Data Acquisition with Swept Source OCT

THORLABS APP HIGHLIGHT

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INTRODUCTION

The Micro-Electro-Mechanical System, Vertical-Cavity-Surface-Emitting Laser (MEMS-VCSEL) swept- wavelength light source is an exceptional technology for optical coherence tomography (OCT), offering significant advantages in terms of high A-scan rates, long imaging depth, and high optical output power. These features make it a highly valuable tool for a wide range of applications in biomedical imaging and industrial inspection. In a previous application note (1), we covered the fundamental aspects of using Thorlabs' light sources for OCT, focusing primarily on the k-Clocking method for data sampling. The k-Clocking approach is widely favored for its simplicity. However, for certain applications, a dual-channel approach can offer significant advantages in the system performance, such as operation at a MHz sampling rate.

In this application note, an alternative sampling method, which is referred to as the "Dual-Channel" approach is introduced and explored. This method differs significantly from the k-Clocking technique, as it requires two input channels for data acquisition instead of one. The text below describes the key differences between the k-Clocking and Dual-Channel methods, as well as how to evaluate the specification requirements for an application and determine the appropriate sampling technique for an OCT system.

MEMS-VCSEL PROPERTIES

A critical parameter in OCT is the optical bandwidth of the light source, as it affects the axial resolution of the OCT images. In general, a broader optical bandwidth enables finer image resolution, allowing for a more detailed visualization of microstructures within a sample. The Thorlabs MEMS-VCSEL light source, which is a single-mode laser (2) - (4), features an exceptionally wide, mode-hop-free tuning range.

By scanning the laser cavity with repetition rates of up to the MHz range, the MEMS-VCSEL achieves a typical optical bandwidth of 100 nm. This can be centered around either 1300 nm or 1060 nm, depending on the specific requirements of the application. The choice of center wavelength directly impacts penetration depth and scattering in different tissues and the image resolution. The 1300 nm wavelength is often preferred for deeper tissue imaging due to its reduced scattering, which allows for larger penetration in non-transparent tissues. In contrast, the 1060 nm wavelength enables slightly higher image resolution and is particularly well-suited for water-rich environments, such as ophthalmology, where reduced water absorption provides better performance i.e. higher sensitivity and thus image contrast, making it ideal for imaging ocular structures like the retina and lens. Both the 1300 nm and 1060 nm wavelengths are also highly suitable for imaging non-biological samples in industrial applications, such as material inspection, surface profiling, and structural analysis.

In Figure 1, a typical optical spectrum of the MEMS-VCSEL light source imeasured with a Thorlabs optical spectrum analyzer is shown. The full optical bandwidth is defined as the spectral range where the optical power remains above -10 dB relative to the peak intensity. This -10 dB threshold is a commonly used criterion for defining the effective bandwidth, ensuring that only the most coherent and useful parts of the spectrum contribute to the image resolution.

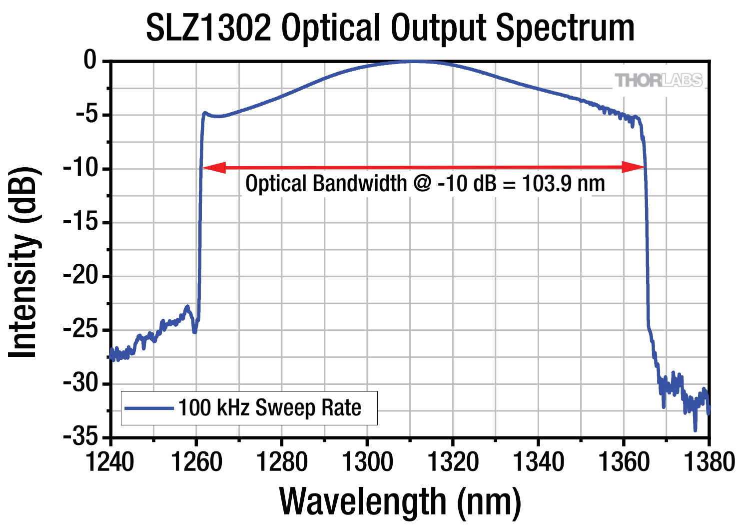

Figure 1: A typical optical spectrum for the SLZ1302 MEMS-VCSEL light source in the 100 kHz sweep mode measured using Thorlabs' OSA202C optical spectrum analyzer. The red arrow shows the optical bandwidth at the -10 dB attenuation threshold. This source has an optical bandwidth of 103.9 nm.

Figure 2 shows a sketch that summarizes the important parameters for a swept-source OCT light source. The laser cavity of the MEMS-VCSEL undergoes both a forward- and backward-wavelength sweep during the tuning process. However, it emits light exclusively during the forward sweep of the laser, and the proportion of this forward sweep to the total repetition cycle is referred to as the duty cycle D. Typically, the duty cycle ranges from 40% to 60%, depending on the sweep rate. The spectral shape is typically exhibiting a Gaussian-like profile, which provides smooth and predictable spectral characteristics in OCT imaging.

As an example, in Figure 3 the typical output signals of an SLZ1002 source with a sweep rate of 1 MHz are shown.

The yellow trace is the digital k-clock signal, which shows either the usable k-trigger signal during the light emission or the dummy clock signal with a clock frequency of 250 MHz during the backward sweep. The digital signal closely resembles a square wave, functioning like a trigger signal for the clock input of the data acquisition card. The time-varying instantaneous frequency of this signal is shown in light blue. The analog k-clock signal can be used in parallel, which is shown in green. This signal is essential for performing Dual-Channel acquisition. Both the digital and the analog k-clock signal are generated by the same internal fiber Mach-Zehnder-Interferometer (MZI), which has a customizable optical path delay. For triggering the data acquisition, one can use the electronic generated sweep trigger (dark blue) or the optical wavelength trigger (red). The latter one is generated by a Fiber Bragg Grating.

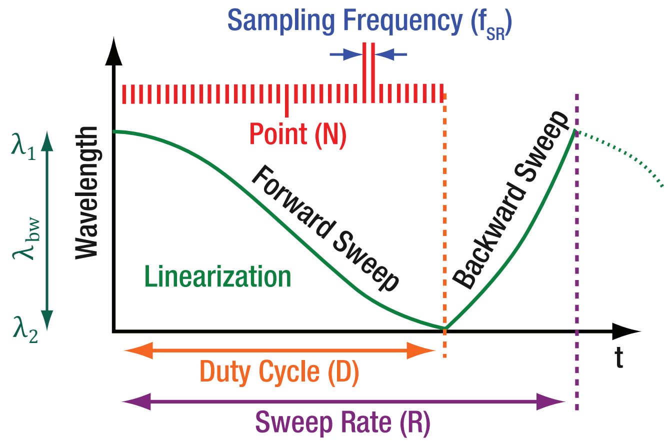

Figure 2: A sketch of important parameters for swept-source OCT. The green curve is showing the wavelength change for one sweep of the MEMS-VCSEL. The sampled bandwidth λ_{bw} is the range between the wavelength λ_1 and λ_2 of the first and the last data point. And the sampling frequency is the inverse time between two data points. Please see the text for more details on each parameter.

Figure 3: Oscilloscope image of the output signals from an SLZ Series MEMS-VCSEL laser source. These are typical signals for an SLZ1002 with a sweep rate of 1 MHz. The repetition rate and the duty cycle are also shown.

The primary distinction between the well-known k-Clocking method and the Dual-Channel approach lies in the type of k-mapping applied to the OCT data. For the processing of OCT data using the Fast Fourier Transform (FFT), the input data must be equidistant in wavenumber k. Utilizing the wavenumber k instead of the frequency f allows for direct extraction of the distance z after the FFT, as is standard for OCT imaging.

Although the optical wavelength sweep of the MEMS-VCSEL is linearized in wavenumber over time, it remains imperfectly linear (see Figure 2), necessitating correction of the sampling data.

Using the k-Clocking approach, the sample points are already sampled equidistant in optical wavenumber intervals because the sampling trigger is generated by the zero-crossing of a reference MZI signal with a delay length of Δl_{MZI}. This relation becomes clear by taking a closer look at the interference signal that is given by

$$I(t)_{\mathrm{MZI}} \propto \cos[k(t) \cdot \Delta l_{MZI}] \,,$$

where $k(t)$ is the changing wavenumber in time.

In comparison, the data points are sampled equidistantly in time when using Dual-Channel acquisition and subsequently mapped from the wavenumber $k(t)$ through a processing step into $k(t')$. At first glance, this makes the data acquisition much easier as the data-acquisition card can sample the data at a fixed sampling rate. However, much more complex data processing is required, as well as high computational power to process a large amount of data.

Several publications, such as (5) and (6), provide detailed explanations of the principles behind this type of acquisition method. In this approach, not only the OCT data needs to be captured, but also a modulation signal from a reference MZI. This is why two channels must be sampled in parallel, giving rise to the name "Dual-Channel" acquisition.

When using a Thorlabs SLZ Series MEMS-VCSEL laser source, the reference MZI signal is conveniently provided. This reference signal, created through a fixed optical delay in the internal MZI, serves as a frequency ruler, which contains all the necessary information about k-linearity. By extracting the phase information from this reference signal, a correction vector can be generated. The k-linearity of the OCT data can be established by interpolation, using the measured compensation vector from the reference signal. Essentially, it allows the system to compensate for any non-linearities in the wavelength sweep, which improves the overall image quality in OCT. A schematic of this Dual-Channel k-mapping processing in comparison to the k-Clocking method is shown in Figure 4.

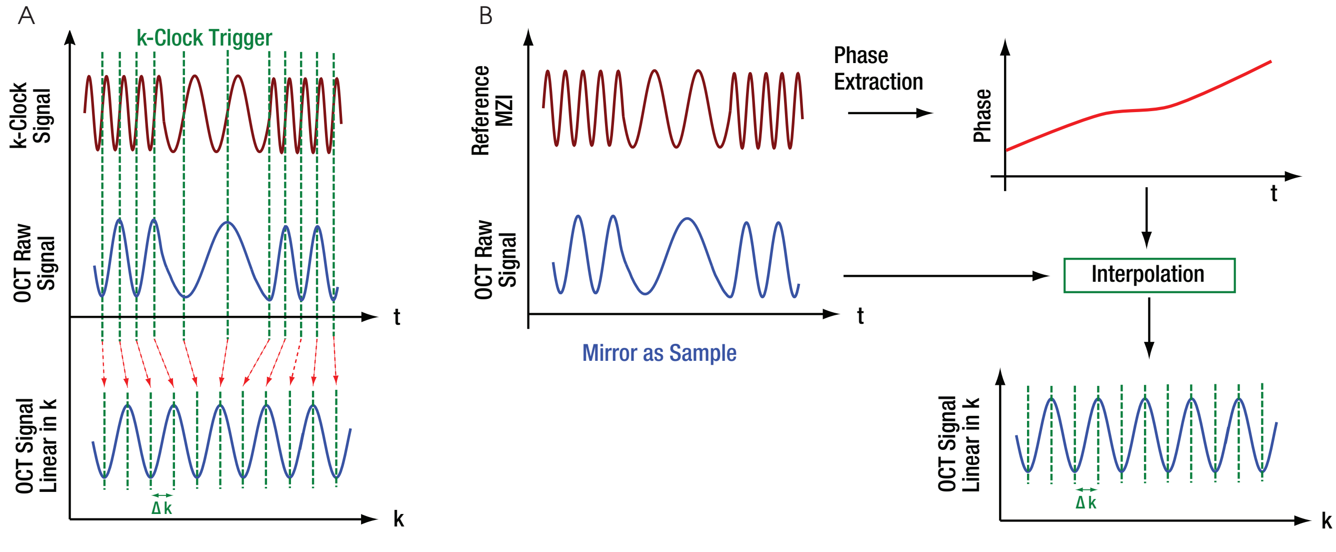

Figure 4: Schematic of the (A) k-Clocking method vs. the (B) Dual-Channel k-mapping processing. The zero-crossings of the k-clock signal act as a sampling trigger. In comparison, the Dual-Channel approach is using the phase of the reference signal to interpolate the OCT-raw data. The output is an interference pattern that is linear in wavenumber.

Our experience has shown that OCT processing works best, if the frequency of the reference signal $f_{modulation}$ is in the range of 0.15 to 0.6 times of maximum sampling frequency. Note that the maximum sampling frequency is half of the sampling frequency f_{SR} due to the Nyquist theorem. For instance, a data acquisition card featuring a sampling frequency of $f_{SR} = 2\mathrm{GS/S}$ can detect frequencies up to 1 GHz. To calculate the necessary MZI delay, one can use the formula

$$f_{modulation} \approx \frac{R \cdot \Delta l_{MZI} \cdot \lambda_{bw}}{D \cdot \lambda_1 \cdot \lambda_2} \cdot l_{ratio} \cdot$$

The linearization ratio l_{ratio} which accounts for the nonlinearity of the wavelength sweep, is an approximation which is determined by

$$l_{ratio} \approx \max(f_{modulation})/\mathrm{average}(f_{modulation})$$

Figure 5 illustrates how a larger signal frequency range, influenced by the linearization ratio, limits the imaging depth.

The linearization ratio typically ranges from 1.1 to 1.5, primarily depending on the sweep rate of the MEMS-VCSEL light source. The sweep rate of the laser R, the sampled optical bandwidth $\lambda_{bw} = \lambda_2 - \lambda_1$ and the duty cycle D are specifications from the light source. A key distinction must be drawn between the full optical bandwidth shown in Figure 1 and the sampled optical bandwidth λ_{bw}. The sampled optical bandwidth refers to the range spanning from the wavelength of the first sampled data point to that of the last. The λ_1 and λ_2, shown In Figure 2 correspond to the red (first sample point) and blue (last sample point) wavelength, and are thus linked to the center wavelength. Only the sampled optical bandwidth affects image quality metrics such as resolution. Notably, the sampled optical bandwidth may be marginally smaller than the full optical bandwidth (typically <5 nm), as noted in the Application Note (1).

The Dual-Channel method offers significant advantages, particularly in enhancing both image depth and the scan rate of the light source. In contrast, the k-Clocking approach is constrained by the frequency limitation of the k-clock signal, which typically reaches a maximum of 1 GHz. This imposes a restriction on the number of data points that can be collected, ultimately limiting both the image depth and the laser's scan rate. The Dual-Channel method, however, is primarily limited by the sampling rate of the data acquisition system, which can exceed 5 GS/s, providing much greater flexibility in terms of data collection and imaging depth.

Despite these differences in data acquisition capabilities, the two methods exhibit comparable performance in terms of resolution and sampling bandwidth. However, it should be emphasized that the Dual-Channel method is more advantageous in terms of general image quality, i.e. image contrast and artifact rejection. Preliminary measurements also suggest that the Dual-Channel approach may offer slightly improved phase stability, potentially providing a subtle but valuable advantage in certain imaging applications.

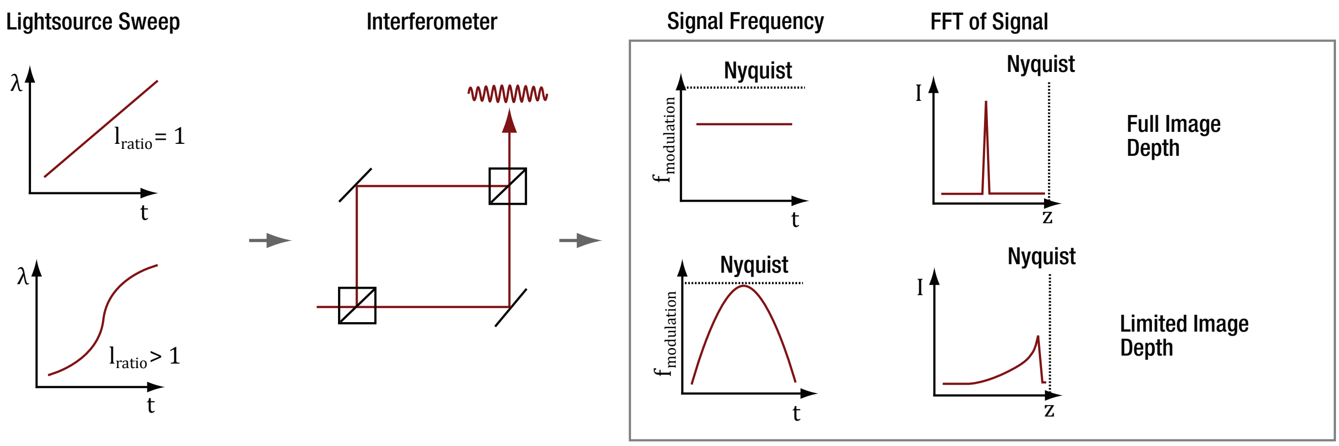

Figure 5: The linearization ratio of the light source directly influences the image depth. Please see the text for a more detailed explanation. According to the Nyquist theorem, the broader frequency range with a high linearization ratio limits the measurable delay in the interferometer.

ESTIMATING THE RESOLUTION IN OCT

The axial distance between two adjacent bins or data points within the processed OCT signal, referred to as the z-spacing Δz, is a critical parameter in OCT systems. It is crucial not to confuse this with axial resolution, δz which refers to the minimum separation between two scatterers that can still be distinguished as distinct. Figure 6 provides a graphical representation by investigating a point spread function of a mirror surface.

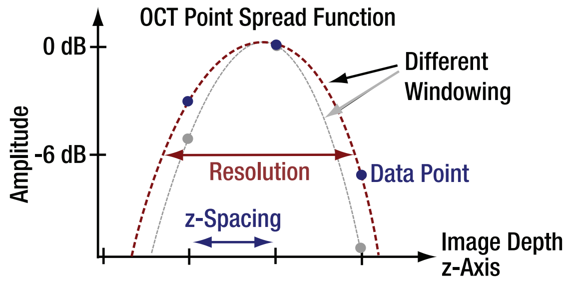

Figure 6: The sketch is showing the relationship between resolution δz and the z-spacing Δz of the point spread function from a mirror surface. Altering the windowing in the FFT modifies the resolution but not the z-spacing, as the latter one depends on the data point spacing. This example is showing two different windowings of the OCT data, before the FFT was performed.

As the z-spacing Δz is related to the optical spectrum via the Fourier transformation, it can be estimated as

$$\Delta z \approx \frac{\lambda_C^2}{2 \cdot \lambda_{bw}},$$

where λ_C is the center wavelength of the optical spectrum (7) - (10). To obtain this expression, the relationship $\Delta z = 2\pi/2\,\Delta k$ can be reformulated by evaluating $dk/d\lambda = d/d\lambda\,(2\pi/\lambda)$. Since we only consider the case in air, the refractive index is set to $n = 1$ in all formulas. The estimation of Δz is consistent across various methods, including k-Clocking and Dual-Channel processing, as it is governed solely by the sampled optical bandwidth λ_{bw}. It is important to note that the term "z-spacing" is not universally standardized in the literature, necessitating careful distinction from the axial resolution δz, which is defined as the full-width at half-maximum (-6 dB for OCT signals) of a single peak in the OCT image (see Figure 6).

In the literature, axial resolution is often expressed in relation to a Gaussian spectral shape. However, this assumption is unlikely in most cases, particularly as we specify the optical bandwidth of our light source based on a -10 dB intensity level rather than relying on a Gaussian profile. A more accurate estimation of axial resolution can be obtained by considering the z-spacing and adding the influence by the spectral shape of the light source and the window function used in the FFT. Thus, the axial resolution can be evaluated as

$$\delta z \approx S_{shape} \cdot \Delta z,$$

where the factor S_{shape} accounts for the spectral influence. Since the window function significantly influences the resolution, Table 1 presents values for S_{shape} corresponding to specific window functions, as reported in (11). Figure 6 illustrates that changing the window function alters the resolution while the z-spacing remains unchanged. This highlights the inherent fact that the axial resolution is always larger than the z-spacing.

Window	S_{shape} (@ -6 dB)
Rectangle	1.21
Hann	2.00 (α = 2)
Tukey	1.57 (α = 0.5)
Hamming	1.81
Dolph-Chebyshev	2.01 (α = 3)
Gaussian	2.18 (α = 3)

Table 1: Value of factor S_{shape} for different window functions (11). This factor can be used for resolution determination, by using the z-spacing. The spectral shape of the light source will also be added to this factor.

As an example, the point-spread-function of a mirror surface was measured with a swept-source OCT system and the linewidth and the z-spacing were measured.

ESTIMATE THE NUMBER OF DATA POINTS

An important practical consideration is the number of data points N, which significantly impacts the processing speed, especially when employing the Dual-Channel approach. With k-Clocking, the number of data points is determined by the delay of the MZI and the sampled optical bandwidth

$$N_{\text{k-clocking}} = \frac{\Delta l_{MZI}}{2 \cdot \Delta z} = \frac{\Delta l_{MZI} \cdot \lambda_{bw}}{\lambda_C^2}.$$

In contrast, for Dual-Channel processing, the number of data points is governed by the time of the forward sweep $t_{sweep} = \frac{D}{R}$ and the time between two sampling points $t_{SR} = \frac{1}{f_{SR}}$ with

$$N_{DC} = \frac{t_{sweep}}{t_{SR}} = \frac{D \cdot f_{SR}}{R}.$$

Here we have the duty cycle D, the sampling frequency f_{SR} and the sampling rate R. Modern data acquisition systems often impose restrictions on the total number of data points, which may, for example, be constrained to multiples of 128. Moreover, the Fast Fourier Transform (FFT) algorithm operates most efficiently when the number of data points is a power of two (i.e. 2^n). These constraints may lead to a reduction in the actual number of sampled points, potentially limiting the effective sampling bandwidth.

ESTIMATING THE IMAGING DEPTH IN SWEPT-SOURCE OCT

A very application-dependent parameter is the imaging depth of an OCT system, which is mainly determined by the light source. The image depth is straightforward to calculate using the k-Clocking approach, as it is directly linked to the delay of the reference MZI

$$\text{Image depth}_{\text{k-clocking}} = \frac{\Delta l_{MZI}}{4}.$$

This relationship becomes clear when examining the connection to the reference MZI signal, which triggers the data points. According to the Nyquist theorem, accurate frequency determination requires two data points, necessitating two complete cycles of the reference MZI signal. Since this signal is directly linked to image depth through the MZI delay, the first factor of 1/2 can be understood. The second factor of 1/2 arises from the fact that OCT measures distance rather than optical path length.

In contrast, determining image depth using the Dual-Channel method requires consideration of the linearization ratio l_{ratio} which was explained earlier in the note and is illustrated in Figure 5. While k-Clocking inherently incorporates this through the sampling rate, the Dual-Channel approach requires explicit use of l_{ratio} where $l_{ratio} > 1$.

In this case, the estimated image depth is given by

$$\text{Image depth}_{DC} \approx \frac{0.5 \cdot N_{DC} \cdot \Delta z}{l_{ratio}} = \frac{\text{dc} \cdot f_{SR} \cdot \Delta z}{2 \cdot l_{ratio} \cdot R} = \frac{\text{dc} \cdot f_{SR} \cdot \lambda_C^2}{4 \cdot l_{ratio} \cdot R \cdot \lambda_{bw}},$$

where f_{SR} is the sampling frequency, R is the sweep rate, and D is the duty cycle.

Since both the number of sampling points and the z-spacing are approximations, the calculated image depth is also an estimate, albeit usually close to the actual value. Importantly, if the sweep rate and duty cycle are fixed, the image depth and resolution are inversely proportional. If this parameter is not adequately considered, modulation frequencies may exceed the sampling frequency, leading to undersampling artifacts. These artifacts arise when signals from deeper regions of the sample generate higher modulation frequencies that exceed the sampling rate of the data acquisition device, resulting in the loss of high-frequency information and the introduction of aliasing. As a result, improper handling of the linearization ratio can significantly degrade the quality of the OCT image by misrepresenting the deeper structures. Therefore, careful consideration of l_{ratio} is essential for accurate imaging, particularly when working with deep tissue or high-resolution applications.

Curious for more? Please Contact oct@thorlabs.com for more information on the specifications and possible customizations. We are eager to discuss your application and learn about your requirements!

SUMMARY

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