Bessel beam spectral-domain high-resolution optical coherence tomography with micro-optic axicon providing extended focusing range

Kye-Sung Lee* and Jannick P. Rolland

CREOL, The College of Optics and Photonics, University of Central Florida, Orlando, Florida 32816, USA
**Corresponding author: kslee@creol.ucf.edu*

Received March 25, 2008; revised May 19, 2008; accepted June 16, 2008;
posted June 27, 2008 (Doc. ID 93704); published July 24, 2008

Endoscopic imaging in tubular structures, such as the tracheobronchial tree, could benefit from imaging optics with an extended depth of focus (DOF) to accommodate the varying sizes of tubular structures across patients and along the tree within the same patient. Yet the extended DOF needs to be accomplished without sacrificing resolution while maintaining sufficient sensitivity and speed of imaging. In this Letter, we report on the measured resolution and sensitivity achieved with a custom-made micro-optic axicon lens designed to theoretically achieve an 8 mm DOF. A measured invariant resolution of ~8 μm is demonstrated across a 4 mm measured DOF using the micro-optic axicon while achieving an invariant sensitivity of ~80 dB with a 25 mW input power. Double-pass Bessel beam spectral-domain optical coherence tomography with an axicon micro-optic lens (i.e., <1 mm in diameter) is, for the first time to our knowledge, demonstrated in a biological sample demonstrating invariant resolution and signal-to-noise ratio across a 4 mm measured DOF, which is compared to Gaussian beam imaging.

OCIS codes: 120.3890, 170.0110.

Optical coherence tomography (OCT) is an imaging modality that provides *in vivo*, noninvasive, high-resolution, cross-sectional images of biological tissues. Given the depth of penetration of a light beam through biological tissue, endoscopic probes are needed to interrogate the condition of wall linings within organs such as the lung, the stomach, the colon, and larger arteries inside the body. After the first endoscopic applications of OCT in 1996, various types of endoscopic OCT instruments were developed [1,2]. Current endoscopic probes use conventional Gaussian beam illumination. Some are moved to the targeted location by pushing and pulling a wire outside the body that is used to guide the catheter. These types of probes operate at low NA and provide ~20 μm lateral resolution. Other probes use higher NA optics; however, they suffer from a shorter depth of focus (DOF) (i.e., measured as the confocal parameter, e.g., ~140 μm confocal parameter for ~5 μm FWHM lateral resolution at 800 nm). For these probes, a fine focus adjustment is required that cannot be achieved by pushing and pulling a wire. A gradient index (GRIN) lens rod-based probe was then presented to focus on a targeted location in a sample by moving a stage outside the body [1]. However, the nonflexible rod limits some endoscopic applications. Thus the development of extended DOF imaging schemes is an active area of OCT research [3,4]. Bessel beam imaging, generated by axicon optics, is distinguished from Gaussian beam imaging in that it enables extension of the focusing range without loss of resolution [5]. The trade-off, however, is a decrease in illumination and collection efficiency and thus associated imaging sensitivity. The use of axicon optics in time-domain OCT was first investigated by Ding *et al.* [6]. Recently, using spectral-domain OCT (SD-OCT), Leitgeb *et al.* proposed a non-dual path imaging scheme for imaging of biological samples [7]. Researchers reported 1.5 μm resolution across a 200 μm DOF.

In this Letter, we report on the feasibility of using a custom-made micro-optic axicon to create a Bessel beam to illuminate and image in biological samples using SD-OCT. We quantify both sensitivity and resolution and demonstrate imaging in biological tissue. An axicon micro-optics lens with an apex angle of 170° illuminated with a 600 μm diameter Gaussian beam (i.e., diameter of the beam at $1/e^2$ intensity) was incorporated into the sample arm of a SD-OCT setup to test its performance. Without lack of generality, a type of SD-OCT (i.e., spectrometer-based OCT) was used for all measurements reported in this Letter. The SD-OCT system consists of a high-power broad bandwidth Ti:sapphire laser centered at 800 nm with 120 nm FWHM (Femtolaser Inc., Integral), a commercial spectrometer with a 3648 line CCD array (Ocean Optics Inc, HR4000), and a broadband custom-made 80/20 (NSF-DARPA/PTAP) fiber coupler that feeds the two arms of a Michelson interferometer. A Fourier-domain optical delay line is used to compensate the overall dispersion of the system [8]. The DOF generated by illuminating the axicon with a 600 μm collimated Gaussian beam (full width at $1/e^2$) is that of a Bessel beam and is calculated to be 8 mm according to $\mathrm{DOF}_B \cong R(\cot\beta - \tan\alpha)$, where R is the incident beam half diameter, α is the edge axicon angle (5°), and β is the beam deviation angle (2.24°) with respect to the optical axis. The axicon gives a central peak radius of 7.7 μm at 800 nm (i.e., the first zero radius from the center of the Bessel beam, which is also the lateral resolution according to the Rayleigh resolution criterion) given by $\rho_0 = 2.4048\lambda/2\pi\sin\beta$, where λ is the central wavelength. The central peak radii along the DOF were compared with the FWHM beam diameters (i.e., a standard lateral resolution metric for Gaussian

beams) formed by a conventional lens of focal length f equal to 8 mm operating at a NA of 0.037 that yields an $1/e^2$ intensity beam waist radius of 6.9 μm (i.e., 8 μm FWHM) at the focus plane. This lens has approximately a 400 μm depth of focus (i.e., two times the Rayleigh range) given by $\mathrm{DOF}_G \cong 2\pi\omega_0^2/\lambda$, where ω_0 is the $1/e^2$ intensity beam waist radius. The lateral resolution, measured as the central peak radius for the axicon and the FWHM for the conventional lens, was measured over the focal region of interest with a CCD imaging camera of 1004 $\times$1002 pixels and 8 μm$\times$8 μm pixel size (Andor Technology Inc., VP885KCS) placed at different axial positions. These positions are recorded on the x axis of Fig. 1(a) as the relative axial distance d with respect to the medial positions of the two optics' respective DOFs. The medial position m_0 of the DOF generated by the axicon is 4 mm from the axicon apex, around which the measured depth of focus is +/−2 mm. The lateral resolution values are reported on the y axis of Fig. 1(a). Results show that the axicon achieves an ~8 μm invariant lateral resolution across a much larger depth range compared with the 0.037 NA conventional lens. The chromatic dispersion created by the 170° apex angle axicon was investigated and found to be negligible, since the wavelength-dependent maximum refraction angle deviation was assessed to be 0.009° (i.e., corresponding to a relative error of 0.4%) over the 120 nm source bandwidth using optical ray-tracing software. We shall now compare the imaging sensitivities for both optics. For Gaussian beam imaging using the conventional lens, the peak sensitivity was conventionally measured at the focal plane of the lens where a mirror was used in place of a biological sample; a 101 dB peak sensitivity was measured with an incident power of 25 mW. This measurement was made with a difference in optical path length of 1 mm between the sample mirror and the reference mirror. For the Bessel beam imaging using the axicon, the independent measurements of illumination efficiency (i.e., resulting from focusing a beam with the axicon) and collection efficiency (i.e., resulting from the light collimated by the axicon from an equivalent scattering sample) are required, from which the sensitivity can be computed. Starting with measuring the intensity along the depth axis after each lens, an illumination efficiency will be derived via normalization. The on-axis intensity distributions at the different axial positions were measured with the CCD camera for both optics and normalized by the incident intensity I_0. Figure 1(b) shows the measured intensities normal-

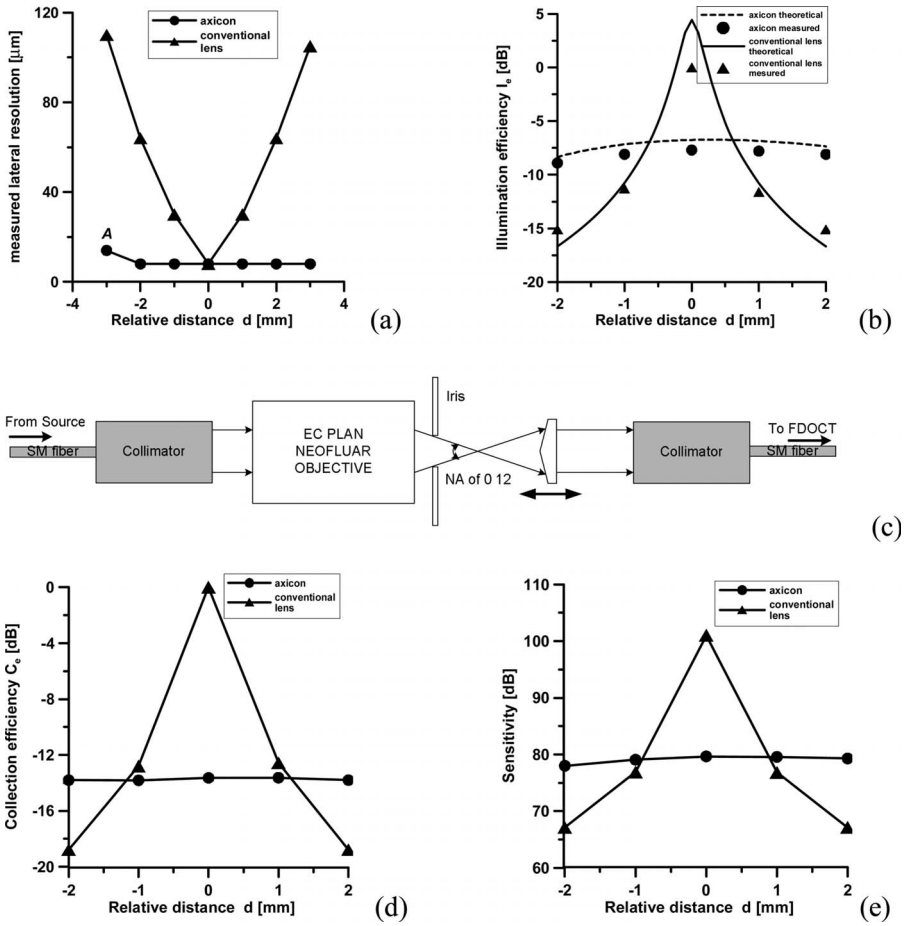

Fig. 1. (a) Measured lateral resolution over focal range parameter d using either a micro-optic axicon of 170° apex angle illuminated by a 600 μm collimated beam or an 8 mm focal length lens operating at a NA of 0.037. Point A corresponds to the focus depth associated with the rounded apex of the axicon lens. (b) Illumination efficiencies versus d, (c) setup to measure collection efficiencies, (d) measured collection efficiencies versus d, (e) measured system sensitivities versus d.

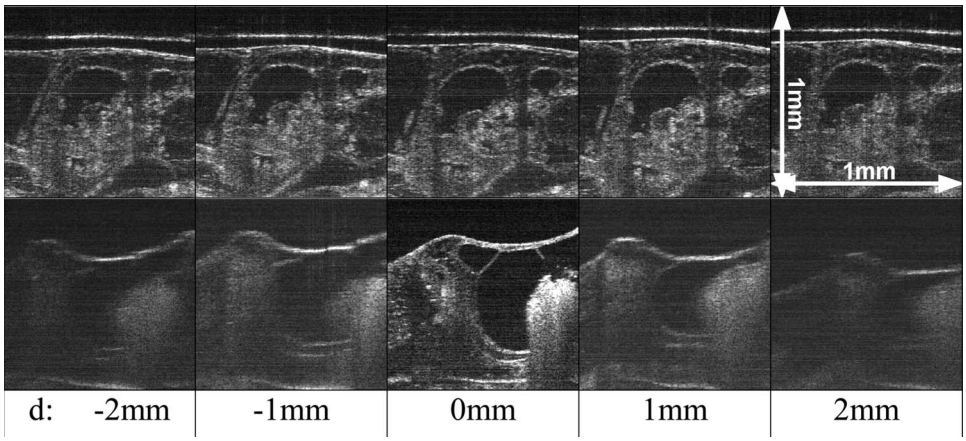

Fig. 2. 1mm × 1 mm SD-OCT images of an African frog (*Xenopus laevis*) tadpole located at various d values acquired using a 600 μm effective diameter axicon micro-optic lens (top row) and a 0.037 NA conventional lens (bottom row).

ized to the peak intensity of the focused Gaussian beam at its focal point. These measured values are in good agreement with the theoretical normalized on-axis intensity distributions given by $I_B(d)/I_G(0) = (4\pi^2\beta^2(d+m_0)/\lambda)\exp[-2\beta^2(d+m_0)^2/R^2]/I_G(0)$ for the axicon and $I_G(d)/I_G(0) = (\pi^2\omega_0^4+\lambda^2 f^2)/(\pi^2\omega_0^4+\lambda^2 d^2)/I_G(0)$ for the conventional lens. These normalized intensity values define the illumination efficiencies, which are reported as a function of the relative distance d. To measure collection efficiencies, we collimated the source and created a 0.12 NA focused beam (using an EC Plan NeoFluar, Zeiss Inc. combined with an iris). This setting simulates an angular distribution of the light after scattering that needs to be collected by the axicon. The beam was then collimated by either the axicon or the conventional lens and coupled to the sample arm of the SD-OCT interferometer as shown in Fig. 1(c). The collected power at the various relative positions d was measured using a power meter connected to the detector arm of the SD-OCT interferometer. Each power value was then normalized by that of the conventional lens at its focal point as shown in Fig. 1(d). These normalized values define the collection efficiencies. In Fig. 1(e) the reported sensitivities for the axicon optics versus d over its DOF were computed by subtracting the measured illumination and collection efficiencies from the measured 101 dB established for the conventional lens. The sensitivity of the Bessel beam SD-OCT is higher than that of the conventional lens when the sample is located beyond $d=1$ mm. Together with the reported resolution in Fig. 1(a), the measured performance demonstrates the advantage of Bessel beam imaging when an extended DOF is required.

Finally, the effectiveness of implementing axicon micro-optics has been exclusively demonstrated by imaging a biological sample, specifically an African frog (*Xenopus laevis*) tadpole, at various positions d acquired with the same incident power of 25 mW and exposure time of 50 μs. The reference arm mirror was adjusted to be within the imaging depth capability of the CCD (i.e., ~2 mm imposed by the CCD resolution), and specifically the reference mirror was located 1 mm in optical path length from the center of the images. The Bessel beam images in Fig. 2(a) show invariant resolution and signal-to-noise ratio over at least a 4 mm focal range, while the Gaussian beam images shown in Fig. 2(b) are already out of focus for $d=1$ mm.

In conclusion, we have shown that a micro-optic axicon (i.e., <1 mm in diameter) can be used for both illumination and collection to perform Bessel beam SD-OCT imaging in a biological sample with a lateral invariant resolution of 8 μm and a sensitivity of 80 dB across a 4 mm DOF that is ten times larger than the DOF achieved with a conventional lens of equivalent resolution at best focus. A 600 μm diameter micro-optic axicon can, in future work, be integrated in a compact-sized endoscopic probe. Such a probe will require no refocusing on a targeted location because of its large focusing range.

This work was supported by the U.S. Army Medical Research and Material Command, the FPCE, and the NSF–DARPA/PTAP. We thank Optical Research Associates for the educational license of CODE V and FEMTOLASERS, Inc., for excellent technical support.

References

1. T. Xie, S. Guo, Z. Chen, D. Mukai, and M. Brenner, Opt. Express **14**, 3238 (2006).
2. J. Rolland, J. O'Daniel, C. Akcay, T. Delemos, K. Lee, K. Cheong, E. Clarkson, R. Chakrabarti, and R. Ferris, J. Opt. Soc. Am. A **22**, 1132 (2005).
3. P. Meemon, K. Lee, S. Murali, and J. Rolland, Appl. Opt. **47**, 2452 (2008).
4. S. Murali, K. Lee, and J. Rolland, Opt. Express **15**, 15854 (2007).
5. J. McLeod, J. Opt. Soc. Am. **44**, 592 (1954).
6. Z. Ding, H. Ren, Y. Zhao, J. Nelson, and Z. Chen, Opt. Lett. **27**, 243 (2002).
7. R. Leitgeb, M. Villiger, A. Bachmann, L. Steinmann, and T. Lasser, Opt. Lett. **31**, 2450 (2006).
8. K. Lee, C. Akcay, T. Delemos, E. Clarkson, and J. Rolland, Appl. Opt. **44**, 4009 (2005).