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Analysis of multimode fiber bundles for endoscopic spectral-domain optical coherence tomography

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Abstract

A theoretical analysis of the use of a fiber bundle in spectral-domain optical coherence tomography (OCT) systems is presented. The fiber bundle enables a flexible endoscopic design and provides fast, parallelized acquisition of the OCT data. However, the multimode characteristic of the fibers in the fiber bundle affects the depth sensitivity of the imaging system. A description of light interference in a multimode fiber is presented along with numerical simulations and experimental studies to illustrate the theoretical analysis.

1. Introduction

Several types of optical biopsy imaging systems have been developed to assist clinicians in the diagnosis of disease. In general, these systems provide high-resolution *in vivo* imaging with the potential for real-time diagnosis. Ideally, they are safe and minimally invasive. One such imaging technique is optical coherence tomography (OCT), which has been adapted to endoscopic designs that are able to image internal organs [1]. The introduction of Fourier-domain OCT, including swept-source OCT and spectral-domain OCT (SD-OCT), has been an important step toward achieving clinically practical *in vivo* OCT imaging systems by increasing the acquisition speed and signal sensitivity compared to time-domain OCT systems [2,3]. Fiber-based SD-OCT systems typically employ a broadband source, a point-scanning geometry with a single optical fiber, a dispersing element, and a linear detector array [4–7]. Alternatively, a swept-source approach can be employed by combining a point detector with a source whose wavelength is scanned in time over a given spectral bandwidth [8,9]. In either case, the spectral dimension encodes the depth information while lateral scanning of the point illumination across the sample allows the reconstruction of a 2D cross section (B-scan) of the tissue.

Implementing OCT with a fiber-bundle probe is a conceptually attractive approach for endoscopic imaging. Using a fiber bundle allows the lateral scanning mechanism to be

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located in the proximal optical assembly, which enables the distal end of the catheter to remain stationary on the sample [10–16]. In addition, faster image acquisition can be achieved via a parallelized acquisition through the use of line and full field illumination geometries [11,12,17]. Lastly, fiber-bundle-based OCT systems are compatible with fiber-bundle-based confocal microendoscopes, which simplifies the implementation of multimodality instruments. In a previous publication, we reported on a system that used a line-illumination profile, a fiber bundle, and a 2D detector to achieve a multimodality imaging instrument capable of switching between fluorescence confocal microendoscopy and parallelized SD-OCT [18]. The motivation for the development of this system was to improve disease diagnosis by combining the complementary information provided by confocal and OCT-based imaging systems.

A number of fiber-bundle-based OCT systems have been investigated [8,11,13–16,19–22]. It is generally agreed that aspects of the fiber bundle, such as core-to-core and modal cross talk [20,21,23,24], as well as alignment-sensitive modal power distributions [25], have an impact on image quality. Other reports have described or modeled the sensitivity falloff in SD-OCT [26,27] and the interaction between modes in an imaging fiber bundle [20,22]. While it is generally accepted and stated that multimode imaging fiber bundles have negative effects on SD-OCT system performance, a detailed mathematical description and explanation of the behavior of a multimode fiber and the subsequent consequences on its use in a common-path, SD-OCT imaging system has not been presented. This paper seeks to provide a clear explanation for why use of multimode fiber bundles in OCT systems degrades image quality and results in significantly reduced depth sensitivity. To accomplish this, a detailed mathematical description of SD-OCT illumination and detection fields in a multimode fiber bundle and subsequent imaging by an SD-OCT spectrometer is presented. The performance degradation described by the mathematical system description is observed and verified by comparing a simulated ideal SD-OCT imaging system with experimental results from a fiber-based system.

2. Light Propagation in a Multimode Fiber

This section presents a mathematical description of the effects of employing a step-index multimode fiber in an SD-OCT imaging system. The description begins with the guided modes in a step-index fiber and culminates in the field distribution exiting a single multimode fiber in a common-path OCT interferometer configuration [Eq. (7)].

A. Guided Modes in an Optical Fiber

The general form of a guided mode field in a step-index cylindrically symmetric fiber is

$$\Psi_n(r, \phi, l, t) = Q_n(r, \phi) \exp(-i(\omega t - \beta_n l)), \quad (1)$$

where n is the mode index, r is the radial coordinate, ϕ is the polar coordinate, l is distance along the length of the fiber, t is time, ω is the angular frequency of the light, β_n is the propagation constant, and Q_n is the transverse profile of mode n . In the weakly guided approximation (valid when a small core/cladding index mismatch exists), which is the case

for most practical fibers, the functional form of the mode profile, Q_n , is found to be the orthonormal linear-polarization (LP) mode solutions of the scalar wave equation [28,29]. The propagation constant, β_n , is determined by solving the transcendental dispersion equation that results from boundary conditions on the electromagnetic field in the fiber. The number of modes that are supported without significant loss in the fiber is determined by the fiber characteristics (indices of refraction and core size) and the wavenumber of light. We begin by considering a single multimode core and expand to the case of the imaging fiber bundle in Section 3.C.

B. Monochromatic Illumination Field at Output of a Multimode Fiber

Consider a monochromatic illumination field, U_I , of wavenumber, σ , at the input face of the fiber. The coupling of this illumination field into mode n of a multimode fiber is given by the overlap integral between the transverse profile of the illumination field and the complex conjugate of the transverse profile of mode n :

$$A_{I,n} = \int_0^{2\pi} d\phi \int_0^\infty r dr Q_n^*(r, \phi; \sigma) U_I(r, \phi). \quad (2)$$

The constant $A_{I,n}$ describes the complex amplitude of the field that is guided in mode n ; the square modulus, $|A_{I,n}|^2$, is proportional to the optical power of the illumination in mode n . The total illumination field, $U_{F,l}$, propagated to the distal end of the fiber is a weighted linear combination of the guided modes,

$$U_{F,l}(r, \phi, l; \sigma) = \sum_n A_{I,n} Q_n(r, \phi; \sigma) \exp(i\beta_n l). \quad (3)$$

Although both the amplitude and propagation constants also depend on wavenumber, for notational simplification the explicit dependence is not included; however, this dependence will be considered when describing the result of using a broadband SD-OCT light source later in the paper.

It is useful to point out here that while the orthonormal fiber modes, Q_n , propagate independently along the length of a perfect fiber, core imperfections result in coupling between modes within a single core and between cores when fiber bundles are employed. The mathematics of coupled-mode theory can be used to describe such effects but only have closed form solutions for waveguides with constant or periodic perturbations. Real fiber geometries are significantly more complex than this, and as a result, accurately modeling mode- and core-to-core coupling is impractical, though some success has been found in simulating these interactions for leached fiber bundles with short interaction lengths (~ 1 cm) [20].

An SD-OCT common-path interferometer geometry is now considered and shown schematically in Fig. 1. The field described by Eq. (3) exits the fiber and propagates through a lens to the reference and sample planes. The reflected fields are imaged back to the distal end of the fiber [30,31]. The reflected fields from the reference and sample planes, respectively, can be written as

$$\begin{aligned} U_{R,n}(r, \phi, l; \sigma) &= R_R \exp(-i(\omega t - \theta_R)) \times \sum_n A_{I,n} Q_{R,n}(r, \phi; \sigma) \exp(i\beta_n l) \\ U_{S,n}(r, \phi, l; \sigma) &= R_S \exp(-i(\omega t - \theta_S)) \times \sum_n A_{I,n} Q_{S,n}(r, \phi; \sigma) \exp(i\beta_n l), \end{aligned} \quad (4)$$

where R_R and R_S represent the reflection coefficients from the reference and sample planes, and θ_R and θ_S represent the accumulated phase of the reference and sample waves as they propagate to the locations of the reference and sample planes. $Q_{R,n}$ and $Q_{S,n}$ represent the fiber mode distributions after propagation to the reference and sample planes. The reflected fields propagate back and are coupled into the fiber. These two fields are coupled into the fiber modes, indexed by n' for detection, with coefficients

$$\begin{aligned} A_{R,n,n'} &= \int_0^{2\pi} d\phi \int_0^\infty r dr Q_{n'}^*(r, \phi; \sigma) R_R \sum_n A_{I,n} Q_{R,n}^\dagger(r, \phi; \sigma) \\ A_{S,n,n'} &= \int_0^{2\pi} d\phi \int_0^\infty r dr Q_{n'}^*(r, \phi; \sigma) R_S \sum_n A_{I,n} Q_{S,n}^\dagger(r, \phi; \sigma), \end{aligned} \quad (5)$$

where $Q_{R,n}^\dagger$ and $Q_{S,n}^\dagger$ represent the reflected illumination mode profiles after return propagation to the fiber. The complete fields associated with the reference and sample surfaces at the distal end of the fiber are given by

$$\begin{aligned} U_{R,n,n'}(r, \phi, l; \sigma) &= \sum_{n'} \sum_n A_{R,n,n'} Q_{n'}(r, \phi; \sigma) \times \exp(-i(\omega t - \theta_R)) \exp(i\beta_n l) \\ U_{S,n,n'}(r, \phi, l; \sigma) &= \sum_{n'} \sum_n A_{S,n,n'} Q_{n'}(r, \phi; \sigma) \times \exp(-i(\omega t - \theta_S)) \exp(i\beta_n l). \end{aligned} \quad (6)$$

The reference and sample fields combine linearly in the fiber and propagate to the proximal end, where the total field, $U_{F,n,n'}$, is expressed as

$$U_{F,n,n'}(r, \phi, l; \sigma) = \sum_{n'} \sum_n [A_{R,n,n'} \exp(i2\theta_R) + A_{S,n,n'} \exp(i2\theta_S)] Q_{n'}(r, \phi; \sigma) \exp(i(\beta_n + \beta_{n'})l). \quad (7)$$

While the guided modes of a fiber are conveniently described in a polar coordinate system, we convert to a Cartesian system from this point forward and drop the time dependence. This will allow for simpler mathematical forms when the dispersing element in the SD-OCT system is introduced and irradiance distributions are calculated. Note that the Cartesian versions of $U_{F,n,n'}$ and $Q_{n'}$ do not have the same functional form as their polar counterparts; however, the notation is maintained for simplicity.

3. Mathematical Description of the SD-OCT Signal

A. Monochromatic Imaging from a Single Multimode Fiber

Mathematical expressions for the irradiance distribution at the detector [Eq. (16)] and the detector signal produced by a common-path SD-OCT system [Eq. (20)] with a single multimode fiber are now derived. The geometry considered and variables used are shown in Fig. 2. Note that in OCT systems employing a single optical fiber, the fiber itself acts as a confocal aperture. When using a fiber bundle, a separate confocal aperture is required and is therefore included in the configuration shown in Fig. 2. For simplicity, the grating is shown

with an unfolded optical path, though a reflective blazed grating is typically used in SD-OCT systems.

The object field distribution at the proximal end of the fiber bundle, $U_{F,n,l}$, is coherently imaged through the system to the confocal aperture and subsequently to the detector plane. There, the detectable irradiance is equal to the square magnitude of the imaged object field distribution. We assume that the point spread function (PSF) of the lens system that images the proximal face of the fiber to this confocal aperture is small relative to the fiber core and the confocal aperture. It is therefore appropriate to model this lens system as an exact mapping, with magnification m_{fa} . For a single fiber SD-OCT system, the field located after the aperture plane is given by

$$U_{ap}(x_{ap}, y_{ap}, l; \sigma) = U_F \left(\frac{x_{ap}}{m_{fa}}, \frac{y_{ap}}{m_{fa}}, l; \sigma \right) \frac{1}{m_{fa}^2} t_{ap}(x_{ap}, y_{ap}) \\ = \sum_{n'} \sum_n [A_{R,n,n'} \exp(i2\theta_R) + A_{S,n,n'} \times \exp(i2\theta_S)] Q_{n'} \left(\frac{x_{ap}}{m_{fa}}, \frac{y_{ap}}{m_{fa}}; \sigma \right) \times \exp(i(\beta_n + \beta_{n'})l) \frac{1}{m_{fa}^2} t_{ap}(x_{ap}, y_{ap}), \quad (8)$$

where t_{ap} describes the transmission of the confocal aperture. If it is necessary to consider the effect of diffraction in imaging from fiber to aperture, the field U_{ap} is described as a linear superposition of diffraction-modified mode distributions Q_n and the following analysis remains valid.

1. Mapping of the Spectral Distribution to Position in the Detector Plane

A second imaging system maps the light in the confocal aperture to the detector plane through a dispersive grating. The spectral dispersion of the grating introduces a wavenumber-dependent shift in the detector plane. If the diffraction grating is oriented such that the central emission wavenumber of the source is mapped to the center of the detector, the relationship between detector position, y , and wavenumber, σ , is given by

$$y_\sigma = f_{gd} \tan \left[\arcsin \left(\frac{1}{p\sigma} - \sin\theta_i \right) - \arcsin \left(\frac{1}{p\bar{\sigma}} - \sin\theta_i \right) \right], \quad (9)$$

where f_{gd} is the focal length of the lens between the grating and detector plane, p is the pitch of the grating, θ_i is the incident angle of the light with respect to the grating normal, and $\bar{\sigma}$ is the central emission wavenumber of the source, such that $y_{\bar{\sigma}}(\sigma) = 0$.

2. Anamorphic Effects of Diffraction Gratings

A grating diffracts light at an angle with respect to the incident beam, which results in an anamorphic effect. If a square grating is the restricting aperture of the system, then it will instead act as a rectangular pupil. If a circular beam underfills the grating, the anamorphic magnification causes the circular pupil to become elliptical. In either case, the grating produces a unique change of lateral magnification in the dispersion direction [32], which yields a coherent PSF (proportional to the Fourier transform of the pupil function) that is scaled differently in the dispersion direction. The inverse of this anamorphic scale factor is the grating angular magnification and is given by

$$m_{\alpha}(\sigma) = \frac{d_i}{d_d} = \frac{\cos\theta_d}{\cos\theta_i}, \quad (10)$$

where d is the beam diameter, θ is the beam angle, and the subscripts i and d refer to the incident and diffracted beams, respectively. The wavenumber dependence on the angular magnification arises because the diffraction angle depends on wavenumber. Because the focal length of the lens imaging the confocal aperture is long compared to the size of the aperture, the small change in incident angle at the grating as a function of position within the confocal aperture is ignored in the following analysis.

3. Imaging to the Detector Plane

The field distribution in the detector plane is calculated by convolving the field at the aperture plane with the coherent PSF of the imaging system, while taking into account the position shift caused by the presence of the diffraction grating. The resulting field on the detector is

$$U_{\text{det}}(x, y - y(\sigma), l; \sigma) = \frac{1}{m_{ad}^2 m_{\alpha}} \iint dx' dy' U_{ap} \left(\frac{x'}{m_{ad}}, \frac{y'}{m_{ad} m_{\alpha}}, l; \sigma \right) \times p_{\text{coh}}(x - x', (y - y(\sigma)) - y'; \sigma), \quad (11)$$

where m_{ad} is the magnification of the imaging system that maps the confocal aperture to the detector plane, and p_{coh} is the coherent PSF, which is given by

$$p_{\text{coh}}(x, y, \sigma) = \mathcal{F}_2 \{ P(x'', m_{\alpha} y'') \}_{x'' \rightarrow \frac{x\sigma}{f_{gd}}, y'' \rightarrow \frac{y\sigma}{f_{gd}}}. \quad (12)$$

Here, $\mathcal{F}_2$ denotes the 2D Fourier transform, with the symbol $\rightarrow$ indicating the transformation from the function variable to the Fourier conjugate variable. The function $P(x'', y'')$ is the pupil distribution in the system, which is assumed to be located at the grating and is properly scaled in the dispersion direction by the angular magnification.

The relationship between the field distribution at the detector and the field out of the fiber is therefore given by

$$\begin{aligned} & U_{\text{det}}(x, y - y(\sigma), l; \sigma) \\ &= \frac{1}{m_{ad}^2 m_{\alpha} m_{fa}^2} \iint dx' dy' \sum_{n'} \sum_n [(A_{R,n,n'} \exp(i2\theta_R) \\ &+ A_{S,n,n'} \exp(i2\theta_S)) Q_{n'} \left(\frac{x'}{m_{fa} m_{ad}}, \frac{y'}{m_{fa} m_{ad} m_{\alpha}}; \sigma \right) \\ &\quad \times \exp(i(\beta_n + \beta_{n'})l)] t_{ap} \left(\frac{x'}{m_{ad}}, \frac{y'}{m_{ad} m_{\alpha}} \right) \\ &\quad \times p_{\text{coh}}(x - x', (y - y(\sigma)) - y'; \sigma). \end{aligned} \quad (13)$$

We define a new function, $Q_{\text{det},n'}$, that represents the wavenumber-dependent spatial profile of the signal in the detector plane for each propagating fiber mode, where

$$Q_{\text{det},n'}(x, y-y(\sigma); \sigma) = \iint dx' dy' Q_{n'} \left(\frac{x'}{m_{fa}m_{ad}}, \frac{y'}{m_{fa}m_{ad}m_{\alpha}}; \sigma \right) \times t_{ap} \left(\frac{x'}{m_{ad}}, \frac{y'}{m_{ad}m_{\alpha}} \right) p_{\text{coh}}(x-x', (y-y(\sigma))-y'; \sigma). \quad (14)$$

With this definition, it follows that

$$\begin{aligned} U_{\text{det}}(x, y-y(\sigma), l; \sigma) &= \frac{1}{m_{ad}^2 m_{\alpha} m_{fa}^2} \\ &\times \sum_{n'} \sum_n (A_{R,n,n'} \exp(i2\theta_R) \\ &+ A_{S,n,n'} \\ &\times \exp(i2\theta_S)) Q_{\text{det},n'}(x, y-y(\sigma); \sigma) \\ &\times \exp(i(\beta_n + \beta_{n'})l). \end{aligned} \quad (15)$$

4. Irradiance at the Detector Plane

The irradiance is proportional to the square modulus of the field at the detector plane. Apart from the constant of proportionality, the irradiance is given by

$$\begin{aligned} I_{\text{det}}(x, y-y(\sigma); \sigma) &= |U_{\text{det}}(x, y-y(\sigma))|^2 \\ &= \frac{1}{m_{ad}^4 m_{\alpha}^2 m_{fa}^4} \sum_{n'=1}^M \sum_{n=1}^M \sum_{m'=1}^M \sum_{m=1}^M (A_{R,n,n'} \exp(i2\theta_R) \\ &\quad + A_{S,n,n'} \exp(i2\theta_S)) (A_{R,m,m'} \exp(i2\theta_R) \\ &\quad + A_{S,m,m'} \exp(i2\theta_S))^* \\ &\times Q_{\text{det},n'}(x, y-y(\sigma); \sigma) Q_{\text{det},m'}^*(x, y-y(\sigma); \sigma) \\ &\times \exp(i(\beta_n - \beta_{n'} + \beta_m - \beta_{m'})l). \end{aligned} \quad (16)$$

It can be shown that Eq. (16) (with dependencies dropped for simplicity) reduces to the following form

$$\begin{aligned} I_{\text{det}} &= \sum_{j=1}^J [c_j + a_j \cos(2(\theta_R - \theta_S) + \varphi_j)] |X_j| \\ &= \sum_{j=1}^J V_j(\sigma) |X_j|, \end{aligned} \quad (17)$$

where j now indexes over all possible combinations of the previous mode indices n , n' , m , and m' . For the case of a single multimode fiber, $J = M^4$, where M is the number of supported modes. The variables c_j , a_j , and φ_j represent an intensity offset, an amplitude of a cosine, and a phase offset of a cosine, respectively. X_j is a spatial distribution that results from the product of the mode distributions $Q_{\text{det},n'} Q_{\text{det},m'}^*$. The phase φ_j has several

contributions, which include: (1) the phase of amplitude products $A_{R,n,n'} A_{S,m,m'}^*$; (2) the phase of the propagation terms $\exp(i(\beta_n - \beta_{n'} + \beta_m - \beta_{m'})l)$; and (3) the phase of X_j , which is spatially dependent. All of the quantities c_j , a_j , ϕ_j , and X_j depend on wavenumber σ . The fundamental dependence for SD-OCT is in the phase difference $2(\theta_R - \theta_S)$, which is proportional to the optical path difference (OPD) between the reference and sample waves and is a linear function of wave-number given by

$$2(\theta_R - \theta_S) = 2\pi\sigma\Delta, \quad (18)$$

where Δ is the round trip OPD between the sample and reference fields. The significance of the wave-number dependence in the other quantities will be discussed at the end of Section 3.B.

The irradiance at the detector plane of the system shown in Fig. 2 is measured by an array of detector elements, indexed in the dispersion direction by k , with individual detector profiles

$$h_{\text{det},k}(x, y, \sigma) = \eta(\sigma) \text{rect}\left(\frac{x}{a}, \frac{y-y_k}{a}\right), \quad (19)$$

where a is the size of the detector pixel elements, and $\eta(\sigma)$ is the spectral quantum efficiency of the detector. For a monochromatic input, the signal out of a detector pixel k , integrated over the exposure time, is proportional to

$$S_{\text{out},k} = \int dt \iint dx dy I_{\text{det}}(x, y - y(\sigma); \sigma) \eta(\sigma) \text{rect}\left(\frac{x}{a}, \frac{y-y_k}{a}\right). \quad (20)$$

B. Polychromatic Imaging Considerations

When the illumination is polychromatic the analysis above still holds, with some modifications. Both the energy distribution within the aperture and the imaging system point response function generate a spatial spread (blurring) at the detector plane [Eq. (16)] that varies with wavenumber. Because of this spatial blurring, each point in the detector plane receives energy over a range of wavenumbers. Although the bandwidth of this spreading is small, it affects the signal analysis. The measured signal out of the detector in the polychromatic case is determined by integrating the previous result in Eq. (20) over wavenumber,

$$S_{\text{out},k} = \int d\sigma \int dt \iint dx dy I_{\text{det}}(x, y - y(\sigma); \sigma) \times \eta(\sigma) \text{rect}\left(\frac{x}{a}, \frac{y-y_k}{a}\right). \quad (21)$$

This expression now contains both the wavenumber dependence of the blurring factors contained within I_{det} and the nonlinear relationship between wavenumber and position on the detector [Eq. (9)] caused by the grating. The nonlinearity between detector location, y_k , and wavenumber is typically corrected by interpolation and resampling of the data. This action can be described by an operator $\mathcal{O}$, the discrete inverse of Eq. (9), which maps

detector position y_k to uniformly spaced wavenumbers σ_k . A discrete Fourier transform of the resampled data yields the final output, $R_{\text{out}}(\Delta)$, which is the OCT A-scan response as a function of OPD, Δ . A point object's OPD is related to its depth, z , via the relationship $\Delta = 2nz$, where n is the refractive index of the media, and the factor of 2 accounts for the round trip path of light in the SD-OCT interferometer. These operations can be written in a shorthand notation as

$$R_{\text{out}}(\Delta) = \mathcal{F}_1 \{ \mathcal{O} \{ [I_{\text{det}} * h_{\text{det}}](y_k) \}_{y_k \rightarrow \sigma_k} \}_{\sigma_k \rightarrow \Delta}, \quad (22)$$

where $[I_{\text{det}} * h_{\text{det}}](y_k)$ represents the convolution of the two functions with output coordinate y_k , and $\mathcal{F}_1$ represents the 1D Fourier transform.

In SD-OCT, it is the modulation depth of the $\cos(2\pi \sigma \Delta)$ term that leads to the OCT image strength at a specific OPD (depth). In a multimode fiber with M modes, Eq. (17) indicates that there are J irradiance terms that have the same fundamental cosine modulation frequency, but with varying amplitudes and phases. Because of the varying phases and amplitudes, the average net modulation depth, which is equivalent to the visibility, $V = (I_{\text{max}} - I_{\text{min}})/(I_{\text{max}} + I_{\text{min}})$, is reduced (i.e., cosine modulation contrast is lost) from what would occur in a single-mode fiber. In addition, since the amplitudes, a_j , and phases, ϕ_j , vary with wavenumber, there is spreading of the axial point response following the Fourier transform operation, which leads to a reduction in axial resolution.

C. Case of a Fiber Bundle

When a fiber bundle, composed of many thousands of multimode fibers (or “cores”), is employed for OCT imaging, additional factors are introduced. Depending on how the illumination profile aligns with the cores of the bundle, there is increased variation in the irradiance distribution at the detector plane that affects imaging performance.

In a point-scanning fiber-bundle OCT system, the point illumination sometimes couples into a single fiber core, while at other times it couples into multiple cores with reduced efficiency. Furthermore, even if a single core is illuminated, there is core-to-core cross talk that results in the propagation of energy through multiple cores. It has been shown that cross talk is significantly suppressed by nonuniformities in the core size, shape, and core-to-core spacing [23,24,33] common to most imaging fiber bundles. Moreover, multimode fibers typically provide high mode confinement in their cores due to their relatively high V-number and NA [28]. Experimental results have demonstrated that cross talk is not a significant issue for endoscopic imaging using typical fiber bundles [24,33]. Nevertheless, core-to-core cross talk can have an effect on performance in SD-OCT systems that utilize a fiber bundle [34]. Due to illumination from multiple cores and core-to-core cross talk, the number of modes contributing to the detected irradiance in Eq. (17) increases ($J > M^2$). This leads to even greater reduction in modulation contrast and reduced OCT system sensitivity. Furthermore, the spatial variation in both the fiber coupling efficiency and modulation cancellation also leads to lateral spatial intensity variation in the OCT image [35] as the point illumination is scanned across the fiber face.

In a parallel line-illumination system, a line simultaneously illuminates the fiber bundle across the full lateral extent of the image field and a slit serves as the confocal aperture. This configuration has even greater potential for multiple cores to contribute illumination to a given point in the object. As the illumination and return signal on the fiber bundle blur with depth, the situation of multiple core contribution is exacerbated. The more defocus there is, the more overlap there is in illumination from neighboring cores, leading to yet more terms, and more phase cancellation in the final modulation. While this effect is present in the point-illumination geometry, it is more significant when using a parallel line-illumination configuration.

4. Inherent Depth-Dependent Signal Sensitivity in SD-OCT

In all SD-OCT systems there is a depth-dependent falloff in sensitivity due to the blur of the irradiance distribution at the detector [36]. The system sensitivity with depth is further reduced in a fiber-bundle OCT system as a consequence of the multimode nature of individual fiber cores and the contributions of multiple illuminating cores. To determine the inherent depth-sensitivity falloff, it is necessary to calculate the Fourier transform of the monochromatic mode distribution, I_{det} , as defined by Eq. (16), assuming a single-mode fiber. The presence of optical aberrations in the system, and the quality of the spectral calibration, are assumed to be negligible effects in this analysis.

To model the inherent depth-sensitivity falloff, a monochromatic LP_{01} detection mode at the output of a single-mode fiber is assumed to be imaged onto the aperture plane, multiplied by the aperture function, and convolved with the imaging PSF to determine the detector irradiance. The resulting $I_{\text{det}}(x, y - y(o); o)$ is then convolved with a rect function that describes the detector pixel geometry. The resulting detector signal, linear in y_k , is converted to σ_k through a sinc function interpolation and nonlinear resampling. The Fourier transform of the resulting distribution yields the depth sensitivity profile. The calculation of depth sensitivity is repeated for a variety of wavenumbers across the bandwidth of the source to assess the average depth sensitivity that is expected.

A calculation of this inherent depth sensitivity was done for a previously reported SD-OCT imaging system [18], which is the same system used to generate and compare the experimental results shown in the next section. The fiber core diameter in this calculation is $2.5 \mu\text{m}$, and the wavelength range of the source is from 803 to 878 nm. Figure 3(a) shows the inherent average depth sensitivity falloff for a single fiber LP_{01} mode in perfect alignment with the confocal aperture. As a comparison the same calculation was done for the LP_{11} mode. The error bars indicate the standard deviation of the calculation across the varying wavenumber of the source. Figure 3(b) shows the depth sensitivity, averaged for 10 different relative shifts, when multiple cores in a fiber bundle are illuminated and fall inside the confocal aperture. Each core receives a fraction of the total energy proportional to the overlap between the core and the image of the line illumination at the fiber plane. The error bars again represent the standard deviation of the calculation across varying source wavenumber. The calculation shows that there is a loss of average sensitivity due to illumination/core misalignment when using a fiber bundle in an SD-OCT system. While there is a slight dependence in this inherent depth sensitivity as a result of modal

characteristics, the primary factor influencing the depth sensitivity is the point response function of the imaging system.

5. Experimental Results

A. System Description

A previously reported system [18] was used to illustrate and validate the concepts presented in this paper. The system is a parallelized, line-illumination, fiber-bundle-based common-path SD-OCT instrument. A 1 m long, 30,000 element imaging fiber bundle with $\sim 2.5 \mu\text{m}$ diameter cores and $\sim 4 \mu\text{m}$ core-to-core spacing (Sumitomo Electric USA, Torrance, CA) was used. We modified the system to use a super luminescent diode (SLD) with a central emission wavelength of 841 nm and a FWHM of 46.9 nm (InPhenix IPSDD0808-3113). At this central emission wavelength, the imaging fiber bundle has a $V = 3.2$, indicating that only two LP modes are supported. A spectral bandwidth of 75 nm was dispersed across the 512 pixels of the 2D CCD camera, which corresponds approximately to the full width at 15% of the SLD's maximum power. The theoretical axial resolution of the system is $6.7 \mu\text{m}$ in air ($4.8 \mu\text{m}$ in tissue), and the theoretical maximum unaliased imaging depth is 1.2 mm in air ($850 \mu\text{m}$ in tissue). A $25 \mu\text{m}$ wide slit was used as the aperture. A spectral calibration procedure [37] was used to obtain OCT data samples on a uniformly spaced grid with respect to wavenumber. Using the SD-OCT system described here, data are acquired, processed, and displayed at 10 frames/s. For point-illumination experiments, the cylindrical lens that forms the illumination line was removed.

To demonstrate the concept of interference transfer through a fiber bundle, two thick glass plates were placed in near contact at the focal plane of the objective (in object space). The plate nearest to the objective was kept fixed. Reflection off the far surface of the first glass plate served as the reference field; the reflection off the near surface of the second glass plate acted as the sample field. Data were collected with and without a fiber bundle in place. Figure 4 shows a comparison of the raw and processed depth image results. Self-interference in the back-thinned CCD silicon substrate leads to low-frequency spatial noise in the raw data [not noticeable in the highly cropped raw data images shown in Figs. 4(a) and 4(b)] that manifests itself as a noise signal at shallow depths in the image domain. This signal is more apparent in the fiber bundle image because of the image enhancement necessary to see the lower strength signal from the planar reflector. In addition, there is structured background noise associated with use of a fiber bundle.

The modulation contrast observed in the fiber-based interferogram [Fig. 4(b)] is clearly less than that without the fiber bundle in place [Fig. 4(a)]. It can also be seen that there is significantly increased lateral variation in the signal intensity with a fiber bundle in place. This is a result of the variable alignment between both the line illumination and the fiber cores, as well as between the confocal slit aperture and the fiber cores. The effect can be exacerbated by core-to-core cross talk. While performance is visibly reduced when using a multimode fiber bundle, this experiment demonstrates that the fiber bundle maintains the basic phase relationship between two interfering waves and transfers the interference of light created by the common-path interferometer.

B. Comparison to an Ideal SD-OCT Interferometer

By comparing the measured irradiance received from the reference and sample surfaces independently it is possible to determine the expected maximum fringe visibility (contrast), which is given by $V_{\max} = (2|A_R||A_S|)/(|A_R|^2 + |A_S|^2)$, where $|A_R|^2$ and $|A_S|^2$ are the reference and sample irradiances, respectively. The value $|A_R|^2$ is measured by removing the sample surface from object space and subtracting the background signal. The value $|A_S|^2$ is determined by subtracting $|A_R|^2$ and the background signal from the total spatially averaged irradiance when the sample surface is reinserted.

Figure 5 shows the loss of fringe visibility as a function of depth for the line- and point-illumination systems. The reduction in $V_{\max}$ is due to decreasing signal amplitude with depth that occurs as a result of beam defocus and from rejection of out of focus light by the confocal aperture. The product of the depth sensitivity from the ideal system performance (Fig. 3) and the measured maximum fringe visibility (Fig. 5) represents what would be the best case depth sensitivity performance for the SD-OCT line- and point-illumination systems used in these experiments.

C. Measurement of Depth Sensitivity

Depth sensitivity measurements were taken for both point- and line-illumination configurations. Two glass microscope slides were used to create the reference and sample surfaces of the interferometer, with the reference surface placed before the focus of the illuminating beam by approximately 25 μm . This was done in an attempt to balance defocusing and power between the reference and sample surfaces. OCT data associated with a single backreflecting surface were recorded as a function of depth by translating the sample surface. The signal intensity in the resulting OCT image was averaged over a finite lateral range and the Fourier transform was taken. The ratios between the contained energies in the resulting signal for each depth location and the corresponding DC signal (an equivalent measurement to fringe visibility) were recorded to generate a plot of the signal sensitivity as a function of imaging depth.

Figure 6 shows the experimental results (measured data points) and the expected depth sensitivity for the LP_{01} (solid line) and LP_{11} (dashed line) modes. As described in Section 5.B, the expected depth sensitivity is the product of the simulated theoretical average depth sensitivity profile due to blurring at the detector [Fig. 3(a)] and the experimentally measured maximum fringe visibility (Fig. 5).

For both the line- and point-illumination profiles, the measured performance with a fiber bundle is significantly degraded from the expected best-case situation. As described in the mathematical development above, for a bundle of multimode fibers the final detected irradiance [Eq. (17)] contains a sum of cosinusoidal modulation terms formed by the interference of reference and sample beam combinations of all the illumination and detection modes from multiple cores. The varying amplitudes and phases of the cosinusoidal terms leads to modulation cancelation and reduced OCT signal sensitivity. Figure 6 shows that the system sensitivity is reduced to about 40% at zero OPD and falls off rapidly with increasing depth.

The sensitivity at zero OPD is reduced as a consequence of the multimode nature of the fiber bundle used. For the case of two supported fiber modes in a single core, a total of 16 individual irradiance terms with varying intensities and phases are imaged to the detector plane [Eq. (17)]. To simulate the performance that would be achievable in this situation, the average modulation depth between 16 phasors was calculated with uniformly distributed random phases and random intensities (cascaded appropriately, such that the total illumination irradiance is split between illumination modes and each illumination split again into detection modes). The resulting cosine modulation was found to be roughly 40% of the maximum possible modulation, which is consistent with the measured data shown in Fig. 6 for both point and line illuminations.

As the object plane is defocused, the illumination spot size increases and a given sample point will receive light from an increasing number of adjacent fiber cores. This exacerbates the modulation contrast reduction by introducing more modulation terms with varying phases. As the number of modulation terms increases with defocus, this manifests as a more severe depth-dependent reduction in system sensitivity. At a depth of approximately 120 μm , the geometric blur causes the illumination from three adjacent cores to overlap.

While the effect of defocus is more significant in the case of line illumination due to the greater number of adjacent cores simultaneously used for illumination and detection, the point-illumination result also shows a reduction in system sensitivity with depth. This is a consequence of optical aberration, imperfections in illumination to fiber core alignment, and core-to-core coupling all causing more than one core to illuminate the object. However, the maximum number of illuminating cores is still fewer than in the line-illumination case, and so the depth sensitivity does not fall off as rapidly.

To further verify that the multimode fiber bundle effects are responsible for the reduction in depth sensitivity, normalized signal intensity as a function of imaging depth was measured using the line-illumination system configured in three different setups.

In the first setup, the fiber bundle was removed and an interferometer with separate reference and sample arms was built. Both the reference and sample arms had identical focusing optics and mirrors. In the second setup, the fiber bundle was inserted and the interferometer from the first setup was placed at the distal end of the fiber bundle. The fiber bundle in this case was used in a single-pass configuration in detection only. In the third setup, illumination and collection of light occurred through the fiber bundle (double-pass). In this case, a common-path interferometer was implemented at the distal end of the fiber bundle. Data were collected in the same manner as those used to generate Fig. 6.

The results from these three experimental configurations are shown in Fig. 7. Figure 7(a) shows that 50% signal falloff occurs at a depth of about 360 μm in the free-space setup, which agrees well with the calculated depth performance shown in Fig. 6(a). In the case of the single-pass fiber-bundle setup, the performance is degraded (50% reduction around 300 μm). This demonstrates that the sum of multiple modes in detection does affect the depth sensitivity. In the double-pass configuration the performance is more significantly degraded

(50% reduction around 90 μm), consistent with the theoretical explanation and the experimental result shown in Fig. 6(a).

D. Depth-Dependent Background Noise

In a multimode fiber, the irradiance cross terms have a βl dependence that affects the phase of the cosine modulation [Eq. (17)]. This dependence varies significantly with wavenumber and is influenced by bending of the imaging fiber. Figure 8(a) shows a portion of raw data that contains five fringes, corresponding to a sample reflectance at a depth of 50 μm . In addition to the desired modulation and lateral variability mentioned previously, there is significant variability in signal intensity in the spectral (vertical) dimension, which results from both spectrally varying cross-modal interference within a single core and coupling of fields between adjacent cores.

By physically moving (jiggling) the fiber bundle during data acquisition, the variability in the signal intensity is reduced as shown in Fig. 8(b). This is a consequence of the fiber motion affecting the modal propagation constants. When the induced time-varying βl differences are sufficiently large within a single camera integration cycle, some of the spectrally dependent variation in the raw signal averages out. Figure 8(c) shows depth plots through the two SD-OCT images obtained with (dotted line) and without (solid line) fiber jiggle. The object reflectance is observed as a peak with unit intensity at the depth of 50 μm . It is evident that the depth-dependent background “noise” is reduced when the fiber is jiggled.

E. Axial Resolution with Depth

It is common in OCT systems to describe the depth resolution by an axial PSF that is the Fourier transform of the source power spectrum. However, in addition to the $2\pi\sigma$ dependence that forms the basis for the fundamental cosinusoidal OCT signal, there is wavenumber dependence in the phase and amplitude of the modulation signal. This additional variation with wavenumber modifies the axial PSF. Figure 9 shows OCT images for a plane reflector at two different depths and plots of the response as a function of depth for a small region around the object. The plots were obtained by projecting a portion of the image data along the lateral direction. The DC component and noise at depths above the object have been zeroed to improve object visibility.

The results show an axial resolution of approximately 10 μm , which is a slight reduction from the axial resolution of 6.7 μm based on the source power spectrum, which may be a result of the spectrally dependent amplitude and phase variations of the cosine modulation. As seen from the results in Fig. 9, the loss of axial resolution does not depend on object depth and does not explain the depth sensitivity fall-off. The images in Fig. 9 also show that the signal to background noise ratio of the object decreases with increasing depth as expected. As with any OCT system, axial resolution can be improved by increasing the bandwidth of the source. However, these results indicate that the spectrally dependent amplitude and phase variations inherent to fiber-based OCT systems are not a significant factor affecting axial resolution.

6. Discussion

One of the potential advantages of using a fiber bundle is the ability to implement a parallelized SD-OCT system to obtain a cross-sectional depth image of tissue with no moving parts or scanning required. However, in a parallelized system multiple fiber cores are illuminated simultaneously. The beams from these cores overlap increasingly with defocus, leading to more interference terms with greater phase variation and a decreasing sensitivity with depth. Our results show a significant loss of overall sensitivity and a more rapid falloff with depth for a parallelized fiber bundle SD-OCT system. The situation is improved for a point-scanning geometry, but there is still a reduction in depth sensitivity from the coupling of the illumination point into multiple cores as it is scanned across the fiber face. Our depth sensitivity measurements are consistent with our analysis, which appears to fully explain the performance degradation observed in a fiber-bundle SD-OCT system where each core supports multiple modes.

In addition to reducing depth sensitivity, the variations across the fiber face in illumination-to-core alignment, core-to-aperture alignment, and cross coupling of light between cores adds lateral structured noise to the OCT images. This structured noise is significant and degrades image quality.

It is possible that the primary problem associated with multimode fibers can be overcome by using a SLD with a wavelength range that enables each fiber in a fiber bundle to act as a single-mode waveguide. However, some amount of cross talk will still cause each core to propagate several fields with different phases. In addition, the issues of multiple modes and defocus will still exist when dealing with parallel illumination systems. Furthermore, commercially available fiber bundles are single mode at a wavelength that is beyond the limit of silicon-based detectors. Structured background noise can be reduced by jiggling the fiber, which introduces a temporal variation in the propagation constant β . If the induced changes are sufficient within a single integration cycle of the detector, some of the wavenumber-dependent variations in the detected signal are averaged out, which in turn reduces the structured background noise that extends over a significant range of depths in the OCT image.

Besides the fundamental problems of reduced sensitivity and increased background noise, there are additional aspects of using fiber bundles in OCT systems. For example, Fresnel reflections from the proximal and distal fiber faces lead to large background signals that use up dynamic range in the detector. Use of anti-reflection coatings, index matching gels, or glass windows glued to the fiber surfaces can significantly reduce this background and improve the signal to background ratio, but residual surface reflections still degrade performance.

Another issue is that commercial fiber bundles currently available and appropriate for endoscopic imaging applications have numerical apertures on the order of 0.35. However, SD-OCT systems designed to image over a depth of a millimeter or more require a low numerical aperture (typically 0.1 or lower) to maintain lateral resolution over a large depth of field. Thus the fiber to tissue imaging optics will ideally have a high magnification equal

to the ratio of the fiber-space and tissue-space numerical apertures. However, a high magnification between fiber and tissue produces large individual fiber pixels in tissue space, which reduces lateral resolution in the final OCT image. One can operate with a lower magnification and reduced numerical aperture in the tissue space, but this results in a waste of illumination light. Although not implemented in this work, interesting techniques have been reported to improve the trade-off between lateral resolution and depth of focus in OCT [38–41], so this may not be a fundamental issue for OCT systems, including those that might employ fiber bundles.

7. Conclusions

The analysis and experimental results show that SD-OCT imaging can be accomplished using a fiber bundle. The basic phase modulation of the signal with wavenumber for a reflector at a specific OPD is maintained in propagation through a fiber bundle. However, the performance of fiber-bundle-based OCT systems is significantly degraded by the multimode nature of the individual fiber cores and the inherent characteristics of using multiple cores. The total phase modulation contrast is reduced by the multiple modes supported in each core and by the multiple cores that contribute to the illumination and detection of light to and from the object. These deleterious effects are exacerbated in a parallelized SD-OCT system where multiple cores along the lateral extent of the fiber and object are illuminated simultaneously. The increasing number of cores that contribute to the illumination of an object point as its depth increases amplify the sensitivity falloff and limit the depth over which an OCT image signal of adequate quality can be obtained. Nevertheless, the advantages of using fiber bundles might outweigh the disadvantages in certain applications and advances in technology or alternative system configurations could result in improved image quality in fiber-bundle-based OCT systems in the future.

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29. It should be noted that the LP modes contain degenerate polarization states, effectively increasing the number of modes that propagate in a fiber. While these modes experience slightly different propagation constants, the difference is negligible for the lengths of fibers used in endoscopic imaging systems. As a result, the degenerate polarization states are phase matched, and energy may couple between them readily.
- 30.

If multiple LP modes are supported in propagation, the output field from the fiber will be spatially inhomogeneous in polarization. The effect of a given object on the polarization of the reflected light, and the consequences of this interaction on the system sensitivity are outside the scope of this manuscript, though extensive literature has been published on polarization-sensitive OCT.

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In addition to this, the lack of globally preferred coordinates means that the LP propagation modes will rotate from core to core, causing polarization-inhomogeneous illumination fields that vary in their ability to interfere spatially with the fields transmitted by adjacent cores. This effect increases the amount of lateral spatial intensity variation in the OCT image, though it may act to reduce the defocus penalty if the introduced modulation terms are of orthogonal polarizations.

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Fig. 1.
Simplified diagram of a common-path SD-OCT interferometer.

Fig. 2.
Simplified system diagram of an SD-OCT imaging path from the proximal end of the fiber to the detector plane.

Fig. 3.

Inherent depth sensitivity falloff with object depth for (a) two individual modes of a single core fiber in perfect alignment with the illumination and confocal aperture, and (b) average performance for a fiber bundle where illumination and aperture alignment can shift relative to the core positions.

Fig. 4. Raw data (top row) and processed depth images (bottom row) for a planar reflector in a free-space SD-OCT system (left column) and a fiber bundle SD-OCT system (right column).

Fig. 5. Maximum fringe visibility as function of depth for the cases of line illumination (left) and point illumination (right).

Fig. 6. Measured and calculated depth sensitivity curves for (a) line- and (b) point-illumination configurations.

Fig. 7.

Normalized signal sensitivity as a function of depth for the SD-OCT system reported in [18] in the case of (a) a free-space setup with separate reference and sample arms, (b) a single-pass fiber-bundle setup with separate reference and sample arms, and (c) a double-pass fiber-bundle setup with a common-path interferometer.

Fig. 8.

Raw spectral data in the line-illumination case, with the fiber (a) held still and (b) attached to an oscillating servo motor arm during data acquisition. The average depth-dependent signal intensity is plotted in (c). The object location is indicated by an arrow.

Fig. 9. Comparing the axial response for a glass slide at two different depths. Data above the slide (nearer to the reference surface) were set to zero to maximize visual contrast.