, and process control, is hampered by its high costs and its large form-factor. Integrated optics has the potential to make OCT devices and components significantly smaller, more functional, and more cost efficient [2-5]. We designed and fabricated integrated-optics components for OCT such as elliptic couplers [3], AWG spectrometers [4, 5], and demonstrated their use in spectral domain OCT.

Compared to spectral-domain OCT, swept-source OCT (SS-OCT) has the advantage of a simpler optical design and a larger imaging depth [6]. Recently, Yurtsever et al. [7] presented an integrated optics interferometer in silicon on insulator and performed SS-OCT depth ranging. Yet, the measured OCT axial resolution was not bandwidth limited and the signal to noise ratio (SNR) was too low for imaging of turbid media.

In this chapter, we demonstrate the design, fabrication, and characterization of an integrated-optics-based SS-OCT systems operating in backscattering and off-axis geometries. For both geometries cross-sectional OCT imaging of a multi-layered tissue phantom is demonstrated.

5.2 OCT chip design and experiment methods

A chip containing optical waveguides is produced using the single strip waveguide geometry from the TriPleX™ technology platform [8, 9]. The waveguide is formed by a single layer of 50 nm of Si₃N₄, deposited on top of an 8 μm thick SiO₂ layer that is thermally grown on a 4 inch Si wafer. The strip waveguide is covered by 8 μm of thermally grown SiO₂. The width of the waveguide is chosen to be 3.4 μm such that the waveguide operates in single mode at 1300 nm wavelength and has a minimum bending loss for TE polarization. The waveguides at the end facets of the chip are tapered down to 1 μm to match the mode field diameter (9.2 μm) of a standard single mode fiber (SMF-28) to achieve optimal fiber-to-chip coupling. Waveguide splitters are made using directional couplers (DCs). A schematic of the on chip waveguide layout and the experimental set-up are shown in Fig. 5.1(a). The chip contains a Michelson interferometer, sample arm, reference arm, two identical DCs for light splitting and balanced detection, a Mach-Zehnder interferometer (MZI) made of 2 identical DCs. The optical path length of the reference arm is chosen such that the zero delay point is

78
positioned 6.9 mm from the edge of the chip. A single OCT chip has a footprint of only 0.4 x 1.8 cm².

The performance of the integrated-optics-based SS-OCT is compared to that of a home build bulk optics SS-OCT system, as shown in Fig. 5.1(b). Both integrated-optics-based and bulk SS-OCT systems use an Axsun swept source with a center wavelength of 1312 nm, 20.9 mW output power, 50 kHz repetition rate, and ~50% duty cycle. The start of the wavelength sweep is detected using light reflected from a fiber Bragg grating (FBG) at $\lambda =1266$ nm (OE Land). For both systems the interference spectrum is detected on a balanced photo detector (Thorlabs, PDB-450C). The signal from the FBG triggers a 500 MHz digitizer (Alazar Tech, ATS9350) that acquires 1088 samples using the Axsun k-clock as external clock signal. The 1088 clock cycles are equivalent to 92 nm of optical bandwidth, which corresponds to a maximum image depth of $z_{\text{max}}=5.09$ mm. OCT depth scans are generated by Fourier transformation of the interference spectrum. To correct for dispersion differences between the sample and reference arm, the spectrum is re-sampled using a second order dispersion correction in the wave-vector vs. frequency relation [10], while keeping the total optical bandwidth fixed.

The bulk SS-OCT system is based on a 90:10 fiber optics splitter with 90% of the light going to the sample arm and 10% going to the reference arm. In the sample arm, light goes through a polarization controller, is focused with a lens and impinges onto a mirror sample. The two arms of the interferometer are fully symmetric to avoid any dispersion mismatch. Via circulators, light from the sample and reference arm are mixed and split by a 50:50 splitter for balanced detection (Fig 5.1(b)).

In the integrated-optics-based SS-OCT systems, light from the swept source is coupled into the chip via a fiber array unit (FAU) based on SMF-28 fibers at 127 µm pitch. A polarization controller is placed on the input fiber of the FAU to couple TE polarized light into the chip. In the chip, light is split into sample and reference arm by the first DC. In the reference arm, light goes to the MZI (constructed from two identical DCs) and is split into two output ports directed to interfere with light backscattered from the sample arm (bar-port) and directed to interfere with light transmitted from the sample arm (cross port). Figure 5.1(a) shows the complete structure of the chip. The output waveguide is used to launch the light onto the sample and in the backscatter mode also collects the light. In off-axis mode, the light is collected by a waveguide located close (10 µm) to the output waveguide, which we call “collection waveguide”. The reference arm power directed towards the backscattered light and towards the off-axis collected light can be tuned by applying a current on the heater located on one arm of the MZI.
The focus of the aspheric lens (Geltech 355200) can be adjusted by changing the distance between the aspheric lens and the chip. In all measurements of the integrated-optics-based OCT systems, unless indicated otherwise, the focus position is set to 0.5 mm after the zero delay point. For this focus position the measured numerical aperture NA of the lens is 0.020±0.001, corresponding to a lateral resolution of 21±1 µm.

In backscattering geometry, light backscattered from the sample goes again through the first DC and is recombined with light from the reference arm in another DC at the left facet of the chip. There it is split and coupled into two fibers of the FAU that are connected to the balanced photo detector.

**a) Integrated-optics-based SS-OCT systems**

![Integrated-optics-based SS-OCT systems](image)

**Figure 5.1:** Schematic of the experimental setup used for (a) the integrated-optics-based SS-OCT and (b) the bulk SS-OCT system. SS=swept source, FBG=Fiber Bragg grating, FAU=Fiber array unit, PC=Polarization controller, BPD=Balanced photo detector, L=lens, PD=Photodiode, ZD=Zero delay, OW=Output waveguide and CW=Collection waveguide.
In off-axis geometry the back-reflected light from the sample is collected by a collection waveguide in close proximity to the output waveguide. The gap between output waveguide and collection waveguide is 10 µm to avoid power coupling between the output and collection waveguide, but the distance is sufficiently close such that the optical path length from the facet of the output waveguide to the focus is similar to the distance from the focus to the facet of the collecting waveguide, thereby maintaining the one-to-one correspondence between optical path length and depth. Light from the collection waveguide and reference arm is recombined in a second balanced detection DC and coupled into two fibers of the FAU to the balanced photo detector.

In integrated-optics-based OCT measurements in off-axis geometry on a mirror reflector, the focus position of the aspheric lens is adjusted to infinity. Due to the non co-linear optical alignment, the mirror angle is adjusted at every depth to get fully back-coupled the light from the reflector sample (e.g. a glass slide or a mirror).

5.3 Results

5.3.1 Directional coupler and MZI structure

Figure 5.2 shows spectral transmission measurements and the derived splitting ratios for the DC used in the chip design shown in Fig. 5.1. The measured spectra at the two output ports of the DC in Fig. 5.2 (a) have a different shape compared to the input spectrum (inset) due to the wavelength dependent splitting ratio, as shown in Fig. 5.2 (b).

**Figure 5.2:** (a) Measurement of the transmitted intensity of the directional coupler in the cross direction (solid lines) and in the bar direction (dashed line). The inset shows the source (input) spectrum. (b) DC splitting ratio for varying wavelength. The dashed line indicates the ideal 50/50 DC performance.
The measured splitting ratio of the DC is 80:20 at the center of the wavelength band (1312 nm) with 80% of the light going to the bar-port and 20% of the light going to the cross-port. The splitting ratio varies from 90:10 (1266 nm) to 75:25 (1358 nm) over the source spectrum. Ideally, all DCs on the chip split the light in a ratio of 50:50 independent of wavelength, however due to fabrication errors this ideal splitting ratio is not reached.

The power splitting ratio of the MZI structure can be derived from the measured splitting ratio of the two DCs that make up the MZI [11]. For a measured DC splitting ratio of 80:20 at the center of the wavelength band (1312 nm), the power splitting ratio of the MZI structure is calculated to be 36:64 (without tuning the heater). Consequently, at the center wavelength, 36% of the total reference arm intensity goes to the backscattering OCT system and 64% of the total reference arm intensity goes to the off-axis OCT system. The power splitting ratio of the MZI varies from 64:26 (1266 nm) to 25:75 (1358 nm) over the source spectrum.

5.3.2 Integrated-optics-based backscattering OCT

Figure 5.3 shows the integrated-optics-based OCT signal in depth for the backscattering geometry, measured for a moveable mirror in the sample arm. The raw data in Fig. 5.3(a) shows poor axial OCT resolution for all depths. After dispersion correction at every depth the OCT signal increases and the axial resolution improves (Fig. 5.3(b)). After correcting for the OCT signal dependence on field and correcting for the SS-OCT system sensitivity roll-off in depth, we obtain a signal decrease as shown with the solid line in Fig. 5.3(b). As can be observed, the decrease of the OCT signal in depth is well described by the combined effects of SS-OCT system sensitivity roll-off in depth and lens focusing.

**Figure 5.3:** Measured integrated-optics-based backscattering OCT signal in depth for a mirror in the sample arm; (a) without dispersion correction, (b) with dispersion correction. The solid line indicates the SS-OCT signal decrease due to the system depth sensitivity and lens focusing, the dashed line indicates the maximum imaging depth ($z_{max}$).
Chapter 5

The OCT axial resolution is determined by taking the full width at half maximum of the OCT signals in Fig. 5.3(a),(b) and is shown in Fig. 5.4. For all depths the dispersion mismatch between the waveguide material and air can be fully corrected. The measured average axial resolution of 12.7±0.5 µm is in excellent agreement with the bandwidth limited axial resolution of 12.5 µm, which is calculated from the reference arm spectrum measured on one port of the balanced detector [12].

Figure 5.4: Measured integrated-optics-based backscattering OCT axial resolution before (filled circles) and after (circles) dispersion correction. The dashed line indicates the bandwidth limited axial OCT resolution.

The integrated-optics-based backscattering OCT sensitivity is measured using a glass plate as a reflector with 4% reflectivity in the sample arm. The measured SNR at a depth of 500 µm is 66 dB. This is 3 dB worse than measured with the bulk optics OCT system (SNR=69 dB) using the same digitizer acquisition settings, sample, depth location, reference arm power, and sample arm power. From the glass reflectivity and the measured SNR we calculate the sensitivity of the integrated-optics-based SS-OCT to be -80 dB.

Figure 5.5: OCT images of the tissue phantom measured with the integrated-optics-based SS-OCT system in backscattering geometry for three different sample depth locations. The dashed line indicates the maximum imaging depth (z_max).
As a demonstration of OCT imaging using the integrated-optics-based backscattering SS-OCT system, two-dimensional images of a layered tissue phantom are obtained by scanning the sample. The tissue phantom consists of three layers of scattering medium ($\mu_s = 4 \text{ mm}^{-1}$, refractive index $n = 1.41$) interleaved with non-scattering tape. Figure 5.5 shows OCT images of the tissue phantom at three different depth locations. For each depth location the dispersion correction is optimized and the focus position adjusted (500 $\mu$m, 2500 $\mu$m, and 3500 $\mu$m, for increasing depth location). All three scattering layers can be clearly observed at the three depth locations.

5.3.3 Integrated-optics-based off-axis OCT

Figure 5.6 shows the integrated-optics-based off-axis OCT signal in depth measured for a moveable mirror in the sample arm with the focus position of the aspheric lens adjusted to infinity and the mirror angle optimized for maximum signal. Similar to the integrated-optics-based backscattering SS-OCT measurements, Figure 5.6 shows the OCT signal in depth before (Fig. 5.6(a)) and after (Fig. 5.6(b)) dispersion correction. The decrease of OCT signal in depth is due to the combined effect of SS-OCT system sensitivity roll-off in depth and the reduced coupling power into the collection waveguide caused by non-optimal optical alignment at every depth. After dispersion correction at every depth the OCT signal increases and the axial resolution improves.

![Figure 5.6: Measured integrated-optics-based off-axis OCT signal in depth for a mirror in the sample arm. (a) Without dispersion correction. (b) With dispersion correction. The dashed line indicates the maximum imaging depth ($z_{max}$).](image)

Figure 5.7 shows the measured integrated-optics-based off-axis-OCT axial resolution before (filled circles) and after (circles) dispersion correction. The measured average axial resolution of 12.9±0.5 $\mu$m is in excellent agreement with the bandwidth limited axial resolution of 12.5 $\mu$m, which is to be expected since the optical paths, and hence the dispersion, in backscattering and off-axis geometry are very similar. The integrated-optics-based off-axis OCT sensitivity is also measured using a glass plate as a
Chapter 5

reflector in the sample arm and at a depth of 500 µm with the focus of aspheric lens at infinity. The measured sensitivity of the integrated-optics-based off-axis SS-OCT is calculated to be -79 dB, similar to the sensitivity in backscattering geometry.

Figure 5.7: Measured integrated-optics-based off-axis OCT axial resolution before (filled circles) and after (circles) dispersion correction. The dashed line indicates the bandwidth limited axial OCT resolution.

As a demonstration of OCT imaging using the integrated-optics-based off-axis SS-OCT system, images of a layered tissue phantom are acquired. The OCT images are shown in Fig. 5.8 (left) with the focus position of the aspheric lens set to 0.5 mm after the zero delay point. For comparison, the image of layered tissue phantom acquired with integrated-optics backscattering SS-OCT system (extracted from Fig. 5.5) is also shown in Fig. 5.8 (right). Similar OCT image quality is observed for both OCT geometries and the three layers of the tissue phantom are clearly observable.

Figure 5.8: OCT images of the tissue phantom measured with the integrated-optics-based off-axis SS-OCT (left) and backscattering SS-OCT (right). In both measurements the focus position is set to 0.5 mm after the zero delay point.
5.4 Discussion

5.4.1 Integrated-optics-based backscattering SS-OCT

For integrated-optics-based SS-OCT in the conventional backscattering geometry, the measured -80 dB integrated-optics-based SS-OCT sensitivity is affected by the following issues:

1) A measured 1.5 dB fiber-to-chip coupling loss and a 3.5 dB chip-to-fiber coupling loss, both mainly due to FAU-to-chip misalignment and small offsets between the fiber and waveguide cores (typically 0.5 μm). In case of perfect alignment, fiber (round core) to waveguide (rectangular core) coupling loss is calculated to be 0.4 dB.

2) Unlike the bulk optics SS-OCT system, which collects most of the back-reflected light from the sample arm via a circulator. Ideally, an integrated-optics-based SS-OCT with a 80:20 DC can only collect 16% of the back-reflected light from the sample (at maximum 25% with a 50:50 DC), which leads to a 6.9 dB reduction in SNR compared to the bulk OCT system. However, this issue potentially can be solved by incorporating an integrated-optics circulator [13] into the integrated-optics SS-OCT system.

3) The 80:20 splitting ratio is not constant over the entire bandwidth of the DCs resulting in sub optimal balanced detection, which leads to a reduction in sensitivity [14]. We measured on the bulk SS-OCT system a 5 dB noise increase due to operating the digitizer at a higher voltage setting, necessary to handle the non-interferometric variation of the signal over the spectrum due to non-ideal balancing. However, we expect that these issues can be solved with the design and fabrication of a 50:50 wavelength-flattened directional coupler [15].

Considering the aforementioned loss processes we estimate that the performance of the integrated-optics-based backscattering SS-OCT system can be improved by 10-15 dB, bringing the performance of integrated-optics-based SS-OCT systems close to that of commercially available bulk SS-OCT systems.

5.4.2 Integrated-optics-based off-axis SS-OCT

Similar to integrated-optics-based backscattering SS-OCT system, the measured sensitivity of -79 dB of the integrated-optics-based off-axis SS-OCT system is reduced by the same processes as for the backscattering.

Furthermore, in the integrated-optics-based off-axis SS-OCT system the reference arm power is higher than that of the integrated-optics-based backscattering SS-OCT system. Consequently, due to the non-ideal splitting ratio of the balanced detector, the measurements have a higher digitizer voltage setting and thus more digitizer noise, as
Chapter 5

mentioned in section 5.4.1. We measured a 3 dB reduction in SNR in integrated-optics-based SS-OCT in off-axis geometry compared to the backscattering geometry. However, we expect that these issues can be solved by tuning down the power of the reference arm (i.e., by the use of a heater in the MZI structure) and/or by having an ideal (wavelength independent) 50:50 directional coupler. Considering all the aforementioned issues, we expect that the sensitivity of the integrated-optics-based off-axis SS-OCT can be improved by 15-20 dB.

Ideally, the integrated-optics-based off-axis SS-OCT can collect as much of the light as in the backscattering geometry albeit that the collection waveguide is close to the output waveguide. The off-axis geometry has the advantage that more power from the sample is collected than in backscattering geometry since the collected light doesn’t have to pass the first DC. This potentially can overcome the disadvantage of having no easily manufacturable integrated-optics circulator available in the integrated-optics-design toolbox. Nevertheless, the collection waveguide and output waveguide must be close enough for the backscattered wave-front from the sample to fully overlap with the mode field of the collection waveguide and thereby obtain a high collecting efficiency. Further investigation on the optimal gap between output waveguide and collection waveguide and performance of OCT in off-axis geometry is necessary for a better quantitative assessment of integrated-optics-based off-axis SS-OCT.

5.5 Conclusion

In conclusion, we successfully designed and characterized a 1300 nm integrated-optics-based SS-OCT system in two geometries: backscattering and off-axis. Imaging of a layered tissue phantom with both integrated-optics-based SS-OCT systems demonstrate the feasibility of integrated-optics-based SS-OCT imaging.

5.6 References


Chapter 5