NRC Publications Archive Archives des publications du CNRC

Optimal signal processing of nonlinearity in swept-source and spectral-domain optical coherence tomography

Vergnole, Sébastien; Lévesque, Daniel; Bizheva, Kostadinka; Lamouche, Guy

This publication could be one of several versions: author's original, accepted manuscript or the publisher's version. / La version de cette publication peut être l'une des suivantes : la version prépublication de l'auteur, la version acceptée du manuscrit ou la version de l'éditeur.

For the publisher's version, please access the DOI link below. / Pour consulter la version de l'éditeur, utilisez le lien DOI ci-dessous.

Publisher's version / Version de l'éditeur:

<https://doi.org/10.1364/AO.51.001701>

Applied Optics, 51, 11, pp. 1701-1708, 2012-04-05

NRC Publications Record / Notice d'Archives des publications de CNRC:

<https://nrc-publications.canada.ca/eng/view/object/?id=e62d5caf-fb81-4307-a5cf-dd1ad6dd6918>

<https://publications-cnrc.canada.ca/fra/voir/objet/?id=e62d5caf-fb81-4307-a5cf-dd1ad6dd6918>

Access and use of this website and the material on it are subject to the Terms and Conditions set forth at

<https://nrc-publications.canada.ca/eng/copyright>

READ THESE TERMS AND CONDITIONS CAREFULLY BEFORE USING THIS WEBSITE.

L'accès à ce site Web et l'utilisation de son contenu sont assujettis aux conditions présentées dans le site

<https://publications-cnrc.canada.ca/fra/droits>

LISEZ CES CONDITIONS ATTENTIVEMENT AVANT D'UTILISER CE SITE WEB.

Questions? Contact the NRC Publications Archive team at

PublicationsArchive-ArchivesPublications@nrc-cnrc.gc.ca. If you wish to email the authors directly, please see the first page of the publication for their contact information.

Vous avez des questions? Nous pouvons vous aider. Pour communiquer directement avec un auteur, consultez la première page de la revue dans laquelle son article a été publié afin de trouver ses coordonnées. Si vous n'arrivez pas à les repérer, communiquez avec nous à PublicationsArchive-ArchivesPublications@nrc-cnrc.gc.ca.

National Research
Council Canada

Conseil national de
recherches Canada

Canada

Optimal signal processing of non-linearity in swept-source and spectral-domain optical coherence tomography

Sébastien Vergnole^{1,3}, Daniel Lévesque¹, Kostadinka Bizheva² and Guy Lamouche^{1,*}

¹ *Industrial Materials Institute, National Research Council Canada, 75 de Mortagne, Boucherville (QC), J4B 6Y4, Canada*

² *Dept. of Physics and Astronomy, University of Waterloo, Waterloo (ON), N2L3G1, Canada*

³ *Current address: Horiba Scientific, 231 rue de Lille, 59650 Villeneuve d'Ascq, France*

**Guy.Lamouche@cnrc-nrc.gc.ca*

We demonstrate the efficiency of the convolution using an optimized Kaiser-Bessel window to resample non-linear data in wavenumber for Fourier-domain optical coherence tomography (OCT). We extend our previous experimental demonstration that was performed with a specific swept-source non-linearity. The method is now applied to swept-source OCT data obtained for various simulated swept-source non-linearities as well as spectral domain OCT data obtained from both simulations and experiments. Results show that the new optimized method is the most efficient for handling all the different types of non-linearities in the wavenumber domain that one can encounter in normal practice. The efficiency of the method is evaluated through comparison with common methods using resampling through interpolation prior to performing a fast-Fourier transform and with the accurate but time consuming discrete Fourier transform for unequally spaced data which involves Vandermonde matrices. © 2012 Optical Society of America

OCIS codes: (170.4500) Optical coherence tomography; (070.4340) Nonlinear optical signal processing

1. Introduction

In Fourier-domain optical coherence tomography (FD-OCT) [1], one usually has to deal with unequally spaced data in wavenumber space (k -space). In swept-source OCT (SS-OCT), the uneven spacing is caused by the nonlinear sweep of the source in wavenumber. In spectral-domain OCT (SD-OCT), it is caused by the detection device (spectrometer), which is typically designed for equally spaced wavelength. The sampling non-linearity in the detected OCT signal can be avoided through hardware modifications by using k -triggering in SS-OCT [2] or by using a custom designed spectrometer that is linear in wavenumber in SD-OCT [3]. These hardware modifications complicate the overall design and increase the cost of the OCT system. Therefore, various interpolations techniques are currently used to resample the unequally spaced raw SS-OCT or SD-OCT data, to obtain evenly spaced data in k -space prior to performing a fast Fourier transform (FFT). The choice of resampling method is quite important since it impacts the processing time and the signal-to-noise ratio. Additionally, it can introduce artifacts in FD-OCT images.

We recently proposed an efficient method based on the convolution by an optimized Kaiser-Bessel window followed by the computation of the FFT [4]. The key element of this method is the optimal selection of the Kaiser-Bessel window parameters that allows efficient processing with a small oversampling factor. This method was validated with experimental data in SS-OCT using a commercial swept-source with a very specific non-linearity. In the current paper, we show that the applicability of the method is very general in handling all types of non-linearities one can encounter in SS-OCT and SD-OCT. This demonstration is performed with simulated data for both SS-OCT and SD-OCT, and experimental data acquired with a SD-OCT system.

Section 2 describes the processing methods used in our study. Section 3 describes the procedure regarding the simulation and the non-linearities. Comparisons of the different methods using point spread functions and processing times are presented in Sections 4 and 5, respectively. Section 6 illustrates our method when applied to 2D images of the human retina, acquired with a research grade SD-OCT system. Conclusions are summarized in Section 7.

2. Processing methods

To deal with non-linearity in k -space data, the methods considered in this study are discrete Fourier transform for unequally-spaced data which involves Vandermonde matrices, linear and spline interpolations followed by FFT, and convolution with a optimized Kaiser-Bessel window followed by FFT. Detailed description of all these methods is provided in [4]. All methods require an accurate calibration vector so that the wavenumber k is *a priori* known for all time sampling points, N . The time consuming Vandermonde matrix method is the

most accurate and is thus our reference. For the other methods, an oversampling factor α is used to resample the data on a finer scale. These other methods also show a strong decrease in the peak intensity with optical depth after the FFT operation, this decrease being different than the OCT sensitivity roll-off related to the design of the SS- or SD-OCT system. A correction, referred to in the following as the roll-off correction, must therefore be applied after the FFT operation, which comprises of multiplying the result with a properly determined vector [4,5]. The correction is directly embedded in the linear interpolation and in the Kaiser-Bessel methods while, in the case of spline interpolation it has to be empirically determined. Also, the windowing of resampled data with a fixed step size in the k -space before the FFT operation is found useful to reduce leakage in the A-scan.

Since it is the central point of the present paper, we briefly review our recently proposed method to resample data through convolution with an optimized Kaiser-Bessel window. Starting with the k vector in the range $k_{min} < k < k_{max}$, the step sizes in wavenumber domain δk and path length domain δz are given by:

$$\delta k = \frac{k_{max} - k_{min}}{N\alpha} = \frac{2\pi}{N'\delta z}, \quad \delta z = \frac{2\pi}{k_{max} - k_{min}}, \quad (1)$$

where $N' = N\alpha$ is the number of points involved in the FFT operation. The convolution of OCT data is made on $N\alpha$ center values of k using the optimal Kaiser-Bessel window as:

$$S_k = \sum_{j=1}^M S_j C_{kj}, \quad (2)$$

where S_k is the vector of the resampled data, S_j is the measured OCT data around the center value k , and the Kaiser-Bessel window is defined by the C_{kj} values:

$$C_{kj} = I_0(\beta \sqrt{1 - (2\kappa/M)^2})/M, \quad \kappa = \left| \frac{k - k^j}{\delta k} \right| \leq \frac{M}{2}, \quad (3)$$

where $I_0()$ is the zero-order modified Bessel function of the first kind, the k^j 's are the neighboring points around the center value k , and M is the upper limit of the convolution window length. As discussed further below, a closed-form expression for the optimal value of the shape parameter β is:

$$\beta = \pi \sqrt{\frac{M^2}{\alpha^2} \left(\alpha - \frac{1}{2} \right)^2 - 0.8}. \quad (4)$$

The convolution of the data with a Kaiser-Bessel window inherently requires a correction after the FFT operation, the roll-off correction, multiplying the result with the vector of index n :

$$\frac{1}{c_n} = \frac{\sqrt{(n\pi M/N')^2 - \beta^2}}{\sin \left(\sqrt{(n\pi M/N')^2 - \beta^2} \right)}, \quad (5)$$

where c_n is the Fourier transform of the Kaiser-Bessel function and complex arithmetics is allowed. In our previous paper, [4] we used a different definition of c_n which corresponds to the reciprocal of our current definition; it provided directly the correction vector. The definition of c_n in Eq. (5) is better since it highlights the relation of the correction vector with the Fourier Transform of the Kaiser-Bessel function. The parameters M and α have to be selected as a trade-off between image quality and processing time.

The optimal value of the shape parameter β for given M and α in Eq. (4) is of primary importance; it allows the use of a small oversampling factor between 1 and 2 while maintaining a good accuracy. It is obtained from a careful analysis of the error in the spatial domain, taking into account the repetition and aliasing from data sampling [6]. To obtain an optimal convolution kernel, Beatty et al. found in the field of magnetic resonance imaging that the Fourier transform of the Kaiser-Bessel function c_n in Eq. (5) should have its central lobe exceeding the passband, i.e. the image content, and reaching the edge of the first stopband to avoid spurious aliasing back into the image. A narrower central lobe leads to increased error through large roll-off correction, and a wider central lobe leads to increasing aliasing amplitude in the passband. A good approach is to allow the central lobe to enter into the stopband with an amplitude below the first sidelobe level. These considerations yield the closed-form expression for β in Eq. (4).

Figure 1 shows the interplay between the passband and stopband using c_n in Eq. 5, for the case of $M = 5$ with an oversampling factor $\alpha = 2$ and 1.2. The effect of non-optimal (arbitrary) and optimal (using Eq. 4) values of β is illustrated. It is noticed that the passband extends on both negative and positive sides up to $N/2$, and so do the stopbands of width N and centered on pN' , with the integer $p = \dots, -2, -1, 1, 2, \dots$. Therefore in each graph, the first vertical line is the edge of the passband and the second line is the beginning of the first stopband, the horizontal line indicating the function amplitude at this point. The transition band between the two vertical lines follows from the oversampling and represents a zone which does not contribute to the aliasing. For $\alpha = 2$, the optimal β value of 11.44 (Fig. 1b) reduces the first sidelobe amplitude in the stopband by about 20 dB from that obtained with a non-optimal β value of 9.0 (Fig. 1a). For $\alpha = 1.2$, the transition band is narrower and using a somewhat similar β value of 11 (Fig. 1c) leads to a central lobe extending in the stopband with an aliasing amplitude at a -40 dB level at the edge. Note that this aliasing amplitude is further increased approximately to a -20 dB level after the roll-off correction related to the amplitude of the central lobe at the edge of the passband. An optimal β value of 8.72 (Fig. 1d) provides a narrower central lobe with a -70 dB sidelobe level at the stopband edge, corresponding to a -50 dB level after roll-off correction. This is suitable if the dynamic range of the OCT is 50 dB or less for large optical depths.

[Fig. 1 about here.]

3. Simulations

To evaluate the efficiency of the different signal processing methods, spectral interferograms having different non-linearities were simulated. For this purpose, simulations were designed and carried out considering a single reflector as the sample, for different optical path delays (OPD). The interferometric signal is then expressed as:

$$S(t) \propto S_0(t) \cos \left[\frac{2\pi}{c} \nu(t) \cdot 2\delta z \right], \quad (6)$$

where $S_0(t)$ is the simulated spectrum, t is the time, c is the speed of light, $\nu(t)$ is the optical frequency which is linked to the wavenumber by $\nu(t) = \frac{c}{2\pi} \cdot k(t)$, and δz is the OPD.

In FD-OCT, the non-linearity of the optical frequency can be described as:

$$\nu(t) = \frac{c}{\lambda_0 + \frac{\Delta\lambda}{2}} + \sum_i^n a_i \cdot t^i, \quad (7)$$

where λ_0 is the center wavelength, $\Delta\lambda$ is the wavelength swept bandwidth, a_i are the non-linear coefficients. The non-linearity as expressed in Eq. (7) reflects the fact the SS-OCT data is experimentally recorded as a function of time, as the wavelength or frequency of the source is swept over time. SD-OCT data is typically not recorded as a function of time, since the spectral components are usually recorded simultaneously. To insure uniformity in the way the data is presented below, the SD-OCT non-linearity is nevertheless also characterized with a time-dependent frequency sweep as in Eq. (7). In fact, this corresponds to an equivalent SS-OCT system that would provide the same non-linearity as the SD-OCT case, which is essentially identical from a signal processing point of view.

In this paper, four sets of parameters corresponding to different types of non-linearities have been simulated. For all cases, S_0 is a Gaussian spectrum with a random additive noise whose standard deviation is equal to 1 % of the maximum amplitude of the spectrum. Here $\lambda_0 = 1300$ nm, $\Delta\lambda = 100$ nm, the full width at half maximum of the Gaussian spectrum is 40 nm, and all the simulations are made with a ~ 30 kHz A-scan rate. In the SS-OCT cases, a 50 MS/s sampling frequency was simulated meaning that the spectral interferograms have around 1500 points. All the non-linearities are similar to those exhibited by commercial source technologies. In fact, they are based on experimental measurements performed with various swept-sources. The non-linearities were inferred from interferograms recorded for a reflector at a known depth, typically 1.5 mm. The manufacturers of the sources are voluntarily not disclosed, since the goal is not to compare sources but to evaluate the efficiency of our proposed method to handle the nonlinearity for various swept-source technologies. In the SD-OCT case, a 1024-pixel spectrometer was simulated and considered linear in wavelength, and thus nonlinear in wavenumber. As mentioned above, the nonlinearity is converted to an equivalent SS-OCT nonlinear sweep by converting pixel number to time. The different cases are described below.

- Low non-linearity swept-source. This case corresponds to a swept-source based on a grating used in conjunction with a rotating mirror. This non-linearity is described by a 4th order polynomial:

$$\sum_i^n a_i \cdot t^i = 4.33 \cdot 10^5 \cdot t + 5.86 \cdot 10^9 \cdot t^2 - 7.17 \cdot 10^{13} \cdot t^3 - 7.09 \cdot 10^{16} \cdot t^4.$$

- Medium non-linearity swept-source. This case corresponds to a swept-source based on MEMS scanner technology. In this case a 10th polynomial is more adapted to represent the non-linearity:

$$\sum_i^n a_i \cdot t^i = 6.84 \cdot 10^4 \cdot t + 2.11 \cdot 10^{11} \cdot t^2 - 4.93 \cdot 10^{16} \cdot t^3 + 8.01 \cdot 10^{21} \cdot t^4 - 8.25 \cdot 10^{26} \cdot t^5 + 5.25 \cdot 10^{31} \cdot t^6 - 2.07 \cdot 10^{36} \cdot t^7 + 4.95 \cdot 10^{40} \cdot t^8 - 6.58 \cdot 10^{44} \cdot t^9 + 3.75 \cdot 10^{48} \cdot t^{10}.$$

- Galvanometer-based swept-source. This case corresponds to a swept-source based on a galvanometer-mounted grating. This non-linearity is described by a 4th order polynomial:

$$\sum_i^n a_i \cdot t^i = 2.12 \cdot 10^4 \cdot t + 5.97 \cdot 10^{10} \cdot t^2 - 1.40 \cdot 10^{15} \cdot t^3 + 1.25 \cdot 10^{18} \cdot t^4.$$

- Spectrometer-based non-linearity. This case corresponds to SD-OCT. This non-linearity is described by a 2nd order polynomial:

$$\sum_i^n a_i \cdot t^i = 8.02 \cdot 10^5 \cdot t + 3.09 \cdot 10^9 \cdot t^2.$$

In all the above expressions, $\sum_i^n a_i \cdot t^i$ is expressed in THz and t is the time in second, leading to the following expression of ν in THz: $\nu(t) = \frac{3 \cdot 10^8}{1300 + \frac{100}{2}} \cdot 10^{-12} + \sum_i^n a_i \cdot t^i = 222.22 + \sum_i^n a_i \cdot t^i$.

4. Point spread function based comparison: error evaluation

To better evaluate the efficiency of each method, we define an error metric. Taking into account that the Vandermonde method is the most exact method, it is considered as our reference method for the imaging quality. As already mentioned, the methods considered are the linear interpolation and the cubic spline interpolation with an oversampling factor α followed by FFT (respectively LIFFT $_{\alpha}$ and SIFFT $_{\alpha}$), and the convolution by an optimized Kaiser-Bessel window of length M with an oversampling factor α followed by FFT (KBFFT $_{M,\alpha}$). Simulated spectral interferogram data were processed using these methods to provide point spread functions (PSFs) for different reflector depths. For each simulation, the error is computed by comparing the PSF obtained with a given method for a given reflector depth with that of the Vandermonde method as:

$$\text{error} = \text{PSF}_{\text{method}, \text{Xmm}} - \text{PSF}_{\text{VDM}, \text{Xmm}} \quad (8)$$

where the error and the PSF are expressed in dB, VDM means Vandermonde method, Xmm is the reflector depth of the considered PSF, and method stands for LIFFT $_{\alpha}$, SIFFT $_{\alpha}$ or KBFFT $_{M,\alpha}$. For each reflector depth location, the error is evaluated as a function of optical

depth between 1 mm and 7 mm. The error defined in Eq. (8) is very sensitive to variations in noise level and to variations in the shape and position of a PSF. In the latter case, any PSF displacement or PSF widening will lead to a sizeable departure from the Vandermonde result, and thus to a large error.

The averaged error $\bar{\epsilon}$ and the standard deviation σ_{ϵ} are computed. The value $\bar{\epsilon}$ gives a global estimation of the error while σ_{ϵ} gives an estimation of the fluctuations of the error over the whole A-scan. The error calculation is illustrated Fig. 2.

[Fig. 2 about here.]

Error evaluations from the PSFs obtained with simulated spectral interferograms for each non-linearity case are presented in Fig. 3.

[Fig. 3 about here.]

The first row presents the different simulated non-linearities. Figure 3(a) shows the low non-linearity, Fig. 3(b) shows the medium non-linearity, Fig. 3(c) shows the galvanometer type non-linearity which has high non-linearity on the edges and smooth non-linearity in the center, and Fig. 3(d) shows the very smooth spectrometer-based (SD-OCT) non-linearity.

The second row presents the LIFFT $_{\alpha}$ method where $\alpha = 1, 2$ and 3. This method does not provide an error-free PSF whatever the oversampling factor and this impacts the image quality. For all kinds of non-linearities, the averaged error and the standard deviation are too large to provide good imaging results.

The third row presents the SIFFT $_{\alpha}$ method where $\alpha = 1.0, 1.5$, and 2.0. It should be noticed that a fractional oversampling factor α was considered for this method to better compare with the KBFFT $_{M,\alpha}$ method. Using this method, $\alpha = 1.0$ provides significant error for depths larger than 2 mm for every kind of non-linearities. Consequently, this method could only be acceptable for low-depth imaging. This is especially relevant for SD-OCT where depths greater than 2-3 mm are experimentally rarely reached up to now. SIFFT $_{\alpha}$ for $\alpha \geq 1.5$ provides an average error close to zero for all PSFs with moderate error fluctuations thus leading to a good quality imaging. As mentioned in Section 2, for this method an empirical roll-off correction has to be made to not underestimate the amplitude at larger depths. This correction is empirical and, consequently, it does not always provide good correction as can be seen in Figs. 3 (j) and (k) where error level is sometimes below 0 dB.

Finally, the fourth row presents the KBFFT $_{M,\alpha}$ method where $\alpha = 1.0, 1.2$ and 1.5. For M , we tried values between 3 and 8 and we found that 5 was the best choice. KBFFT $_{5,1.0}$ leads to large error for all PSFs and, therefore, it should not be used. However KBFFT $_{5,\alpha}$ with $\alpha \geq 1.2$ provides nearly error-free PSFs with very small error fluctuations.

Therefore, among the considered methods, error-free PSFs leading to good image quality are obtained for SIFFT $_{\alpha}$ for $\alpha \geq 1.5$ and KBFFT $_{5,\alpha}$ with $\alpha \geq 1.2$. Methods of choice should

have the smallest possible oversampling factor α to provide smaller matrices to process and thus to ask for shorter processing time. The processing times of each method are compared in the next section.

5. Processing time

Regarding processing time, our implementation of the Vandermonde matrix approach is not fully optimized, but it is not expected that further optimization would reduce the computational time below that of the other methods. The LIFFT $_{\alpha}$ and SIFFT $_{\alpha}$ methods are optimized with a pre-processing of the weighting factors and roll-off correction. However, the SIFFT $_{\alpha}$ method further requires the calculation of the second derivative of each OCT signal which makes the resampling more time consuming. Also, for this method, the roll-off correction can only be determined empirically. This is made once by comparing the amplitude roll-off as a function of optical path with that of the Vandermonde matrix method. The comparison provides a depth-dependent correction factor that can be approximated by a polynomial. Regarding the KBFFT $_{M,\alpha}$, the coefficients C_{kj} and c_n in Eqs. (3) and (5) are calculated once for processing all OCT signals. Also in all methods, $N' = \alpha N$ is the number of points involved in the FFT operation, and our FFT algorithm only requires that N' is the product of small prime numbers, possibly after padding the vector with a few zeroes. Our implementation relies on the algorithm FFTPack [7].

We have shown in the previous section that KBFFT $_{5,1,2}$ provide nearly error-free PSFs. Figure 4 presents the ratio of processing time of each method to that of the KBFFT $_{5,1,2}$ method. Notice that the pre-processing time of each method is not considered since it is negligible for a large number of A-scans (typically more than 500).

[Fig. 4 about here.]

Based on the results from the previous section, the methods that provide error-free PSFs are depicted in green, the other methods in red. One exception is SIFFT $_{1,0}$ which is depicted in pink since it can be considered a good method for low depth imaging.

Only LIFFT $_1$ and KBFFT $_{5,1,0}$ are faster methods than KBFFT $_{5,1,2}$, but these methods provide large averaged error and error fluctuations. All other methods are slower than our proposed method including SIFFT method which is widely used in literature. It means that the KBFFT method is the optimal choice considering the trade-off between imaging quality and processing time.

6. Imaging

In our previous study, we have experimentally shown using OCT imaging with one kind of non-linearity in SS-OCT that KBFFT $_{5,1,2}$ was the optimal method to process the data [4].

Here we apply this processing to a SD-OCT setup where experimental data were acquired *in vivo* from the healthy human retina, and the data are processed with the different methods mentioned above. The human retina images were acquired with a SD-OCT system, the design of which was discussed in detail in previous publications [8,9]. Briefly, the SD-OCT system is based on fiber optic Michelson interferometer interfaced to a superluminescent diode (SLD) based light source (Superlum). To counteract to negative effect of water absorption on the OCT axial resolution in the retina, the spectral output of the SLD was custom modified for high power emission at 980 nm, where the water absorption has a local maximum [9]. The OCT system was integrated into a slitlamp biomicroscope. The interference signal was measured with a high performance spectrometer (P&P Optica), outfitted with a 1024 pixel InGaAs camera (SUI, Goodrich Corp.) with 92 kHz readout rate. The SLD spectrum was centered at ~ 1020 nm and its base ranged from 940 nm to 1100 nm, corresponding to spectral resolution of ~ 5 μm axial. The lateral resolution was ~ 15 μm in the human retina. The signal to noise ratio was ~ 95 dB at 1.5 mW power of the imaging beam incident on the cornea. OCT images (1000×512) were acquired from the foveal region of the human retina (shown in Fig. 6).

Before evaluating the error on the PSFs, the roll-off correction for the SIFFT method should be made whereas it is directly embedded in the other methods. This is made once by comparing the amplitude roll-off as a function of optical path with that of the reference method using the Vandermonde matrix, providing a vector c_n as a polynomial fit. Here, it is a 4th order polynomial given by: $c_n = 1.01 - 4.87 \cdot 10^{-2} \cdot \delta z + 9.44 \cdot 10^{-2} \cdot \delta z^2 - 4.60 \cdot 10^{-2} \cdot \delta z^3 - 1.42 \cdot 10^{-2} \cdot \delta z^4$.

Error evaluation on the measured PSFs is presented in Fig. 5.

[Fig. 5 about here.]

The non-linearity is shown Fig. 5(a) and could be described as a 2nd order polynomial in THz: $\nu(t) = 271.3 + 3.84 \cdot 10^6 \cdot t + 1.23 \cdot 10^{11} \cdot t^2$. As in Section 3, the non-linearity is provided as a function of time, which corresponds to an equivalent SS-OCT system with the same non-linearity. Overall, Figures 5(b-d) show that over a small depth range (~ 1 mm), the averaged error is small (< 1 dB) for all methods and the fluctuations of the error are relatively small except for LIFFT₁ method where fluctuation errors are greater than 3 dB. It means that whatever the method used in that specific experimental configuration it will provide a good imaging quality. Nevertheless, the KBFFT_{5,1.2} method exhibits the best results with an error close to 0 dB and an error fluctuation < 0.2 dB for $\alpha \geq 1.2$. As it is also the fastest among the accurate methods, it is thus the method of choice to process SD-OCT data.

An example of retina imaging is presented in Fig. 6 using the same set of data processed with the methods SIFFT_{1,0} and KBFFT_{5,1.2}.

[Fig. 6 about here.]

The method $\text{SIFFT}_{1.0}$ is chosen for the comparison since it is often used in the literature. Both methods provide good imaging quality. Few differences can be observed from the comparison of both images over such a small depth range. A slight increase of the noise level in the upper part of the image can be observed for the $\text{SIFFT}_{1.0}$ method; this region is located far from the zero-delay position that is at the bottom of the image. This increase is due to an improper roll-off correction highlighting once again the difficulty of providing a good implementation with the SIFFT method. More important, $\text{KBFFT}_{5,1.2}$ is nearly twice faster than $\text{SIFFT}_{1.0}$. Also, the roll-off correction is directly embedded in the processing using KBFFT. Therefore, $\text{KBFFT}_{5,1.2}$ is not only faster than $\text{SIFFT}_{1.0}$ but also simpler to implement.

7. Conclusion

We expanded our previous study to typical non-linearities that one can encounter in FD-OCT. A comparison of various processing methods to resample data having different kinds of non-linearity in FD-OCT was performed. The different methods used are the discrete Fourier transform for unequally spaced data which involves Vandermonde matrices, resampling the data with linear interpolation or spline interpolation prior to FFT, and resampling the data through convolution with a Kaiser-Bessel function followed by FFT.

The simulations were made using a low non-linearity, a medium non-linearity, a galvanometer-based non-linearity, and spectrometer-based non-linearity. It was shown that resampling the data through convolution using an optimized Kaiser-Bessel function was the optimal method offering the best trade-off between imaging quality and processing time. Indeed, $\text{KBFFT}_{5,1.2}$ provides a nearly error-free resampling with a processing time 1.8 times faster than the $\text{SIFFT}_{1.0}$ widely used in the literature. Moreover, its implementation is simpler since the roll-off correction is directly embedded whereas an empirical roll-off correction is required with the SIFFT method.

Regarding the experimental result in SD-OCT, the small depth imaging implies that no big difference exists between $\text{SIFFT}_{1.0}$ and $\text{KBFFT}_{5,1.2}$ methods in imaging quality. However, $\text{KBFFT}_{5,1.2}$ remains the method of choice due to its faster execution and its simpler implementation.

Acknowledgments

The authors would like to thank S. Hariri and E. Song for assistance with the acquisition and processing of the human retina data. This research was supported by the Genomics and Health Initiative of the National Research Council Canada and the National Sciences and Engineering Research Council of Canada.

References

1. A. F. Fercher, C. K. Hitzenberger, G. Kamp, and S. Y. El Zaiat, "Measurement of intraocular distances by backscattering spectral interferometry," *Optics Communications* **117**, 43–48 (1995).
2. M. A. Choma, K. Hsu, and J. A. Izatt, "Swept source optical coherence tomography using an all-fiber 1300-nm ring laser source," *Journal of Biomedical Optics* **10**, 044009 (2005).
3. Z. Hu and A. M. Rollins, "Fourier domain optical coherence tomography with a linear-in-wavenumber spectrometer," *Opt. Lett.* **32**, 3525–3527 (2007).
4. S. Vergnole, D. Lévesque, and G. Lamouche, "Experimental validation of an optimized signal processing method to handle non-linearity in swept-source optical coherence tomography," *Opt. Express* **18**, 10446–10461 (2010).
5. D. Hillmann, G. Huttman, and P. Koch, "Using nonequispaced fast fourier transformation to process optical coherence tomography signals," in "Optical Coherence Tomography and Coherence Techniques IV," , *Proc. SPIE* **7372**, 73720R (2009). .
6. P. J. Beatty, D. G. Nishimura, and J. M. Pauly, "Rapid gridding reconstruction with a minimal oversampling ratio," *IEEE Transactions on Medical Imaging* **24**, 799–808 (2005).
7. P. N. Swartztrauber, "Vectorizing the FFTs," in "Parallel Computations," G. Rodrigue, ed. (Academic Press, New York, NY, 1982), pp. 51–83.
8. P. Puvanathan, P. Forbes, Z. Ren, D. Malchow, S. Boyd, and K. Bizheva, "High-speed, high-resolution fourier-domain optical coherence tomography system for retinal imaging in the 1060 nm wavelength region," *Opt. Lett.* **33**, 2479–2481 (2008).
9. S. Hariri, A. A. Moayed, A. Dracopolos, C. Hyun, S. Boyd, and K. Bizheva, "Limiting factors to the oct axial resolution for in-vivo imaging of human and rodent retina in the 1060nm wavelength range," *Opt. Express* **17**, 24304–24316 (2009).

List of Figures

1	(Color online) Plot of the Fourier transform of the Kaiser-Bessel window (c_n in Eq. (5)) with non-optimal (a,c) and optimal (b,d) values of β , for the case $M = 5$ and $\alpha = 2$ (a,b) and $\alpha = 1.2$ (c,d). PB denotes the passband zone and SB is the first stopband zone	13
2	(Color online) Example of an error calculation: (a) PSF at 3 mm processed with the Vandermonde method (green plot) and the LIFFT ₁ method (red plot), (b) error evaluated from the subtraction of the two PSFs over the whole A-scan.	14
3	(Color online) Simulated non-linearities (1 st row (a-d)). Error evaluated from comparison to the Vandermonde method: LIFFT _{α} processing (2 nd row (e-h)), SIFFT _{α} processing (3 rd row (i-l)). KBFFT _{M,α} processing (4 th row (m-p)). Error bars correspond to $2\sigma_{error}$. Pink area in SD-OCT correspond to depths which are, up to now, rarely reached with SD-OCT systems.	15
4	(Color online) Processing time ratio evaluated by comparison to our optimized proposed method (KBFFT _{5,1.2}). Based on the previous analysis of Fig. 3, green color identifies good quality imaging methods, while red color identifies bad quality imaging methods. Pink color identifies a method which is efficient for low-depth imaging.	16
5	(a) Experimental non-linearity of the spectrometer used in our setup. Error evaluation for LIFFT (b), SIFFT (c) and KBFFT (d) methods.	17
6	Example of imaging in SD-OCT processed with (a) SIFFT _{1,0} and (b) KBFFT _{5,1.2} . Tic marks are 0.5 mm spaced in X-direction and 0.2 mm spaced in Y-direction. Gray-scale is log. scale. The dynamic range is 30 dB.	18

Fig. 1. (Color online) Plot of the Fourier transform of the Kaiser-Bessel window (c_n in Eq. (5)) with non-optimal (a,c) and optimal (b,d) values of β , for the case $M = 5$ and $\alpha = 2$ (a,b) and $\alpha = 1.2$ (c,d). PB denotes the passband zone and SB is the first stopband zone

Fig. 2. (Color online) Example of an error calculation: (a) PSF at 3 mm processed with the Vandermonde method (green plot) and the LIFFT₁ method (red plot), (b) error evaluated from the subtraction of the two PSFs over the whole A-scan.

Fig. 3. (Color online) Simulated non-linearities (1st row (a-d)). Error evaluated from comparison to the Vandermonde method: LIFFT _{α} processing (2nd row (e-h)), SIFFT _{α} processing (3rd row (i-l)). KBFFT _{M,α} processing (4th row (m-p)). Error bars correspond to $2\sigma_{error}$. Pink area in SD-OCT correspond to depths which are, up to now, rarely reached with SD-OCT systems.

Fig. 4. (Color online) Processing time ratio evaluated by comparison to our optimized proposed method ($\text{KBFFT}_{5,1.2}$). Based on the previous analysis of Fig. 3, green color identifies good quality imaging methods, while red color identifies bad quality imaging methods. Pink color identifies a method which is efficient for low-depth imaging.

Fig. 5. (a) Experimental non-linearity of the spectrometer used in our setup. Error evaluation for LIFFT (b), SIFFT (c) and KBFFT (d) methods.

Fig. 6. Example of imaging in SD-OCT processed with (a) $\text{SIFFT}_{1.0}$ and (b) $\text{KBFFT}_{5,1,2}$. Tic marks are 0.5 mm spaced in X-direction and 0.2 mm spaced in Y-direction. Gray-scale is log. scale. The dynamic range is 30 dB.