Practical considerations for OCT applications

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Abstract. OCT measurements have a fundamental trade-off between the ability to resolve small details and the range over which the measurement is consistent. A measurement which is able to resolve small details is able to do so over a small range. A measurement which is consistent over a larger range is not able to resolve small details. While the axial resolution of the OCT measurement is determined by the optical bandwidth of the source, the lateral resolution (spot size) is determined by the focusing optics and the characteristics of the Gaussian beam. The spot size and the depth of field of a Gaussian beam are directly related in such a way that there is a trade-off between the spot size (details which can be resolved) and depth of field (distance over which the spot size is maintained). In this paper we analyze and discuss in detail the trade-off between the spot size and depth of field in OCT measurements. Some techniques to mitigate this limitation are mentioned and one is applied to measurements of cervical tissue with and without cervical intraepithelial neoplasia (CIN).

Keywords. OCT, optical coherence tomography, optical metrology, Gaussian beam, cervical intraepithelial neoplasia, CIN.

1. Introduction
Since its very beginnings, the suitability of Optical Coherence Tomography (OCT) technology to biomedical applications has been discussed [1]. OCT equipment currently allows the acquisition of tomographic images with micrometric resolution at high acquisition rates and with high sensitivity. OCT is now routinely used in ophthalmology to examine retinal pathologies [2–5]. It is being actively studied for use in dermatology [6], gastroenterology [7], urology and gynecology [8,9] OCT is also being explored for uses in industrial and non-destructive evaluation (NDT) applications [10].
The operating principles of OCT [11] are analogous to ultrasound in that waves are launched into a sample and a backscattered signal holds information on characteristics of what is inside. In the case of OCT, light is used rather than high frequency sound waves. The smaller wavelengths of light allow OCT a higher resolution, but as light is easily scattered or absorbed the penetration depth is limited to a few mm in biological applications, a limit which is not always applicable to industrial and NDT applications. Ultrasound’s terminology for single point scans (A-Scan), linear scan (B-Scan) and area scan (C-Scan) are used by OCT as well.

It is quite important to note that OCT can be reliably calibrated in all 3 measurement dimensions using secondary standards, therefore, metrological traceability is possible to establish whenever measurements need to be compared between instruments and laboratories. [12–14]

2. Experimental Setup
There are four variants of OCT, Time Domain (TD-OCT), Spectral Domain (SD-OCT) and Swept Source (SS-OCT) and Linear (L-OCT) [15] but while they share the same general block diagram as shown in Figure 1 there are differences which lead to advantages and disadvantages, depending on the use case, as shown in Table 1.

![Figure 1. Generic OCT block diagram shared by all four forms of OCT: TD-OCT, SD-OCT, SS-OCT and L-OCT.](image)

In all forms of OCT, the axial resolution of the A-scan is inversely proportional to the optical bandwidth of the source, as shown in equation 1. The source is incoherent in the case of TD-OCT, SD-OCT and L-OCT and coherent swept in the case of SS-OCT. The focusing optics locate the focus of the source beam at some distance inside the sample with a spot size and depth of field, which, along with the effects of the optical bandwidth on these values, will be discussed below. The distance to the reference mirror controls where in the A-scan the focus is resolved. How the A-scan is performed depends on the technique. TD-OCT and L-OCT perform the A-scan in Cartesian space while SD-OCT and SS-OCT scan in the Fourier domain and require an FFT back to Cartesian space to complete the A-scan. The location of TD-OCT’s mirror is scanned to perform its A-scan. At the detector, the reference and sample beams are combined and then detected with a photodetector. The detected signal traces out an interference pattern as the mirror is scanned which is the A-scan. The axial range is determined by how far the mirror can move. The sampling rate of the data acquisition system determines whether the full resolution is accessible. L-OCT does not combine the reference and sample beams; rather, at the detector, it expands each separately and superimposes them at an angle on a linear CCD array producing the A-scan interference pattern, which is directly read out. The length of the CCD array determines the axial range and the CCD array resolution determines whether the full axial resolution is accessible. SD-OCT combines the reference and sample beams before passing them through a diffraction grating onto a linear CCD. An interference pattern is again produced on the CCD, but this time it is in Fourier domain and an FFT of this pattern produces the A-scan. As an FFT
is involved, scales are reversed and it is the resolution of the CCD array that this time determines the axial range and the length of the CCD array which determines whether the full axial resolution is available. SS-OCT combines the reference and sample beams and detects them together with a photodetector as the wavelength of the laser is swept. The detected signal traces out an interference pattern, which, after the full sweep completes, an FFT converts to the A-scan. As an FFT is involved, scales are reversed and it is the sampling period (1/sampling rate) which determines the axial range (subject to the coherence length of the laser) while the length of the data set determines whether the full axial resolution is available.

Table 1. Key Application Differences

|         | Speed                  | Axial Range                                      |
|---------|------------------------|--------------------------------------------------|
| TD-OCT  | Slow – mirror moves    | Medium (m) – Limit is reference mirror’s movement range |
| L-OCT   | Fast – No moving parts | Short (mm) – Limit is the length of CCD array     |
| SD-OCT  | Fast – No moving parts | Short (mm) – Limit is resolution of CCD           |
| SS-OCT  | Medium – swept laser   | Long (km) – Limit is sampling rate vs speed of laser’s sweep |

Figure 2 is a common figure, showing the position which OCT inhabits in the axial resolution / penetration depth parameter space between confocal microscopy and ultrasound. While this is accurate, it doesn’t tell the complete story and alone can be misleading as it implies that OCT has voxels (3D volume elements) of few to tens of microns on a side over a few (1 to 5) millimeters of range. However, in fact, the figure only specifies the axial resolution, which is independent of and typically different from the lateral resolution (spot size). The penetration depth is not even a function of OCT specifically, rather a more general limitation of light penetrating tissue at visible to low infrared wavelengths.

A better basis with which to explore OCT applications might be Figure 3, which shows the spot size vs the depth of field of OCT broken into three classes with a small design representing the beam near focus of each functional space. The allowed spot size and depth of field combinations lie on the solid line. The three classes are OCM (Optical Coherence Microscopy), where the beam diverges rapidly away from focus, Traditional OCT, where the spot size is never quite as small as one wants over the depth of field that one has and Coherent LIDAR, where the beam is large but essentially unchanging over long distances. In all of these classes, the axial resolution can be considered the same. The spot size is the lateral resolution at focus and the depth of field is the axial distance over which the lateral resolution remains within a factor of 1.41 of the spot size.

Figure 2. The relative locations of OCT, Confocal Microscopy and Ultrasound in Penetration Depth – Axial Resolution space.
The leftmost class of OCT has been termed OCM (Optical Coherent Microscopy) and is OCT used in the confocal extreme, with the extremely short depth of field exploited for range selection by passing through a pinhole as in confocal microscopy but with the added benefit of coherence gated detection [16].

The rightmost class of OCT applications is coherent LIDAR which is the opposite extreme where the huge depth of field makes imaging or ranging distance limited only by the characteristics of each form of OCT [17–19]. TD (Time Domain) is limited by the distance over which the reference mirror can be moved. L (Linear) is limited by the length of its CCD. SD (Spectral Domain) is limited by the resolution of its CCD array. SS (Swept Source) is limited by the sampling rate of its analog to digital convertor and coherence length of its laser. This modality (SS for Lidar) has a lot in common with Optical Frequency Domain Reflectometry (OFDR) which is used in fiber measurements [20–22].

The class in the middle is traditional OCT, which covers the space from small spot size and limited depth of field to a spot size of a few 10’s of microns and a depth of field of a few hundreds of millimeters. This is the traditional OCT ground, where decisions must be made as to the depth of field (the axial distance over which a B-scan remains in focus) and spot size (what can be resolved in the focused region) [23].

For a Gaussian optical spectral shape, the axial resolution $\Delta z$ is given by equation 1, where $\lambda_0$ is the source central wavelength and $\Delta \lambda$ is the width at half maximum [24]:

$$\Delta z = 2\ln2 \frac{\lambda_0^2}{\pi \Delta \lambda}$$  

The spot size and depth of field are independent of $\Delta z$ and are related as shown in equation 2 as derived from [25]. Other notable relations are the confocal parameter ($b$) is a commonly used term for the depth of field, the Rayleigh range, $z_r$, which is half the depth of field and $w_0$, the beam width radius which is half the spot size.

$$\text{depth of field} = \frac{\pi \text{(spot size)}^2}{2\lambda}$$

![Figure 3. Calculated spot size vs depth of field for Gaussian beams.](image_url)
3. Results and discussion

Figure 4 shows the calculated spot size vs depth of field over the range of traditional OCT. In Figure 3 this relation was shown on a log-log scale on which the relation is a straight line. In Figure 4 it is useful to see the relation in linear space to appreciate how fast the lateral resolution is lost.

![Graph](image1)

**Figure 4.** Calculated depth of field vs spot size at 1310 nm. (i) Depths of field to cover up to the fully allowed 3.6 mm depth (ii) zoom of the smaller spot sizes. (a) to (e) refer to Table 2.

Figure 4(i) shows the calculated depth of field up to 3.6 mm, which is the full imageable axial range of the Thorlabs Telesco commercial SD-OCT system. Correspondent with this depth of field is a spot size of 55 μm. In other words, with the depth of field chosen to be 3.6 mm the lateral resolution at focus (the middle horizontal of the B-scan) would be 55 μm and would remain close to this value to the top and bottom, only expanding to be 1.41 times as large at the top and bottom. Figure 4(ii) is the zoom of the six smaller spot sizes, from Figure 4(i).

The entire B-scan would be in the same focus, albeit a rather coarse focus. It is in testament to the limited usefulness of such a coarse focus that the lens sets offered do not include this option (See Table 2).

As one moves down the curve to the intermediate spot size / depth of field combinations, choices must be made, questions answered. What needs to be resolved and over what distance and what is more important? This is important because a single B-scan will now produce a swath of focused data bordered top and bottom by out-of-focus data, for example Figures 6(i), (ii) and (iii).

The location top-to-bottom of the swath can be controlled by the relative distance of the reference and the focusing optics. The focusing optics place the swath at some depth in the sample and adjusting the reference distance places the B-scan window over the swath.

Figure 5 shows the calculated beam shape of some specific choices of spot size and depth of field. Figure 5(i) shows the 3.6 mm depth of field in green along with other values with better spot size. The legend identifies the spot sizes (in μm, on top) and depths of field (in μm, on the bottom). As Figure 5(i) is roughly the top to bottom range of a full 3.6 mm axial range of a Thorlabs B-scan, the size of the swaths for each choice can be imagined by observing the vertical distance between colored squares that mark the depths of field. Figure 5(ii) expands on some of these better focused values, showing 500 μm, a typical top-to-bottom B-scan range for imaging tissue. Some of these (20 μm (d), 13 μm (c) and 7 μm (b)) correspond to lens sets which Thorlabs offers commercially (see Table 2) and illustrate the types of choices that must be made when choosing a commercial lens set or building your own.
Figure 5. Half the calculated Gaussian beams for various calculated spot sizes. (i) has a 4 mm window top to bottom and (ii) is a 500 µm window subset. The squares on the curves show the extents of the depths of field. The legend values are the spot sizes in µm. (a) to (e) refer to Table 2.

Specifically, look at the pink (d) and blue (c) curves in Figure 5(ii) which is 500 µm top to bottom. Assume for the moment that you are imaging a thin layer of tissue that has a thickness of 500 µm. The most faithful imaging of the range would be the pink curve which neatly covers the 500 µm with a spot size 20 µm. However, imagine that really you were interested in details a little smaller than what the 20 µm is allowing. Looking at the data in 5(ii), you might choose the blue curve lens set, which, despite having a smaller depth of field than the 500 µm range, the lateral resolution only gets worse near the top and bottom and is actually much better near the center, 13 µm at the focus. This could be better for you if the improved spot size was truly necessary to decide something while giving enough of a hint of something at the top and bottom so you could re-scan with the focus re-positioned if necessary.

Table 2. Some interesting spot-size and depth-of-field pairs from the plots of Figures 4 and 5.

| Spot size (µm) | Depth of field (µm) | Notes |
|---------------|---------------------|-------|
| (a) 5         | 29.98               | Spot size equal to Z resolution |
| (b) 7         | 58.76               | Lens 1 LSM02 |
| (c) 13        | 202.64              | Lens 2 LSM03 |
| (d) 20        | 479.63              | Lens 3 LSM04 |
| (e) 55        | 3627.22             | Depth of field covers entire Z range |

One solution would be to simply scan your tissue sample multiple times with a better spot size but smaller depth of field lens set with the focus placed at different locations in the sample in an organized fashion and afterwards combine the images to create a composite super image [26]. An example of this procedure is shown in Figures 6(i-iv) where fresh cervical tissue was scanned three times with the focus changed to different depths and the distance to reference adjusted to bring the image back into the center of the B-scan window each time to produce the images in 6(i), 6(ii) and 6(iii). The three images were sliced horizontally and stitched together using Photoshop to produce the 6(iv) super
image. The lens is the Thorlabs LSM02 lens with a spot size of 13 μm and depth of field of 204.2 μm. The top-to-bottom distances are 3.6 mm in 6(i), (ii) and (iii) and slightly larger in 6(iv) (because of stitching). In 6(i), the focus is towards the bottom, and reveals an area with CIN-III (0 mm – 4 mm). In 6(ii), the focus is in the middle, and reveals inflammation and a lack of epithelial tissue (4 mm-5 mm). In 6(iii), the focus is high, and reveals normal ectocervical epithelial (5.5 mm– 8.0 mm). In 6(iv), all the information is visible in one super image.

Figure 6. (i), (ii), (iii) show three B-scan measurements of cervical tissue at the same location but with the focus at different depths. (iv) combines the three into one composite super image.

There are other creative optics and math intensive solutions being proposed to the spot size / depth of field issue. These either take multiple images (each at a different focus) at the same time and combined as above (the difference being that all the data is taken during one single B-scan rather than multiple B-scans) [27,28], or artificially focus the beam in post processing by using phase differences created by multiple optical paths to the sample being imaged [29–31], or use Bessel beams generated by axicon optics which can have longer depths of fields at smaller spot sizes albeit with added side-lobes [32–34].

Finally, as mentioned earlier, better axial resolution comes from increased source wavelength range (Δλ) as shown in equation 1. However, from equation 2, as the wavelength increases, the depth of field for a fixed spot size will shrink, while the spot size for a fixed depth of field will grow. Table 3 shows the effect of a wavelength range of 200 nm centered at 1310 nm (from 1210 nm to 1410 nm) on the depth of field, keeping the spot size fixed and on the spot size, keeping the depth of field fixed for a beam whose spot size is 13.00 μm at 1310 nm.

In practice, however, the spot size, the depth of field as well as the position of the focus all change as the cascading effects of the changed wavelength on upstream optical elements propagate to the end. Fig. 7 shows a Zemax simulation of a simple system. A 3.55013 mm Grintech LFRL-100 GRIN lens is fed by a Corning SMF-28 single mode fiber. This produces, at 1310 nm, a Gaussian beam focused 928 μm after the output of the GRIN lens to a 13.00 μm spot size with a 202.64 μm depth of field.
wavelength range of 200 nm is assumed, resulting in a start wavelength of 1210 nm and end wavelength of 1410 nm. The change in the beam parameters (spot size, depth of field and \( z_0 \), the distance from the output of the GRIN lens to the beam focus) relative to the values at the midpoint wavelength 1310 nm, are shown in Table 3.

**Figure 7.** Zemax simulation of a system using a GRIN lens to focus the output of a single mode fiber.

In the simulation, the spot size changes by around 2 \( \mu \text{m} \) over the wavelength range, about twice what the calculations for fixed depth of field showed, while the depth of field changed around 32 \( \mu \text{m} \), similar to the values calculated for a fixed spot size. Finally the simulation shows that \( z_0 \) changes by around 8 \( \mu \text{m} \), small, but still enough to deteriorate the effective axial resolution, being more than twice as large as the 3.8 \( \mu \text{m} \) axial resolution given by equation 1.

**Table 3.** Calculated and Zemax simulated differences in spot sizes, depths of field and focus locations at the start and end wavelengths for a beam with a spot size of 13.00 \( \mu \text{m} \) at 1310 nm (depth of field of 202.64 \( \mu \text{m} \)).

| \( \lambda \) (nm) | Calculated \( \Delta \)Spot size (\( \mu \text{m} \)) | Calculated \( \Delta \)Depth of field (\( \mu \text{m} \)) | Simulated \( \Delta \)Spot size (\( \mu \text{m} \)) | Simulated \( \Delta \)Depth of field (\( \mu \text{m} \)) | Simulated \( \Delta z_0 \) (\( \mu \text{m} \)) |
|-------------------|--------------------------------|--------------------------------|-------------------------------|--------------------------------|------------------|
| 1210              | -0.51                         | +16.75                         | -1.02                         | -16.26                         | -4.89            |
| 1410              | +0.49                         | -14.37                         | +1.02                         | +16.19                         | +3.68            |

4. Conclusions
The typical manner of evaluating the usefulness of OCT is to compare its axial resolution and penetration depth in tissue to that of Ultrasound and Confocal Microscopy. In reality, OCT is useful over a broader range of applications than this would imply as seen by examining its spot size to depth of field relationship. Three spaces appear, OCM, which enhances Confocal Microscopy, traditional OCT, and large beam coherent lidar applications. Traditional OCT requires dealing with the very real tradeoff between spot-size and depth-of-field which changes very rapidly throughout its range. One way to lead with this limitation is to repeat measurements with the location of the focus at different depths and stitch together a composite super image. This technique was applied to human cervical tissue.

Acknowledgements
The authors would like to acknowledge financial support from the Brazilian agencies FAPERJ, CAPES and CNPq.

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