Characterizing the point spread function of retinal OCT devices with a model eye-based phantom

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Abstract: We have designed, fabricated, and tested a nanoparticle-embedded phantom (NEP) incorporated into a model eye in order to characterize the point spread function (PSF) of retinal optical coherence tomography (OCT) devices in three dimensions under realistic imaging conditions. The NEP comprises a sparse distribution of highly backscattering silica-gold nanoshells embedded in a transparent UV-curing epoxy. The commercially-available model eye replicates the key optical structures and focusing power of the human eye. We imaged the model eye-NEP combination with a research-grade spectral domain OCT system designed for in vivo retinal imaging and quantified the lateral and axial PSF dimensions across the field of view in the OCT images. We also imaged the model eye-NEP in a clinical OCT system. Subtle features in the PSF and its dimensions were consistent with independent measurements of lateral and axial resolution. This model eye-based phantom can provide retinal OCT device developers and users a means to rapidly, objectively, and consistently assess the PSF, a fundamental imaging performance metric.

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References

1. A. Agrawal, M. A. Gavrielides, S. Weininger, K. Chakrabarti, and J. Pfefer, “Regulatory perspectives and research activities at the FDA on the use of phantoms with in vivo diagnostic devices,” Proc. SPIE 6870, 687005, 687005-8 (2008).
2. R. J. Nordstrom, “The need for validation standards in medical imaging,” Proc. SPIE 7567, 756702, 756702-7 (2010).
3. IEC International Standard 62464–1:2007, “Magnetic resonance equipment for medical imaging—part 1: determination of essential image quality parameters” (International Electrotechnical Commission, Geneva, Switzerland, 2007).
4. IEC International Standard 61391–1:2006, “Ultrasound—Pulse-echo scanners—part 1: techniques for calibrating spatial measurement systems and measurement of system point-spread function response” (International Electrotechnical Commission, Geneva, Switzerland, 2006).
5. ISO/IEC International Standard 9919:2005(E), “Medical electrical equipment—particular requirements for the basic safety and essential performance of pulse oximeter equipment for medical use” (International Organization for Standardization, Geneva, Switzerland, 2005).
6. ISO International Standard 8600–5:2005, “Optics and photonics—medical endoscopes and endotherapy devices—part 5: determination of optical resolution of rigid endoscopes with optics” (International Organization for Standardization, Geneva, Switzerland, 2005).
7. T. G. van Leeuwen, D. J. Faber, and M. C. Aalder, “Measurement of the axial point spread function in scattering media using single-mode fiber-based optical coherence tomography,” IEEE J. Sel. Top. Quantum Electron. 9(2), 227–233 (2003).
8. W. Drexler, U. Morgner, F. X. Kärtner, C. Pitris, S. A. Boppart, X. D. Li, E. P. Ippen, and J. G. Fujimoto, “In vivo ultrahigh-resolution optical coherence tomography,” Opt. Lett. 24(17), 1221–1223 (1999).
1. Introduction

Standardized test methods for image quality assessment enhance the ability to perform objective, consistent and quantitative medical imaging tasks [1,2]. Well-characterized bench test methods can facilitate evaluation of specific imaging system parameters and comparison of different device designs. Monitoring imaging performance over time can provide quality assurance during extended periods of clinical use and research data collection, especially when imaging-based biomarkers are involved. For these and other reasons, standard image quality test methods for most major medical imaging modalities, such as magnetic resonance imaging and ultrasound, have been established and readily available in international consensus documents [3,4]. These methods often involve phantoms that replicate some physical properties of the target tissues and enable reliable assessment of properties such as resolution, contrast detectability, and signal uniformity. In spite of the many benefits of image quality standards, few have been established for optical diagnostic/imaging approaches [5,6].

The widespread adoption of optical coherence tomography (OCT) as a clinical technique for imaging the retina has created a strong need for validated evaluation methods. Currently, retinal OCT image quality is typically assessed qualitatively with in vivo imaging studies. However, such assessments are time consuming and costly, and they do not provide the validated, quantitative data that are necessary for establishing performance standards. Thus, the OCT field will benefit from tissue-phantom-based test methods that provide quantitative, verifiable performance measures and rigorous characterization of image quality.

Resolution is arguably the most fundamental image quality characteristic for medical imaging, as it defines the sharpness of an image. In OCT, the physics of axial and lateral resolution are generally decoupled. Lateral resolution is typically governed by Gaussian beam propagation, focusing, beam scanning pattern, and aberrations of the OCT optical system and the eye, while axial resolution depends primarily on the coherence length of the light source, along with dispersion and absorption in the optical system. Standard approaches for assessing
OCT system axial resolution have typically involved imaging a specular surface such as a mirror [7]. Lateral resolution is typically calculated based on the profile of the OCT illumination beam [8] or measured with a resolution target [9]. These approaches are limited in that they do not characterize variations in resolution throughout the 3D imaging volume, and they are difficult to implement in a geometry that simulates the human eye.

One of the key methods of describing resolution is the point spread function (PSF), as the image formed by an instrument is the convolution of the PSF with the true dimensions of a structure. Our group and others have designed and fabricated phantoms incorporating high-contrast, near- or sub-resolution particles to enable characterization of axial and lateral PSFs across the entire three-dimensional OCT image field. In our prior work we introduced a nanoparticle-embedded phantom (NEP) comprising gold nanoshells embedded in silicone [10], as these plasmonic nanoparticles have been shown to generate strong backscattering signals at 1300 nm [11]. Other approaches have included resin-based phantoms with silica microspheres [12] and submicron iron oxide particles [13,14]. This prior work also demonstrated that NEP-based PSF measurement can be used to improve the resolution of OCT images. Tomlins et al. investigated an alternate approach that involves inscribing sub-resolution defects in fused silica substrates with a femtosecond laser [15]. While the aforementioned research involved OCT systems near 1300 nm, studies by Davis et al. and Ralston et al. assessed the resolution of systems in the 800 nm range [16,17]. These articles described the use of 1 μm diameter TiO$_2$ particles suspended in silicone, although a significant assessment of the accuracy of this approach was not provided.

In order to assess OCT systems for retinal imaging—the predominant clinical application—an approach that accounts for the physiological optics of the human eye is necessary. This feature has been has been incorporated in recent studies involving tissue-simulating phantoms which approximate retinal morphology and optical properties. In one study, 50-μm-thick layers of silicone embedded with absorbers and scatterers were incorporated in a water-filled chamber with a lens [18]. The other study investigated a commercially-available model eye with a custom retinal phantom fabricated from rubber adhesive and 60-μm-layers of polypropylene film [19].

Here we describe the development and evaluation of a novel test method for quantitative evaluation of retinal OCT imaging performance. This method involves an NEP located at the retinal plane of an artificial model that replicates the focusing characteristics of the human eye. Validation is achieved through comparison with standard methods for axial and lateral resolution determination. Results include three dimensional PSF maps that graphically illustrate spatial variations in axial and lateral resolution.

2. Materials and methods

2.1. Nanoparticle-embedded phantom

Our NEP design consists of gold-silica nanoshells sparsely and randomly distributed in a highly transparent polymer. We followed a similar procedure as in our previous work [10] to fabricate the NEP in the current study, but with two key modifications: we used a different transparent polymer and performed a different molding process.

Previously we used silicone as the transparent polymer, but here we use a UV-curing epoxy (Light Weld 4-20577, Dymax Corp., Torrington, CT). The epoxy had similar transparency to silicone but with less inherent scattering, which can obscure the appearance of the nanoshells, particularly for OCT source wavelengths around 800 nm versus longer source wavelengths. We measured the absorption coefficient ($\mu_a$) of epoxy and silicone with a spectrophotometer (UV-3101PC, Shimadzu, Columbia, MD) using test samples of known thickness. The absorption spectra of the polymers, with water as a reference, are shown in Fig. 1. The epoxy has exceptionally low $\mu_a$ ($<$0.005 mm$^{-1}$) around the primary retinal OCT wavelength of 800 nm, while also remaining below 0.01 mm$^{-1}$ at longer wavelengths (i.e., 1000 nm) where OCT is now also being performed. The refractive index of this epoxy is 1.47,
Fig. 1. Absorption spectra of polymers used in NEPs (silicone: previous NEP, epoxy: current NEP). Water absorption is shown as a reference.

determined from measurements of OCT optical thickness with the test samples. These OCT images of the epoxy test samples also verified the absence of inherent scattering.

The nanoshells used in this version of the NEP had a silica core diameter of 213 nm and 19 nm gold shell thickness and were fabricated at Rice University as an aqueous suspension [11] with \( \sim 10^9 \) particles/mL\(^{-1}\). The desired concentration of nanoshells in the phantom was \( 10^7 \) mL\(^{-1}\), which corresponds to a mean spacing of 46 \( \mu \)m between particles and thus ensures that multiple nanoshells do not contribute to the PSF measurement at any given location. Ten \( \mu \)L of the nanoshell suspension was pipetted into a 1” diameter Delrin mold and the water was allowed to evaporate. Approximately three grams of epoxy was then mixed in with the nanoshells, first using a probe sonicator for 30 seconds to ensure minimal particle aggregation, then using a glass rod for two minutes to ensure a homogeneous distribution of particles. The mixture was then placed in a vacuum chamber for 20 minutes to purge air bubbles. We then placed a 1” diameter transparent acrylic sphere on top of the Delrin mold, creating a concave surface on the phantom matching the curvature of the inner retinal surface. The phantom mold was then placed underneath a UV LED curing lamp (CS2010, Thorlabs, Newton, NJ) for 20 minutes to cure. After curing, the acrylic sphere was removed and the finished NEP in the Delrin mold, shown in Fig. 2, could then be placed inside the model eye. The edges of the NEP were slightly damaged during removal of the acrylic sphere, but the central area (>10 mm\(^2\)) was smooth and intact. The NEP was 3.8 mm thick at the center, thin enough to allow adjustment of its depth within the eye.

Fig. 2. (a) Fabricated NEP in Delrin mold. (b) Model eye. Scale bar is 5 mm.

2.2. Model eye

We used a commercially-available model eye (OEMI-7, Ocular Instruments, Inc., Bellevue, WA) which mimics emmetropic human vision with a total power of 60 diopters. As shown in part (b) of Fig. 2 and in Fig. 3, the model eye realistically simulates the appearance and geometry of the key ocular structures, including the cornea, pupil, crystalline lens, aqueous and vitreous media, as well as the retinal surface. The cornea and lens are composed of
poly(methyl methacrylate) (PMMA, \( n = 1.49 \)), and deionized water (\( n = 1.33 \)) serves as the aqueous and vitreous media. The eye is modular, with the anterior chamber detachable from the posterior and the retina removable from the posterior chamber. This last feature allowed us to easily replace the included retina, which has an anatomically-correct surface appearance but no relevant volumetric structure, with our NEP.

![Fig. 3. (a) Engineering drawing and (b) schematic of the model eye.](image)

2.3. Laboratory OCT imaging and analysis

2.3.1. OCT setup

A custom-designed Fourier domain OCT (FDOCT) system (Physical Sciences, Inc., Andover, MA) [20] was used to evaluate the NEP in detail. This OCT system, designed for \textit{in vivo} retinal imaging, uses a broadband superluminescent diode (SLD) operating at a center wavelength of 855 nm with 56 nm full-width at half maximum (FWHM) spectral bandwidth, which for an ideal Gaussian spectrum yields a FWHM coherence length of 6 \( \mu \)m in air. The OCT beam is collimated to a 2 mm diameter, suitable for a variety of imaging tasks involving small or large pupils. This beam is scanned laterally, in horizontal (X) and vertical (Y) directions, by two galvanometer-mounted mirrors. Reflected light from the eye enters a transmission grating spectrometer, which employs a high-speed line scan camera operating at 28 kHz. A graphical processing unit (GPU)-based algorithm enabled acquisition of B-scan images with 1024 A-scans and 1024 points/A-scan at 18 frames/second. The spectrometer had a fixed axial (Z) range of 1.95 mm optical depth, thus the Z sampling interval was 1.95 \( \mu \)m per pixel in an A-scan.
The model eye was placed at the pupil conjugate in front of the OCT system, which represents the ideal position for clinical imaging since the scanning beam is pivoting about a fixed point upon entering the eye’s pupil. We first calibrated the X and Y galvanometer angles to lateral distance across the retina surface using 100 μm and 500 μm precision slits in place of the NEP at the focal plane of the model eye. With the NEP reinstalled, we raster scanned the beam over rectangular areas of the NEP surface, centered on the optical axis, and acquired a stack of B-scans during each raster scan. The B-scan width corresponds to the X dimension, and the B-scans were stacked along the Y dimension. We performed raster scans of two different sizes: 0.5 mm X × 0.1 mm Y (narrow scan) and 1.5 mm X × 0.15 mm Y (wide scan). The narrow scan consisted of 512 A-scans per B-scan and 128 B-scans, so the lateral sampling intervals were 1 μm in X and 0.8 μm in Y. The wide scan had 1024 A-scans per B-scan and 128 B-scans, yielding 1.5 and 1.2 μm lateral sampling intervals in X and Y, respectively. All sampling intervals were smaller than the expected minimum focused spot size of the OCT beam at the retina.

2.3.2. Validation of NEP-based PSF measurements

The lateral PSF of retinal OCT can be directly accessed via beam profiling, which we performed with a camera-based system (Beamstar FX-50, Ophir-Spiricon, North Logan, UT). With 9.9 μm × 9.9 μm camera pixels and a 22X objective lens as the camera’s front optic, the beam intensity could be spatially sampled at 0.45 μm intervals in both X and Y. In order to profile the beam being focused by the model eye, we mounted the anterior chamber of the model eye on the front of a water-filled tank with a submerged turning mirror to direct the beam upward to the camera’s front optic, which was just beneath the water’s surface. As a reference, we also computed idealized lateral beam profiles using ZEMAX-EE software (Radiant ZEMAX LLC, Bellevue, WA) to simulate the OCT beam propagating through the model eye and incident on the NEP. The software’s physical optics propagation (POP) feature provides a more accurate rendering of a real Gaussian beam in an optical system than geometric ray tracing.

We validated the axial PSF using a glass plate submerged in the same water-filled tank and measuring the specular reflection from the plate’s front surface. To minimize systematic error in lateral PSF width estimation, we also verified the X and Y spatial sampling intervals by imaging precision slits submerged in this tank.

2.3.3. Data analysis

We used data visualization software (Slicer Dicer, Pixotec LLC, Renton, WA) to create 3D renderings of B-scan stacks. We then used freely available code [21] written in MATLAB (Mathworks, Natick, MA) to automatically locate individual particles in the OCT data volumes. We wrote MATLAB code to quantify the PSF FWHM dimensions from each particle and create false-color maps of the PSF values over the imaged region. Similar MATLAB code quantified the lateral PSF widths from the beam profiling measurements and the axial PSF widths from the glass plate measurements.

2.4. OCT imaging with a commercially-available clinical device

We acquired images of the model eye-NEP with a commercial retinal FDOCT device, to provide an initial demonstration of the PSF measurement capabilities in a clinical setting. This device was capable of acquiring raster scans with ~10 μm spatial sampling intervals both in X (between A-scans) and in Y (between B-scans). The Z sampling interval was ~4 μm, in physical depth units (i.e., optical depth divided by tissue refractive index). We acquired raster scans with dimensions of 5.3 mm in X and 1.3 mm in Y.

Because the X and Y spatial sampling was on the order of the expected lateral PSF width, we limited our identification of nanoparticles to only those which visually appeared in both three adjacent A-scans and three adjacent B-scans. This approach enabled robust lateral PSF width estimation under the sparse sampling conditions by fitting each lateral PSF to a 2D Gaussian surface whose FWHM was then calculated.
3. Results and discussion

3.1. Laboratory OCT imaging and analysis

A representative B-scan and a 3D rendering of the B-scan stack from a narrow scan (0.5 mm X × 0.1 mm Y) are shown in Fig. 4. The Z dimension for both images in this figure has been truncated at an optical depth of ~1000 μm, below which no particles were visible. The B-scan is a raw image with logarithmic intensity scaling of the brightness. For the 3D rendering, the intensity in the local region enclosing each particle was normalized to that particle’s maximum intensity, thereby allowing uniform visualization of every particle’s PSF unbiased by intensity variations. The focal point (defined as the depth where the lateral PSF area is a minimum) is at approximately 340 μm optical depth, resulting from arbitrary positioning of the NEP within the eye. The beam focusing characteristics are readily apparent via the lateral size of the PSF.

![Fig. 4. (a) Representative B-scan and (b) 3D rendering of a B-scan stack acquired from a 0.5 mm X × 0.1 mm Y rectangular surface area of the NEP. Color scale is intensity in arbitrary units.](image)

3.1.1. Lateral PSF

Two-dimensional lateral cross sections of the PSF at different depths relative to the focal point are presented in Fig. 5. Images from representative individual nanoshells in the NEP [parts (a)-(c)] are similar to the corresponding beam profiler images [parts (d)-(f)]. The non-circular shape of the PSF above and below the focal point indicates astigmatism, wherein the focal length differs in the two lateral directions. The OCT system introduces astigmatism by using spherical mirrors off-axis for beam focusing. Observing the nearly identical astigmatism in the beam profiler images confirmed the NEP does not contribute to the aberration. We also confirmed that the model eye does not introduce its own astigmatism by replacing the model eye with a glass achromatic doublet and observing a similar amount of astigmatism in the NEP images (data not shown). Slight differences between corresponding NEP and beam profiler images result from experimental error in positioning the model eye with respect to the OCT beam during the two separate imaging sessions.

Figure 6 contains graphs of the lateral PSF dimensions versus depth. X and Y FWHM dimensions measured from the ~300 particles visible in the narrow scan of the NEP are compared to beam profiling results. Theoretical estimates from ZEMAX POP are shown for reference. The beam waist sizes are quite similar from all the methods. Here the astigmatism is apparent in the difference of the depth of the minimum PSF width between the X and Y
dimensions and as a deviation from the theoretical PSF width, most noticeably above the focus in the X direction. This deviation is coupled with the appearance of horizontal sidelobes in the 2D PSF images in part (a) of Fig. 5, likely caused by diffraction effects that narrow the central lobe and thus reduce the PSF FWHM. Most of the random variations in the NEP-derived width values are likely related to speckle, wherein the coherent mixing of the nanoparticle’s reflection with the reference beam can lead to random fluctuations in the PSF shape. Increased variability at increased depths results from the expected reduction in signal-to-noise ratio with deeper particles.

To also examine the dependence of lateral PSF across the OCT field of view, Fig. 7 shows false color maps of lateral PSF width versus X and Z for a wide scan (1.5 mm X × 0.15 mm Y). We have not mapped the lateral PSF versus Y because the scan yields no meaningful spatial dependence of the PSF across the 0.15 mm in Y. Instead, these maps combine the data points from the entire Y scan range. The color in Fig. 7 represents the X FWHM of the lateral PSF in part (a) and the Y FWHM in part (b). Linear interpolation of the discrete data points provides the spatial continuity in these maps, although the scatter in the PSF widths causes a heterogeneous appearance, particularly at greater depths. In this example, the NEP was positioned closer to the eye’s anterior such that the focal point was at about 700 μm optical depth, somewhat deeper than during the narrow scan. Again, the astigmatism is apparent by the difference in the apparent focal depth between the X and Y cross-sections of the PSF. The
A deeper focal point leads to noticeably larger values of the Y PSF width of approximately 14 μm, near the surface. In addition, the X width becomes slightly smaller (~4 μm) than the Y width (~5 μm) at the focal point, similar to the effect seen with the narrow scan in Fig. 6. But the wide scan offers the opportunity to see another effect, wherein the focal point rises by approximately 100 μm in optical depth across the 1.5 mm width of the scan, suggesting slight tilt and/or decentration of the OCT beam with respect to the optical axis of the eye.

### 3.1.2. Axial PSF

In Fig. 8, the axial (Z direction) PSF widths measured from the NEP are compared with values obtained using a glass plate. Unlike Fig. 6, here the horizontal axis is simply optical depth, because the focal point is irrelevant to the axial PSF width. Also, the vertical axis represents the FWHM in air and thus is independent of the refractive index. There is excellent agreement between the measurement methods. Furthermore, both results show a similar increase in axial PSF width with depth, indicating that the NEP material demonstrates no excess dispersion over water. This increase is primarily attributable to spectral interferogram k-space resampling error, which becomes more pronounced as the fringe frequency increases with depth, along with incomplete dispersion matching between sample and reference arms of the OCT interferometer. We also note that the axial PSF widths do not reach the ideal value of 6 μm primarily because the SLD output spectrum is not truly Gaussian.

### 3.2. OCT imaging with a commercially-available clinical device

In a 5.3 mm × 1.3 mm raster scan, we identified 106 nanoparticles with which to analyze the PSF. Surface plots of a single lateral PSF and its Gaussian surface fit are shown in Fig. 9 as an example. The lateral and axial PSF FWHM values are shown in Fig. 10 as a function of depth and X position. The lateral PSF values represent the average of the X and Y widths from each nanoparticle. We can see the clear depth dependence of both lateral and axial PSF, as expected from effects observed with our laboratory system. Though we might have expected dependence of the PSFs on lateral position due to off-axis beam propagation, there is little, if
any, systematic variation of the lateral PSF over the entire 5 mm scan width. The axial PSF also remains constant over the scan width, another expected result.

![Surface plots of (a) an example of a lateral PSF and (b) Gaussian fit to this lateral PSF.](image)

Fig. 9. Surface plots of (a) an example of a lateral PSF and (b) Gaussian fit to this lateral PSF.

![Lateral and axial PSF widths as a function of (a) depth and (b) X position. Solid lines represent polynomial fits to the data.](image)

Fig. 10. Lateral and axial PSF widths as a function of (a) depth and (b) X position. Solid lines represent polynomial fits to the data.

4. Discussion and conclusions

We have demonstrated the application of a model eye-NEP approach to obtain PSF measurements on two different OCT systems: a highly configurable laboratory system and a “black box” clinical system. The laboratory system provided a platform to corroborate phantom-based PSF measurements with those obtained with well-known methods. However, clinical systems typically do not offer the end-user access to the fine spatial sampling (i.e., smaller scan lengths) which provides the most accurate PSF representation. It may be possible to mitigate this spatial sampling limitation using a data analysis approach proposed by Williams et al [14]. Also, commercial OCT devices could benefit from comprehensive PSF analysis with the model eye-NEP during preclinical device testing, when the engineer or technician can perform high resolution scans.

Some properties of the model eye and NEP impact clinical applicability. The refractive indices of the model eye’s PMMA cornea and lens (n = 1.49), and the epoxy-based NEP (n = 1.47) are higher than those of the human eye’s cornea, lens, and retina (n = 1.35-1.39). The cornea and lens differences are mitigated since both the model eye and the human emmetropic eye possess similar dioptric powers and aqueous and vitreous media (n = 1.33-1.34); therefore, both eyes share similar f-numbers and corresponding Gaussian beam focusing characteristics. The main consequence of the high refractive index of the epoxy is that the focal point occurs deeper in the phantom than it would in the human retina. However, we can...
compensate for this by applying a scaling factor of $n_{\text{retina}}/n_{\text{epoxy}}^2 \approx 0.63$ to the original optical depth, to correct for increases in both the OCT pathlength and the Gaussian beam focal length to obtain an effective physical depth in the retina. In addition, the large index mismatch between the water and epoxy gives rise to a large OCT peak which can obscure PSF measurements near the NEP surface.

Also, the OCT image contrast of the NEP is much higher than the typical contrast between retinal structures. Therefore, the NEP cannot be used to test OCT sensitivity to subtle, low-contrast features in real retinas. However, the NEP provides a benchmark in the limit of spatial resolution under idealized conditions—an approach incorporated in medical imaging standards—and to obtain details of the PSF that would not be apparent with low contrast.

In conclusion, we have developed a tool for rapid, comprehensive, and objective PSF characterization of retinal OCT devices, based on a realistic model eye containing an NEP placed at the position of the retina. This phantom allows accurate dimensions of the PSF to be obtained, along with subtle features and variations in the PSF across the desired field of view. OCT developers and users can quantify the spatial resolution of their instrument in three dimensions under idealized conditions. The effects of deviations from the ideal, such as misalignment of the eye, can also be studied in a highly controlled manner. Such a tool has the potential to enhance preclinical bench testing of retinal OCT devices for research, manufacturing and regulatory purposes, as well as to ensure consistent and high quality imaging with these OCT devices routinely during clinical use.

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