Performance comparison between 8- and 14-bit-depth imaging in polarization-sensitive swept-source optical coherence tomography

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Abstract: Recently the effects of reduced bit-depth acquisition on swept-source optical coherence tomography (SS-OCT) image quality have been evaluated by using simulations and empirical studies, showing that image acquisition at 8-bit depth allows high system sensitivity with only a minimal drop in the signal-to-noise ratio compared to higher bit-depth systems. However, in these studies the 8-bit data is actually 12- or 14-bit ADC data numerically truncated to 8 bits. In practice, a native 8-bit ADC could actually possess a true bit resolution lower than this due to the electronic jitter in the converter etc. We compare true 8- and 14-bit-depth imaging of SS-OCT and polarization-sensitive SS-OCT (PS-SS-OCT) by using two hardware-synchronized high-speed data acquisition (DAQ) boards. The two DAQ boards read exactly the same imaging data for comparison. The measured system sensitivity at 8-bit depth is comparable to that for 14-bit acquisition when using the more sensitive of the available full analog input voltage ranges of the ADC. Ex-vivo structural and birefringence images of equine tendon indicate no significant differences between images acquired by the two DAQ boards suggesting that 8-bit DAQ boards can be employed to increase imaging speeds and reduce storage in clinical SS-OCT/PS-SS-OCT systems. One possible disadvantage is a reduced imaging dynamic range which can manifest itself as an increase in image artifacts due to strong Fresnel reflection.

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1. Introduction

High bit-depth data acquisition (DAQ) boards, typically 12, 14 or 16, are commonly used for imaging through scattering tissue with high dynamic range in swept-source optical coherence tomography (SS-OCT) [1]. High imaging speed, typically ~50-400 kHz A-scan rate [2–6], is required to acquire entire three-dimensional sets providing a greatly improved flexibility in image data analysis and visualization. However, the imaging speed of SS-OCT systems used in biological laboratory and clinical medicine is currently limited by digital acquisition with the development of high-speed swept sources [7,8]. For example, Wang et al. [9] recently demonstrated a high-speed wavelength-swept laser with a tuning range of 104 nm and a repetition rate of 403 kHz.

One solution to this limit is to employ lower bit-depth, e.g., 8-bit depth DAQ, which can be used to lower instrument cost and/or increase acquisition speed. This also can be used to reduce data volume by reducing overall bandwidth [4,10]. Yasuno et al. [11] have used 8-bit analog-to-digital converters (ADCs) for faster data acquisition and total cost reduction and
achieved a dynamic range of 52dB by using the phase-shifting method in the Fourier domain OCT. Huber et al. [4] also used an 8-bit, 5 GS/s digital oscilloscope to achieve a high imaging speed of 370,000 A-scans per second in their SS-OCT system and achieved a 37dB image contrast in the 8-bit image. They have also compared the 8 and 14 bit-depth images by using a second 14-bit digitizer (200MS/s, Gage Applied Technologies, US) and showed that the 8-bit image was comparable with the 14-bit image, however the comparison was made without a quantitative analysis.

A recent report [10] shows that image acquisition at 8-bit depth allows high system sensitivity with only a minimal drop of 0.6dB in the signal-to-noise ratio (SNR) compared to higher bit-depth systems by simulations and experimental measurements, where data was digitized natively at 14 bits and then the data was numerically rounded to generate data sets at reduced bit-depths. Wieser et al. [12] reported a similar result by artificially bit-reducing the data sampled with a 12-bit ADC and further pointed out that significant image degradation occurs at an effective number of bits smaller than 7 and real world 8-bit ADCs are suitable for OCT imaging. However, this artificial bit-reduced transformation in post-processing ignored the difference in DAQ noise and quantization noise among true DAQ boards at different bit-depths. DAQ noise is due to electronic noise within the DAQ circuitry and quantization noise to distortions caused by the finite bit-depth of the board. Hence these noise sources may degrade the system sensitivity/dynamic range at some full analog input voltage ranges (FIVR) of the ADC.

In this paper, both 8- and 14-bit DAQ boards were used in a polarization-sensitive SS-OCT (PS-SS-OCT) system at 1.3μm wavelength to compare the system performance including sensitivity, OCT structural and birefringence image quality. Images from equine tendon and human finger skin are presented.

2. Experiments

The PS-SS-OCT data acquisition system used is adapted from a Michelson Diagnostics OCT microscope EX1301. The PS-SS-OCT system follows the scheme reported by Yamanari et al. [8,13–15]. The system is shown in Fig. 1 and described here briefly. The source used is a wavelength-swept laser (HSL-2000-10-MDL, Santec, Japan) with a centre wavelength of 1315nm, a wavelength range of 157nm, an FWHM of 128nm, a wavelength scanning rate of 10kHz, a duty cycle of about 60%, and an average output power of 10mW. This leads to an axial resolution of about 10μm in air.

![Fig. 1. Schematic diagram of the system. SS: wavelength-swept source, PC: polarization controller, LP: linear polarizer, EOM: electro-optic modulator, FC: fiber-optic circulator, BS: non-polarizing beamsplitter, PBS: polarization beamsplitter, H and V: balanced photo-detectors for horizontally and vertically polarized optical signals, respectively.](image)

The light is polarized by a linear polarizer and then modulated continuously by an electro-optic modulator (PC-B3-00-SFAP-SFA-130-UL, EOsapce, US) operating at 6.67MHz. The
modulated laser light is split into the reference and sample arm and recombined and detected by two balanced photoreceivers (1817-FC, New Focus, US). The minimum noise-equivalent power (NEP) of these detectors is quoted to be 2.5pW/√Hz from DC to 10MHz. The detected horizontally and vertically polarized optical signals are low-pass filtered from DC-8MHz and divided into two parts electrically. The two signals are sampled at 20MS/s simultaneously with 14-bit (M2i.4022, Spectrum GmbH, Germany) and 8-bit (M2i.2031, Spectrum GmbH, Germany) resolution. The input impedance is set at 1MΩ for both ADCs. The two hardware-synchronized high-speed DAQ boards read exactly the same imaging data for comparison. Therefore, the comparison is not related to the stability of the sample used. However, the use of temporally stable samples could provide more clear OCT images. The theoretical description and the date processing procedure of the PS-SS-OCT system have already been previously described [13,16]. In brief, the EOM shifts a portion of the signal to the frequency of the polarization modulation. The modulated and non-modulated signals are numerically demultiplexed. From the two photoreceivers in the PS detection section in Fig. 1, four complex OCT signals are obtained by a single A-scan. The depth-resolved Jones matrices are algebraically calculated from these signals [13],

\[
J_{\text{measured}} = \begin{pmatrix}
-\left(\tilde{I}_x^* + \frac{\tilde{I}_{x1}}{J_1(A_0)}\right) & \tilde{I}_x^* - \frac{\tilde{I}_{x1}}{J_1(A_0)} \\
-\left(\tilde{I}_v^* + \frac{\tilde{I}_{v1}}{J_1(A_0)}\right) & \tilde{I}_v^* - \frac{\tilde{I}_{v1}}{J_1(A_0)}
\end{pmatrix},
\]

where \(\tilde{I}_{x1}, \tilde{I}_x^*, \tilde{I}_{v1}, \tilde{I}_v^*\) shows the complex conjugate of the horizontally polarized non-modulated, first-order, vertically polarized non-modulated, first-order OCT signals, respectively, and \(J_1(A_0)\) is the first-order Bessel function of the first kind evaluated at \(A_0 = 2.405\) radians. \(J_{\text{measured}}\) is the overall measured sample Jones matrix which is determined both by the true sample Jones matrix, \(J_{\text{sample}}\), and the single-mode fiber components, i.e.

\[
J_{\text{measured}} = J_{\text{out}} \cdot J_{\text{sample}} \cdot J_{\text{in}},
\]

where \(J_{\text{in}}\) is a Jones matrix to characterize the section from the EOM to the sample surface, \(J_{\text{sample}}\) is a depth-dependent double-pass Jones matrix of the sample, and \(J_{\text{out}}\) is a Jones matrix from the sample surface to the non-polarizing beamsplitter in Fig. 1.

In order to compensate the fiber-induced birefringence in the sample arm fiber, the Jones matrix at the sample surface is used as a reference matrix to calculate the birefringence in the sample. The Jones matrix at the surface of the sample can be expressed as

\[
J_{\text{surface}} = J_{\text{out}} \cdot J_{\text{in}}.
\]

Therefore, the double-pass phase retardance \(\eta\), fast axis orientation, \(\theta\) and diattenuation \(D\) of the sample can be obtained from the matrix diagonalization of the following equation [17,18],

\[
J_{c,m} = J_{\text{measured}} \cdot J_{\text{in}}^{-1} = J_{\text{out}} \cdot J_{\text{sample}} \cdot J_{\text{out}}^{-1} = J_U \begin{pmatrix}
p_1 e^{i \eta/2} & 0 \\
0 & p_2 e^{-i \eta/2}
\end{pmatrix} \cdot J_U^{-1},
\]

where \(p_1, p_2\) are two transmittances of the eigenpolarizations of the sample, and \(J_U\) is a general unitary matrix, whose columns are the fast and slow eigenpolarizations of \(J_{c,m}\). \(\theta\) is extracted from these eigenpolarizations. The degree of the phase retardance can be extracted through the phase difference of the resulting diagonal elements, and the diattenuation \(D\) from their magnitudes, i.e. \(D = (p_1^2 - p_2^2)/(p_1^2 + p_2^2)\). Polarization-insensitive OCT intensity is determined
as the summation of horizontally and vertically detected zeroth-order OCT intensities, i.e. 
\[ 10 \cdot \log_{10}(|\hat{I}_{h0}|^2 + |\hat{I}_{v0}|^2). \]

3. Results

3.1 Noise measurements

The DAQ noise of the two DAQ boards were measured and compared initially. The DAQ noise was measured by terminating the ADC input at 50 ohms and digitizing the signal with four set FIVRs of ± 0.2V, ± 1V, ± 2V and ± 5V. These FIVRs were selected as they were available for both ADCs. The 50 ohms termination was used instead of short-circuit to match the 50-ohm output impedance of the balanced photoreceivers used in our system. The recorded data is then converted using the FFT technique for spectral analysis. Figure 2 shows the measured noise power spectra in dB using a reference level of 1 volt-squared per Hz.

![Fig. 2. DAQ noise measurements at different set FIVR for 14-bit DAQ (left) and example ADC counts at the set FIVR of ± 0.2V for 8-bit DAQ (right), respectively.](image)

It can be clearly seen from Fig. 2 that the measured DAQ noise for the 14-bit DAQ in Fig. 2(left) increases proportionally as the FIVR increases. This is in agreement with the fact [10] that DAQ noise increases proportionally as the maximum voltage range increases since it is due to electronic noise within the DAQ circuitry. However, the 8-bit DAQ shows only a few ADC spike counts at these set FIVRs. An example at the set FIVR of ± 0.2V is shown in Fig. 2 (right). This is because the 8-bit DAQ noise is limited to one quantization level, which is much larger than 14-bit at the same set FIVR. Therefore, the FFT noise analysis at this sampling rate and sample length (1024) could not be applied.

The receiver noise was then measured by digitizing the signal from the detector using the two DAQs while the optical signals were blocked. The results are shown in Fig. 3. It can be seen that the measured receiver noise is approximately the same for different FIVRs for both DAQs, suggesting that the receiver noise is dominant in the noise sources. The receiver noise measured using the 14bit is ~20dB greater than the measured DAQ noise for the set FIVR of ± 0.2V while it is close for the set FIVR of ± 5V as shown in Fig. 2. The receiver noise could not be measured using the 8bit at higher set FIVRs (> ± 2V) since the receiver noise was limited to one quantization level of the board. Figure 3(c) shows the noise standard deviation of the measured DAQ plus receiver noise by using both boards. It can be seen clearly that the measured noise standard deviation is between 1.2mV and 3mV by the 14-bit at different set FIVR but it increases proportionally with the set FIVR of the 8-bit ADC, which suggests that the quantization noise of the 8-bit ADC is dominant in the noise sources when the set FIVR > ± 2V. Quantization noise is induced through distortion in converting an analog signal to a digital signal with a finite bit-depth of the DAQ board. A formal analysis of the properties of quantization noise show that the equivalent noise standard deviation is given by [10]

\[ \sigma_{eq} = \frac{V_{\text{max}}}{\sqrt{32^p}}, \]
where $V_{\text{max}}$ is the maximum measurable voltage signal and $b$ the number of bits in the DAQ. The quantization noise levels of the two DAQ boards are calculated by using this noise model and are shown in Fig. 3c. The standard deviation of the measured DAQ plus receiver noise is larger than the values calculated from Eq. (5) for the 8-bit ADC, although the variation trend versus the set FIVR is the same. This is because the amplitude of the DAQ plus receiver noise signals crosses only a few (2 or 3) quantization levels. The noise model in Eq. (5) breaks down and quantization noise increases significantly for small signals when the amplitude of the signal does not cross several quantization levels [10].

Figure 3c also shows that the standard deviation of the measured DAQ plus receiver noise is larger than the values calculated from Eq. (5) for the 14-bit ADC. This is not due to the breakdown of the noise model since the amplitude of the DAQ plus receiver noise signals crosses >10 quantization levels at these FIVRs. This is because the DAQ plus receiver noise rather than quantization noise is dominant in the noise sources in the measurement.

3.2 System sensitivity

System sensitivity specifies the highest possible attenuation in the sample arm, i.e. smallest possible back reflection, which can be detected. The system sensitivity analysis below uses only one of the polarized light channels. The OCT structural image analysis uses one channel with the polarization modulator disabled. The OCT birefringence image analysis uses both polarization channels with the polarization modulator enabled. Typically the detection threshold is set where the SNR reaches 1. For good image quality in biomedical applications the sensitivity should be $\sim 95$dB [12]. Therefore, the system sensitivity, characterized as SNR, was measured and compared at all available FIVRs for both DAQ boards in our system. An optical mirror was placed near the zero depth in the sample arm as a test target. Two neutral density filters were used to control the sample-illuminating light power. The reference power was adjusted to maximize the SNR. The average noise floor was calculated in a region where there was no signal component.

Figure 4 shows the measurement result at a reference light power of $\sim 0.3$mW and a probing light power of $\sim 3$mW to the sample. The maximum change of measured sensitivities...
versus available FIVRs was 10dB and 27dB for the 8- and 14-bit boards respectively. It can be clearly seen that the sensitivity at 8-bit depth is close to that at 14-bit at FIVRs of 0.1 to 1V with a drop of ~3dB, but it reduces faster when the FIVR was larger than 1V. This indicates that image quality is fully comparable between the two boards at the FIVRs of 0.1 to 1V, but the imaging dynamic range is reduced at the FIVRs of >1V compared to the 14-bit depth. Therefore, as a consequence of using 8-bit ADCs, the FIVR, i.e. the preamplifier gain of the ADC, has to be set more carefully than with 14-bit in order to use the full available dynamic range.

The measured maximum sensitivity was 106.6dB for both DAQ boards at a reference power of ~0.3mW, which agrees with the calculated system sensitivity of ~107dB in the system when shot-noise, detector-noise, relative intensity noise (RIN) and ADC quantization-noise were taken into account. The shot-noise-limited sensitivity [19] was calculated to be 113.9dB for the system. The difference of 7.3dB between the shot-noise-limited and experimental values can be attributed to imperfect balanced detection, because the beamsplitter in the PS detection section and fiber-optic components in Fig. 1 lead to a wavelength-dependent split ratio between the balanced detector channels that differs significantly from the ideal value of 50%, thus reducing the effectiveness of RIN suppression [20]. The difference of 7.3dB is also due to using only one of the polarized light channels in the measurement of the system sensitivity.

3.3 OCT structural image quality

In order to compare the quality of 8 and 14 bit-depth imaging in our system, we took image sets of the human finger skin. Figure 5 shows the comparison of OCT structural images of the same sample (human finger skin) acquired with 14-bit and 8-bit DAQ boards respectively at the same set FIVR value of ± 0.4V. The raw image signal was real-time monitored during the imaging in order to make sure that the analog input fully covered the set FIVR of the two ADCs. Hence the 14-bit image data can be used to acquire reduced bit-depth data for image quality comparison by using the artificial bit-reduced transformation approach. It can be seen that there was little observable qualitative difference between the images taken at 8- and 14-bit depth. To quantitatively compare raw signals read out directly from the DAQs and also the resulting images, we use the root mean squared error (RMSE) and the mean absolute errors (MAE) [10]. The RMSE and MAE differences between the 8- and 14-bit raw signals were 0.01V and 0.08V, respectively. The RMSE and MAE differences between the native 8- and 14-bit depth images in Fig. 5 were 7.1dB and 2.2dB, respectively. The errors are likely to be negligible for both qualitative assessment and quantitative analysis.
We also compare the resulting image quality when the 14-bit depth image data is artificially bit-reduced during post-processing and this is shown in Fig. 5(top). However, in agreement with the results reported in [10,12], we also observe that in our system that real-world 8-bit image shows more artifacts e.g. horizontal stripes suggestive of fixed frequency interference than the one acquired by numerically truncating to 8-bits from the native 14-bit image, especially in low backscatter areas of the image. This most likely reflects differences in the hardware performance of these boards and the fact that the electronic bandwidth of the 8-bit board must be much higher than that of the 14-bit board to support the much higher maximum sampling rate (100 MS/s vs 20 MS/s). The RMSE and MAE differences between the real-world 8 bit and numerically truncated raw signals were 0.01V and 0.09V respectively. There is slight difference in the MAE from the comparison between the real-world 8- and 14-bit raw signals due to the rounding error in the bit-reducing. The RMSE and MAE differences between the images produced by native 8-bit and numerically truncated 8-bit raw signals were 6.8dB and 2.2dB respectively. The slight difference in the RMSE from the comparison between the real-world 8- and 14-bit images is due to the rounding error in the bit-reducing. The RMSE and MAE differences between the images produced by native 14-bit raw signal and numerically truncated reduced-bit signal are also included (bottom). Image size is 4 (transversal) × 2.5mm (axial).

The RMSE and MAE differences between the images produced by native 14-bit raw signal and numerically truncated reduced-bit signal are shown in Fig. 5 (bottom). It shows that the RMSE is larger or equal to the MAE at these reduced bit-depths. This is because the RMSE gives a relatively high weight to larger errors in measuring the average magnitude of the error, while the MAE uses an equal weight in the average. Therefore, the RMSE will always be larger or equal to the MAE and the greater the difference between them, the greater the variance in the individual errors in the sample. If the RMSE is equal to MAE, then all the errors are of the same magnitude.

It should be noted that the image thresholding was not used in the post-processing for images in Fig. 5 in order for comparison. However, it is normally used in OCT for good image visualization.

Figure 6 shows the image sets taken from human finger skin and an equine tendon sample to demonstrate the effect of different set FIVRs of the ADCs on the system sensitivity/dynamic range. Left column is for 8-bit and right column is for 14-bit. It can be clearly seen in Fig. 6(a) that the 14-bit image quality is much greater than the 8-bit at the set FIVR = ± 5V for both ADCs. However, at a lower FIVR good quality images can be obtained by the two ADCs as shown in Fig. 6(b)-(c). The RMSE differences between the 8 and 14 bit-depth images in Fig. 6 (b)-(c) were 4.7dB and 4.2dB respectively, and the MAEs were 1.9dB and 1.8dB, respectively. The errors are likely to be negligible for both qualitative assessment and quantitative analysis.
It should be also noted that there were several white vertical stripe artifacts on both images but these were more pronounced in the native 8-bit depth image in Fig. 6 (c), although identical detector voltage waveforms were acquired. This is due to signal saturation errors in 8-bit depth imaging arising from strong Fresnel reflections at the sample surface. In the system, the set FIVR for 8-bit depth imaging was lower in order to achieve similar sensitivity, i.e. the dynamic range at 8-bit is lower than that at 14-bit. This is in good agreement with the measurement results shown in Fig. 4. For example, in order to achieve ~100dB system sensitivity, the set FIVR can be any available value equal or less than 10V for the 14-bit board, but the maximum is 1V for the 8-bit board. Therefore, one possible disadvantage of using 8-bit instead of 14-bit DAQ board in the system is a reduced imaging dynamic range. The balance between the imaging dynamic range and system sensitivity may be carefully optimized depending on the sample scattering properties. This is in agreement with the result reported in [12]. The dynamic range is the ratio in signal strength between strongest and weakest reflection which can be measured simultaneously within one A-scan. Biomedical OCT images often have a dynamic range of ~35dB, so an OCT should provide 40-50dB [12].

3.4 OCT birefringence image quality

The phase retardance image quality between the two ADCs was also compared by using the equine tendon sample. The set FIVR was ± 0.4V and ± 5V for the 8- and 14-bit ADC respectively in order to achieve similar system sensitivity (~107dB). Figure 7 shows the obtained intensity and phase retardance images. The RMSE and MAE differences between the 8 and 14 bit-depth phase retardance images were 0.57 and 0.65 radians, respectively. The 14-bit raw data was also numerically truncated to 8-bit, which was used to calculate the phase retardance. The RMSE and MAE differences between the native 14 bit-depth and truncated 8

![Fig. 6. Images taken using 8-bit (left) and 14-bit (right) DAQ boards respectively in the PS-SS-OCT system at different set FIVR. (a)-(b): for human finger skin; (c) for an equine tendon sample. Image size is 4 (transversal) × 2.5mm (axial).](image-url)
bit-depth phase retardance images were 0.40 and 0.55 radians, respectively. The difference in RMSE & MAE from the comparison between the native 8- and 14-bit phase retardance images suggests that quantization noise produces the majority of the observed difference but that DAQ noise in the digitizer hardware makes a significant contribution also.

Fig. 7. Intensity (a) and phase retardance (b) images for the equine tendon sample obtained using 8-bit (left) and 14-bit (right) DAQ boards respectively in the PS-SS-OCT system. Image size is 4 (transversal) × 0.8mm (axial). The phase retardance images are shown in gray scale from black (−180°) to white (180°). (c): Example single A-scan comparison of the two retardance images.

These differences in RMSE and MAE are large compared to the full phase retardance measurement range of −π to π. This is possibly because the errors in 8-bit depth OCT signals were enlarged in the calculation process of phase retardance by using both the non-modulated and modulated OCT signals [13], while the structural OCT images in Fig. 5 and 6 were directly obtained by using optical interference signals (non-modulated OCT signals). Figure 7(c) shows an example of A-scan comparison of the two retardance images.

Therefore, we suggest that in structural image the DAQ noise is not a big problem anywhere we have a strong reflectivity because the signal is high. But in PS-SS-OCT the DAQ noise in the horizontal or vertical channel might produce larger effects. This is because the zeroth- or first-order signal in the horizontal or vertical channel might become weak at points where the retardance is low, i.e. at multiples of 2π, even though the backscatter is high. This is similar to single input state PS-OCT, such as time-domain (TD) PS-OCT [21–26]. Also the retardance is wrapped into the range −π to +π which might increase the apparent fractional difference as the mean retardance is constrained always to lie in this range.

4. Conclusion

We have shown via a direct comparison of true 8 and 14-bit AD converters that the 8-bit depth imaging can be used instead of 14-bit depth imaging in the PS-SS-OCT system with
negligible error for both qualitative assessment and quantitative analysis. We have shown that high system sensitivity can be achieved using both DAQ boards but it drops faster with the full input voltage range for 8-bit than for 14-bit DAQ board. Similar quality structural and phase retardance images can be obtained by using either 8-bit or 14-bit DAQ board provided that care is taken in the balance between the imaging dynamic range and system sensitivity. We also find that the fractional rms difference between images derived from 8 or 14-bit data is markedly higher for PS-SS-OCT phase retardance images than for reflectivity images however the difference is still not grossly evident to the eye. Some image artifacts are more obvious on 8-bit images due to signal saturation. The incidence of such artifacts depends on the sample scattering properties especially the surface Fresnel reflectivity profile.

Two caveats should be noted. Firstly our system sensitivity is about 7dB lower than the shot-noise limited value. Hence, some differences might become apparent if RIN noise is fully suppressed. However, theoretically this is not expected [10]. Secondly the conclusion may only be valid for the particular ADC hardware considered here and so the conclusions might change if we try a genuine GS/sec card since the noise level generally increases with sampling rate. Our study is aimed at comparing two equivalent cards that differ only in their bit-depths. High-speed cards may be noisier and make OCT images much worse but this is not fundamentally a limitation imposed by the bit-resolution. So a GS/s 8-bit will perform as well as the input detection electronics will allow. It is a useful piece of future work to systematically compare all available high-speed ADC’s for OCT imaging.

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