

REPORT

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Point-Spread Function Phantoms for Optical Coherence Tomography

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ABSTRACT

Optical coherence tomography (OCT) is maturing as a powerful imaging modality in medical and non-medical fields. As the use of this technique increases, so too do the requirements for consistent and robust characterisation methods for fundamental system performance. An important baseline metric in this respect is the system 3D point-spread function (PSF) that ultimately limits the optical resolution of an instrument. In this article, we present a theoretical description of the OCT lateral and axial PSF and outline design considerations for constructing PSF measurement phantoms using clear resin, doped with a low density of micro-spheres. The theoretical treatment identifies a $\sqrt{2}$ enhancement of the lateral resolution due to re-combination of the reference and sample fields. Furthermore, we study the potential PSF degrading effects of optical dispersion and spherical aberration. It is shown that the dispersion coefficient of any material chosen for PSF phantoms must be carefully considered with respect to the OCT coherence length. Additionally, it is found that spherical aberration due to refraction at the surface of a phantom becomes important for numerical apertures of 0.4 or more at depths greater than 0.5 mm, which may be important in optical coherence microscopy. An important design consideration for PSF phantoms is the target ‘point’ size, that is required to be less than 15% of the PSF full-width at half maximum. The phantoms consist of a clear epoxy resin, doped with a low density of monodispersed silica micro-spheres. The preparation, measurement and analysis of such phantoms is discussed and we demonstrate how these phantoms can be used to characterise the PSF of an OCT instrument throughout the B-Scan image plane, yielding quantitative information about the spatial distribution of resolution.

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1 Introduction

High resolution, non-invasive imaging of the internal micro-structure of materials is increasingly important in many fields, including clinical medicine[1] micro-manufacturing[2] among others. Optics is playing an important role in the development of suitable technologies, for example the non-invasive, three-dimensional imaging modality, optical coherence tomography (OCT)[3]. This emerging technique is becoming increasingly widespread as a tool for medical diagnosis[4, 5], and its use can be expected to grow as a number of commercial instruments begin to come to market. With the increasing success of OCT, there is equally a growing requirement for consistent and robust characterisation of fundamental system performance. This is true from both regulatory and user confidence perspectives. There are numerous figures of merit reported in the literature that might be used to determine the quality of a given OCT instrument or veracity of the image data, for example texture[6], scattering[7] and principle components analysis[8]. The most widely accepted metrics are however, system absolute sensitivity and full width at half maximum (FWHM) of the source coherence function, often stated as the axial resolution or point-spread function (PSF), or by implication as the source spectral FWHM bandwidth. The subject of this article is concerned with presenting a robust test method for determination of OCT PSF in both lateral and axial dimensions and its spatial distribution throughout an imaging plane (B-Scan). Previous authors have used the knife edge method and flat reflective surface[9] to characterise lateral and axial PSF respectively. However, these methods do not lend themselves to measurement of the spatial characteristics of these properties. Moreover, it is intrinsically difficult to measure at precisely the beam waist using the knife edge method. We demonstrate a robust and reproducible epoxy resin PSF phantom applied to characterise a commercial OCT instrument. Discussion of the preparation of such phantoms and detailed experimental analysis is left to a further article. The purpose of this article is to present basic theory of OCT resolution and the associated well known expressions. Furthermore, potential resolution limitations introduced through refractive index dispersion and surface refraction induced spherical aberration are discussed.

2 Theory

One of the unique properties of OCT is that the PSF in the axial and lateral dimensions are effectively de-coupled. This is because in the limit of most implementations, they are governed by independent physical phenomena of optical coherence and diffraction respectively. This differs from other optical sectioning modalities such as confocal microscopy, where the axial and lateral PSF are both diffraction limited. At this point it is worth noting the subtle difference between PSF and resolution in the context of OCT imaging. The PSF describes how an infinitely small point is smeared out by the optics and processing of an OCT imaging system, we assume this to be in the absence of scattering. However, the resolution defines how close two features can be and just be resolved[10]. When imaging tissue there is generally significant scattering that gives contrast between tissue types. In this situation there is also significant speckle that is inherent to the coherent detection of OCT. It has previously been shown[11] that the speckle size is dictated by the system optics, and hence this will govern the effective resolution. Unlike previous investigations[9] we have preferred to evaluate the system PSF in the absence of scattering in order to establish a baseline performance metric. This has advantages from the perspective of quality testing, although perhaps does not reflect exact system performance when imaging through turbid media such as human tissue.

2.1 The OCT Signal

The OCT intensity signal $I(\omega)$ incident upon a large detector is readily described as a function of angular frequency ω [3]

$$I(\omega) = \int_{-\infty}^{\infty} E_s E_s^* d\rho + \int_{-\infty}^{\infty} E_r E_r^* d\rho + 2 \int_{-\infty}^{\infty} \Re(E_s E_r^*) d\rho, \quad (1)$$

where E_r and E_s represent the electric field in the reference and sample arms of a Michelson interferometer typical of OCT systems. Disregarding losses in the interferometer and beam-splitter transmission coefficients, the sample and reference fields can be written in terms of an incident time dependent (t) field $E_{in}(\omega, t) = s(\omega) \exp(i\omega t)$ with field spectral profile $s(\omega)$,

$$E_r(\omega, t, \rho) = E_{in} \xi_r(\rho) \exp(i\phi) \quad (2)$$

and

$$E_s(\omega, t, \rho) = E_{in} \Pi(\omega, \rho - \rho_0) \xi_s(\rho). \quad (3)$$

In Eq. 2, it is assumed that the reference mirror is a spatially uniform 100% reflector of infinite extent and that it is illuminated with an identical spatial field profile $\xi_r(\rho) = \xi_s(\rho) = \xi(\rho)$ as the sample $\Pi(\omega, \rho - \rho_0)$. The lateral position of the sample beam is denoted ρ and its offset from the centre of the sample is $\rho - \rho_0$. The phase term $\phi = \omega \Delta\tau$ accounts for any length mis-match $\Delta\tau$ between the reference and sample paths. In OCT imaging, the first term of Eq. 1 is typically small and the second term is a readily subtracted DC term. Therefore, the part of interest is the third ‘measurement’ term and hence, Eq. 1 may be simplified to become

$$I(\omega) \approx 2 \int_{-\infty}^{\infty} \Re(E_s E_r^*) d\rho \quad (4)$$

that upon substitution of Eq. 2-3 and setting $\phi = 0$, becomes

$$I(\omega) \approx 2S(\omega) \Re \left[\int_{-\infty}^{\infty} \Pi(\omega, \rho - \rho_0) \xi(\rho)^2 d\rho \right] \quad (5)$$

where $S(\omega) = s(\omega)^2$ is the source spectral intensity profile. An OCT axial scan (A-Scan) $\tilde{r}(t)$ is determined either from the Fourier transform of Eq. 5, i.e. Fourier domain or swept source OCT,

$$\tilde{r}(t) = \mathcal{F}[I(\omega)] = \frac{1}{\sqrt{2\pi}} \int_{-\infty}^{\infty} 2S(\omega) \Re \left[\int_{-\infty}^{\infty} \Pi(\omega, \rho - \rho_0) \xi(\rho)^2 d\rho \right] \exp(i\omega t) d\omega, \quad (6)$$

or from its integral over all frequencies evaluated at different values of ϕ , i.e. time domain OCT,

$$\tilde{r}(t) = \int_{-\infty}^{\infty} 2S(\omega) \Re \left[\int_{-\infty}^{\infty} \Pi(\omega, \rho - \rho_0) \xi(\rho)^2 d\rho \right] \cos(\phi) d\omega. \quad (7)$$

Eq. 7 therefore has the form of a Cosine transform which can for our purposes be treated equivalently as the related Fourier transform. Furthermore, these equations clearly show the de-coupled nature of axial and lateral resolution in OCT.

2.2 Axial Resolution

Eq. 5 is useful for elucidating the origin of Axial resolution in OCT. In both Eq. 6 and Eq. 7, the A-Scan is an integral transform of the product of the source spectrum and an

object function $\Pi(\omega, \rho_0)$ describing the sample. It is well known[12] that multiplication of two functions in the frequency domain is equivalent to a convolution in the time domain. Therefore, the OCT A-Scan is the result of convolving the source spectral intensity profile with an object function. At a given lateral position ρ_0 the object function can be represented by a simple summation model[13] over N discrete interfaces,

$$\Pi(\omega, \rho_0) = \sum_{j=1}^N r(\rho_0)_j \exp[i\varphi(\rho_0)_j]. \quad (8)$$

The optical thickness τ_j of each layer is represented by the phase term $\varphi_j = \omega\tau_j$ and the field reflectivity of the j th interface is given by r_j . Therefore, the A-scan function $\tilde{r}(t)$ has a magnitude proportional to the Fresnel coefficient of each interface within the sample, or for a scattering sample, the back-scattered field reflectivity. A mirror-like sample, representing a single interface ($N = 1$), will manifest as a single peak in the A-Scan, representative of the axial PSF. The PSF width is determined by the coherence length δ_z of the light source employed[14]. For a Gaussian source function, the coherence length is defined in terms of the PSF FWHM. Assume a light source spectral intensity profile with peak intensity S_0 ,

$$S(\omega) = S_0 \exp \left[-\frac{(\omega - \omega_0)^2}{w_\omega^2} \right]. \quad (9)$$

In Eq. 9 ω_0 is the source centre frequency and w_ω is the 1/e half-width of the source spectrum. The frequency FWHM $\Delta\omega$ is related to w_ω in the usual way

$$w_\omega = \frac{1}{2} \frac{\Delta\omega}{\sqrt{\ln(2)}}. \quad (10)$$

From Eq. 6 for a single reflective interface and ignoring amplitude coefficients, the PSF or temporal coherence function is given by Fourier transformation of the source spectrum.

$$\text{PSF}_z(t) = \mathcal{F}[S(\omega)] \propto \exp \left(-\frac{t^2}{W_t^2} \right), \quad (11)$$

which is a Gaussian with 1/e half-width W_t , defined as

$$W_t = \frac{1}{2} \frac{2}{w_\omega}. \quad (12)$$

The factor of 1/2 in Eq. 12 occurs due to the double pass of the probe light through the sample, which is typical for reflection mode interferometry and is implicit in OCT imaging. Following the general form of the relationship in Eq. 10, the FWHM of the PSF, or coherence time t_c , can be written in terms of W_t as

$$t_c = W_t \sqrt{\ln(2)} \quad (13)$$

and substituting for W_t and then w_ω leads to

$$t_c = 4 \frac{\ln(2)}{\Delta\omega}. \quad (14)$$

The frequency FWHM can be related to its wavelength equivalent by approximating the differential operator as finite steps in frequency $\Delta\omega$ and in wavelength $\Delta\lambda$, with speed of light c and source centre wavelength λ .

$$\frac{\Delta\omega}{\Delta\lambda} \approx \frac{d\omega}{d\lambda} = -2\pi \frac{c}{\lambda^2} \quad (15)$$

Solving 15 for $\Delta\omega$, substituting into Eq. 14 and multiplying by c yields the usual expression for coherence length,

$$\delta_z \approx 0.44 \frac{\lambda^2}{\Delta\lambda}. \quad (16)$$

This expression is a measure of the source coherence length in terms of electric field amplitude and is widely used in the OCT literature to define the axial resolution in terms of the source bandwidth $\Delta\lambda$.

2.2.1 Spectral Shaping

There exist a number of super-luminescent light emitting diode sources whose spectral profile is approximated well by a Gaussian function. However, alternative broadband light sources such as frequency swept lasers and super-continuum light sources typically do not have such well-behaved shapes, and as a result suffer from side-bands in the coherence function that manifest as ghosting and blurring in OCT images. With the advent of frequency domain detection in OCT[15], it is commonplace to de-convolve the light source spectrum from the OCT signal and apply a well-behaved window function, whose Fourier transform is free from side-bands[16, 17]. An example of a typical window is the Hann function[18]

$$H(\omega) = \cos^2 \left[\frac{\pi(\omega - \omega_0)}{2\Delta\omega} \right]. \quad (17)$$

The corresponding coherence length is

$$\delta_z^{\langle \text{Hann} \rangle} = \frac{\pi c}{\Delta\omega} = \frac{\lambda^2}{2\Delta\lambda}. \quad (18)$$

2.2.2 Effect of Refractive Index on Axial Resolution

It is worth noting that Eq. 16 and Eq. 18 assume values for λ and $\Delta\lambda$ in a vacuum. The influence of sample refractive index on the coherence length is included by dividing δ_z by the sample refractive index n_2 . However, it is the vacuum axial PSF that is actually measured directly. Therefore, for the purpose of comparison all values for δ_z presented in what follows are quoted for a vacuum, with no adjustment made for refractive index.

2.3 Dispersion

Refractive index, however, does play a role in the loss of axial image resolution in real materials. Typically, n_2 is not a constant, but varies as a function of optical wavelength, causing source light of different wavelengths to travel at differing speeds within the sample. This effect, known as dispersion, leads to broadening of the axial PSF of an OCT system that degrades the image resolution as a function of depth[19]. This can be seen by writing a simplified expression for the spectral OCT signal $I(\omega)$ from a bulk mono-layer as a function of angular frequency[20].

$$I(\omega) = S(\omega) \exp[-i2\phi(\omega)]. \quad (19)$$

The phase term $\phi(\omega) = \beta(\omega)z$ describes light propagation within the sample. For a dispersive medium, the propagation function $\beta(\omega) = n(\omega)\omega/c$ can be written as a Taylor series expansion about the source centre frequency ω_0 .

$$\phi(\omega) = \left[\beta_0 + \beta_1(\omega - \omega_0) + \frac{1}{2}\beta_2(\omega - \omega_0)^2 \dots \right] z \quad (20)$$

The coefficients β_m are defined in the usual way[21] for a Taylor series

$$\beta_m = \left[\frac{d^m \beta}{d\omega^m} \right]_{\omega=\omega_0} \quad (m = 0, 1, 2, \dots). \quad (21)$$

For most applications of practical interest it is sufficient to expand up to second order. In this case, the Fourier transform of Eq. 19 can be evaluated analytically as

$$|\mathcal{F}[I]| = \exp \left[\frac{-w_\omega^2 (t + 2z\beta_1)^2}{4w_\omega^4 z^2 \beta_2^2 + 4} \right], \quad (22)$$

that has the form of a Gaussian function, whose 1/e width parameter W_β is written,

$$W_\beta^2 = \frac{4w_\omega^4 z^2 \beta_2^2 + 4}{w_\omega^2}. \quad (23)$$

Therefore, it is possible to write the resolution broadening factor $\eta(z)$ as the ratio of the temporal envelope width W_β at depth z into a dispersive medium, with the non-dispersive width W_t , where W_t has already been defined in Eq. 12.

$$\eta(z) = \sqrt{w_\omega^4 z^2 \beta_2^2 + 1} \quad (24)$$

Eq. 24 assumes that higher order dispersion can be neglected. However, Hillman and Sampson[19] have previously shown that for ultrahigh resolution OCT systems, 3rd order and higher dispersion terms play an important role in distorting and broadening the axial PSF. For general use, however, Eq. 24 can be used to provide a simple estimate of the effect of dispersion.

2.4 Lateral Resolution

The lateral resolution in OCT is governed by the same physical principles as in standard light microscopy. The sample and reference beams will always have a finite spatial extent, limited by diffraction. The spatial characteristics of these illuminating fields are accounted for in Eq. 2-7 by the expression $\xi(\rho)$, that at the focus of a Gaussian beam becomes

$$\xi(\rho) = \exp \left[-\frac{\rho^2}{w_0^2} \right]. \quad (25)$$

The beam waist parameter w_0 gives the field 1/e radius in the focal plane of the beam. The integral inside the real part (denoted $\Re[\cdot]$) of Eq. 5 takes the explicit form of a convolution, from which it is evident that images formed in the lateral plane are a convolution of the object function with the product of the sample and reference illuminating spatial field profiles. Assuming an infinitely narrow object function in the lateral dimension,

$$\Pi(\rho) = \begin{cases} 0, & |\rho| > 0 \\ 1, & \rho = 0, \end{cases} \quad (26)$$

and the spatial illumination function Eq. 25, then the lateral detection PSF in terms of field amplitude is given by

$$\text{PSF}_{xy}(\rho_0) = \exp \left[-\frac{\rho^2}{w_\rho^2} \right] \quad (27)$$

where the 1/e field radius is $w_\rho = w_0/\sqrt{2}$, representing a $\sqrt{2}$ resolution improvement over the illuminating Gaussian beam. Conveniently, OCT often employs a single mode optical fibre coupled light source, and hence a Gaussian spatial illumination profile is generally valid. Under these conditions, assuming a monochromatic light source, the depth varying transverse PSF 1/e radius $w_{xy}(z)$ is given in terms of electric field amplitude by[22]

$$w_{xy}(z) = w_\rho \sqrt{1 + \left(\frac{z}{z_R}\right)^2}, \quad (28)$$

reducing to w_ρ at the focus ($z = 0$). The Rayleigh range of the incident field $z_R = \pi w_0^2 \lambda^{-1}$. Eq. 28 is related to the effective detection numerical aperture (NA_ρ) of the objective lens by evaluating $w_{xy}(Z)$ at some distance Z , where $Z/z_R \gg 1$. In this case, Eq. 28 reduces to

$$w_{xy}(Z) \approx w_\rho \frac{Z}{z_R}. \quad (29)$$

Assuming the paraxial approximation, the effective NA is equal to the angle of divergence $\text{NA}_\rho \approx \alpha \approx w_\rho/z_R$, and

$$2w_\rho = \frac{\sqrt{2}\lambda}{\pi \text{NA}} \approx \frac{1.22}{\sqrt{2}} \frac{\lambda}{2\text{NA}}, \quad (30)$$

thus demonstrating equivalence between the Gaussian beam approximation and Abbe's rule for a diffraction limited spot focussed through a circular aperture. The numerical aperture of the illuminating optical field is $\text{NA} = \sqrt{2}\text{NA}_\rho$. For consistency with Eq. 16, we express the lateral field PSF FWHM δ_{xy} as

$$\delta_{xy} = 2w_\rho \sqrt{\ln 2}. \quad (31)$$

2.4.1 Effect of Refractive Index on Lateral Resolution

Regarding the effect of refractive index on the lateral PSF width, inspection of Eq. 30 shows that w_0 is linearly dependent upon both λ and the NA. Within a medium of refractive index n_2 these become n_2/λ and n_2/NA respectively and therefore, the effect of refractive index is cancelled, leaving the PSF width unchanged.

2.5 Spherical Aberration

While optical dispersion can degrade OCT axial resolution, wavefront aberrations can cause degradation of the lateral resolution. Even in the case of ideal optics, in general spherical aberration will be introduced by focussing through the refractive index discontinuity at the surface of a sample, unless some index matching scheme is implemented using water immersion optics for example. This can be corrected for at specific depths within a known medium by using specially designed optics, or dynamically by implementing adaptive optics wavefront correction[23]. Assuming that no attempt has been made to correct for spherical aberration, this may limit the OCT lateral resolution. Török *et al* have previously found analytical solutions to Maxwell's vector equations for the case of focusing through single[24] and multiple[25] planar layers of different refractive index. This was later applied to the problem of finding the PSF of a conventional or fluorescence microscope by Haeberlé *et al*[26]. Aberration upon focussing through a refractive index discontinuity has also been

treated by Booth *et al*[27] following a scalar approach. In this latter method, the aberrated intensity point-spread function in the plane of the Gaussian focus, at depth z , is written

$$|h(r)|^2 = \left| \int_0^1 \exp[ik\psi(z, \varrho_n)] J_0(krNA\varrho_n) \varrho_n d\varrho_n \right|^2 \quad (32)$$

where $k = 2\pi/\lambda$ is the wavenumber, ϱ_n is the normalised radial co-ordinate, r is the physical radial co-ordinate and J_0 is the zero order Bessel function of the first kind. The numerical aperture $NA = n_1 \sin \alpha$ is defined in terms of the angular semi-aperture of the focussing objective in the first medium α and the corresponding refractive index n_1 . The angular semi-aperture in the second medium β is related to that in the first medium via Snell's law ($n_1 \sin \alpha = n_2 \sin \beta$) by the expression $\beta = \arcsin(NA/n_2)$. Aberration induced by refraction at the interface between the two media is described by an aberration function

$$\psi(z, \varrho_n) = f(\varrho_n)zNA \quad (33)$$

where

$$f(\varrho_n) = \sqrt{\csc^2 \beta - \varrho_n^2} - \sqrt{\csc^2 \alpha - \varrho_n^2} \quad (34)$$

and $\csc x = 1/\sin x$ is the cosecant of x . In order to examine the PSF at the actual beam focus, defocus must be removed from the above expression. This is accomplished by replacing ψ by a modified expression ψ' in Eq. 32.

$$\psi'(z, \varrho_n) = \psi(z, \varrho_n) - [A_{00} + A_{20}Z_{20}(\varrho_n)]zNA \quad (35)$$

whereby $A_{n0} = B_n(\alpha) - B_n(\beta)$ are Zernike coefficients and

$$B_n(\gamma) = \left[1 - \frac{n-1}{n+3} \tan^2\left(\frac{\gamma}{2}\right) \right] \frac{\tan^{n-1}(\gamma/2)}{2(n-1)\sqrt{n+1}}. \quad (36)$$

Z_{20} is the zero kind Zernike circle polynomial of order two that describes defocus in the optical system.

$$Z_{20}(\varrho_n) = \sqrt{3}(2\varrho_n^2 - 1) \quad (37)$$

Departure of a focussed spot from the diffraction limit due to wavefront aberration is often expressed in terms of the Strehl ratio Σ [14],

$$\Sigma = \frac{|h(0)|^2}{|h'(0)|^2}, \quad (38)$$

where h and h' represent the un-aberrated and aberrated PSFs respectively. A Strehl ratio greater than 0.8 is considered to be diffraction limited, however below this the optical performance of a system will become considerably impaired.

2.6 Point-Spread Function

As seen from Eq. 6-7, the images obtained from an OCT imaging system are a convolution of the PSF in the relevant dimension with the sample object function in that same dimension. In order to determine the PSF experimentally, it follows that a sample with an object function substantially smaller than the PSF is required. Practically, however, the sample will always have a finite size, it is therefore important to determine the limits on this with regard to PSF measurement. A single particle of finite diameter Δ can be reasonably

modelled by a top-hat function $\tilde{\Pi}(t) = \Phi(t + \Delta/2) - \Phi(t - \Delta/2)$, where $\Phi(x)$ is the unit step function, defined such that

$$\Phi(x) = \begin{cases} 0, & x < 0 \\ 1, & x \geq 0. \end{cases} \quad (39)$$

For a Gaussian source spectrum as in Eq. 9, convolution with $\tilde{\Pi}(t)$ can be written analytically as

$$\text{PSF}(t) * \tilde{\Pi}(t) = \sqrt{\frac{\text{erf}\left[\frac{1}{2}(\Delta - 2t)\sqrt{2/W_t^2}\right] + \text{erf}\left[\frac{1}{2}(\Delta + 2t)\sqrt{2/W_t^2}\right]}{2\text{erf}\left[\frac{1}{2}\Delta\sqrt{2/W_t^2}\right]}} \quad (40)$$

where $\text{erf}(x)$ is the error function and $*$ is the convolution operator. In the axial dimension, the PSF is not diffraction limited in the conventional sense for optical imaging and therefore $\tilde{\Pi}(t)$ is a good approximation for a transparent particle. However, a particle having lateral dimensions of the same order as the wavelength of the illuminating field will result in a complex backscattered field structure. In this case, the extent of validity of Eq. 40 is yet to be fully investigated.

3 Design Consideration for PSF Phantoms

To realise a practical PSF phantom for OCT, our group have opted for a design consisting of an epoxy resin matrix (Araldite DBF with HY951 hardener, Aeropia Resins, UK) containing a low density of mono-dispersed spherical glass particles (Whitehouse Scientific Inc., USA)[28]. When designing such phantoms, the principle consideration is that the particle size is sufficiently small to allow direct PSF measurement in both lateral and axial dimensions of the target instrument. Our OCT system (EX-1301, Michelson Diagnostics Ltd, UK) has a Hann window band-limited source FWHM of $\Delta\lambda = 75$ nm at a centre wavelength of $\lambda = 1310$ nm and hence from Eq. 18 an expected FWHM coherence length of $\delta_z = 11.4$ μm . The PSF FWHM is expected to be $\delta_{xy} = 8.4 \pm 0.2$ μm for measured illumination numerical aperture of $\text{NA} = 0.059 \pm 1 \times 10^{-3}$ in agreement with previously reported specifications[29]. In order to estimate a suitable particle size to measure the PSF of our OCT system, Eq. 40 was used to simulate PSF measurements with particles in the size range 0.1 to 10 μm assuming PSF widths of 8.0 and 10.0 μm . According to the numerical results shown in Fig. 1, the particle should have a diameter $< 15\%$ of probe PSF FWHM. For the system used for the present work, this limits the sphere diameter to < 1 μm . Another important design consideration is optical dispersion. The refractive index of the cured epoxy resin was measured using an Abbe refractometer (Abbe 60, Bellingham and Stanley, UK) over the wavelength range 436 nm to 1600 nm with an uncertainty of 10^{-4} . The measured dispersion curve is shown in Fig. 2. From this data we were able to obtain an estimate of $\beta_2 = 7 \times 10^{-26}$ $\text{s}^2\text{m}^{-1}\text{rad}^{-1}$ for the epoxy. This is compared with 2×10^{-26} $\text{s}^2\text{m}^{-1}\text{rad}^{-1}$ for water. Using Eq. 24 the broadening of the axial PSF from our system was examined for penetration depths of up to 2 mm in both the epoxy and water, Fig. 3. The dispersive nature of our resin in the 1300 nm waveband leads to PSF broadening that is approximately 10 times that of water for our system. However, up to 2 mm the effect is still small. For a typical penetration depth of around 1 mm, broadening is less than 1% and is therefore not expected to make any significant contribution to the results. Because the particles are suspended within a medium of differing refractive index to the surrounding medium, i.e. air, it is important also to assess the impact of spherical aberration on the lateral PSF. In

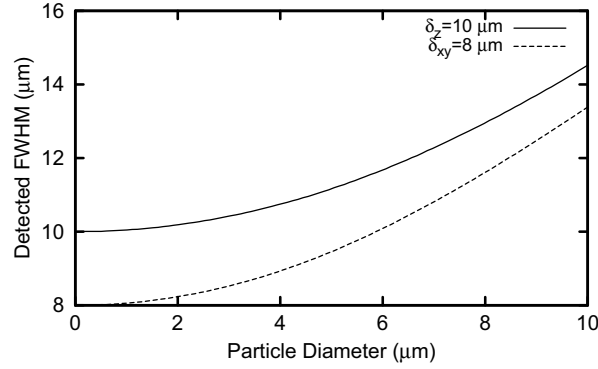


Figure 1: Simulated measurement of the axial PSF width (solid line) and the lateral PSF width (dashed line) as a function of particle diameter using Eq. 40 for axial and lateral PSF widths of $10 \mu\text{m}$ and $8 \mu\text{m}$ respectively.

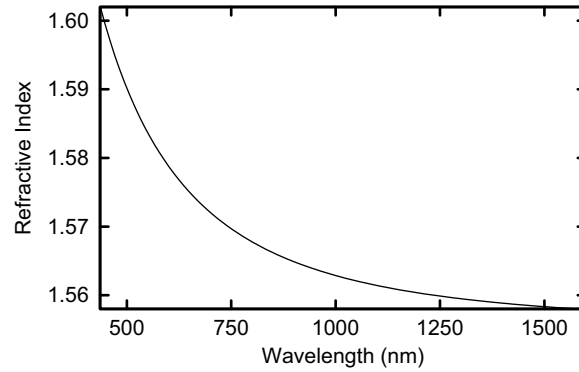


Figure 2: Refractive index of cured DBF resin measured over the wavelength range 435.8 nm to 1600 nm.

Fig. 4 we show the depth within our epoxy to which an optical instrument may focus and maintain a Strehl ratio of 0.8 as a function of the objective numerical aperture. Clearly, for $\text{NA} < 0.4$ and $z < 0.5 \text{ mm}$ spherical aberration can be neglected. In fact, for most practical imaging depth ranges, an NA less than 0.2 will incur no significant aberration penalty due to focussing through a refractive index mis-match.

4 Phantom Preparation

Our PSF phantoms are formed from an epoxy resin matrix (Araldite DBF) and hardener (Araldite HY951), doped with a low density of mono-dispersed pure silica microspheres (Whitehouse Scientific MS0001). In the present work we have chosen to use $1 \mu\text{m}$ diameter particles, which according to our previous theoretical results[30] is sufficiently small to measure axial and lateral PSFs down to $7 \mu\text{m}$ (field amplitude FWHM) respectively. In order to avoid speckle we have preferred to maintain an average point spacing of approximately $30 \mu\text{m}$, this was achieved by using a 0.003% volume ratio of micro-spheres to resin. The point spacing was verified from the OCT data. The particles were mixed into the resin hardener

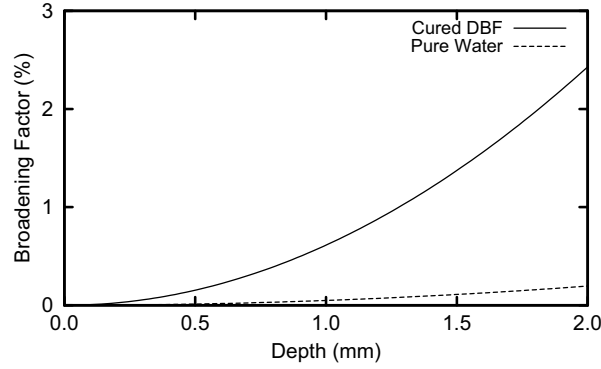


Figure 3: % broadening of axial PSF width as a function of physical depth propagated into DBF resin (solid line) and water (dotted line) plotted according to Eq. 24.

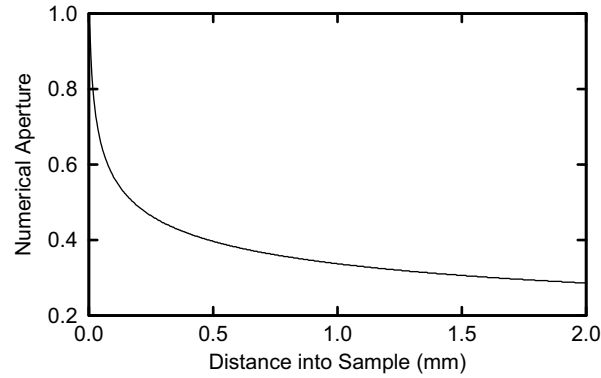


Figure 4: Depth within our epoxy to which an optical instrument may focus and maintain a Strehl ratio of 0.8 as a function of the objective numerical aperture. Epoxy refractive index is taken to be $n_2 = 1.56$.

prior to the addition of resin and distributed within the medium using an overhead stirrer. Whilst stirring, the resin-particle composite was held in an ultra-sonic bath to prevent agglomeration of the small spheres. When mixed, the resin was added into the composite at a ratio of 10:1 resin to hardener. Stirring was continued for a further 5 minutes to ensure final sample homogeneity. The final stage of preparing the composite mixture was pouring into 25 mm diameter cylindrical moulds to a nominal depth of 15 mm and degassing. The phantom mixture was held in the meld under a reduced pressure of 0.15 bar for 10 minutes and then left to cure at 23 deg for 12 hours. Room temperature curing was followed by a period of 6 hours heated cure at 70 deg, required to harden an uncured ‘skin’ that formed on the surface of the samples. It was found that the surface of the cured resin caused a stray reflection that saturated the OCT detection system, causing strong artefacts in the image plane that masked the single scattering points within the medium body. Therefore, our samples were polished to a flatness of few visible wavelengths across the 25 mm diameter by our optical workshop, verified using a Fizeau interferometer (Mk 2 Interferometer, Zygo, USA) and demonstrated by the typical result in Fig. 5. Over the surface it was essential to obtain a perfectly specular finish with no significant scratches or pits. The high quality

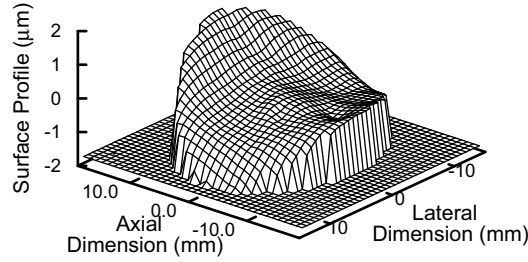


Figure 5: Surface flatness over the total area of the 25 mm diameter phantoms.

surface finish made it possible to substantially reduce the sample surface reflection by tilting the sample to approximately 5 deg from normal incidence such that the specular reflection component was directed away from the OCT detection optics. The surface profile was measured over a 1.8 mm lateral distance using a white light surface profilometer (Coherence Correlation Interferometer, Taylor Hobson, UK). A typical example of the surface profile is shown in Fig. 6.

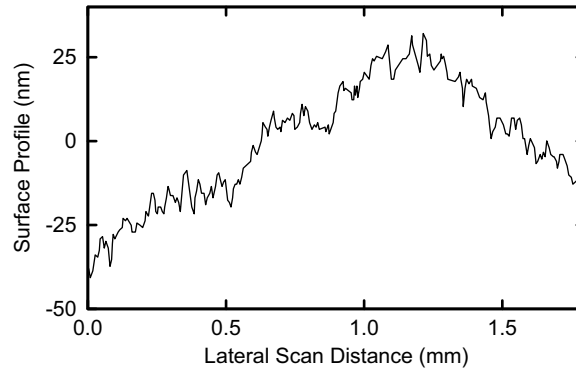


Figure 6: Typical phantom surface profile indicating local surface height variations of less than 100 nm.

5 Materials & Methods

PSF measurements of a commercial OCT instrument (EX-1301, Michelson Diagnostics Ltd, UK) were obtained using the phantoms described above. This OCT instrument utilises a swept laser source (HSL-2000, Santec, Japan) operating at a source centre wavelength of $\lambda_0 = 1310$ nm, over an acquisition bandwidth of 150 nm. The nominal axial PSF FWHM is $11.4 \mu\text{m}$, estimated from the effective coherence length of a Hann window applied over the source bandwidth as previously described[30]. The source spectrum was measured using an optical spectrum analyser (OSA) (MS9717A, Anritsu, Japan). The spectral acquisition band used by the OCT system was estimated by matching the OCT system internal reference

spectrum with the source spectrum measured directly using the spectrum analyser. The lateral resolution was measured to be $8.4 \pm 0.2 \mu\text{m}$ using a beam profiler. Beam profile measurements were used to determine the lateral resolution from images of the system spot size at 11 positions along the optical axis at a nominal distance of 15 mm from focus. Images were obtained using an InGaAs digital camera (Alpha NIR, Indigo Systems, USA) placed in the sample beam path. This was translated along the optical axis and images acquired at 1 mm intervals, Fig. 7. It was necessary to attenuate the OCT system optical power by

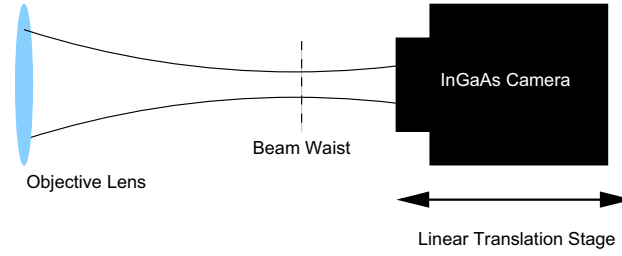


Figure 7: The beam profile is measured by placing an InGaAs camera in the OCT sample beam after the beam waist in the region of linear beam divergence. The camera was translated along the optical axis and beam profile measurements were obtained at 1 mm intervals over a range of 10 mm.

placing a fibre optic variable attenuator between the source and OCT microscope in order to avoid saturating the camera. Images were acquired using an integration time of $25 \mu\text{s}$. Measurements were repeated three times in order to estimate the uncertainty associated with the final result. The acquired data was calibrated for bad pixels, uniformity and linearity as described by Fatadin *et al*[31]. The $1/e$ half-width of the field at each position was estimated by taking the square root of each intensity beam profile and fitting a Gaussian function to the data. The $1/e$ radii were plotted as function of position along the optical axis, to which a straight line was fitted. The slope of the line provided an estimate of the system numerical aperture (NA), from which it was possible to estimate the OCT field PSF FWHM δ_{xy} using standard expressions for a Gaussian beam[30],

$$\delta_{xy} = \sqrt{2 \ln 2} \frac{\lambda_0}{\pi \text{NA}}. \quad (41)$$

A Gaussian beam approximation is valid in this case because our OCT instrument launches its source light from a single mode optical fibre. In order to measure sufficient data points to gain insight into the statistical distribution of results, measurements of the phantom were obtained over a three-dimensional volume. The volume comprised a series of B-Scans, each 4 mm wide by 3 mm in depth, with an A-Scan spacing of $4.4 \mu\text{m}$. The inter-B-Scan spacing was $4 \mu\text{m}$ over an acquisition range of 1 mm. The acquired data volume was processed to obtain the axial and lateral PSFs as follows. Initially the data was prepared as a series of sequential, depth resolved B-Scans, scaled linearly in field amplitude. This was achieved in the usual manner for a spectral domain OCT system. The spectral interference pattern corresponding to each A-Scan was divided by a reference spectrum, re-sampled from linear wavelength to linear frequency space, multiplied by a Hann window function and fast Fourier transformed (FFT) into the spatial domain. The linear field amplitude A-Scan was simply the magnitude of the FFT. A key component of the point extraction is background subtraction, which removes persistent image artefacts and interface signals from the dataset, leaving only the data points of interest. An estimate of the background signal is obtained

by taking a rolling mean of 20 adjacent B-Scans relative to each B-Scan. Here it is assumed that the background signal is slowly varying compared to the signal due to individual points that changes completely every few B-Scans, depending upon the B-Scan spacing and particle size. A background signal is then subtracted from each B-scan, leaving only the data-points of interest. Following background subtraction, a binary image of each B-scan was formed by setting all points greater than 90% of maximum field amplitude to one and all else to zero. The binary image makes it possible to estimate the location of each point in the original data, rejecting those with low magnitude. A 21×21 pixel data area was extracted around each located point in the original dataset, from which the 10th lateral and axial cross-sections were taken as an estimate of the PSF in each dimension. To each of these a Gaussian expression of the form

$$g(t) = A \exp \left[-\frac{(t - t_0)^2}{w^2} \right] + g_0 \quad (42)$$

was fitted using a non-linear least squares estimation based upon QR decomposition[32]. In Eq. 42, A represents the optical field amplitude of the detected signal, t_0 is the peak location and g_0 accounts for any systematic offset due to detection noise. The parameter of interest for PSF estimation is w which represents the $1/e$ half-width of the peak. This is related to the peak full-width at half maximum (FWHM) Δw by the expression

$$\Delta w = 2w\sqrt{\ln(2)}. \quad (43)$$

Although the PSF of an arbitrary spectral window is not strictly Gaussian[17], we have found that for well behaved functions such as a Hann window a Gaussian fit enables an accurate estimation of the peak FWHM, which is the parameter of interest. Typically, in the region of three-thousand points contribute to our PSF measurements. The axial and lateral PSF is estimated from a histogram of all values of Δw for a given dimension.

6 Results & Discussion

6.1 Source Spectrum Measurements

The source spectral bandwidth utilised by the OCT instrument was determined by comparing the instrument reference spectrum with the actual spectrum measured using an OSA. The resultant plots are shown in Fig. 8 which indicates that the OCT system acquires spectral data across a full bandwidth of 150 nm. For the purpose of data processing, a Hann window of the form

$$H(\omega) = \cos^2 \left[\frac{\pi(\omega - \omega_0)}{2\Delta\omega} \right] \quad (44)$$

was applied to the acquired spectral data. This avoids side-lobes in the OCT image data that can manifest as ghosting and blurring in B-Scan images. The corresponding FWHM coherence length was calculated to be $\delta_z = 11.4 \mu\text{m}$.

6.2 Beam Profile Measurements

Measurements of the instrument beam profiles are shown in Fig. 9. These show the increased spot diameter as the system is moved away from focus. Evident also is a slight asymmetry to the spot, particularly in Fig. 9a-c. This is likely to be because the system utilises four individual optical channels, each focussed at different depths, with a nominal $250 \mu\text{m}$ spacing

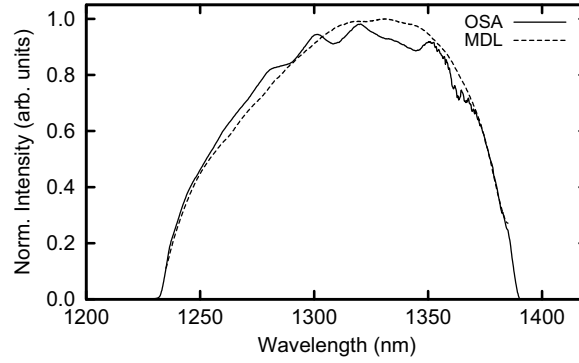


Figure 8: The OCT system source spectra as measured using an optical spectrum analyser (solid line) compared with the internal reference spectrum used by the MDL EX-1301 OCT instrument (dashed line).

between adjacent beam waists. Each channel also appears to be offset laterally, leading to the observed asymmetry. At larger distances from focus, the contribution from multiple channels is apparent as an increased uncertainty in the beam profile widths. However, this was found to have no effect upon our measurements of the beam divergence, the average of which is plotted in Fig. 10, along with the associated error bars representing 95% confidence. The linear fit can be seen to pass through the data points and yielded a system numerical aperture of $NA = 0.059 \pm 0.002$ and a corresponding focussed field illuminating spot FWHM of $7.1 \pm 0.2 \mu\text{m}$ in agreement with the theoretical predictions of Holmes *et al*[29] for this instrument.

6.3 PSF Phantom Measurements

In order to locate the micro-particles embedded within our phantoms, a background subtraction is necessary as described above. The result of this procedure is shown in Fig. 11 whereby strong image artefacts due to top-surface reflections and inter-reflections within the OCT instrument itself have been eliminated. Background removal does not always remove 100% of the background signal, as evidenced by the clustering of points along straight lines in Fig. 12b where all of the points acquired from an entire image volume have been plotted. Therefore, only the highest 10% magnitude points are selected for PSF measurement in an attempt to reduce noise in the analysis. This is demonstrated in Fig. 12a whereby a square has been drawn around each point identified by the 90% threshold. In Fig. 12, Clearly there are a large number of points that are visible but do not contribute to the final analysis. This conservative selection of PSF data-points serves to reject points with a low signal to noise ratio and background noise artefacts. A typical example of an extracted point is shown in Fig. 13. The plot shows that the point consists of multiple data points, indicating the possibility of achieving a successful Gaussian fit. This is further re-enforced in Fig. 14, where typical cross-sections are plotted in the axial and lateral dimensions including the fitted Gaussian curves. These plots demonstrate that there are sufficient data points within the selected peaks to obtain a fit of Eq. 42 to the data. Occasionally, despite the amplitude threshold, clumped points or background noise is selected for fitting. These can lead to a fit whose estimated width parameter or centre are outside the bounds of the 21×21 pixel selection grid and consequently such data points are also automatically rejected. Such rejection is valid

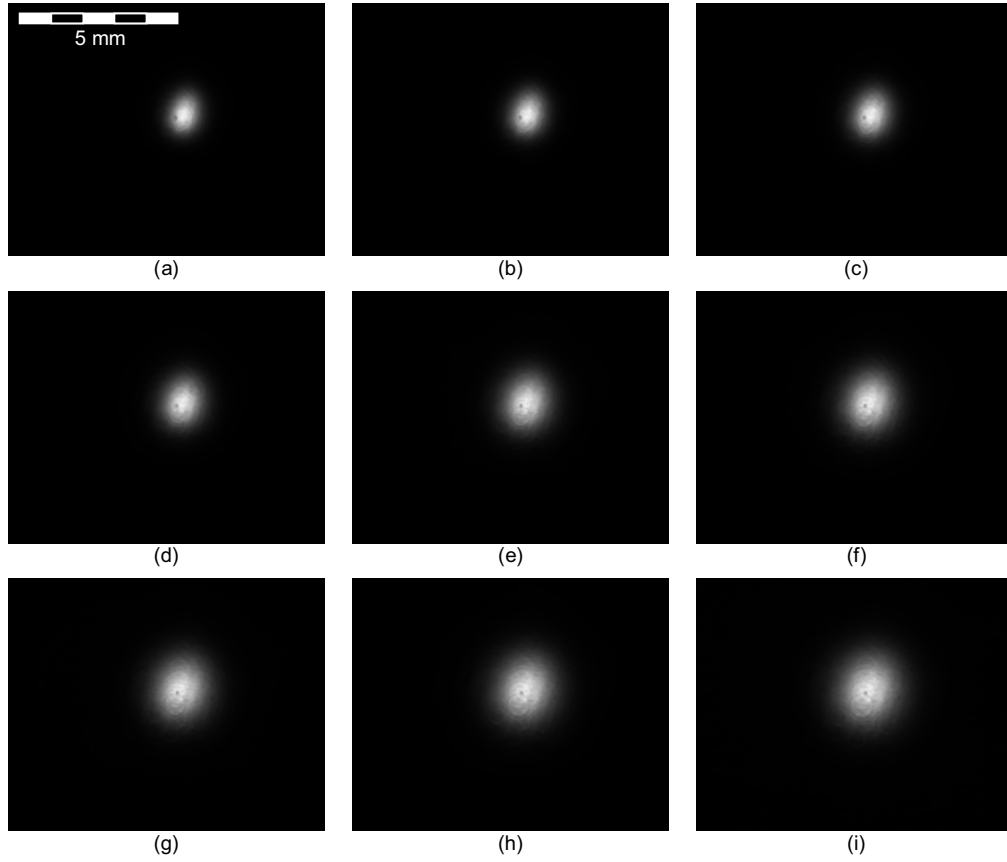


Figure 9: Beam profile measurements obtain at positions along the optical axis relative to the first measurement (a) 0 mm, (b) 1 mm, (c) 2 mm, (d) 3 mm, (e) 4 mm, (f) 5 mm, (g) 6 mm, (h) 7 mm, (i) 8 mm.

since it is known that any valid single point will be confined well within the selection grid bounds. Whilst the cross-sections in Fig. 14a and Fig. 14b correspond with the central axes of the selected data square in Fig. 13, it is true that they may not pass precisely through the centre of the sampled data peak. However, by obtaining cross-sectional samples from a large number of points and finding their width, we find that the statistical distribution associated with these measurements is Gaussian, demonstrated by the histograms plotted in Fig. 15a and Fig. 15b. The histogram peak locations correspond with the mean field PSF FWHM, yielding values for δ_z and δ_{xy} of $10.9 \pm 2.0 \mu\text{m}$ and $8.9 \pm 2.0 \mu\text{m}$ for the axial and lateral dimensions respectively, where the uncertainties represent 95% confidence. It is clear from Fig. 15a that the axial PSF width has a symmetric Gaussian distribution. This indicates that there is no significant axial PSF degradation over the image plane outside of the normal statistical uncertainty, that is likely dominated by the sampling each peak at not exactly its centre. In our related previous article[30] optical dispersion was discussed as the principle mode of axial resolution degradation and an expression for the depth dependent

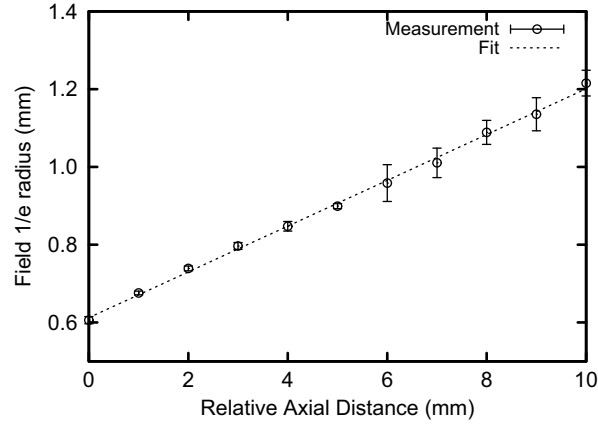


Figure 10: Beam field radius measurements (points) and linear fit to the experimental data (dotted line). The error bars represent twice the standard deviation.

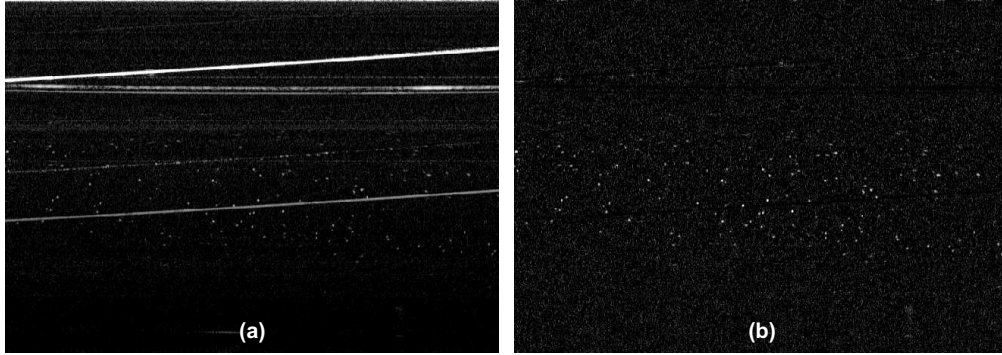


Figure 11: A typical OCT B-scan of a point-spread function phantom showing (a) the image before subtraction of constant noise terms and (b) after background subtraction.

resolution degradation factor was derived,

$$\eta(z) = \sqrt{\left(\frac{\pi c \Delta \lambda}{\lambda_0^2 \sqrt{\ln 2}}\right)^4 z^2 \beta_2^2 + 1}. \quad (45)$$

where c is the speed of light in a vacuum, $\Delta \lambda$ is the source FWHM bandwidth in wavelength, z is the physical depth into the medium and β_2 is the second order dispersion coefficient. The cured epoxy used to construct our phantoms has a dispersion coefficient $\beta_2 = 7 \times 10^{-26} \text{ s}^2 \text{ m}^{-1} \text{ rad}^{-1}$ at 1310 nm. Therefore, using our measurement of the PSF width above, the resolution degradation at the maximum imaging depth range of 1.0 mm is $\eta(z = 1.0 \text{ mm}) = 2\%$ or in physical units, peak broadening of approximately 200 nm. Hence, dispersion is not expected to be a significant contribution to the axial PSF measurement. The large outlying points are statistically insignificant and are predominantly due to residual background artefacts not sufficiently removed by the subtraction procedure outlined above. There is a 0.5 μm discrepancy between the phantom measured axial PSF FWHM and that expected from the source spectrum measurements, indicating that our estimation of the source acquisition

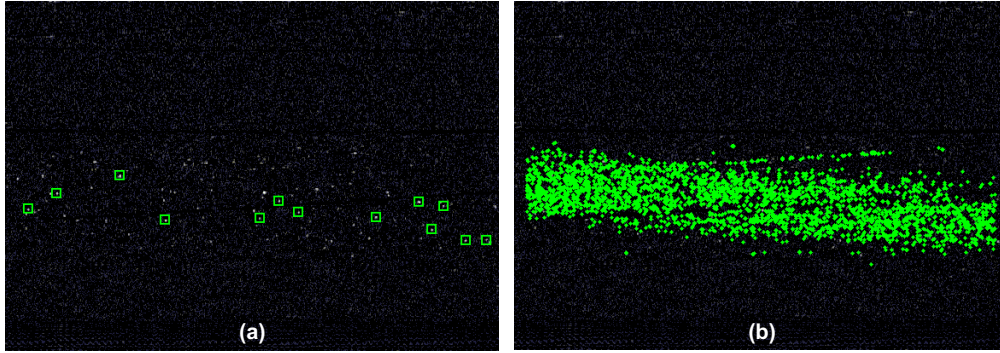


Figure 12: Points identified in (a) a single B-Scan and (b) an entire volume, from which the point-spread function is determined.

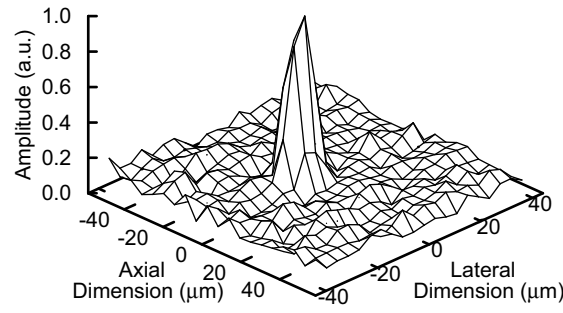


Figure 13: A typical 21×21 pixel data square extracted from the linear B-scan data and containing a signal from a single micro-sphere.

bandwidth is approximately 5 nm too small, which is unsurprising given the differences in shape between the measured and reference spectra in Fig. 8. Additionally, in processing the OCT data, the source spectrum is removed from the spectral interference pattern by dividing by the reference spectrum prior to application of the Hann window. The difference between the two spectra means that the Hann window is applied to a spectrum with some residual shape. This shape distorts the pre-shaped spectrum which can effect the resultant coherence length. Using PSF phantoms, the actual PSF is measured, thereby removing the ambiguity of such effects that remains in simple theoretical calculations. In the lateral dimension, it can be seen from Fig. 15b that there is a clear asymmetry in the distribution of measurements, forming a tail extending to larger FWHM values to the right of the main Gaussian peak. This additional contribution from larger PSF widths can be explained by the fact that PSF measurements are obtained over a depth range that extends significantly outside of the focus. Therefore, there is a contribution of results that are significantly greater than the focussed PSF. This has important implications because it makes it feasible to quote a mean lateral PSF width or resolution over the B-scan acquisition range as opposed to quoting the resolution at the focus of the objective, that in general will be higher than that achieved throughout the imaging plane.

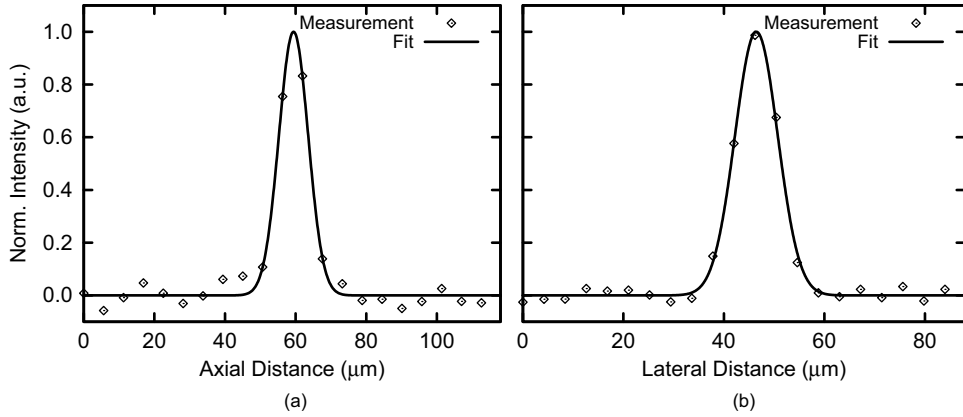


Figure 14: Typical cross-sections of a single point in the (a) axial and (b) lateral dimensions. The measured data (points) is approximated by a Gaussian fit (solid line).

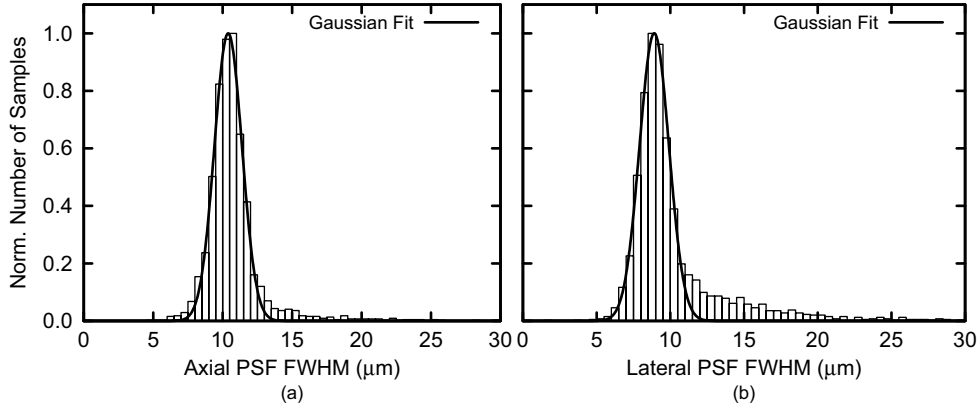


Figure 15: Histograms for the FWHM of the (a) axial and (b) lateral point-spread functions. The solid line represents a Gaussian fit to the distribution.

6.4 Spatial Distribution of Point-Spread Function

Measurements made using such PSF phantoms as those presented here have the advantage over alternative techniques that they can be used to analyse the spatial distribution of PSF throughout the B-scan image plane. This is demonstrated in Fig. 16a and Fig. 16b which show the spatial distribution of the axial and lateral PSF. A feature of interest that is particularly prominent in the axial plot (Fig. 16a), is the clustering of relatively large PSF values along straight lines in the image plane. This line corresponds with background artefacts, visible in Fig. 12, that have not been completely removed by the background subtraction. However, the axial PSF FWHM is generally uniform over the image plane. Large points identified in the spatial distribution can be identified as large outliers in the statistical distribution shown in Fig. 15a and are clearly a negligible contribution to the final result. However, as a function of depth, the lateral PSF is observed to vary considerably, whilst the axial PSF remains constant, in agreement with the statistical distributions.

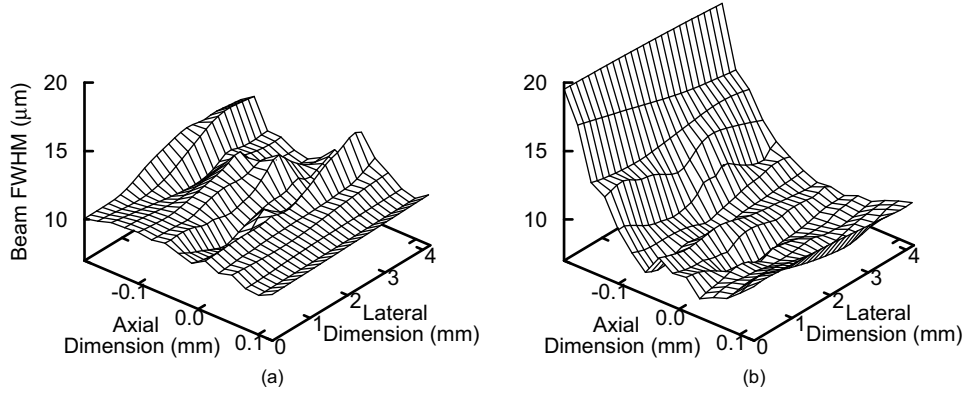


Figure 16: Spatial map of the lateral PSF over the B-Scan image plane in (a) the axial dimension and (b) the lateral dimension.

6.5 Characterisation of Multiple Channels

The OCT system under test in the present study comprises of four interferometer channels, focussed at successive depths. Using the PSF phantoms, it is possible to characterise the PSF performance of each channel. Composite datasets of multi-channel, lateral PSF results are shown in Fig. 17. These shown how the multiple focus technique maintains a lateral

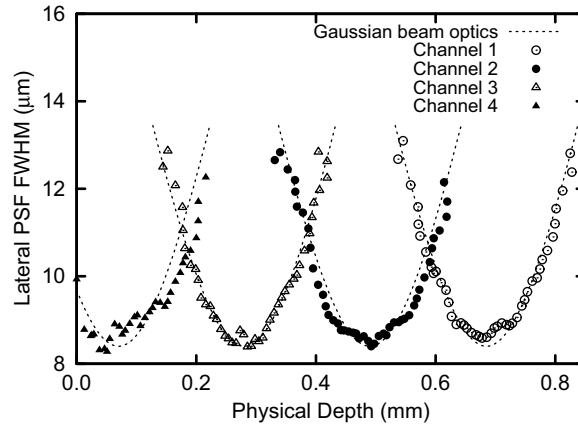


Figure 17: Measured field PSF FWHM as a function of physical distance along the axial dimension. The results clearly show the four channels with a focus spacing of approximately $230 \mu\text{m}$ along the optical axis. The expected curves for a Gaussian beam are shown for comparison (solid line).

resolution between 8 and $12 \mu\text{m}$ over a depth of 1 mm. We see no significant degradation of the lateral PSF width as a function of depth, indicating that spherical aberration caused by surface refraction is not a significant contribution. This is in agreement with predictions from our previous paper, that suggests a numerical aperture (NA) greater than 0.4 is required before spherical aberration becomes a significant source of resolution degradation at depths

greater than 0.5 mm.

7 Conclusion

In this report we have given a theoretical background for PSF in an OCT and introduced a set of design considerations for OCT PSF measurement phantoms. We have shown that in the lateral dimension, OCT achieves a factor $2^{-1/2}$ resolution enhancement over that expected from the illuminating field width. In addition, we have investigated potential sources of PSF degradation, namely optical dispersion and spherical aberration. In the OCT system and materials considered here, dispersion was found to be negligible. However, Eq. 24 can be used to show that for the resin matrix chosen here, dispersion will begin to become important for coherence lengths of approximately $5\text{ }\mu\text{m}$ at depths of around 0.5 mm . However, because the refractive index of our resin is well known over a wide spectral band, it would be straight-forward to mathematically remove the resin induced dispersion in post-processing. Nonetheless, future effort will explore materials having lower dispersion over the waveband of interest. Furthermore, it has been shown that for an imaging system having a numerical aperture less than 0.2 as is common in OCT, spherical aberration will generally be insignificant. For numerical apertures approaching 0.4 , spherical aberration will become a significant contribution to PSF distortion for imaging depths greater than 0.5 mm . This may become an important consideration for optical coherence microscopy (OCM)[33]. For the OCT system described in this report, the axial and lateral PSF was accurately evaluated by embedding glass microspheres of $1\text{ }\mu\text{m}$ diameter into an epoxy resin. We have shown that the particle diameter must be less than 15% of the incident PSF FWHM in order to obtain accurate measurements. Therefore, the $1\text{ }\mu\text{m}$ spheres used here are valid for characterising OCT instruments with axial and lateral PSF widths $> 7\text{ }\mu\text{m}$, where this represents the electric field FWHM for consistency with Eq. 16. In addition to the theoretical study, we have presented a robust resin based PSF phantom for OCT and demonstrated how this can be used to measure the axial and lateral PSF of an OCT instrument. We have demonstrated the ability to map spatially the PSF throughout the imaging plane and observed that for the present system, dispersion is negligible in the axial direction and spherical aberration is not significant in the lateral direction. We are presently investigating the application of an anti-reflection coating to the phantom surface to avoid the requirement to tilt the sample in order to remove the strong surface reflection. Furthermore, we are exploring the possibility of using surface replication techniques to avoid polishing each sample. In the present work it is necessary to process the results from raw data, scaled linearly in field amplitude. This is non-ideal since access to such data cannot always be guaranteed in a commercial instrument and therefore work is underway to develop phantoms that work with arbitrary scaling. Furthermore, real OCT images are viewed on log scale, therefore the validity of defining resolution in terms of the FWHM of the field amplitude PSF is perhaps questionable. However, our phantoms enable a physical realisation of the well known theoretical expressions, forming a baseline measurement from which further analyses can be conducted.

8 Acknowledgements

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