Complete complex conjugate resolved heterodyne swept-source optical coherence tomography using a dispersive optical delay line

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Abstract: Swept-source optical coherence tomography (SSOCT) provides a substantial sensitivity advantage over its time-domain counterpart, but suffers from a reduced imaging depth range due to sensitivity falloff and complex conjugate ambiguity. Heterodyne complex conjugate-resolved SSOCT (HCCR-SSOCT) has been previously demonstrated as a technique to completely resolve the complex conjugate ambiguity, effectively doubling the falloff limited imaging depth, without the reduction in imaging speed associated with other CCR techniques. However, previous implementations of this technique have employed expensive and lossy optical modulators to provide the required differential phase modulation. In this paper, we demonstrate the use of a dispersive optical delay line (D-ODL) as the reference arm of an OCT system to realize HCCR-SSOCT. This technique maintains the existing advantages of HCCR-SSOCT in that it completely resolves the complex conjugate artifact and does not reduce imaging speed, while conferring the additional advantages of being low cost, maintaining system sensitivity and resolution, not requiring any additional signal processing, and working at all wavelengths and imaging speeds. The D-ODL also allows for hardware correction of unbalanced dispersion in the reference and sample arm, adding further flexibility to system design. We demonstrate the technique using an SSOCT system operating at 100kHz with a central wavelength of 1040nm. Falloff measurements performed using a standard OCT configuration and the proposed D-ODL demonstrate a doubling of the effective imaging range with no sensitivity or resolution penalty. Feasibility of the technique for in vivo imaging was demonstrated by imaging the ocular anterior segments of healthy human volunteers.

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OCIS codes: (170.4500) Optical coherence tomography; (230.4110) Modulators.

References and links

1. D. Huang, E. A. Swanson, C. P. Lin, J. S. Schuman, W. G. Stinson, W. Chang, M. R. Hee, T. Flotte, K. Gregory, C. A. Puliafito, and J. G. Fujimoto, “Optical coherence tomography,” Science 254(5035), 1178–1181 (1991).
2. M. R. Hee, J. A. Izatt, E. A. Swanson, D. Huang, J. S. Schuman, C. P. Lin, C. A. Puliafito, and J. G. Fujimoto, “Optical coherence tomography of the human retina,” Arch. Ophthalmol. 113(3), 325–332 (1995).
3. J. G. Fujimoto, S. A. Boppart, G. J. Tearney, B. E. Bouna, C. Pitris, and M. E. Brezinski, “High resolution in vivo intra-arterial imaging with optical coherence tomography,” Heart 82(2), 128–133 (1999).
4. M. A. Choma, M. V. Sarunic, C. H. Yang, and J. A. Izatt, “Sensitivity advantage of swept source and Fourier domain optical coherence tomography,” Opt. Express 11(18), 2183–2189 (2003).
5. R. Leitgeb, C. K. Hitzenberger, and A. F. Fercher, “Performance of Fourier domain vs. time domain optical coherence tomography,” Opt. Express 11(8), 889–894 (2003).
6. J. F. de Boer, B. Cense, B. H. Park, M. C. Pierce, G. J. Tearney, and B. E. Bouma, “Improved signal-to-noise ratio in spectral-domain compared with time-domain optical coherence tomography,” Opt. Lett. 28(21), 2067–2069 (2003).
7. M. Wojtkowski, A. Kowalczyk, R. Leitgeb, and A. F. Fercher, “Full range complex spectral optical coherence tomography technique in eye imaging,” Opt. Lett. 27(16), 1415–1417 (2002).
1. Introduction

Optical coherence tomography (OCT) [1] is a non-invasive imaging modality that provides micrometer scale resolution of tissue structures over depth ranges of a few millimeters. The technique has found a number of biomedical applications, most notably in ophthalmic [2] and cardiovascular [3] imaging. Fourier-domain OCT (FDOCT) is an improvement to OCT that provides a dramatic sensitivity advantage over traditional time domain techniques [4–6]. In FDOCT, the reference arm is held stationary and a spectrally resolved interferometric signal is acquired as function of wavenumber. The sample’s depth can then be retrieved from the acquired FDOCT, the reference arm is held stationary and a spectrally resolved interferometric signal is provided a dramatic sensitivity advantage over traditional time domain techniques [4–6].

**References**

1. G. Tearney, B. Bouma, and J. Fujimoto, “High-speed phase- and group-delay scanning with a grating-based phase control delay line,” Opt. Lett. 22(23), 2632–2634 (2008).
2. B. Hofer, B. Povazay, B. Hermann, A. Unterhuber, G. Matz, and W. Drexler, “Dispersion encoded full range Fourier-domain optical coherence tomography using an electro-optic phase modulator,” Opt. Lett. 32(23), 3453–3455 (2007).
3. A. Rollins, S. Yazdanfar, M. Kulkarni, R. Ung-Arunyawee, and J. Izatt, “In vivo full range Fourier domain optical coherence tomography,” Appl. Phys. Lett. 90(5), 054103 (2007).
4. Y. K. Tao, M. Zhao, and J. A. Izatt, “High-speed complex conjugate resolved retinal spectral domain optical coherence tomography using a fiber optic switch,” Opt. Lett. 32(20), 2918–2920 (2007).
5. H. Wang, Y. Pan, and A. M. Rollins, “Extending the effective imaging range of Fourier-domain optical coherence tomography using sinusoidal phase modulation,” Opt. Lett. 32(20), 2918–2920 (2007).
6. B. Hofer, B. Povazay, B. Hermann, A. Unterhuber, G. Matz, and W. Drexler, “Dispersion encoded full range Fourier-domain optical coherence tomography,” Opt. Lett. 22(23), 2632–2634 (2008).
7. J. A. Izatt and M. A. Choma, “Theory of optical coherence tomography,” in Optical Coherence Tomography: Technology and Applications, W. Drexler and J. Fujimoto, eds. (Springer, New York, 2008), pp. 47–72.
8. T. Klein, W. Wieser, C. M. Eigenwillig, B. R. Biedermann, and J. A. Huber, “Performance of reduced bit-depth acquisition for optical frequency domain imaging,” Opt. Express 19(4), 3044–3062 (2011).
9. B. J. Vakoc, S. H. Yun, G. J. Tearney, and B. E. Bouma, “Elimination of depth degeneracy in optical frequency-domain imaging with frequency shifting,” Opt. Express 12(20), 4822–4828 (2004).
10. R. K. K. Wang, “In vivo full range Fourier domain optical coherence tomography,” Appl. Phys. Lett. 90(5), 054103 (2007).
11. J. Zhang, J. S. Nelson, and Z. Chen, “Removal of a mirror image and enhancement of the signal-to-noise ratio in Fourier-domain optical coherence tomography using an electro-optic phase modulator,” Opt. Lett. 30(2), 147–149 (2005).
12. A. Bachmann, R. Leitgeb, and T. Lasser, “Heterodyne Fourier domain optical coherence tomography for full range probing with high axial resolution,” Opt. Express 14(4), 1487–1496 (2006).
13. M. V. Sarunic, B. E. Applegate, and J. A. Izatt, “Real-time quadrature projection complex conjugate resolved Fourier domain optical coherence tomography,” Opt. Lett. 31(16), 2426–2428 (2006).
14. B. J. Vakoc, S. H. Yun, G. J. Tearney, and B. E. Bouma, “Elimination of depth degeneracy in optical frequency-domain imaging through polarization-based optical demodulation,” Opt. Lett. 31(3), 362–364 (2006).
15. A. B. Vakhitn, K. A. Peterson, and J. D. Kane, “Resolving the complex conjugate ambiguity in Fourier-domain OCT by harmonic lock-in detection of the spectral interferogram,” Opt. Lett. 31(9), 1271–1273 (2006).
16. Y. Yasuno, S. Makita, T. Endo, G. Aoki, M. Itoh, and T. Yatagai, “Simultaneous B-M-mode scanning method for real-time full-range Fourier domain optical coherence tomography,” Appl. Opt. 45(8), 1861–1865 (2006).
17. L. An and R. K. Wang, “Use of a scanner to modulate spatial interferograms for in vivo full-range Fourier-domain optical coherence tomography,” Opt. Lett. 32(23), 3423–3425 (2007).
18. R. A. Leitgeb, R. Michaely, T. Lasser, and S. C. Sekhar, “Complex ambiguity-free Fourier domain optical coherence tomography through transverse scanning,” Opt. Lett. 32(23), 3453–3455 (2007).
19. R. K. K. Wang, “In vivo full range Fourier domain optical coherence tomography,” Appl. Phys. Lett. 90(5), 054103 (2007).
20. Y. K. Tao, M. Zhao, and J. A. Izatt, “High-speed complex conjugate resolved retinal spectral domain optical coherence tomography using a fiber optic switch,” Opt. Lett. 32(20), 2918–2920 (2007).
21. H. Wang, Y. Pan, and A. M. Rollins, “Extending the effective imaging range of Fourier-domain optical coherence tomography using a fiber optic switch,” Opt. Lett. 33(22), 2632–2634 (2008).
22. B. Hofer, B. Povazay, B. Hermann, A. Unterhuber, G. Matz, and W. Drexler, “Dispersion encoded full range Fourier-domain optical coherence tomography,” Opt. Express 17(1), 7–24 (2009).
23. J. A. Izatt and M. A. Choma, “Theory of optical coherence tomography,” in Optical Coherence Tomography: Technology and Applications, W. Drexler and J. Fujimoto, eds. (Springer, New York, 2008), pp. 47–72.
24. G. J. Tearney, B. E. Bouma, and J. G. Fujimoto, “High-speed phase- and group-delay scanning with a grating-based phase control delay line,” Opt. Lett. 22(23), 1811–1813 (1997).
25. A. Rollins, S. Yazdanfar, M. Kulkarni, R. Ung-Arunyawee, and J. Izatt, “In vivo video rate optical coherence tomography,” Opt. Express 3(6), 219–229 (1998).
26. O. Martinez, “3000 times grating compressor with positive group velocity dispersion: application to fiber compensation in 1.3–1.6 μm region,” IEEE J. Quantum Electron. 23(1), 59–64 (1987).
27. A. M. Rollins and J. A. Izatt, “Optimal interferometer designs for optical coherence tomography,” Opt. Lett. 24(21), 1484–1486 (1999).
28. Y. Chen, D. M. de Bruin, C. Kerbage, and J. F. de Boer, “Spectrally balanced detection for optical frequency domain imaging,” Opt. Express 15(25), 16390–16399 (2007).
29. M. Wojtkowski, V. Srinivasan, T. Ko, J. Fujimoto, A. Kowalczyk, and J. Duker, “Ultra-high-resolution, high-speed, Fourier domain optical coherence tomography and methods for dispersion compensation,” Opt. Express 12(11), 2404–2422 (2004).
30. R. Huber, D. C. Adler, and J. G. Fujimoto, “Buffered Fourier domain mode locking: unidirectional swept laser sources for optical coherence tomography imaging at 370,000 lines/s,” Opt. Lett. 31(20), 2975–2977 (2006).
31. B. D. Goldberg, B. J. Vakoc, W.-Y. Oh, M. J. Suter, S. Waxman, M. I. Freilich, B. E. Bouma, and G. J. Tearney, “Performance of reduced bit-depth acquisition for optical frequency domain imaging,” Opt. Express 17(19), 16957–16968 (2009).
32. W. Wieser, B. R. Biedermann, C. M. Eigenwillig, T. Klein, and R. A. Huber, “FDML based multi-spot OCT at 4,100,000 A-scans and 4 Gvoxels per second,” presented at SPIE Photonics West, San Francisco, CA, Jan. 23–28, 2010.
33. T. Klein, W. Wieser, C. M. Eigenwillig, B. R. Biedermann, and R. Huber, “Megahertz OCT for ultrawide-field retinal imaging with a 1050 nm Fourier domain mode-locked laser,” Opt. Express 19(4), 3044–3062 (2011).
Fourier transform of this spectral interferogram. FDOCT can be realized in two ways, either through the use of a broadband source and spectrometer (spectral-domain, or SDOCT) or a frequency swept laser and high bandwidth detector (swept-source, or SSOCT). Both FDOCT techniques suffer from an inherent (sample independent) reduced imaging depth range, typically limited to between 1 and 5mm. Optical attenuation from absorption and scattering in tissue typically limit how much light is recovered from depths beyond a few millimeters, and thus for many applications this inherent reduced depth range is not the limiting factor in determining the practical imaging depth. However, several important OCT applications would benefit from extended imaging depths, including ophthalmic imaging of the anterior segment, intrasurgical imaging, small animal imaging, and catheter imaging of coronary arteries. Extending the imaging range of FDOCT has thus been an area of interest for which a number of techniques have been developed [7–22]. These techniques include phase shifting using a PZT-mounted reference arm [7] or electro-optic phase modulator [8], heterodyne SSOCT [9–11], instantaneous acquisition of phase separated interferograms using 3x3 interferometers [13] or polarization encoding [14], harmonic lock-in detection of phase modulation [15], imparting a phase ramp across a B-scan with B-M mode scanning [16] and pivot-offset scanning [17–19], sinusoidal phase modulation [20], rapidly switching between multiple reference arms [21] and dispersion encoding [22]. Unfortunately, all of these techniques are accompanied by drawbacks in the form of reduced sensitivity, reduced axial resolution, reduced imaging speed, required lateral oversampling, increased system complexity, increased cost and/or increased signal processing overhead. In addition, most of these techniques produce incomplete suppression of the complex conjugate artifact, resulting in distracting “ghost” images.

Arguably the most effective of these methods is heterodyne complex-conjugate resolved SSOCT (HCCR-SSOCT), which resolves the ambiguity by shifting the peak sensitivity position away from electronic DC, such that positive and negative displacements from that position can be discerned. As this technique shifts, rather than suppresses, the complex conjugate, it completely resolves the artifact. In addition, HCCR-SSOCT does not result in any reduction in imaging speed or require lateral oversampling. In this method, one or two active elements acting as frequency shifters, usually acousto-optic modulators (AOM’s) [9,10] (though electro-optic modulators, EOM’s, have been used [11]) are used to apply a differential modulation frequency between the sample and reference arms. While effective, this technique is limited in that such modulators are expensive and require careful alignment. More significantly, active frequency shifters tend to have appreciable insertion losses, resulting in reduced sensitivity, and restricted optical bandwidth, resulting in spectral distortion and broadening of the axial point-spread function. In addition, processing of the acquired data requires either hardware demodulation [10] or significant post-processing [9].

In this work, we present a method of realizing HCCR-SSOCT using a dispersive optical delay line (D-ODL) as an alternative to AOM’s or EOM’s. This technique confers the same advantages as traditional HCCR-SSOCT in that it doubles the inherent imaging range without sacrificing imaging speed or requiring lateral oversampling. Furthermore, this technique bears no resolution penalty, incurs little to no sensitivity loss, is low-cost and easy to implement, requires no additional signal processing and can be designed to support broad wavelength ranges and arbitrary imaging speeds. As an additional benefit, the D-ODL also allows for hardware dispersion management, which reduces dependence on software dispersion compensation algorithms and adds flexibility to system design.

2. Theory

2.1. Background

All FDOCT systems suffer from an inherent limited imaging depth range due to two factors. The first of these stems from the fact that SSOCT extracts depth information from the Fourier transform of a spectral interferogram. As the spectral interferogram can only be recorded as a real signal, its Fourier transform is necessarily Hermitian symmetric about the zero pathlength
difference position (ZPD), which occurs at DC after Fourier transformation of the spectral interferogram. Consequently, positive and negative displacements about the ZPD position cannot be unambiguously resolved, giving rise to mirror image artifacts. These artifacts can be avoided by placing the zero pathlength difference position outside of the sample, which results in two non-overlapping mirror images of the sample being acquired in the positive and negative frequencies. This technique resolves the complex ambiguity at the expense of halving the useful imaging range.

The complex conjugate ambiguity would not pose such a problem if it were not for the fact that the total imaging range is also limited by a phenomenon known as sensitivity falloff. The instantaneous linewidth of the swept laser can be thought of as a sampling function that interrogates the intrinsic spectral interferogram. The spectral interferogram is sampled by, and thus convolved with, the instantaneous laser linewidth, which results in reduced fringe visibility when the fringe period approaches the linewidth. As smaller fringe periods (i.e. higher fringe frequencies) correspond to deeper imaging depths, this reduced visibility results in decreasing sensitivity with increasing imaging depth.

The combination of the complex conjugate ambiguity and sensitivity falloff results in a depth sensitivity profile that is well approximated by the single-sided Fourier transform of the average instantaneous linewidth of the laser over an entire sweep. Assuming a central wavelength of $\lambda_0$ and a Gaussian linewidth function with a full-width at half-max (FWHM) of $\delta_\lambda$, the depth at which the sensitivity is reduced by one half, $z_{6\text{db}}$, is given by [23]

$$z_{6\text{db}} = \frac{\ln(2) \lambda_0^2}{\pi \delta_\lambda} \quad (1)$$

For complex conjugate resolved SSOCT techniques, like HCCR-SSOCT, the sensitivity profile is given by the two-sided Fourier transform, and ranges from $-z_{6\text{db}}$ to $+z_{6\text{db}}$, thus doubling the inherent imaging range. The deepest resolvable single-sided depth, $z_{\text{max}}$, is determined by the spectral sampling interval, $\delta_\lambda$, and the central wavelength, $\lambda_0$, according to the relation [23]

$$z_{\text{max}} = \frac{\lambda_0^2}{4\delta_\lambda} \quad (2)$$

Because the signal acquired in HCCR-SSOCT is up-shifted in frequency, image information is shifted to deeper depths. As such, the maximum imaging depth, $z_{\text{max}}$, must also be increased by increasing the spectral sampling density.

2.2. Heterodyne Swept-Source Optical Coherence Tomography

Heterodyne SSOCT refers to a technique wherein the spectral interferogram frequencies are shifted by mixing with a high frequency carrier. This frequency mixing is achieved by applying a net frequency shift to one arm of the interferometer, either by shifting just the reference arm or differentially shifting both arms. As a result, the spectral interferogram will be frequency shifted to be centered at the beat frequency of the applied shifts ($\omega_D$), with positive frequencies appearing above $\omega_D$ and negative frequencies appearing below $\omega_D$, thus resolving the complex conjugate ambiguity. This technique is described in greater detail in references [9,10].

The technique we describe in this work differs slightly from prior implementations of HCCR-SSOCT in the manner in which the frequency shift is achieved. To create the frequency shift, a dispersive optical delay line (D-ODL) is constructed such that it produces a phase delay that varies linearly with wavelength, but a group delay that is constant. To demonstrate how applying this wavelength-dependent phase delay creates a frequency shift, we begin with the derivation in [10] and modify it accordingly.

In standard SSOCT, the photocurrent signal generated at the photodiode as a function of the instantaneous laser wavenumber, $i(\hat{k})$, is related to
\[ i(k) \propto S(k) \left[ R_n + \sum_{n=1}^{N} \sqrt{R_n} \cos(2k[z_n - z_r]) + 2 \sum_{n=1}^{N} \sqrt{R_n R_m} \cos(2k[z_n - z_m]) \right] \] (3)

where \( k \) is any wavenumber in the source’s sweep range, \( S(k) \) is the source’s power spectral density, \( R_n \) is the reflectivity of the \( n^{th} \) of \( N \) reflectors located at axial position \( z_n \), and \( R_m \) is the reflectivity of the reference mirror located at \( z_r \). The first two terms inside the brackets are the non-interferometric (DC) terms, the third term is the cross-interference term of interest, and the final term is the auto-correlation artifact.

For simplicity, we neglect the spectral envelope, DC and autocorrelation terms and only consider the carrier of the interferometric cross term. To simplify further, let us only consider the photocurrent due to a single reflector, \( i_n(k) \). We also now account for the wavelength-dependent phase delay by recasting the phase pathlength of the reference reflector as a function of wavenumber, \( \phi_{\text{ref}}(k) \):

\[ i_n(k) \propto \cos(2k[z_n - z_{\phi,n}(k)]) \] (4)

Note that here we have used the subscript \( \phi \) to indicate phase pathlengths. The critical design feature of the D-ODL is that the phase delay varies linearly with wavelength. If we define the phase pathlength at the center wavelength to be \( z_{\phi,0} \) and the slope of the delay to be \( M \) (in mm/nm), we can express this pathlength as

\[ z_{\phi,n} = z_{\phi,0} + M(\lambda_n - \lambda_0) \] (5)

Re-expressing this pathlength as a function of wavenumber gives

\[ z_{\phi,n}(k) = z_{\phi,0} + M \left( \frac{2\pi}{k_0} - \frac{2\pi}{k} \right) \] (6)

Inserting this into Eq. (4) and simplifying yields

\[ i_n(k) \propto \cos(2k[z_n - z_{\phi,n} - 2\pi M \left( \frac{1}{k_0} - \frac{1}{k} \right)]) \] (7)

Here, the final term in the cosine represents a wavelength-dependent differential phase shift, which is defined to be zero for the center wavelength. The axial position shift created by this phase shift can thus be defined as \( \Delta z_D = 2\pi M/k_0 \), and Eq. (7) can be re-expressed as

\[ i_n(k) \propto \cos(2k(z_n - z_{\phi,n} - \Delta z_D) + 4\pi M) \] (8)

This expression is of the same form as Eq. (4), but with the cross-interference terms shifted by \( \Delta z_D \) and the addition of a constant phase term. The \( \Delta z_D \) term here is analogous to the \( z_0 \) term discussed in reference [10], except that the shift is applied linearly in wavenumber. Previously described HCCR techniques [9,10] apply the frequency shift in the time domain, that is, with a constant frequency. As swept-source lasers typically sweep non-linearly in time (with respect to wavenumber), these techniques require additional processing, either in the form of hardware demodulation [10] or post-processing in software [9]. In contrast, the frequency shift applied by the D-ODL occurs directly in the wavenumber domain, and as such, is encoded in the same time-base as the frequency sweep of the laser. As a result, no hardware demodulation or complicated signal processing is required to process the HCCR-SSOCT data. All that is required is that the time-encoded photocurrent be resampled to be linear in wavenumber, which is a usual processing step in SSOCT.

A critical feature of HCCR techniques is that the sensitivity falloff is centered about \( \Delta z_D \), and not DC. For this technique, while the D-ODL generates a wavelength-dependent phase delay, group delay remains constant. As a result, the frequency shift imposed on the spectral interferogram does not affect the group pathlength difference between the reference and the sample arm. As sensitivity falloff depends on group pathlength, and not phase pathlength, the falloff profile remains centered about the ZPD position, which is shifted from DC to \( \Delta z_D \).
It is important to note that since signal acquired in this technique is still real, the Fourier transform of the photocurrent remains Hermitian symmetric. However, positive and negative displacements from the ZPD position now appear on either side of the $\Delta z_D$ position. The complex conjugate still exists, but now produces an identical full-range image in the negative frequency space centered about $-\Delta z_D$, thus resolving the ambiguity. In addition, the DC and autocorrelation artifacts remain centered at DC and can be removed by high-pass filtering.

2.3. Grating-based Dispersive Optical Delay Line

As described above, any optical system that produces a linear, wavelength-dependent phase delay and constant group delay can be used to create the frequency shift required to realize HCCR-SSOCT. One can conceive of numerous optical systems to create a wavelength-dependent phase delay, with varying degrees of complexity. Here, we present a compact and efficient grating-based D-ODL, similar in design to the rapid scanning optical delay lines (RSOD’s) that were used in the fastest iterations of time-domain OCT systems [24,25]. However, this D-ODL differs from an RSOD in that the scanning mirror in the image plane of the grating lens is replaced with a mirror at a fixed angle. When carefully designed, such a D-ODL provides a linear, wavelength-dependent phase delay and constant group delay. By controlling the slope of this wavelength-dependent phase delay through careful selection of design parameters, the D-ODL can be used to realize HCCR-SSOCT with a precisely tuned $\Delta z_D$. Figure 1 shows a schematic of one such D-ODL, and a plot of the phase pathlength versus wavelength through the system.

RSOD’s have been used extensively in TDOCT due to their ability to provide rapidly scanned group delays, while also providing precise control of the Doppler frequency and dispersion compensation [24–26]. Design considerations are described in detail in reference [25], so only the relevant parameters, the free-space phase and group pathlengths, will be discussed here.

Rollins et al. [25] provide a detailed derivation of the phase pathlength through an RSOD. However, our implementation differs slightly in that the mirror is stationary and the wavelength sweeps rapidly in time. We define $\lambda_0$ to be the center wavelength of the entire sweep, $\lambda_c$ to be the center of the instantaneous linewidth, and $\lambda$ to be any wavelength propagating through the system. Then, following the derivation in [25], the phase shift for a single pass through the D-ODL (double pass of the grating) as a function of $\lambda$ is given by

Fig. 1. Left: Schematic of the optical delay line. $L_{col}$: Collimating lens, DG: Diffraction grating, $L_{ODL}$: compound achromatic lens with focal length $f_{ODL}$, M: gold mirrors, $\theta$, mirror angle. Red, green and blue lines represent ray traces at wavelengths of 1090nm, 1040nm, and 990nm respectively. Right: Plot of ray-tracing derived phase pathlength difference through the system as a function of wavelength demonstrating a nearly perfect linear relationship.
\[ \phi(\lambda) = \frac{4\pi f_{\text{ODL}} (\lambda - \lambda_0)}{p\lambda} \]  

(9)

where \( \theta \) is the mirror angle, \( f_{\text{ODL}} \) is the lens focal length and \( p \) is the grating pitch. This expression was derived using the grating equation and the small angle approximation. Specifically, this small angle approximation was used to compute the diffraction angle as a function of wavelength, as follows:

\[ \theta_d(\lambda) = \arcsin \left( \frac{\lambda - \lambda_0}{p} \right) \approx \frac{\lambda - \lambda_0}{p} \]  

(10)

It was found that the use of the small angle approximation introduces less than 0.02% error over the range of diffraction angles used in our experiments (−30mrad to 30mrad). Next, Eq. (9) is expressed as a function of angular optical frequency:

\[ \phi(\omega) = \frac{4\pi f_{\text{ODL}} (\omega_0 - \omega)}{p\omega_0} \]  

(11)

Ultimately, we are interested in the phase delay, defined as

\[ t_\phi = \frac{\phi(\omega)}{\omega} \]  

(12)

Note that the phase delay is defined according to the phase shift of the instantaneous central wavelength. Combining Eq. (11) and Eq. (12) yields

\[ t_\phi(\lambda_c) = \frac{2\theta f_{\text{ODL}} (\lambda_c - \lambda_0)}{pc} \]  

(13)

The phase delay is thus a function of \( \lambda_c \), which is swept rapidly in time. The free-space phase pathlength difference, relative to the central wavelength, is then also a function of \( \lambda_c \):

\[ \Delta \lambda(\lambda_c) = \frac{2\theta f_{\text{ODL}} (\lambda_c - \lambda_0)}{p} \]  

(14)

As expected, the free-space phase pathlength varies linearly with the instantaneous wavelength. To clarify, this pathlength difference is applied in a single pass through the D-ODL (double pass of the grating). The fact that the double-pass mirror in the D-ODL also serves as the reference mirror in the OCT interferometer, causing the D-ODL to be double passed (quadruple pass through the grating) is accounted for by the factor of 2 in Eq. (8).

Recalling that we defined \( M \) as the slope of the wavelength-dependent phase pathlength, we can now define \( M \) in terms of design parameters of the D-ODL:

\[ M = \frac{d t_\phi(\lambda_c)}{d\lambda_c} = \frac{2\theta f_{\text{ODL}}}{p} \]  

(15)

Similarly, we can also express \( \Delta \lambda_D \) in terms of these parameters:

\[ \Delta \lambda_D = M\lambda_0 = \frac{2\theta f_{\text{ODL}} \lambda_0}{p} \]  

(16)

As discussed above, while the D-ODL creates a linear phase delay, it also creates a constant group delay. The group delay can also be defined from the phase shift from the relation

\[ t_g = \frac{\partial \phi(\omega)}{\partial \omega} \]  

(17)

Combining Eq. (11) and Eq. (17) yields
As expected, the group delay is constant with respect to the instantaneous optical frequency. Thus, both the phase and group pathlength difference can be expressed as simple functions of the slope parameter \( M \):

\[
\Delta \phi (\lambda_i) = M (\lambda_i - \lambda_0) \quad (19)
\]

\[
\Delta \lambda_g = -M \lambda_0 \quad (20)
\]

### 3. Methods

For a proof-of-principle experiment, we constructed an SSOCT system with a reference arm that was interchangeable between a standard configuration and a D-ODL. Sensitivity and falloff measurements were performed with each system and compared. The complex conjugate suppression ratio (CSR) was also measured. The feasibility of this technique for \( \textit{in vivo} \) imaging was demonstrated by imaging the anterior segment of healthy human volunteers.

#### 3.1. Optical Delay Line

The laser used in the SSOCT system, had a measured \( z_{\text{d}} \) length of approximately 5.4mm (see below, Fig. 3, top). The depth sensing range for this system was limited by the amplifier and digitization bandwidths, resulting in an electronic 3dB bandwidth from 17MHz to 500MHz. This corresponded to an axial sensing range from 0.35mm to 10.6 mm. Thus, we chose \( \Delta \lambda_0 \) to be approximately 5mm to center the falloff profile in the depth scan. A reflective, ruled grating with 600 grooves/mm (\( p = 1.67\mu \text{m} \)) and a 100mm compound achromatic lens (\( f_{\text{ODL}} \approx 104.2\text{mm} @ 1040\text{nm} \)) were used to construct the D-ODL. From Eq. (16) above, the required mirror angle, \( \theta \), was 2.1 degrees. The D-ODL was modeled in ray tracing software (Zemax) to model the optical pathlength as a function of input wavelength (Fig. 1, left). A linear fit was applied to the ray tracing data and fit with an \( R^2 \) value greater than 0.99999 (Fig. 1, right).

#### 3.2. SSOCT system

The SSOCT system we constructed is shown schematically in Fig. 2. The source used was a swept-source laser (Axsun Technologies) with a central wavelength of 1040nm, sweep bandwidth of 100nm, repetition rate of 100kHz, 46% duty cycle and average output power of 20mW. Using a balanced coupler and a fiber-optic circulator (AC Photonics), a balanced Michelson fiber interferometer was constructed [27]. The sample arm consists of two scanning galvanometers (Cambridge technologies) and a compound objective lens. In compliance with the ANSI standard for ocular exposure to laser light (ANSI Z136.1), power incident on the sample arm was attenuated to 1.8mW. Two interchangeable reference arms were tested: a standard reference arm (SRA) and the dispersive optical delay line (D-ODL). Light returning from the sample and reference arms interfered in the fiber coupler and was directed to an 800MHz balanced receiver (NewFocus 1607). Output of the balanced receiver was high-pass and anti-alias filtered, amplified and digitized at 1GS/s.

#### 3.3. Digitization, wavenumber calibration and resampling

As described above, the frequency shift applied by the D-ODL occurs directly in the wavenumber domain, and thus the only processing step required is that the time-encoded photocurrent to be sampled to be linear in wavenumber, a usual processing step in SSOCT.

The Axsun laser used in the system described above operated with a 46% duty cycle. This means that, even though the laser supported an A-scan rate of 100kHz, the sweep period was only 4.6\( \mu \)s. The internal k-clock of the Axsun laser output a calibration signal with 1376 periods over the 4.6\( \mu \)s sweep, with a frequency that varied between 250 and 350MHz. If the spectral interferogram is sampled once per clock period, as is its intended design, this...
allows for a 3.7mm free-space imaging range. However, as the signal in HCCR-SSOCT techniques is up-shifted, the depth imaging range must be extended. This can be achieved in two ways. One option is to construct a k-clock with a longer pathlength mismatch and trigger the acquisition off of this longer clock. The alternative is to use the digitizer’s internal clock and digitize the k-clock to create a calibration signal. The former approach can be problematic as the pathlength mismatch required would be very long, and thus the visibility of the calibration signal would be strongly attenuated. Thus, we elected to digitize the laser’s internal k-clock along with the output of the balanced receiver. Both signals were digitized at 1GS/s in a dual channel, 8-bit digitizer card (Alazar Technologies, ATS9870). It is important to note that, in general, the two-fold increase in digitization bandwidth required for HCCR-SSOCT would increase shot noise (and thereby reduce sensitivity) by about 3dB. However, for this particular implementation, the theoretical sensitivity was unchanged, as even when using the SRA, the full 1GS/s digitization rate (500MHz bandwidth) was required to digitize the k-clock.

Wavenumber recalibration was carried out as follows. Both the clock and receiver signals were digitized using a custom designed LabVIEW VI. The clock signal was then up-sampled and the zero-crossings of the first derivative were recorded. The receiver data was then resampled to be linear with respect to the intervals of these zero-crossings. It was found that the laser sweep was sufficiently stable that, for real time image display in LabVIEW, only the clock corresponding to the first A-scan for every B-scan needed to be processed, and signals for all other A-scans could be resampled according to the calibration from this first A-scan. However, every clock signal was recorded and stored for use in post-processing. Due to the wide bandwidth of the k-clock (250 to 350MHz), the system $z_{\text{max}}$ varied over the laser sweep between 10.6mm and 14.8mm. As a result, image quality and axial resolution degrade rapidly beyond a depth of 10.6mm, as fringe frequencies corresponding to these depths begin to alias.

4. Results

4.1. Sensitivity and Axial resolution

Using the system depicted in Fig. 2, a sensitivity of 96.1 dB was measured (61.0dB measured with a $-35.1$dB calibrated reflector) with an average power of 1.8mW incident on the sample. Neglecting coupling losses, the theoretical shot-noise limited sensitivity expected was

![Fig. 2. SSOCT system schematic. Black lines represent optical fiber. Dashed lines and boxes represent the interchangeable standard reference arm (SRA) and optical delay line (D-ODL). Blue lines represent electrical connections. C: circulator, Lcoll: collimating lens, G: galvanometer scanning mirrors, Lobj: Objective lenses, M: mirror, BR: balanced receiver, HP/AA: high-pass and anti-alias filter, RFA: RF amplifier, A2D: digitizer.](image)
102.6dB [4]. The 6.5dB discrepancy can be explained by coupling losses (measured to be 3.5dB), amplification noise (0.7dB) and receiver intensity noise due to imperfect balancing in the balanced detection scheme [28]. A similar sensitivity was measured using a standard reference arm (96.4dB). Axial resolution, measured as the FWHM of the point-spread function from a mirror reflector, was found to be transform-limited in both cases at 6.7 microns without spectral shaping. Spectral shaping with a hamming window yielded an axial resolution of approximately 10 microns in both cases. We thus experimentally verified that the use of the D-ODL does not result in a loss of sensitivity or axial resolution.

4.2. Falloff
Sensitivity falloff was measured using both the SRA and D-ODL. Measurements using the SRA showed a single sided $z_{6\text{dB}}$ of approximately 5.4mm (Fig. 3, top). Measurements using the D-ODL yield an imaging range over which the sensitivity falloff is less than 6dB of approximately 10.1mm, ranging from 0.5mm to 10.6mm (Fig. 3, bottom). The limited electronic bandwidth of the amplifier and digitizer accounts for the slight reduction in the imaging range of the HCCR-SSOCT system from the expected imaging range of 10.8mm.

4.3. Complex conjugate suppression ratio
The performance of CCR techniques are often quantified by their complex conjugate suppression ratio, (CSR); that is, the ratio between the amplitude of a calibrated reflectance peak to the amplitude of its residual artifact. As heterodyne techniques shift, rather than suppress, the complex conjugate, residual artifacts would only arise if a portion of the reference light traversed the D-ODL without undergoing a phase shift. We verified experimentally that no residual artifact existed, at least within the dynamic range of our system. To increase the system dynamic range, 30 A-scans of a reflector at the same position

![Fig. 3. Sensitivity falloff measured using the SRA (top) and D-ODL (bottom). Red horizontal lines are drawn at a sensitivity level of $-6\text{dB}$ from the peak. The 6dB imaging ranges were $\sim 5.4\text{mm}$ with the SRA and $\sim 10.1\text{mm}$ with the D-ODL.](image)
were averaged. Figure 4 shows the average of these 30 A-scans acquired with a reflector positioned approximately ~1.8mm from the ZPD position. As the phase modulation shifts the peak by 5 mm, the peak appears at + 6.8mm, and residual artifacts would appear at + 1.8mm and −1.8mm. As seen in Fig. 4, no residual artifact is discernable. As such, we quantify the CSR to be at least the ratio of the A-scan peak at + 6.8mm to the standard deviation of the noise floor around + 1.8mm, measured to be 61.9dB.

4.4. Anterior segment imaging

To demonstrate the feasibility of this method for in vivo ophthalmic imaging, the anterior segments of three healthy volunteers were imaged. Figure 5 shows two representative anterior segment b-scans, acquired with 1000 a-scans/b-scan, registered and averaged over 10 frames for a total acquisition time 100ms. The image in Fig. 5 (top left) was acquired with the volunteer’s contact lens and cornea positioned near the focus of objective lens. As a result, the cornea appears bright and continuous, and corneal layers can be readily discerned, as shown in the enlargement in Fig. 5 (bottom left). However, as the depth range of the HCCR-SSOCT system exceeds the depth of focus of the objective lens, defocus results in poor visibility of the crystalline lens. The image in Fig. 5 (right) was acquired in the same volunteer with the
objective lens focus positioned near the posterior surface of the crystalline lens. This image demonstrates simultaneous imaging of all of the refractive surfaces of the anterior segment, from the anterior surface of the cornea to the posterior surface of the lens. In this case, the crystalline lens is readily visualized, but corneal visibility is significantly degraded due to both the limited depth of focus of the objective and the steep curvature of the cornea.

Figure 6 (Media 1 and Media 2) contains two movies depicting renderings of anterior segment data collected from two additional healthy volunteers. Figure 6 (left, Media 1) displays a volume where the focus of the objective lens was positioned near the cornea, similar to the b-scan shown in Fig. 5 (top left). Figure 6 (right, Media 2) displays a volume where the objective lens focus was positioned near the anterior surface of the crystalline lens. As a result, as was the case for the b-scan portrayed in Fig. 5 (right), the steep curvature of the cornea results in poor corneal visibility. In this case, the subject’s eyelid (located anterior to the cornea) is visualized, confirming that the inability to resolve the cornea was due to the corneal curvature, and not sensitivity falloff.

This limited depth of focus can be addressed through the use of a low numerical aperture (NA) objective, at the expense of lateral resolution. To demonstrate this, we replaced the original objective (NA~.019) with a lower power objective (NA~0.14). Figure 6 (left) shows a

Fig. 6. Single frame excerpts from Media 1 (left) and Media 2 (right) depicting 3D data sets of in vivo anterior segment volumes. Each volume consisted of 1000 x 2304 x 150 samples and was acquired in 1.5s.

Fig. 7. Projection views of in vivo anterior segment volumes acquired with standard (left) and low NA (right) objective lenses. Each volume consisted of 1000 x 2304 x 150 samples and was acquired in 1.5s.
projection of a volume acquired with the original objective, wherein the cornea and pupil are clearly visible but the crystalline lens is not visualized. Figure 7 (right) shows projection of a volume acquired with the low NA objective, demonstrating simultaneous visualization of all of the refractive surfaces of the anterior segment, from the anterior surface of the cornea to the posterior surface of the lens.

5. Discussion

5.1. Requirements and Applicability

The technique we describe is applicable to all SSOCT systems, regardless of the laser central wavelength or sweep rate. The grating-based D-ODL detailed above can be designed for virtually any optical wavelength by appropriate selection of the grating pitch and diffraction angle. Furthermore, this technique is not restricted to the grating-based D-ODL design, and in fact will work with any delay line that creates a wavelength-dependent phase delay and constant group delay. As the phase modulation acts directly in the wavenumber domain, the laser sweep speed does not factor into the design, and the D-ODL will provide the appropriate phase modulation for any sweep speed. In fact, for lasers with variable tuning speeds, no adjustment of the delay line is necessary when varying imaging speed.

5.2. D-ODL Performance and Design Considerations

While the design of the D-ODL used in these experiments is relatively simple, a few performance notes and design considerations are worth mention. First, as the axial position shift (\(\Delta z_D\)) scales with the lens focal length \(f_{ODL}\) and the mirror angle \(\theta\), for a given lens aperture, increasing one of these parameters requires decreasing the other. Thus, the maximal shift that can be applied will ultimately depend on the lens aperture, and is relatively invariant on the choice of focal length or angle. A standard one inch lens was sufficient to create the ~5mm axial position shift in our experiments, although larger shifts would likely require a larger diameter lens.

Careful selection of the lens is important, as aberrations in the system can result in low throughput, wavelength-dependent coupling efficiency and the introduction of higher order dispersion. Furthermore, there is a trade-off in the selection of the lens focal length, as short focal lengths are more prone to optical aberrations, while long focal lengths can significantly and inconveniently extend the reference delay. Note that the focal length is traversed 8 times in a double-pass through the D-ODL. In our experiments, we found that the use of a matched pair of 200mm focal length achromatic doublets provided efficient coupling while introducing minimal higher order dispersion. A second-order software dispersion compensation algorithm [29] was sufficient to correct for any residual unmatched dispersion between the sample arm and D-ODL.

Finally, the choice of grating may be important in certain applications. As the beam diffracts off of the grating four times in a double-pass through the ODL, the diffraction efficiency of the grating may be critical for power-starved systems. This is especially true for high-speed systems where reference power is used to compensate for the low transimpedance gain of high bandwidth photoreceivers. Our system employed a ruled aluminum grating, which was blazed at a slightly larger angle than was ideal (22°1’ versus an ideal blaze of 19°18”). When optimized, our maximal D-ODL throughput was approximately 8%, which provided more than 100\(\mu\)W of reference power but was highly polarization sensitive. For applications where throughput or polarization insensitivity is critical, custom volume phase holographic gratings can be used to achieve highly efficient, polarization-insensitive performance.

5.3. Digitization Considerations

Huber et al. [30] and Goldberg et al. [31] have demonstrated that, for SSOCT systems, 8-bit digitization results in only a marginal reduction in image quality and SNR as compared to digitization at higher bit depths. This is in large part due to the dynamic range that is
conserved by the attenuation of the source’s spectral shape via balanced detection. However, imperfect balancing, which occurs as a result of the chromatic dependence of the coupler splitting ratio, can result in considerable residual DC artifact. In fact, for the source and couplers used in these experiments, we found that this residual DC artifact dominated over weak and moderate signals. This required the use of the higher input ranges on the digitizer, which, due to the shallow bit depth, resulted in quantization noise dominating over shot noise and limiting SNR. Fortunately, because the signals of interest are shifted away from DC in HCCR-SSOCT, this residual DC artifact can be removed by high-pass filtering before digitization. After high-pass filtering, lower input ranges could once again be used, and quantization noise was no longer limited SNR.

An important requirement for implementing HCCR-SSOCT is available detection and digitization bandwidth. The minimum bandwidth required for a heterodyne SSOCT system is the sum of the frequency corresponding to the standard “homodyne” $z_{\text{max}}$ and the shift frequency corresponding to $\Delta z_D$. Therefore, the required bandwidth for heterodyne systems will typically be at least twice that of homodyne systems. Furthermore, because the DC and autocorrelation artifacts remain centered at the baseband, bandwidths even higher than twice the homodyne bandwidth are beneficial. With the recent and continuing development of swept-source lasers with sweep rates in the megahertz regime [32,33], the viability of heterodyne techniques for such high speed applications will ultimately depend on the continuing development of high bandwidth digitization technology.

5.4. Balanced detection pathlength matching

It is important to note that the two fiber paths from the balanced coupler to the balanced receiver must be pathlength matched for this system. While this is not ordinarily a concern for SSOCT systems operating at speeds on the order of tens of kilohertz or less, at faster speeds, unmatched fiber lengths in the receiver arms may result in undesirable image artifacts. This occurs when the spectral interferogram fringe wavelengths approach the fiber pathlength mismatch, such that the orthogonal components in each detection arm are no longer in phase. For a 4.6µs sweep, a fringe signal with 1000 periods will have a wavelength of 1.32 meters, corresponding to approximately 0.9 meters of fiber. Mismatches on the order of this length will cause the fringes in the two arms of the balanced detection to move in and out of phase, with the relative phase shift depending on the fringe frequency. As a result, the fringes will be constructively and destructively superimposed as a function of frequency, resulting in an undesirable signal modulation along the depth scan. This phenomenon can be avoided by ensuring the fiber lengths in the detection arms are well-matched.

5.5. Hardware dispersion management

An additional advantage conferred by the use of RSOD’s in time-domain OCT was the ability to manage dispersion in hardware, which was achieved by displacing the grating from the focal plane of the lens [26]. This feature is preserved in the D-ODL design described here, allowing for increased flexibility in design of the interferometer. This is particularly useful as the D-ODL reference arm is inherently very long. Rather than constructing an equally long sample arm, the excess length of the reference arm can be matched by adding fiber to the sample arm, and correcting the resulting dispersion with the D-ODL. When using this method, software dispersion compensation would still be required to correct residual and higher order dispersion, but this is easily implemented using the techniques described in [29].

6. Conclusion

The use of a dispersive optical delay line to achieve complex conjugate resolved heterodyne swept-source optical coherence tomography has been demonstrated. This technique confers numerous advantageous over alternative techniques for complex conjugate rejection in that it completely resolves the complex conjugate artifact, does not reduce imaging speed, sensitivity or resolution, is low cost, requires no additional processing, works for any wavelength or sweep rate, and also allows for hardware dispersion management. The extended imaging
depth that can be achieved using this method is valuable to several important OCT applications, especially ophthalmic imaging of the anterior segment and endoscopic imaging of coronary arteries.

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