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PAPER

Fibre Bundle Optical Coherence Tomography Using Angled Divergent Illumination.

Hal Dorrington^{1,*}, Michael R. Hughes¹ and Adrian Podoleanu¹

¹Applied Optics Group, School of Engineering, Mathematics and Physics, University of Kent, Canterbury, United Kingdom

*Author to whom any correspondence should be addressed.

E-mail: H.G.Dorrington@tuta.com

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Abstract

Fibre imaging bundles, along with chip-on-tip cameras, form the basis of conventional forward-viewing endoscopes, which do not currently have depth-scanning capability. Clinically deployed endoscopic Optical Coherence Tomography (OCT) devices are typically side-viewing, with two-dimensional scanning achieved by rotation and pull-back, which is suitable for imaging along tubular, tract-like organs, such as the airways. Forward-viewing endoscopic OCT could be useful for many other applications, but such devices are difficult to miniaturise due to the need for two-axis scanning at the fibre-tip. Fibre bundles are a potential scanner-free alternative for image relay, but previous work has delivered images with poor signal-to-noise ratio, particularly when the bundle is in only one arm of the OCT interferometer. In this paper, we present an adapter that provides an interferometer at the distal tip of a fibre bundle, such that the bundle is in a common-path configuration, with sample and reference fields relayed together. The sample is illuminated by an angled divergent beam directly from a single-mode fibre, and imaged by a 4f telescope onto the bundle. A reference beam is similarly delivered onto the face of the bundle from a second single-mode fibre, with a diffuser placed flush with the bundle face to improve power injection under angled illumination. This allows for full-field swept-source OCT, delivering a forward-view in a compact cylindrical package in-line with the bundle. The adapter could potentially enable conventional endoscopes to perform depth-resolved imaging. The approach is verified by imaging a phantom at a volume rate of 0.5 Hz, demonstrating the feasibility of acquiring OCT images through a fibre bundle using the compact adapter.

1 Introduction

Optical Coherence Tomography (OCT) [1] is an interferometric 3D imaging technique that has been applied to a wide range of medical applications. It is capable of high-resolution, real-time imaging at depths of up to 1-2 mm in tissue, with speeds of up to multiple volumes per second. Tissue is typically illuminated with a beam of non-ionising, low-power, near-infrared (NIR) light, and the backscattered light collected and analysed with an interferometer. 2D cross-sectional scans through the tissue (B-Scans) can be assembled into 3D volumes, from which en-face slices (C-Scans), parallel to the tissue surface, can also be viewed.

OCT interferometers are readily implemented in fibre, enabling endoscopic OCT (e-OCT), which has been extensively explored [2]. Most e-OCT systems demonstrated previously, and those in routine clinical use, are side-viewing, whereby the beam exits the OCT fibre probe at around 90° to the probe axis via a prism or a mirror at the distal end. Rotation of the fibre assembly or just the prism or mirror, combined with a pull-back of the probe, allows for acquisition of 3D images. Side-viewing OCT is straightforward to miniaturise but is only well-suited to imaging luminal organs such as the gastrointestinal tract, airways and vasculature.

Forward-viewing e-OCT is desirable for clinical applications in which the side-viewing scheme is not well-suited [3]. Typical forward-viewing e-OCT requires either 1D or 2D lateral scanning of the beam emitted from a fibre-end, which is hard to achieve within the size constraints of an endoscope. Ideally, an e-OCT device should be small enough to either deploy through the working

channel of conventional endoscopes, or to be integrated alongside macroscopic imaging and instrument channels, which in either case limits the outer diameter of the device to less than a few millimetres. Various approaches have been suggested, such as piezo or magnetic actuators to vibrate the fibre-tip [4-6], but it is challenging to achieve both high resolution and a large field-of-view (FOV) in a small package, with FOVs typically on the order of 1 mm^2 . Alternative scanning regimes have been suggested [7], delivering superior FOVs of up to 2.25 mm^2 , but in this case from a 4 mm wide probe.

Fibre imaging bundles are thin and flexible image-relays consisting of several thousand optical fibre cores, each effectively acting as a single pixel. Scanner-free white light endoscopic imaging was traditionally performed using fibre bundles. Currently, following the development of compact chip-on-tip cameras, endoscopy is now commonly practised either with fibre bundle or camera-based endoscopes. To date, fibre bundle systems can provide superior image resolution, whilst chip-on-tip camera endoscopes are generally cheaper and less susceptible to damage from bending or crushing [8]. In fibre bundle endoscopy, illuminated tissue at the distal end is typically imaged onto the bundle face by a small focusing element, such as a micro-objective, a 4f telescope, or a GRIN lens [9]. In conventional use, the image relayed to the proximal end is then viewed directly through an eyepiece or imaged onto a camera. Such endoscopes are generally forward-viewing, delivering high resolution macroscopic images. However, they are incapable of providing depth-resolving scans. Hence, it would be desirable to endow such endoscopes with depth-scanning capability, e.g. by adapting to perform OCT. In general, development of fibre bundle imaging techniques is an active area of research. Recent scanner-free techniques have been developed, such as lensless holographic imaging [10], enabling high-resolution imaging over a large field-of-view. In addition, diffraction-limited beam focusing has also been demonstrated by adaptive wavefront shaping [11]. More broadly, recent advances in meta-structured optical fibres have highlighted the potential of fibre-based photonic devices for highly compact imaging and sensing applications [12].

Fibre bundles have been proposed as a means of achieving forward-viewing e-OCT without the need for a distal scanning mechanism [13]. They are particularly attractive owing to their widespread adoption in endoscopy. However, the image quality that has so far been achieved with fibre bundle OCT systems is sub-clinical. There are several difficulties involved with using fibre bundles for OCT. Each fibre core in the bundle, depending on the design, is between 2 and $12\text{ }\mu\text{m}$ in diameter, and between 0.3 and 0.6 in numerical aperture (NA), meaning each core supports multiple modes at visible or NIR wavelengths. In conventional white light endoscopy or fluorescence imaging, the incoherence of the light means that modal effects can be ignored, and each core generates an approximately Gaussian-shaped intensity pattern on the camera. However, with coherent light, each core instead forms a small speckle pattern, representing a superposition of modes, typically degrading image quality and potentially limiting the effective lateral resolution [14]. Additionally, and of pertinence to OCT, modal dispersion occurs within cores, leading to group delay variations and resulting in degradation of the axial resolution and the introducing of repeating artefacts in depth through the OCT image [15, 16]. Multimodal effects in fibre bundles are also bending-dependent, as such the quasi-random behaviour of modes when the bundle is in use makes it difficult to compensate for by calibration [13].

These difficulties make it undesirable to place the fibre bundle solely in one arm of the OCT interferometer; instead, it is preferable to use a common-path arrangement, in which both the object and reference fields are brought into superposition before entering the bundle. The reported common-path methods are typically based on having either Fizeau [17] or Michelson [18] interferometers at the distal end. However, each presents difficulties; a Michelson interferometer is hard to miniaturise, due to requiring a beamsplitter at the distal end, while reported Fizeau interferometers have delivered poorer signal-to-noise ratio and have the caveat of difficulty in controlling reference power. Furthermore, it would be beneficial to deliver light separately from the bundle, so that there is only a single pass through the bundle.

In this paper, we present a new approach to forward-viewing e-OCT using a fibre bundle, which is particularly suitable for implementation in a very compact package. This approach presents a means of transforming a fibre bundle into a depth-resolved imaging instrument, endowing with OCT functionality. In theory the adapter could also be applied to camera-based endoscopes. While the approach presents a scanning-free method of forward-viewing e-OCT, due to difficulties in fibre bundle OCT imaging, the performance metrics of the presented prototype do not compete with current state-of-the-art fibre-scanner systems. However, a significant merit of the proposed method is its capability to endow widespread fibre bundle and camera-based endoscopes with depth-scanning capability. The proposed method utilises divergent full-field illumination from single-mode fibres, taken from the output of a fibre coupler connected to a wavelength-tunable

source. This allows OCT reconstruction via full-field swept-source OCT (FF-SS-OCT) [19], whereby full-field camera images are acquired sequentially for each wavelength. Critically, the design allows for the bundle to be common-path without needing to introduce a beamsplitter at the distal end. Secondly, relative powers in the sample and reference arms can be selected by choosing a coupler with the desired coupling ratio, and adjusted as required using optical attenuators prior to reaching the device. Below, we evaluate this device as a step towards viable OCT imaging through fibre bundles.

2 Method

2.1 Adapter and system design

As shown in Fig. 1(a), the novel design for the interferometer sits at the distal end of a fibre bundle, such that the bundle is in a common-path configuration but without requiring a distal beamsplitter. Light is delivered from the tips of two single-mode fibres, taken from the output of a 99/1 coupler (AFW Technologies - FOSC-2-840-01-L-1-2), denoted as *Obj* and *Ref* respectively. A unit magnification 4f-telescope, of two identical lenses, L1 and L2, with focal lengths f and diameter D , images the sample onto the face of the bundle. The tip of the object fibre is positioned at the edge of L1, at an angle α with respect to the optical axis, and projects a diverging, angled beam onto the tissue, whose centre is incident at angle α . The reference fibre, positioned at the edge of L2 and at the opposite side of the adapter, is tilted at the same angle α , and delivers a similarly diverging and angled reference beam onto the bundle face. The reference beam comes into superposition with the image formed by the telescope at the bundle face, allowing the bundle to be in an extremely compact common-path arrangement. By securing a common-path, the detrimental effects caused by multimodal coupling which are exhibited when the bundle is only on one arm of the interferometer are mitigated, similarly to existing common-path designs. The principal benefit of the newly proposed design over established common-path methods is the potentially very compact geometry, not easily achieved with a Michelson design [18]. While a Fizeau design is similarly compact, our design features the delivery of sample illumination and the delivery of a reference field both directly from single-mode-fibre, not possible with the Fizeau design [17]. In this sense, major caveats of each existing design are addressed by the proposed method.

A photograph of the adapter at the distal end of a bundle is shown in Fig. 1(c). Hypodermic needles mounted onto XYZ translation stages are used to house the fibre-tips and deliver them to the edges of the adapter, enabling the position and angle of the fibre-tips to be precisely controlled for evaluation purposes. In a future finalised clinical design these would be fixed in place.

The compact design necessitates off-axis illumination of the sample and off-axis injection of the reference wave into the bundle. Off-axis illumination of tissue has been explored previously in OCT for other purposes and has benefits in terms of reduced specular reflection from the tissue surface and greater penetration depth due to selective detection of scattered photons [20]. The angle of illumination at the centre of the object, α , is determined by D and f , and is given by $\alpha = \arctan(D/2f)$. Identical plano-convex lenses with diameter $D = 4$ mm and focal length $f = 8$ mm (Edmund Optics, Part No. 47-852), are used, giving the angle $\alpha = 14^\circ$. The lenses are mounted inside a 3D printed casing with an outer diameter of 8 mm, with a smaller diameter aluminium tube used as a spacer to separate the lenses.

For validation experiments, an 18,000 core leached fibre bundle (Schott - Part No. 1549304), with imaging diameter, $d = 1.05$ mm, is used. Schott leached bundles have a sparse and highly regular honeycomb arrangement of cores, exhibiting less cross-coupling and hence better suitability for full-field OCT than wound or fused bundles. The cores have a numerical aperture (NA) of approximately 0.5 [18], enabling good acceptance of light from off-axis. The pitch of the cores in the bundle is given by $p \approx 8.0$ μm ; assuming unit magnification of the distal telescope, this imposes a limiting resolution of 16.0 μm and a highest resolvable spatial frequency of $f = \frac{1}{2p} = 63$ mm^{-1} .

A schematic of the system set-up is provided in Fig. 1(b), along with a photograph showing parts of the system in Fig. 1(d). Illumination is provided from a swept-source laser (Superlum BroadSweeper-840 BS0002) with a central wavelength of 850 nm. The swept-source delivers 5 mW of power which is then boosted (using Superlum BB-840-HP-OEM) to 35 mW power input into the coupler. A 99/1 coupler is used, as this coupling ratio was found to best match object and reference powers at the camera sensor, which is generally required for good signal-to-noise ratio (SNR) and efficient use of the camera bit depth. As the power returned from the object is much lower than the power returned from the reference arm, several couplers with ratios up to 99/1 were tested, from which a 99/1 ratio was found to deliver the closest match of powers at the camera sensor, albeit with slightly less power in the object field. A free-space section, including a variable attenuator, was hence introduced into the reference arm, and the reference power attenuated to approximately

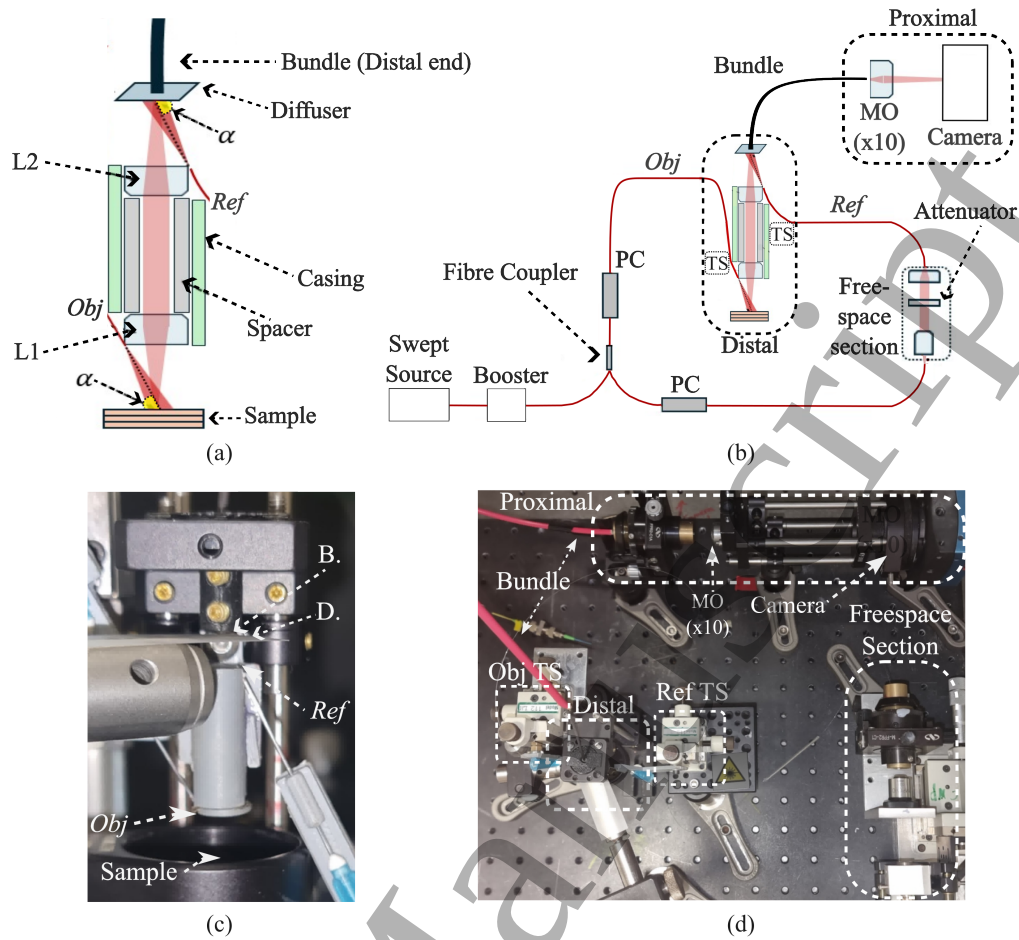


Figure 1. (a) Schematic of interferometer at distal end of bundle. The sample illumination is projected onto the sample at angle α , depicted in yellow, and the reference light is injected into the bundle at the same angle. L1 and L2: lenses. (b) Schematic of OCT system. MO: microscope objective; PC: polarization controller; TS: Translation Stage, used to facilitate positioning of *Obj* and *Ref*. (c) Photograph of adapter at distal end of bundle. B: Bundle; D: Diffuser. Tips of object and reference fibres labelled as *Obj* and *Ref*. (d) Photograph of parts of the system. MO: microscope objective; TS: Translation Stage.

match powers at the sensor. The free-space section also allowed for the reference path length to be adjusted, enabling performance characterisation with increasing optical path difference (OPD). A $10\times$ microscope objective (MO) forms an image of the proximal face of the bundle directly onto the sensor of a high frame rate camera (Mikrotron MC1632) with $14\ \mu\text{m}$ square pixels. The tube length is adjusted to achieve a magnification factor of 10 and hence approximately 6 sensor pixels across a single core-to-core spacing. In-line polarisers, denoted PC in Fig. 1(b), are positioned in each arm of the fibre coupler, and are adjusted to maximise interference contrast at the camera.

2.2 Diffused angled injection of fibre bundle cores

Angled illumination of the bundle requires particular attention due to the strong dependence of modal coupling efficiency on the incidence angle. At larger angles there is reduced total power coupled as well as a preferential coupling into higher-order modes with greater azimuthal number. At the prescribed angle of $\alpha = 14^\circ$, the power injection efficiency into the fibre bundle, without any diffuser in place, is $\ll 0.1$ (relative to on-axis illumination). In the experimentally acquired plot of mean pixel intensity on the camera against angle of fibre in Fig. 2(a), for the case without diffuser in place in dashed line, the injection efficiency at $\alpha = 14^\circ$ is ≤ 0.1 , relative to near on-axis illumination at 5° .

The total coupling efficiency of the reference light is not a practical concern, since the reference power must be attenuated anyway to avoid saturation of the camera, in our case achieved by using a 1% tap and an adjustable attenuator in the free-space section of the reference path. However, at large angles, power is preferentially coupled into higher-order modes, resulting in a weak OCT signal with repeated ghost artefacts (i.e. spurious duplications of features at different depths) [13, 15]. To combat this, a diffusing film (Luminit P1 $40^\circ - 0.254\ \text{mm}$) with a 40° diffusing angle and a

thickness of 0.254 mm, was positioned directly on the bundle face, resulting in a significant fraction of the reference beam power now entering the bundle close to on-axis. Due to the increased coupling efficiency at this angle, a large fraction of the power now sits in lower order modes. Several 0.254 mm thick Luminit diffusers with differing scattering angles were considered. With power only in the reference arm, coupling into cores was found to more closely resemble single-mode coupling as the diffusing angle was increased. However, with power only in the object arm, intensity decreased substantially with increasing diffusing angle, relative to the intensity with no diffuser in place, dropping to 24% for 40°, 13% for 50° and 4.6% for 80° diffusing angles. A 40° diffusing angle was deemed to provide a suitable compromise of improved reference coupling whilst preserving a suitable object intensity.

To demonstrate the effects of the diffuser, images were acquired with power only in the object and reference fibres in turn, both with and without the diffuser in place. In both cases the object and reference fibres were inclined at an angle of 15°. The images with only reference power are shown, cropped to an 110 $\mu\text{m} \times 110 \mu\text{m}$ FOV, in Fig. 2(b). Without the diffuser in place, Fig. 2(b, left), it is hard to distinguish individual cores due to dominance of higher-order modes and potentially some cross-coupling between cores. With the diffuser in place, Fig. 2(b, right), a number of cores appear to exhibit Gaussian-like profiles, although some higher-order mode patterns can still be seen at some cores. To analyse coupling with and without the diffuser in place, the radial profile of intensity coupled into each core, averaged across all cores was calculated, from which the profile was found to more closely resemble single-mode-like coupling with the diffuser in place. Line profiles were extracted across the images with only reference power, in the direction perpendicular to the inclined fibre, such that the angle of injection remained approximately 15° across the profile. The line profiles with and without the diffuser in place are shown in Fig. 2(c). With the diffuser in place, shown in solid blue line, the intensity distribution is less uniform, as can also be seen qualitatively in Fig. 2(b). However, we observe higher intensity peaks at a number of cores, indicating better suitability for OCT. Without the diffuser, there are no such high intensity peaks, and instead intensity is broadly distributed within and between cores, representing higher modal coupling and potentially cross-talk, making the illumination unsuitable for OCT.

To characterize modal coupling of light injected at 15°, with and without the diffuser in place, the radial profile of intensity coupled into each core, averaged across all cores, was calculated to generate plots of relative intensity vs distance from core centre, which are given in Fig. 2(d). Without the diffuser in place, shown in dashed red line, more power is distributed into the outer regions of cores, compared with when the diffuser is in place, shown in solid blue line. This can be used to infer that less power is coupled into higher-order modes with the diffuser in place. Due to there being less power coupled into higher-order modes, the effects of higher-order coupling, such as ghost bands, are reduced with the diffuser in place. In Fig. 2(d), a simulated curve representing on-axis injection is also shown, in dashed black line, which was calculated using a core diameter of 5.4 μm and a core NA of 0.4, at a central wavelength of 850 nm.

To evaluate the loss of uniformity due to the diffuser, a 2D Gaussian filter with $\sigma = 10$ pixels (13.7 μm) was applied to each image, and the coefficient of variation from the central 200 $\mu\text{m} \times 200 \mu\text{m}$ region calculated, giving 0.49 without the diffuser and 0.58 with the diffuser in place. However, due to the improved modal coupling with the diffuser in place, this loss of uniformity was deemed a necessary compromise. For smaller angles, in the range of 5° – 10°, the overall intensity of the reference image is lower with the diffuser in place, as demonstrated in Fig. 2(a). However, at angles larger than 10°, the mean intensity is greater with the diffuser in place.

To evaluate the effect of the diffuser on lateral resolution, a United States Air Force (USAF) resolution target was imaged, with power in the object arm only, with and without the diffuser in place. The USAF target was imaged at group 5 and group 4, which are shown cropped to a 0.4 mm $\times$ 0.4 mm FOV in Fig. 2(e), without the diffuser (left column) and with the diffuser (right column). Without the diffuser, the highest resolvable line pair is group 5, element 2, with 35.9 lp/mm, equivalent to 28 μm resolution, as shown in Fig. 2(e, bottom left). With the diffuser in place, only up to group 4, element 3 can be resolved, with 20.16 lp/mm and 50 μm resolution, as shown in Fig. 2(e, top right). This loss in lateral resolution is therefore a compromise we accept in order to achieve a better coupling of the reference wave.

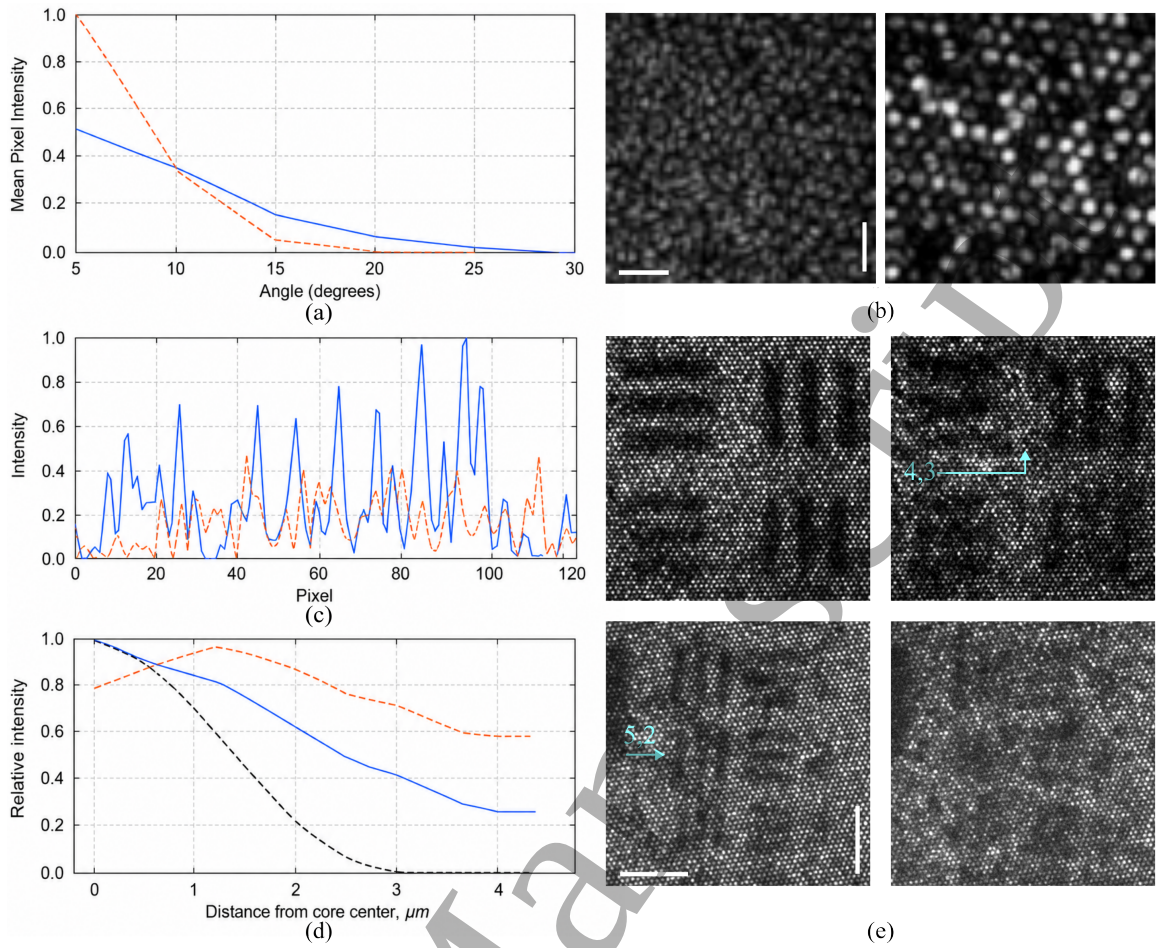


Figure 2. Characterisation of effect of diffuser. (a) Intensity of proximal end of bundle imaged onto camera with varying angle of reference beam, without diffuser (red, dashed) and with diffuser in place (blue, solid), both normalised to the intensity at 5° without the diffuser in place; (b) camera image when only reference beam is incident, illuminating bundle at 15° angle, cropped to $110 \mu\text{m}$ square FOV with $20 \mu\text{m}$ scale bars, illustrating modal behaviour without diffuser (left) and with diffuser (right); (c) line profile along central 120 pixels of images with only reference power, without diffuser (red, dashed) and with diffuser in place (blue, solid); (d) relative intensity vs distance from core centre, with light injected at 15° , without diffuser (red, dashed), and with diffuser in place (blue, solid), along with simulated curve under on-axis injection (black, dashed); (e) camera image with object beam only, with USAF target as sample imaged at group 4 (top row), and group 5 (bottom row), without diffuser (left column), with highest resolvable line pair, group 5 element 2 labelled, and with diffuser in place (right column), with highest resolvable line pair, group 4 element 3 labelled. Images cropped to a 0.4 mm square FOV with $100 \mu\text{m}$ scale bars.

2.3 Imaging by FF-SS-OCT

To acquire an OCT volume, the swept-source laser, centred at $\lambda_0 = 850 \text{ nm}$, with linewidth, $\delta\lambda = 59 \text{ pm}$, is tuned over a bandwidth range of $\Delta\lambda = 47 \text{ nm}$, while a stack of images is acquired by the camera. The number of images captured in a stack, M , determines the number of points along an A-scan, given by $M/2$, as well as the axial range, z_{max} , given by $z_{\text{max}} = \frac{1}{4} \frac{\lambda_0^2}{2\Delta\lambda} \cdot M$. A fundamental decay curve of SNR with depth exists, determined by $\delta\lambda$, and M should be selected such that $\frac{\Delta\lambda}{M} \leq \delta\lambda$, otherwise potentially degrading the depth decay curve. By calculating $\frac{\Delta\lambda}{\delta\lambda}$ we arrive at 800 frames as a suitable value, providing the maximum possible integration time without potentially infringing on the source-limited decay of SNR in depth. With a camera frame rate of 400 Hz, a stack of 800 images across the wavelength range is acquired in 2 s, with each exposure having an integration time of approximately 2.49 ms. Hence, acquisition of volumes is achieved at a rate of 0.5 Hz. In theory, with $M = 800$, the expected axial range is 1.54 mm, while the expected 6 dB decay depth is 1.35 mm. The 35 mW of power delivered to the coupler leads to 15 mW of power in the object beam, and, after free-space injection, 60 μW in the reference beam. The object power is diminished due to connector losses, while the reference beam is attenuated in free-space to match approximately the power returned from the object to the camera sensor.

A 3D OCT volume can then be recovered by taking a 1D Fast Fourier Transform (FFT) along

the spectral direction for each spatial pixel, under the assumption that the wavelength scanning is linear in wavenumber and that dispersion mismatch between the interferometer arms is negligible. In practice, while source tuning was found to be approximately linear in wavenumber, a non-negligible dispersion imbalance was present, which could have the effect of degrading axial resolution and lowering SNR. However, due to the adapter geometry, it was not possible to obtain calibration spectra from a mirror, making it difficult to perform a reliable dispersion compensation, which was hence not performed. Instead, experiments were focused on the proof-of-concept demonstration of OCT imaging, and the effects of unmatched dispersion ignored. In a future revision, a reliable dispersion compensation method that does not require a calibration could be implemented, such as using numerical optimization techniques [21, 22], or image-based iterative methods [23].

For imaging, a sample is placed approximately at the focus of the distal telescope, as shown in Fig. 1(a,b). As a sample we used a phantom consisting of a rough-metal perforated disc directly atop of several layers of adhesive polymer tape. Stacks are initially acquired with first the reference and then the object arms blocked in turn; these stacks are used for background subtraction. An imaging stack is then acquired with both arms unblocked. For both imaging and background stacks, for each pixel on the camera, the spectra (i.e. each line along the spectral direction of the stack) is windowed with a Hamming filter, an FFT is performed over the windowed stack, and finally the absolute value of the FFT is taken, returning a real-valued stack. The background stacks, processed in the same way, are then subtracted from the imaging stack. Each en-face slice of the resultant volume is then smoothed with a 2D Gaussian filter with $\sigma = 10$ pixels ($13.7 \mu\text{m}$) to remove the fibre bundle core pattern. Finally, the mean along the depth axis in the stack is computed and subtracted from each depth slice to remove any fixed pattern noise common across all depths.

3 Results

3.1 SNR and axial resolution decay

In swept-source OCT the instantaneous linewidth of the source, in our case, $\delta\lambda = 59 \text{ nm}$, determines a decay of sensitivity with depth (i.e. with optical path difference, OPD, between the interferometer arms). To assess the dependence of SNR and axial resolution with OPD, a further set of spectra were acquired for a range of OPD values by varying the free-space length in the reference arm. From this dataset, a plot was generated of A-scan amplitude and resolution against OPD, shown in Fig. 3. The decay of signal amplitude, and hence SNR, with OPD is plotted relative to the value at close to zero OPD. A 6 dB SNR fall-off occurred at an OPD of 1.66 mm, corresponding to an imaging depth of 0.83 mm. The axial resolution at close to zero OPD and at the -6 dB OPD point are $34.15 \mu\text{m}$ and $40.25 \mu\text{m}$, respectively. The A-scan at close to zero OPD was used to obtain a value for SNR from the surface of the phantom, while noise was calculated by computing the standard deviation of the last 50 points of the A-scan, giving a SNR of 45 dB.

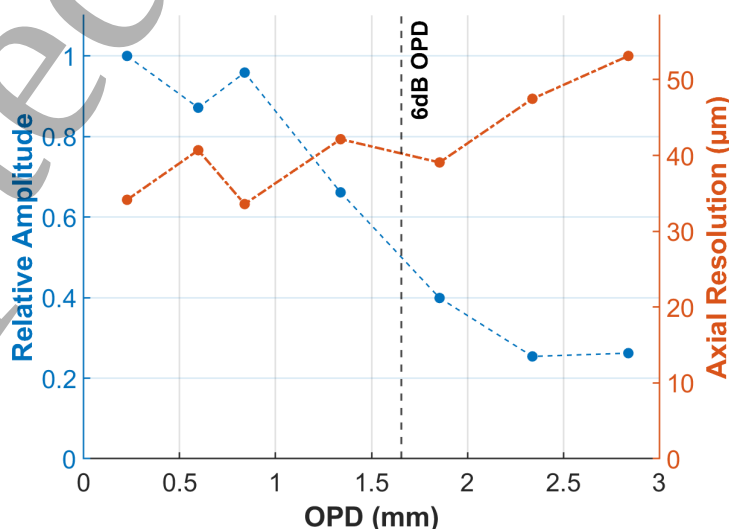


Figure 3. Variation of SNR (blue) and axial resolution (red) with OPD. The SNR decays by 6 dB at 1.66 mm.

3.2 Apparent sample tilt

Due to the sample and reference beams being tilted with respect to the optical axis, it is possible for the OPD between the two arms of the interferometer to vary across the face of the bundle, and hence a tilt or distortion can occur in the OCT images. However, if the object and reference fibres are positioned on opposite sides of the adapter, both making the same angle with the tissue/bundle, and if the lenses form a unity magnification telescope, as illustrated in Fig. 4(a), then the length variations across the sample and reference beams are identical, and hence there should be no variation in OPD. The diverging section of the object path is illustrated from the tip of the object fibre, at O_{Obj} , to the imaged plane on the sample, at points A, B, C, along rays with lengths l_A , l_B and l_C , while the reference path covers lengths from the tip of the reference fibre, at O_{Ref} , to the face of the bundle at points A', B', C', along rays with lengths l'_A , l'_B and l'_C ,

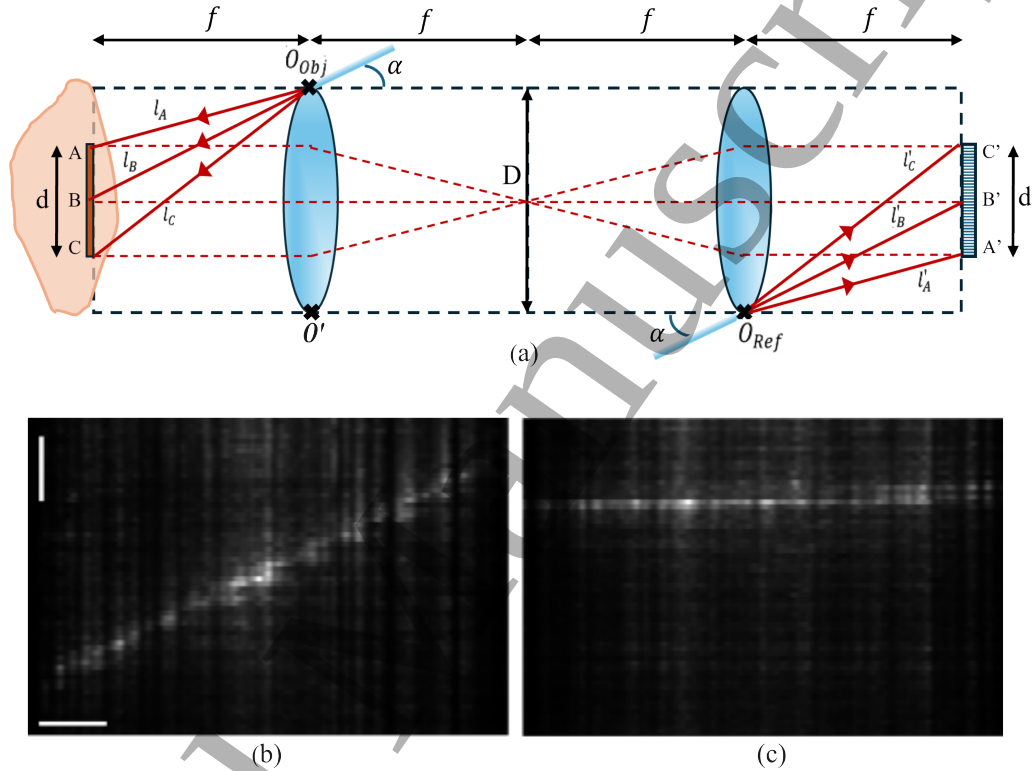


Figure 4. (a) Illustration of the OPD between object and reference paths within the adapter, as evaluated for rays landing at points A', B', C' on the bundle face. l_A , l_B and l_C represent the lengths from O_{Obj} to points A, B, C, at the object, which are equivalent in length to the rays l'_A , l'_B and l'_C , from O_{Ref} to A', B' and C' on the bundle face; (b) OCT image obtained from phantom at region across rough-metal surface, for mismatched case with both fibres on same side of adapter; (c) OCT image across same region of phantom, obtained with correctly positioned fibres at opposite sides of the adapter, as in (a). Images are 0.7 mm in width and 0.5 mm in depth, with 100 μ m scale bars.

With correct positioning of the fibres, as shown in Fig. 4(a), the lengths covered by diverging sections of the reference and object beams are equal for all positions along the bundle in the plane of the figure, that is $l'_A = l_A$, $l'_B = l_B$ and $l'_C = l_C$, and so there is no OPD variation along the face of the bundle. However, as an extreme example, we consider a case of the object fibre being instead positioned at O' , on the same side of the adapter as the reference fibre, where position O' is shown in Fig. 4(a). If we define OPD as the object path minus the reference path, then the OPDs at the two points along the bundle A' and C' are instead given by $OPD(A') < 0$ and $OPD(C') > 0$. This creates a tilted representation of the sample, with negative relative OPD at A' and a positive relative OPD at C'. The expected apparent sample tilt is exactly twice the illumination angle, giving $\theta = 28^\circ$.

To demonstrate this experimentally, OCT volumes were acquired of the phantom, taken from a region across the rough-metal surface. OCT volumes were acquired for the two extreme cases, mismatched (i.e. with object fibre at O') and correct positioning (i.e. with object fibre at O_{Obj}), from which B-Scans are shown in Fig. 4(b) and Fig. 4(c) respectively. As expected, an apparent OPD tilt of the sample was observed for the mismatched case, with an apparent tilt of the sample of 26° , close to the predicted value of 28° . Conversely, an approximately flat sample orientation is observed in the correctly-aligned case, confirming that the true sample shape is captured provided

the two fibres are aligned as in Fig. 4(a).

3.3 OCT Imaging

To assess OCT imaging, the same phantom, consisting of a rough-metal perforated disc directly atop of several layers of adhesive polymer tape, was imaged. The perforated pattern of the sample is demonstrated in photographs shown in Fig. 5(a), while a sketch showing the depth structure at a single hole is given in Fig. 5(c). Scattered light is returned by the tape at areas underneath holes, providing a feature in depth to be resolved by OCT.

OCT imaging was performed with both fibres at their correctly-aligned positions. A volume was rendered with 400 en-face depth slices, and 500×500 lateral pixels, giving approximately 83 resolvable points across a B-Scan, with each A-scan having 400 points. Two separate averaging operations were then performed: along depth and along B-Scans in the y-direction. In depth, every 10 en-face slices were averaged, generating 40 en-face frames, from which two selected frames are shown in Fig. 5(b). The first en-face image (left) shows the metal plate with holes, the second image (right) shows the polymer under-layer directly below the holes.

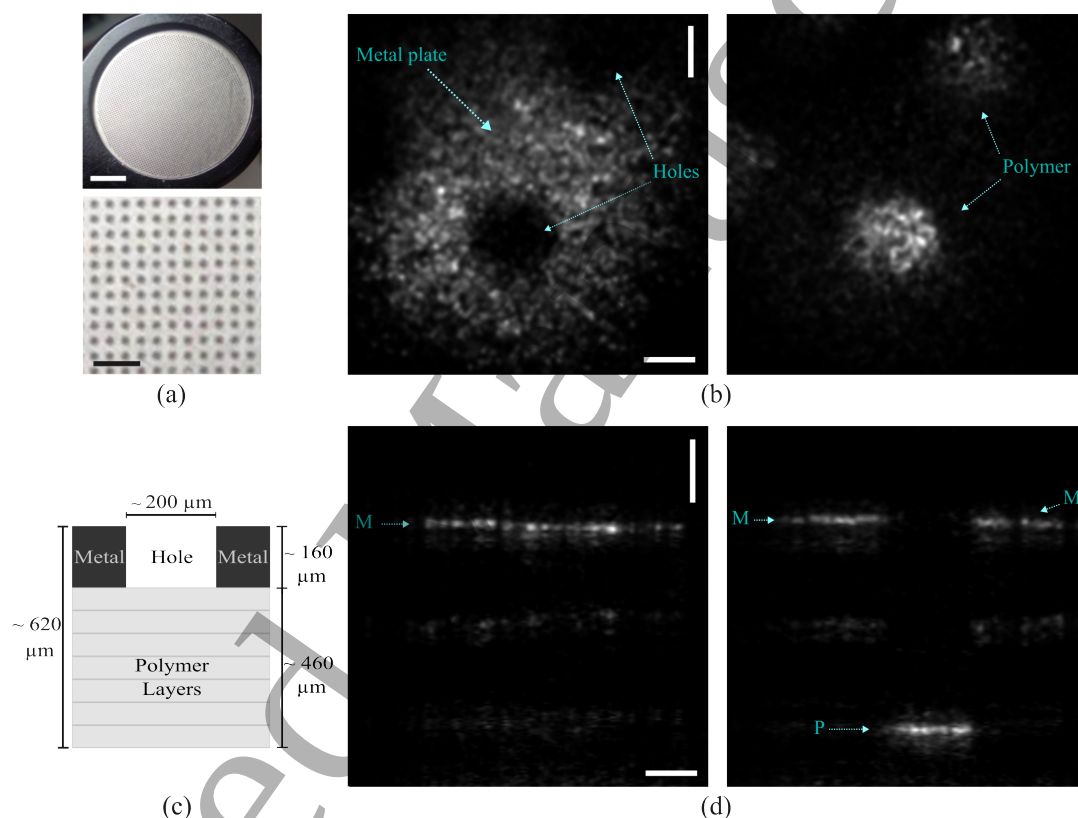


Figure 5. OCT imaging results from phantom. (a) Photographs of phantom, demonstrating the perforated pattern of holes, with 5 mm scale bar (top) and 1 mm scale bar (bottom); (b) selected en-face frames of OCT volume, image size of $0.7 \text{ mm} \times 0.7 \text{ mm}$, with $100 \mu\text{m}$ scale bars, from metal plate with holes (left), and polymer under-layer (right). (c) sketch of a hole region of the phantom, showing a sample feature in depth to be resolved by OCT; (d) selected B-Scans from OCT volume, for region with no holes (left), and from region across a hole (right) revealing polymer under-layer. B-Scans are 0.7 mm in width and 1.2 mm in depth, with $100 \mu\text{m}$ lateral scale bar and $200 \mu\text{m}$ axial scale bar. M: Metal plate; P: Polymer.

Along B-Scans in the y-direction, every 10 B-Scans were averaged, leaving 50 B-Scans. Two selected B-Scans are presented in Fig. 5(d). The first B-Scan (left) is along a region without a hole, and hence the polymer layers are occluded, with only the rough metal surface observed. The second B-Scan (right) is taken across a hole and a signal is seen corresponding to the bottom surface of the lowest layer of tape, perhaps resulting from scattering from deposited fine material, i.e. dust. The individual layers of tape are not resolved, suggesting the system sensitivity is below that of the tape's reflectivity, which is qualitatively low, given that the tape is optically clear. Within the B-Scans, a faint signal is seen approximately halfway between the two sample layers, which is potentially a multiple reflection induced from a cavity in the swept source.

4 Discussion

4.1 Imaging metrics

Imaging metrics of our system can be compared with existing full-field fibre bundle OCT designs, given that such methods are most suited to adapting conventional fibre bundle and camera-based endoscopes to perform OCT. Proximal-scanned fibre bundle OCT systems using a photodetector in place of a camera to detect the relayed scan, are less simple to equip to endoscopes, while fibre-scanning e-OCT systems require especially fabricated probes. Comparisons of our approach with point-scanning fibre-bundle systems, and with state-of-the-art e-OCT fibre scanning probes, are given in appendix A. In general, greater performance metrics are reported from state-of-the-art fibre-scanning e-OCT systems, as is expected. Currently, flying spot SS-OCT systems deliver better quality images than their full-field SS-OCT counterparts, mainly due to loss of confocality, which leads to cross-talk between lateral pixels. In addition, fast cameras are needed to reach the same volume acquisition speed. As a result, the collected data is too large for real-time OCT imaging, instead volumes are acquired and the data transferred to a PC for signal processing and image reconstruction. These factors should be taken into consideration when viewing our tables.

A limited number of full-field OCT designs have been reported, of which only a few are suitable to endoscopy, with several reports being unsuitable due to using either rigid bundles or bulk distal optics. In comparison with suitable designs, our system performs well in several metrics. To convey this, a table is presented comparing our approach with existing full-field fibre bundle OCT designs, given in table 1. The methods most suitable for endoscopy are displayed in bold text. In comparison to the other suitable design, i.e. Ford et al. [17], our system delivers superior SNR and imaging depth, as well as a slight improvement in speed.

Table 1. Metrics comparison with reported fibre bundle FF-OCT designs.

Method	SNR/Sens	Lateral	Axial	FOV	Axial range	Speed	Diam
Our approach	45 dB	50 μm	34 μm	0.7 mm	0.83 mm	0.5 Hz	8 mm
Tang et al. [24]							
Bulk components, common path using a glass plate	NR	1.7 μm	$\sim 1 \mu\text{m}^\dagger$	NR	$\geq 60 \mu\text{m}^\S$	NR	$\sim 50 \text{ mm}$
Latrive et al. [25]							
Common path using a glass plate, rigid endoscope	80 dB*	3.5 μm	1.8 μm	1.5 mm	$\sim 100 \mu\text{m}$	$\sim 1 \text{ Hz}$	2 mm $^\beta$
Ford et al. [17]							
Bundle in the sample arm only	16 dB	$\sim 30 \mu\text{m}$	$\sim 30 \mu\text{m}^\ddagger$	1.4 mm	0.4 mm	$\sim 0.33 \text{ Hz}^\dagger$	NR
Oh et al. [26]							
Distal interferometer with bulk components	77.4 dB**	3.6 $\mu\text{m}^\parallel$	1.1 μm	0.5 mm	NR	0.125 Hz	Bulk

Column headings: SNR/Sens: SNR/sensitivity; Lateral: lateral resolution; Axial: axial resolution in air; FOV: width of field-of-view; Axial range: -6 dB imaging depth; Speed: rate of volume acquisition; Diam: total diameter of distal components. NR: not reported. Bulk: bulk optics, not suitable for endoscopy. Qualifying symbols: * after averaging (20 images); ** after averaging (800 images); † theoretical value; ‡ estimated from the reported coherence length; § demonstrated imaging depth; † potential acquisition rate based on 100 images acquired in $\leq 3 \text{ s}$; $^\parallel$ value after smoothing was 4.1 μm ; $^\beta$ rigid bundle. SNR and sensitivity values are reproduced according to the convention used in each cited study and are not necessarily directly comparable.

A 6 dB decay in SNR was measured at an imaging depth of 0.83 mm, less than the source-limited expected depth of 1.35 mm. This could be an effect of non-linear tuning, and more work would be required to establish if the depth is impacted by effects due to the common-path design. However, as seen in Table 1, the adapter presented represents an axial range superior to existing reported methods, albeit less than existing fibre-scanning methods, which have imaging depths typically on the order of 1-3 mm [4-7].

Our B-Scans contained, before averaging, approximately 83 lateral points, i.e. 83 cores across 500 pixels, spanning a sample width of 0.7 mm, and a total en-face FOV of approximately 0.5 mm^2 , representing values of pixel density and FOV which are relatively low compared with existing fibre bundle full-field OCT methods and scanning-based methods [5-7, 27]. Assuming unitary magnification, the FOV is necessarily limited by the bundle diameter, as is the case for existing fibre bundle OCT designs [17, 18]. In our case, the limiting FOV imposed by the bundle diameter was 1.05 mm, however, to increase power in the object image, the object fibre was brought closer to the sample, reducing the area of illumination and hence the effective FOV to 0.7 mm. The fundamental limiting FOV of 1.05 mm could be increased, either by using a wider bundle, or by

adjusting the magnification of the adapter telescope. While adjusting magnification would then affect the sampling-limited resolution from the bundle core spacing, due to the use of the diffuser, the lateral resolution was only approximately 50 μm , greater than the core limiting 16 μm , and it could therefore be possible to redesign the telescope to a larger potential FOV without any impact on resolution in practice.

The lateral resolution is larger than values presented in scanning probe methods, which are typically on the order of 15 μm to 20 μm [4-7], and larger than the core spacing in the bundle, which generally limits the lateral resolution in competing fibre bundle OCT techniques [17,18]. However, the lateral resolution offers an improvement over typical resolutions of ultrasound, which is currently often used for surgical and biopsy guidance. There may be ways to mitigate the losses of lateral resolution, as well as power losses, caused by the diffuser. For instance, an ultra-thin diffuser, as manufactured by 'WaveFront Technology', with a 50 μm thickness (20% of the currently used 250 μm thick diffuser), could be used. Another strategy could be to apply a diffusing layer directly to the bundle, which could reduce the loss of resolution, as has been demonstrated for single fibres, or a more intricate strategy could be to explore the effects of alternative scattering profiles, such as circular or square profiles.

An axial resolution of 34 μm was achieved, comparable to the competing full-field fibre bundle OCT method suitable to endoscopy [17], though worse than typically obtained in fibre-scanning systems, which is often as good as 10 μm [4-6, 27]. This may be a result of modal dispersion in the fibre bundle, and further work is needed to investigate whether this is a fundamental limitation. In theory, the expected axial resolution of the source is approximately 9 μm . Experimentally, when bypassing the bundle and adapter entirely, utilising an in-bulk beamsplitter, with the output superimposed directly onto the camera sensor, an FFT A-scan produced an axial resolution of approximately 14 μm . In the proposed system, no dispersion calibration was performed, so it is difficult to discern what portion of this loss of axial resolution from 14 μm to 34 μm is due to the bundle. By applying a calibration-free method of dispersion compensation [21-23], the axial resolution and SNR may perhaps be improved.

4.2 Practical considerations

In the OCT images, no ghost bands were observed above noise levels, suggesting that the common-path design has effectively suppressed any OCT signals that would have typically resulted from coupling into higher-order modes. OCT images were achieved without any prior calibration of the bundle, i.e. of bending configurations prior to imaging, and without any physical restrictions of the bundle when imaging, which was instead free-hanging and bent during all experiments. Therefore, OCT can be expected to work under all bending configurations, and hence is expected to work in real-life application in endoscopy, whereby the bundle is sporadically bent. In extreme bending conditions there is likely to be increased cross-coupling and hence degradation to the OCT image quality, although such severe bends could perhaps be avoided in real-life application.

The current speed of 0.5 Hz is too slow for real-time or in-vivo imaging, in which motion artefacts would be induced. Faster speeds could be achieved by increasing the camera frames-per-second (fps), at the cost of either reducing integration time or reducing the number of frames. Reducing integration time may cause a reduction in SNR, though only if the current SNR is limited by either shot noise or camera detector noise. If SNR is instead limited by artefacts induced by the bundle, then reducing integration time would not reduce SNR. The achievable increase in acquisition speed that is possible, whilst maintaining adequate sensitivity, remains to be established. We have not conducted a SNR-budget analysis, as this was beyond the scope of the paper. Quantitative analyses of SNR in full-field fibre bundle OCT systems have been conducted and reported elsewhere [17, 26]. Alternatively, reducing the number of frames would have the effect of reducing the axial range below the source-limited axial range of 1.54 mm, which could be a sensible option, considering the current 0.83 mm depth 6 dB decay.

The prototype of the adapter is 8 mm in diameter, due to the width of the lenses and the surrounding casing structure, somewhat oversized for endoscopic use. However, the width could be reduced by decreasing the lens diameter, either using conventional lenses (with 1 mm diameter lenses commercially available) or gradient index optics. Reducing the lens diameter would, for a given focal length, also reduce the required angle of illumination onto the sample and bundle. The angle α scales with diameter and focal length of the lenses, as determined by equation $\alpha = \arctan(D/2f)$ in section 2.1. If NA is reduced, α is smaller, and an improved modal purity of the injected reference wave may occur, albeit at the cost of reduced power in the object image.

With reduction in size, the sensitivity of the system to the alignment of the fibres at the angle α , and the tolerances needed to maintain the OCT signal quality should be considered. The principal

effect of changing α is the effect on modal coupling of the reference wave into the bundle. In our set-up, with the diffuser in place, a similar modal coupling is achieved across the bundle face, which due to the angled divergent injection, incorporates differences in injection angle of approximately $\pm 3^\circ$, suggesting a good tolerance to discrepancies from $\alpha = 14^\circ$ exists with the diffuser in place. The current method permits fine control over the positions and angles of the fibre-tips, such that securing the correct angle within this tolerance is easily achieved, and could be secured at smaller scales as the adapter size is reduced. The correct angles could also be fixed in a finalised clinical version. In Fig. 4, we demonstrated an OCT signal obtained with non-aligned object and reference angles, which qualitatively had a similar signal strength and fidelity to the correctly aligned case, suggesting there is a large tolerance to the relative alignment of angles between the object and reference paths. The true sample shape and orientation is sensitive to their alignment, which may become harder to secure at smaller diameters, however, any offset could easily be calibrated.

4.3 Component redesign

Considering most non-retina clinical applications of OCT use longer wavelengths for enhanced penetration, it could be beneficial to adapt the system to this wavelength using a high-speed InGaAs camera along with a 1300 nm swept source, for which FF-SS-OCT has been demonstrated [29]. At 1300 nm we would also expect less multimoding of cores [14], as well as reduced cross-talk between cores [15]. However, very high-speed InGaAs cameras are less readily available and so this may mean imaging at slower speeds. 1300 nm cameras are also typically noisier, and therefore the resulting sensitivity would need to be evaluated. Alternatively, a higher performing NIR camera for use at 800 nm may deliver improved images from tissue. The camera used (Mikrotron MC1632), has a quantum efficiency of $\leq 30\%$ at 850 nm, however there are commercially available cameras with efficiencies $\geq 50\%$ (Dhyana 400 BSI V3), which may deliver a greater OCT sensitivity.

A current limitation is also the number of pixels in the image, resulting from the limited number of cores in the bundle. Achieving a greater pixel density would be possible with a higher resolution bundle, with smaller core pitch and greater core density, such as Fujikura's FIGH-40-920G with 40,000 cores across approximately 0.9 mm diameter, and FIGH-10-350S with a pitch of $p \approx 6 \mu\text{m}$. However, due to a closer core packing, these bundles exhibit increased inter-core cross-talk [14], and hence the possibility of using such bundles with the adapter would need to be established.

5 Conclusion

This paper has presented a proof-of-concept of a promising new approach to forward-viewing e-OCT, demonstrating OCT functionality of a common-path fibre bundle interferometer, achieving a compact geometry in-line with the bundle via a novel off-axis arrangement of divergent beams. However, due to current limitations in sensitivity, clear images from a biological sample have not yet been demonstrated.

With further refinement, miniaturisation and attention to sensitivity and speed, the adapter could be developed towards integration with endoscopes. In essence, the adapter has the capability of transforming a fibre bundle into a depth-resolving instrument, which with further development could be well-suited to forward-viewing endoscopy.

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Author contributions

H.D. conducted the research and wrote the paper, M.H. supervised the research and wrote the paper, A.P. conceptualised and supervised the research and edited the paper.

Data availability

Data underlying the results presented in this paper are not publicly available at this time but may be obtained from the authors upon reasonable request.

Conflict of interest

A. Podoleanu and M. Hughes are inventors on patent US11717154B2 “Imaging apparatus and method” assigned to University of Kent.

Appendix A - Existing forward-viewing endoscopic OCT methods

A1 Proximal-scanned fibre bundle OCT

Performance metrics of proximal-scanned fibre bundle OCT methods. Those suitable to endoscopy are displayed in bold text. Other methods are unsuitable, either due to having bulk distal optics, not easily miniaturized, e.g. a Michelson interferometer [18], which requires assembly of components sideways to the endoscope thereby exceeding the bundle’s diameter, or due to using a rigid bundle [13].

Table A1. Comparison of point-scanning fibre-bundle OCT configurations and metrics.

Method	Sens./SNR	Lat. res.	Ax. res.	FOV	Axial range	Speed	Diam.
Wurster et al. [16]	86 dB	72–120 μm^*	8 μm	0.5 mm	NR	0.2 Hz[¶]	~1.5 mm
Ford et al. [18]	25 dB	$\geq 10.6 \mu\text{m}^*$	$\geq 8 \mu\text{m}^*$	NR	1 mm	NR	Bulk
Han et al. [27]	NR	$\geq 10 \mu\text{m}^*$	9 μm	2 mm[†]	0.4 mm[‡]	NR	$\geq 2 \text{ mm}^\gamma$
Xie et al. [13]	NR	$\geq 12 \mu\text{m}$	$\sim 10 \mu\text{m}$	2.8 mm [†]	2.8 mm [§]	NR	3.2 mm ^{β}

NR: not reported. Bulk: bulk optics, not suitable for endoscopy. Qualifying symbols: * estimated from fibre/core dimensions or reported system parameters; [†] demonstrated image width rather than a separately specified system FOV; [‡] demonstrated imaging depth; [§] imaging depth achievable with fibre-bundle length >900 mm; [¶] calculated volume acquisition rate; ^{γ} diameter of bundle, inferred from en-face dimensions; ^{β} rigid bundle.

A2 Fibre-scanner e-OCT systems

Performance metrics of fibre-scanning e-OCT methods. Those suitable to endoscopy are displayed in bold text, while [6] is less suitable due to having a diameter exceeding the 3.7 mm diameter of typical endoscope working channels.

Table A2. Comparison of distal fibre-scanning endoscopic OCT configurations and metrics.

Method	Sens.	Lat. res.	Ax. res.	FOV	Axial range	Speed	Diam.
Wu et al. [7]	NR	25 μm	6 μm^*	1.5 mm	NR	0.9 Hz	3 mm
Wurster et al. [6]	96 dB	12 μm	12 μm	1.3 mm	>1.17 mm	0.5 Hz	4 mm
Schulz-Hildebrandt et al. [28]	NR	11 μm	7.5 μm^*	0.8 mm	NR	6 Hz	1.6 mm
Liang et al. [5]	~100 dB	5 μm	9 $\mu\text{m}^\dagger$	0.9 mm	2.1 mm[‡]	5.9 Hz	3.3 mm

NR: not reported. Qualifying symbols: * axial resolution reported in air; [†] axial resolution reported in tissue; [‡] imaging range reported in air.

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