Silicon photonic integrated circuit swept-source optical coherence tomography receiver with dual polarization, dual balanced, in-phase and quadrature detection

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Abstract: Optical coherence tomography (OCT) is a widely used three-dimensional (3D) optical imaging method with many biomedical and non-medical applications. Miniaturization, cost reduction, and increased functionality of OCT systems will be critical for future emerging clinical applications. We present a silicon photonic integrated circuit swept-source OCT (SS-OCT) coherent receiver with dual polarization, dual balanced, in-phase and quadrature (IQ) detection. We demonstrate multiple functional capabilities of IQ polarization resolved detection including: complex-conjugate suppressed full-range OCT, polarization diversity detection, and polarization-sensitive OCT. To our knowledge, this is the first demonstration of a silicon photonic integrated receiver for OCT. The integrated coherent receiver provides a miniaturized, low-cost solution for SS-OCT, and is also a key step towards a fully integrated high speed SS-OCT system with good performance and multi-functional capabilities. With further performance improvement and cost reduction, photonic integrated technology promises to greatly increase penetration of OCT systems in existing applications and enable new applications.

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1. Introduction

Optical coherence tomography (OCT) [1] is one of the most successful new biomedical optical imaging modalities over the past 20 years, and has become a standard diagnostic tool in ophthalmology where it is used to make clinical decisions on 10’s of millions of patients per year. OCT is also an emerging clinical imaging technology for intravascular imaging, cancer detection, as well as many other medical and non-medical applications [2]. Fourier-domain OCT (FD-OCT) has significant speed and SNR advantages over earlier time-domain OCT [3–5]. However, FD-OCT also has limitations including range dependent signal roll-off and complex-conjugate artifacts [6]. There are two complementary implementations of FD-OCT, spectral-domain OCT (SD-OCT) [7] and swept-source OCT (SS-OCT) [8]. SS-OCT uses a frequency-swept laser [9], which offers better spectral resolution than the spectrometer used in SD-OCT systems and therefore has longer imaging range and less sensitivity roll-off than SD-OCT. By detecting frequency-encoded backscattered light in time using an interferometer and a high speed, balanced photo-receiver, SS-OCT enables higher imaging speeds and has the potential for being more compact, lower cost and higher performance in
many applications. Significant effort has been made to improve the image quality and speed of SS-OCT. The recent development of high speed swept lasers such as Fourier-domain mode locked lasers (FDML) [10,11] and vertical-cavity surface-emitting lasers (VCSEL) [12, 13] achieves megahertz A-line rate acquisition and real-time video rate volumetric imaging [14], enabling many new applications. Another important area in OCT technology development is functional OCT, including OCT angiography (OCTA) [15–17], Doppler OCT [18–21] and polarization-sensitive OCT (PS-OCT) [22, 23]. OCTA and Doppler OCT provide label-free visualization and quantification of microvasculature and blood flow, respectively. PS-OCT can assess depth-resolved tissue birefringence, which can be used to differentiate tissue type and microstructural features such as collagen organization [23–25].

In addition to improving OCT imaging speed, image quality and augmenting its functional capabilities, miniaturization and cost reduction will be critical for developing new clinical applications as well as accelerating the adoption of existing ones. There are several research efforts aimed at this objective. Nguyen et al [26] developed an integrated interferometer in TriPleX technology platform for SS-OCT, but used a separate light source and a detector. Integrated optics spectrometers for SD-OCT have also been demonstrated [27, 28]. Yurtsever et al. [29] reported a photonic integrated Mach-Zehnder interferometer for SD-OCT with a separate SLED and spectrometer, and a miniaturized silicon photonic integrated interferometer for SS-OCT with an independent light source and photo detector [30]. Despite these efforts, a fully integrated high speed SS-OCT receiver or transceiver with advanced functionalities has not been demonstrated.

Silicon photonics is an important low-cost, high-volume, multi-functional platform for integrated optics because it leverages the infrastructure and ecosystem developed by the CMOS industry to reduce per unit manufacturing cost, reduce size, and offers exceptional functionality and performance by integrating many advanced optical functions with high yield. Furthermore, it has high index contrast both horizontally and vertically, enabling extremely compact photonic circuits and excellent polarization handling elements. Recently, silicon photonics integrated circuits have made major advances due to the maturation of key building blocks, such as low loss waveguides [31] and low loss fiber-to-chip coupling [32]. The development of highly efficient on-chip polarization rotators [33] and polarization beam splitters [34] heralded the implementation of polarization diversity circuits. By growing Ge on Si, on-chip detectors can be made with high responsivity [35]. Discrete component level type of performance is readily achieved and this is driving a transformation in the fiber optics telecom transceiver’s cost, size, power consumption and complexity.

Acacia Communication Inc. (Maynard, MA) has developed and commercialized silicon photonics integrated circuit (PIC) technology, that among other things, can enable a fully integrated coherent receiver that performs dual polarization, dual balanced, in-phase and quadrature (IQ) detection, which can detect the full vector field (amplitude, phase, and polarization) of the optical signal. One embodiment of this Acacia PIC technology was demonstrated in a telecommunications context in the world’s first high-speed silicon transceiver PIC operating at 100Gb/s [36]. This manuscript reports the demonstration of Acacia state-of-the-art coherent receiver technology for SS-OCT. As representative applications which show the functionality of the coherent receiver, we demonstrate full-range OCT to remove the complex-conjugate artifacts, polarization diversity to eliminate signal and polarization fading artifacts, and PS-OCT for tissue birefringence imaging. The coherent receiver is a critical OCT technology for enabling these functions in a cost effective and compact manner. Silicon photonic integrated receiver technology promises to enable wide spread availability of advanced OCT methods and accelerate research and clinical applications in multiple areas.
2. Methods and results

2.1 Silicon photonic integrated receiver

A schematic diagram of the integrated coherent receiver is illustrated in Fig. 1. The sample and reference inputs are first separated into X and Y polarization channels by a polarization beam splitter. The reference signal of each polarization channel is further split into in-phase and quadrature components and mixed with the signal input, then coupled into four separate dual-balanced photodetectors whose outputs are directly coupled to trans-impedance amplifiers (TIA). The entire system, from the fiber input up to and including the photodetectors, is contained in a single silicon photonic integrated circuit measuring 3.4mm x 2.7mm. The receiver is designed to have a center wavelength of 1550nm and a 3dB bandwidth that is more than two times the C band. The sample arm input fiber splice, fiber-to-chip coupling, on-chip polarization splitter, on-chip polarization rotator and photodetector responsivity have a combined optical loss of ~4dB. Each hybrid is designed for phase accuracy of less than 2.5 degrees away from the 90-degree target over the C band. The dual balanced receivers have a common-mode rejection ratio (CMRR) of ~20dB.

![Schematic diagram of the silicon photonic integrated coherent receiver for SS-OCT.](image)

There are multiple possible operational modes enabled by this coherent receiver. For example, it can be used for complex-conjugate suppression to realize full-range OCT, if the I and Q channels of one or both polarization state are used; or for polarization diversity detection and PS-OCT, if the X and Y channels of one or both quadrature components are used. Finally another mode would be to enable simultaneous PS-OCT and complex-conjugate suppression by using all four channels. In general, using all four channels simultaneously allows the measurement of full optical electric field vector signal. The coherent receiver has powerful performance advantages over the conventional single channel balanced receiver for OCT imaging because it offers this wide range of functionalities.
2.2 Fourier-domain mode locked (FDML) laser

We used a FDML laser with a center wavelength of 1550nm as the swept light source for the SS-OCT system. The schematic of the FDML is shown in Fig. 2 and is similar to the one reported in [37]. The laser consists of a 1550nm semiconductor amplifier (SOA, BOA 1004, Thorlabs, Inc.) as the gain medium, two optical isolators (AC Photonics), and a fiber Fabry-Perot tunable filter (FFP-TF, Lambda Quest with a Finesse of 775). A 55.56kHz sinusoidal waveform was used to drive the FFP-TF. Two lengths of 158m of dispersion compensation fibers (DCF, LLWBDK-C, OFS) were used to compensate the dispersion introduced by the SMF-28 fibers inside the FDML cavity. The backward sweep was copied and delayed by 9µs for buffered FDML operation [11]. The laser output was post-amplified by a second SOA (BOA 1004, Thorlabs, Inc.) outside the laser cavity, which was modulated to produce a unidirectional sweep at a rate of ~111.12kHz. The average laser output was ~50mW. The sweeping bandwidth was up to 120nm, although narrower sweep bandwidth was used for the measurements reported here due to the receiver bandwidth. The dispersion was carefully matched in the FDML laser in order to achieve as close to zero net dispersion as possible. However, residual dispersion mismatch caused the laser to have excess intensity noise because of the round trip time mismatch from the FFP-TF drive period as the laser wavelength was swept [37].

![Fig. 2. Schematic of the 1550nm FDML swept light source used for SS-OCT imaging with the integrated receiver. DCF: dispersion compensation fiber; ISO: isolator; FFP-TF: Fabry-Perot tunable filter; SOA: semiconductor amplifier; PC: polarization controller.](image)

2.3 Full-range OCT

The in-phase and quadrature outputs of the coherent receiver can be used to perform full-range OCT. Without loss of generality, we denote the two channels by \(I_x\) and \(Q_x\). Compared to previous methods for complex conjugate suppression, such as using a 3-by-3 coupler [38], Hilbert transform [39–42], frequency shifters [43], polarization-based optical demodulation [44, 45], bulk-optics quadrature implementation [46] or iterated algorithms based on dispersion imbalance [47], the coherent receiver requires no additional hardware change in the interferometer, is independent of polarization, requires no alignment, simplifies post-processing, and is compact.

The intensity signals acquired by the coherent receiver in the spectral domain can be represented as:

\[
I(k_m) = S(k_m) \left( |E_{ref}|^2 + |E_{sample}|^2 + |E_{ref}| |E_{sample}| \left( \cos(2\Delta x k_m + \varphi) + j \sin(2\Delta x k_m + \varphi) \right) \right) \tag{1}
\]

where \(S(k_m)\) is the power spectrum density of the light source at wavenumber \(k_m\), \(|E_{ref}|\) and \(|E_{sample}|\) are the amplitude of the reference and sample signals, respectively. \(\Delta x\) is the path-length mismatch between the reference and sample, and \(\varphi\) is a phase term. After fast Fourier
transform (FFT), and neglecting the DC and autocorrelation terms, we obtain the complex-conjugate free interference signals in the spatial domain:

\[ I(x_n) \approx S(x_n) \odot \left| E_{\text{rep}} \right| E_{\text{sample}} \delta(x_n - \Delta \tau) \]  

(2)

where \( \odot \) is the convolution operation.

Figure 3 shows the system diagram used for full-range OCT. The majority (95%) of the FDML laser output was delivered to the OCT system, and the remaining 5% output was connected to an optical spectrum analyzer or a high speed photodetector (DET08CFC, Thorlabs) to monitor the laser performance. An 80/20 coupler was used to split the light into the sample (80%) and reference paths (20%). Circulators were used in the sample and reference arms to deliver light to the sample or mirror and back-couple the returned light to the receiver. By bypassing the sample path and sharing the same reference mirror, we built a Mach–Zehnder interferometer (MZI) with the same path length as the OCT system in order to acquire a sweep calibration trace and resample the interference signals from constant time interval to constant frequency or wavenumber. The calibration signal was acquired before the imaging experiment and was used throughout the experiment. The returning paths from the sample and reference arms were directly connected to the signal and local oscillator (LO) input of the coherent receiver, respectively. A polarization controller was placed before the LO input to align the polarization state to the slow axis of the PM fiber. A polarization controller on the sample path was also used to maximize the signal input to X polarization channels.

The differential output of each channel was first converted to a single-ended output by an amplifier (MAX4444, Maxim Integrated), low pass filtered (160MHz, VLF-160 +, Mini-circuits), and sampled by a high speed 12-bit ADC digitizer (ATS9360, Alazar Technologies Inc.) at 500MHz with an internal clock, yielding an imaging range of 3.1mm limited by the low pass filter. The ADC was triggered by the laser synchronization output and 7936 samples were acquired per A-scan to acquire the original and buffered frequency sweeps. The sample arm was either a bench-top, low NA scanning microscope or a fiber-optic catheter using a distal micromotor beam scanner. The system used custom software to synchronize the scanning with the laser sweep and data acquisition. The axial resolution was measured to be 16\(\mu\)m in air. The transverse image resolution of the microscope scanner was 18\(\mu\)m. The catheter has a distal scanning micromotor for endoscopic imaging and the transverse resolution was 20\(\mu\)m. Details about the micromotor imaging probe are in reference [48].

Interference spectra from the Ix and Qx channels were first resampled in post processing using cubic spline interpolation from equal time to equal intervals in wavenumber using the calibration MZI data. After resampling, the Ix and Qx channels were combined into a complex signal \(X = Ix + jQx\) and standard OCT image processing was performed. Specifically, background subtraction was performed using the average of all spectra along the lateral direction of each B-scan. Then, each spectrum was reshaped using a Hamming window, followed by numerical dispersion compensation, multiplying the spectrum by a second order polynomial phase vector whose coefficients were optimized to minimize the point spread function of a single reflection. Finally, each spectrum was zero-padded to 4096 points and Fourier transformed to generate A-lines. The amplitude of each A-line was logarithmically compressed to display structural images, following standard OCT display conventions.
With the full-range OCT set-up (Fig. 3), the sensitivity of the coherent receiver was measured to be 94dB with 26mW power incident on the sample. The sensitivity was measured using an attenuated reflection from a mirror using the scaled signal relative to the background level. The measured sensitivity is approximately 10dB away from the shot noise limit and was mainly limited by excess intensity noise from the FDML laser as well as losses in the receiver.

Figure 4(A) shows the PSF of a mirror over a ± 3mm imaging range. Similar results would be obtained using the Qx channel. The sensitivity rolled off by 6dB at 2mm range in air. The sensitivity roll-off was determined by the instantaneous linewidth of the FDML laser. Figure 4(B) shows the PSF obtained after
suppressing the complex conjugate using the quadrature signal. The complex conjugate suppression ratio was measured to be ~28dB. The complex conjugate suppression was likely limited by residual phase variation between the I and Q channels as a function of wavelength as well as wavelength dependent variation in the internal split ratios in the receiver.

Figure 5 shows examples of full-range OCT images of human lip in vivo acquired using the micromotor catheter. Imaging was performed under a protocol approved by the institutional review board of M.I.T. With conventional single detection channel processing (Fig. 5(A)), the FFT of the acquired real spectrum generates mirror symmetric images with respect to the zero delay. Because the sample was imaged across the zero-delay position, the image exhibited the complex conjugate artifact and positive vs. negative delays cannot be distinguished. With quadrature signal processing, the FFT of the complex spectrum enabled the complex conjugate artifact to be suppressed (Fig. 5(B)). Quadrature detection allows the sample to be imaged close to zero delay for high sensitivity imaging, mitigating the possible effects of sensitivity roll-off. In addition, the effective imaging range is doubled and allows for reduced ADC digitization speeds, albeit requiring a second ADC channel.

2.4 Polarization diversity detection

The two polarization channels of either the in-phase or quadrature components (or both) can be used for polarization diversity detection. For this experiment, we used only the in-phase components. Polarization diversity detection is especially important for catheter-based OCT such as cardiovascular imaging and endoscopic imaging, where polarization artifacts are generated by the rotating optical scanner in the probe which causes the polarization state to vary as a function of the angular direction of the beam. Characterization of tissue typically relies on the OCT signal intensity, which can vary depending on the polarization state of the incident light. We demonstrated polarization diversity detection in OCT imaging of swine coronary artery ex vivo using the micromotor catheter probe.

A fresh normal swine heart was obtained from an external vendor and the left main coronary artery was dissected and stored in Dulbecco's Modified Eagle's Media (DMEM, Cellgro, Corning Inc.) at 4 degrees C prior to the experiment. Imaging was performed with the micromotor imaging catheter scanning at a rotation rate of 150Hz and a pullback speed of 1mm/sec. Figure 6 shows polarization diversity OCT imaging of a swine coronary artery ex vivo. Cross-sectional OCT images show the intima, media and adventitia layers characteristic of the normal healthy coronary artery (Fig. 6(C)). Polarization artifacts are present in the individual polarization channels (Figs. 6(A)-6(B), (D-E)) and manifest as signal poor regions which occur at angular positions where the rotary beam scanning causes the polarization of the light from the sample path to become orthogonal to the light from the reference path. The polarization fading effects are especially evident in the en face image (Figs. 6(D) and 6(E)) extracted from the depth indicated in Fig. 6(A), and are a potential confounding artifact for clinical applications. By compounding the intensity of the two orthogonal polarization channels, using the root sum of squares, polarization diversity detection can be achieved.
where the polarization artifacts are effectively removed and the sensitivity is also enhanced (Figs. 6(C) and 6(F)) since both polarizations are detected.

![Fig. 6. Polarization diversity catheter-based OCT imaging of a normal swine left main coronary artery ex vivo. Cross-sectional intensity images from individual polarization detection channels (A-B) and after combining the two channels (C). (D-F) Corresponding en face OCT images from a volumetric data set. Polarization artifacts can be seen in each polarization detection channel but were effectively removed by compounding the two orthogonal polarization channels. SB: side branch. The cross-sectional images consist of 738 A-lines per B-scan and their longitudinal position is indicated by the dotted line in (D). The en face view images are generated from a volumetric data sets consisting of 1000 B-scans (1000 x 738 A-scans) by extracting the pixel values at a depth indicated by the arrow head in (A).](image)

2.5 PS-OCT

PS-OCT uses depth-encoded polarization multiplexing of two incident polarization states in order to remove the effects from varying birefringence in the optical fiber before the sample, combined with polarization resolved detection to detect phase retardance [49, 50]. Only a minor modification of the OCT system in Fig. 3 is required in order to perform PS-OCT. Specifically, an 11m PM fiber was added to the sample arm to encode two orthogonal polarization states of illuminating light into different delays (separated by 1.5mm in air). An 11m SM fiber was used in the reference arm to match the sample path length. The two polarization channels of the in-phase components (denoted by Ix and Iy) from the coherent receiver were acquired simultaneously by the ADC card. All other components of the system in Fig. 3 were unchanged. Since only two ADC channels were available, it was not possible to detect the quadrature components. Therefore the PS-OCT system had a sensitivity penalty of 3dB, although this sensitivity could be recovered if a four channel ADC were used.

For PS-OCT post-processing, in addition to standard OCT signal processing, Jones matrix analysis was employed to extract tissue birefringence [50, 51]. Briefly, the Jones matrix of each pixel in the image was multiplied by the inverse of the tissue surface Jones matrix to cancel the unknown illumination Jones matrix, generating a matrix whose eigenvalues are the same as that of the tissue Jones matrix [51]. The double-pass phase retardation was extracted as the absolute phase difference between the two eigenvalues. A Jones averaging method which did not require phase stability was used to reduce speckle [50].

PS-OCT was demonstrated using human coronary artery imaging ex vivo. The coronary artery specimen was collected from a cadaveric heart and was fixed by 10% neutral buffered formalin before imaging. The protocol was approved by the institutional review board of
M.I.T. Figures 7(A) and 7(B) show intensity and phase retardation images of a human left anterior descending artery sample acquired by the bench-top scanner. Intra-frame (3-by-5 kernel) and inter-frame (n = 3) Jones averaging was adopted for the phase retardation image, but not for the intensity image. A fibrous plaque is present and shows high retardance. The type of plaque is evident based on previous studies comparing OCT images with plaque histology [52]. The strong birefringence signal may be related to high collagen or smooth muscle cell content within the fibrous plaque [25]. PS-OCT is potentially useful for assessing the vulnerability of atherosclerotic plaques [25].

![Intensity image](image1)

![Phase retardation image](image2)

Fig. 7. PS-OCT imaging of a human left anterior descending artery specimen ex vivo. (A) Intensity image. (B) Phase retardation image. FP: fibrous plaque. The images have 1000 A-scans per B-scan.

3. Discussion

There are several limitations in this study which should be discussed. First, there was residual dispersion in the 1550nm wavelength FDML light source and this created excess high frequency intensity noise. The dispersion caused the circulating laser field in the FDML to become desynchronized with the transmission peak of the swept fiber Fabry Perot on successive round trips of the cavity, resulting in high frequency, large amplitude fluctuations in the output. In addition, the FDML sweep repetition rate was limited and therefore OCT beat frequencies were low compared to the available receiver bandwidth. This meant that the excess noise on the FDML was detected by the receiver. Although the receiver used dual balanced detection, small mismatches in the 3dB split ratio or wavelength dependence during the sweep meant that the excess intensity noise was not fully cancelled. Although we used a low pass filter after the TIAs to reduce the excess noise, the system sensitivity was ~10dB worse than the shot noise limit, with ~4dB attributable to the receiver excess loss. Using an improved swept light source with lower excess noise should improve the sensitivity by ~6dB.

We expect that the receiver design can be improved to have a reduced excess loss of ~2dB, cover greater bandwidth (>200nm) and have better CMRR (>25dB).

In addition, due to the ADC limitations, we could only acquire two signal channels from the coherent receiver at one time. This compromised functionality as well as sensitivity because the input OCT signal was split into IQ as well as polarization channels, and it was not possible to detect all channels simultaneously. However, it is straightforward in the future to utilize a 4-channel ADC acquisition system. This will make it possible to perform simultaneous full-range and PS-OCT imaging, and to essentially measure the full vector electric field of OCT optical signals.

Finally, the study was performed on 1550nm wavelengths, while most OCT applications are at 1300nm wavelengths. It is straightforward to scale the design at 1550nm to the 1300nm
wavelength range. This will make the integrated receiver technology compatible with existing OCT for biomedical imaging applications in scattering tissues. Extending silicon coherent receiver technology to the 1050nm wavelength range used for ophthalmology is a bit more challenging, but fabrication methods which use SiN or other materials should make this achievable.

To our knowledge, this is the first study to demonstrate a silicon photonic integrated coherent receiver for SS-OCT. The integrated coherent receiver provides increased functionality in a compact and robust form, without the need for large numbers of fiber optic components, complex alignment protocols or polarization maintaining components. The in-phase and quadrature (IQ) heterodyne detection functionality of the coherent receiver provides good complex conjugate suppression ratio (~28dB), and a compact and robust way to realize full-range OCT. Previous methods for complex conjugate suppression in OCT imaging have been relatively complicated or indirect. These methods required non-standard hardware for the interferometer compared to conventional OCT [38, 41–43, 46], precise polarization adjustment [44, 45] or bulk optics alignment [46], or involved sophisticated post-processing [47]. The coherent IQ receiver does not require custom interferometer design, has simple post-processing algorithms and will be the simplest solution for full-range OCT in the future. Although the current suppression ratio may not be high enough to suppress bright reflections, further improvement is possible with improved device design as well as by computationally compensating the phase errors between the quadrature signals [44, 45].

The coherent receiver also simplifies polarization diversity detection and PS-OCT imaging. Most previous PS-OCT systems employed bulk-optic components for polarization diversity detection, which are susceptible to misalignment. We previously demonstrated a single mode (SM) fiber based polarization diversity detection scheme [50], but this set-up required pre-calibration to remove the polarization ambiguity from the SM fibers. Lee et al [53] described a PM coupler based polarization diversity detection method. However, it required careful splicing of the PM coupler to the polarization beam splitter in order to align the PM fiber axes, and potential misalignment could generate polarization beating artifacts in the image. The integrated coherent receiver eliminates bulk optic components as well as the polarization uncertainty of the SM fibers, and provides a compact solution to polarization diversity detection and PS-OCT imaging.

With direct access to the instantaneous phase of the interference signal, the coherent receiver promises to facilitate novel applications. The phase information may be potentially useful for techniques such as interferometric synthetic aperture microscopy (ISAM) [54], spectroscopic OCT [55], image artifact removal by enabling access to both I and Q signals, image enhancement by coherently averaging multiple spectra, etc. The ability to measure polarization as well as phase means the coherent receivers essentially measure the vector state of the OCT signal field.

In addition, OCT imaging can potentially benefit from the high electrical bandwidth that is possible due to the highly compact integrated nature of the device, photodetectors and TIAs. Coherent receivers used in fiber optic telecommunications can achieve electrical bandwidths well in excess of 20GHz and ADC speeds in excess of 60 GS/s [36] could be used in combination with this receiver in OCT applications. With a high speed ADC and a long coherence laser such as VCSEL [12], such bandwidth may allow extremely long range imaging (centimeter range). The high speed coherent detection also makes it possible to better compensate for various signal distortion effects such as chromatic dispersion and polarization mode dispersion (PMD) by using digital signal processing techniques.

SS-OCT has speed and performance advantages over SD-OCT and can enable ultrafast imaging (100s kHz – MHz) with excellent image quality. As this study shows, SS-OCT enables multifunctional OCT which full range, polarization diversity detection and PS-OCT. The ability to detect the electric field vector of the OCT signals promises to enable many new and novel applications. An integrated coherent receiver is a critical technology for realizing the full capabilities of SS-OCT. Because it enables a compact and simple implementation, it
can facilitate a wide range of future studies which will use full range, polarization sensitive or vector based detection of OCT signals.

Perhaps the most important advantage of silicon integrated photonics technology is that it can dramatically reduce the size and cost of OCT. Integrated photonics is having a revolutionary impact in telecom and datacom. Integration makes the cost of adding additional optical functions very low, since most of the cost in traditional approaches is due to packaging and testing and the cost scales with interfaces and packages. However in integrated photonics, almost all the interfaces are lithographically defined and placed in one package, thereby dramatically reducing cost. For this reason, integrated photonics is a key technology powering next generation telecom and datacom transceivers [56, 57]. It is widely accepted that the only way to continue to reduce the costs in the telecom and datacom markets is through optical integration. We predict that forces which are driving the use of photonics integrated circuits to achieve cost reductions, size reductions, and offering increased optical performance in the telecom and datacom fiber optic communication markets will also have a significant impact on OCT, biomedical optics, sensors for chemical analysis, life sciences and many other markets. In the not too distant future, with advances in silicon integrated photonic technology, it will be possible to fully integrate an entire high-performance OCT system, including a tunable laser, k-clock, optical delay, isolators, and eventually even a phased array for beam steering on a single chip.

4. Conclusion

We demonstrated a miniaturized, low-cost, monolithic and fully packaged silicon photonic integrated SS-OCT coherent receiver with dual polarization, dual balanced, in-phase and quadrature detection with applications for full-range OCT, polarization diversity detection and PS-OCT. The coherent receiver enables measurement of the full vector optical electric field, and is a key component for a fully-integrated SS-OCT system with multi-functional capabilities. Many important clinical as well as non-medical applications have been limited by the cost, size and availability of OCT technology. Cost reduction, miniaturization and increased OCT system functionality, promise to have a revolutionary impact which will accelerate the pace of clinical research, enabling advances in multiple clinical specialties and improving patient care.

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