Coupling and cross-talk effects in 12–15 µm diameter single-mode fiber arrays for simultaneous transmission and photon collection from scattering media

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We examine signal degradation effects in fiber arrays from fiber-to-fiber coupling and from cross talk attributable to backscatter from the sample medium originating from adjacent fibers in the array. An analysis of coupling and cross talk for single-mode fibers (SMFs) operating at 1310 nm with different core diameters, interaction lengths, core center spacing, and numerical apertures (NAs) is evaluated. The coupling was evaluated using beam propagation algorithms and cross talk was analyzed by using Monte Carlo methods. Several multimode fiber types that are currently used in fiber image guides were also evaluated for comparative purposes. The analysis shows that an optimum NA and core diameter can be found for a specific fiber center separation that maximizes the directly backscattered signal relative to the cross talk. The coupling between fibers can be kept less than −35 dB for interaction lengths less than 5 mm. The calculations were compared to an experimentally fabricated SMF array with 15 µm center spacing and showed good agreement. The experimental fiber array without a lens was also used in a coherent detection configuration to measure the position of a mirror. Accurate depth ranging up to a distance of 250 µm from the tip of the fiber was achieved, which was five times the Rayleigh range of the beam emitted from the fiber. © 2007 Optical Society of America

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1. Introduction

Optical fibers have proven to be very useful for minimally invasive optical diagnostic applications in medicine such as fluorescence,1 Raman scattering,2 two-photon microscopy,3 laser Doppler flowmetry,4 and optical coherence tomography.5–7 In many cases fiber arrays can increase the acquisition rate of collected information and make the technique more suitable for clinical applications.

When used for clinical applications it is desirable to be able to bend and twist the fiber or fiber array to allow it to be positioned in various parts of the body. In some applications, such as coherent imaging, higher contrast signals are obtained by using single-mode fibers (SMFs) that do not transfer power between different fiber modes. SMF arrays are not readily available at longer wavelengths that have a greater penetration depth in highly scattering tissue samples. Therefore it may be necessary to custom fabricate arrays of SMFs that operate at near-infrared wavelengths (1300–1500 nm).

Many fiber and fiber array systems of interest in medical applications require transmitting light to the sample and then collecting it. A variety of transmission and collection geometries for these configurations have been documented1,2,8,9; however, the fibers used in these studies typically had large diameters (200–600 µm) with separate transmit and receive fibers. In imaging applications it is often useful to have fiber arrays with much smaller center spacing. These requirements have been met in part in recent experiments by mechanically scanning a single beam across the end face of a multimode fiber image guide with 12 µm fiber center spacing.10,11 In another experiment a two-photon laser scanning fluorescence microscope was successfully demonstrated by using a fiber image guide that was single mode at 440 nm with 2.4 µm diameter core fibers separated by 4.0 µm.3 However, to our knowledge, single-mode propagation in flexible high-density fiber arrays at near-infrared wavelengths has not been previously demonstrated.

This paper examines the performance of fiber...
arrays consisting of small-diameter (<15 μm) SMFs operating in the 1300 nm wavelength range. SMFs can be bent in small radii (~1 cm) and twisted without transferring power between modes as occurs in multimode fibers. This makes them desirable for flexible endoscope applications that require a single-mode beam, such as coherent detection processes. Operation at 1300 nm allows greater optical beam flexibility and makes them desirable for transferring power between modes as occurs in multimode fibers. This technique requires the refractive index distribution \( n(x, y, z) \) of the fiber and the input wave field \( u(x, y, z = 0) \) that is launched into the fiber. The refractive index distribution for the fiber array includes the core index, the cladding index, the substrate index, and the index of the cement that secures the fibers to the substrate. The input field used is a Gaussian profile that matches the mode field diameter (MFD) of the lowest-order stable mode in the fiber. The MFD (\( 2\omega \)) is determined by the fiber core diameter, wavelength, and numerical aperture (NA) and can be computed using the relation

\[
2\omega = d_{co}[0.65 + 1.619V^{-3/2} + 2.87V^{-6}],
\]

where

- \( d_{co} \) is the fiber center separation.
- \( V \) is the relative power coupling efficiency between waveguide modes.

The photon propagation from and the collection by the fiber array is analyzed without a lens. The lens was not included so that the degrading effects of inaccurate coupling, mode matching, and lens aberrations would not enter into the results and therefore allows us to establish an upper bound on the performance of the fiber array. Without using a lens the depth of focus is equivalent to the Rayleigh range of the Gaussian beam emitted from the fiber [Fig. 3(a)]. If a lens is used and configured in a 1:1 imaging system the beam will be focused over a distance that is twice the Rayleigh range of the fiber alone [Fig. 3(b)]. There are advantages in 1:1 imaging systems in reducing aberrations over systems with other magnification values, and in this system the center spacing between the fibers in the array will set the spatial resolution of the system. A fiber array without a lens may be appropriate for applications where the plane of interest is near the tissue surface.

Field coupling between closely spaced fibers is well understood and can be accurately modeled by using established beam propagation methods.\(^{12-14}\) This technique requires the refractive index distribution \( n(x, y, z) \) of the fiber and the input wave field \( u(x, y, z = 0) \) that is launched into the fiber. The refractive index distribution for the fiber array includes the core index, the cladding index, the substrate index, and the index of the cement that secures the fibers to the substrate. The input field used is a Gaussian profile that matches the mode field diameter (MFD) of the lowest-order stable mode in the fiber. The MFD (\( 2\omega \)) is determined by the fiber core diameter, wavelength, and numerical aperture (NA) and can be computed using the relation

\[
2\omega = d_{co}[0.65 + 1.619V^{-3/2} + 2.87V^{-6}],
\]
with \( V = (\pi d_{\text{oc}}/\lambda) \text{NA} \) representing the \( V \) number or normalized frequency of the fiber.

Our beam propagation analysis employed a commercial simulator that uses bidirectional field propagation techniques to account for reflection and resonant effects that might arise in the structure.\(^{15}\) The power in different fibers was computed as a function of the interaction length \( (L) \) to determine the compromise between packaging and coupling constraints. It is desired to have the length of the reduced fiber section long enough to securely mount on a probe tip package but short enough to limit beam coupling between fibers. The remaining length of fiber will have a larger separation distance in the endoscope and not suffer from appreciable coupling effects. For our analysis the coupling between a fiber of interest and two fibers on either side are considered with each fiber separated by a distance \( d_s \).

Light transmitted into the scattering sample from a fiber is treated as photon propagation with a MC model\(^ {16} \) to determine the percentage of light remitted as a function of location and propagation angle, thus determining the recoupling of scattered light into the launch and adjacent fibers. Light emitted from a fiber is assumed to have a Gaussian spatial profile and extends over an angular range of \( \alpha_{\text{max}} \) corresponding to the NA of the fiber. The NA also limits the collection angle of the backscattered photons. Since the results from the model will later be compared to the experimental data by using an Intralipid-10% solution as a tissue phantom, the parameters for this solution were used for the MC analysis. The Intralipid-10% solution has a refractive index of 1.34, an anisotropy coefficient of \( g = 0.4 \), a scattering coefficient of \( \mu_s = 100 \text{ cm}^{-1} \), and an absorption coefficient of \( \mu_a = 20 \text{ cm}^{-1} \) at a wavelength of 1310 nm.\(^ {17,18} \)

In the MC analysis each photon is initially assigned a weight of unity. When the photon is launched, there is a refractive index mismatch at the interface between the fiber and the solution. The reflection loss at the interface reduces the photon weight by a factor of \( 1 - r_{sp} \), where \( r_{sp} \) is equal to the Fresnel reflection coefficient \( n_s/n_n \).\(^ {11,12} \) The MC simulation uses \( 10^7 \) launch photons to compute the statistical results.

Since the receiving area of a fiber core is fairly small and the scattering characteristics are radially symmetric, the collection area can be extended by using the area of a ring with the ring width equal to the fiber core diameter and a radius \( (r) \) equal to the separation between the launch fiber and the receiving fiber of interest. This increases the number of collected backscattered photons to improve the accuracy of the analysis with a manageable computational time.\(^ {19} \) After counting the photons collected within the ring and applying the appropriate weight, the value is scaled by a factor of \( A_{\text{core}}/A_{\text{ring}} \) to determine the percentage of photons coupled back into a fiber. The number of backscattered photons received by a fiber is in general calculated as a function of \( r \) and \( \alpha \). \( \alpha \) is the trajectory angle of a photon, relative to the fiber normal, backscattered into a fiber with the value of \( \alpha < \alpha_{\text{max}} \) with \( \alpha_{\text{max}} \) determined by the NA of the fiber as mentioned earlier. Summing the weights of the photons received by a fiber and dividing by the total number of launched photons \( (10^7) \) gives the fractional energy and power received by a fiber.

### 3. Simulation Results

The parameters for five different types of fibers that were used in the simulation are listed in Table 1. The multimode fibers MMF A and MMF B have much
higher NAs than are normally found in SMFs (0.55 for MMF A and 0.25 for MMF B). In addition, the core diameters and NAs for these fibers are typical of those found in commercially available fiber image guides (available from Schott North America, Inc.). SMF A is Corning SMF 28 fiber with a \( V \) number of 2.36 and is used to form the experimental fiber array. SMF B is a commercially available single-mode fiber (Stocker Yale BIF-RC-1310-L2) with a \( V \) number of 2.30 that has a larger NA and a smaller core diameter than SMF A. Fiber C is a fiber with a larger NA but the same core diameter as SMF A to evaluate the effect of larger NA on backscatter performance. This fiber has a \( V \) number that is just beyond the cutoff for single-mode operation.

All calculations were performed at a wavelength of 1310 nm, and the refractive index was assumed to have a step profile. It was also assumed that an optical cement with a refractive index of 1.435 surrounds the circular fiber cladding sections. This cement index is less than the cladding index of all the fibers analyzed and corresponds to a commercially available cement (Angstrom Bond OG 134). The results for the beam propagation analysis to determine the coupling to adjacent fibers and throughput as a function of interaction length \( L \) are shown in Fig. 4. The coupling power was computed assuming that the fiber of interest was surrounded by two fibers on either side in a linear array. The coupling was calculated as a function of the interaction length \( L \) for different types of fiber to determine the maximum permissible lengths of reduced diameter fiber sections that can be used in the probe tip of an array package. The results indicate that less than \(-35\) dB of coupling to adjacent fibers results if the interaction length \( L \) was kept at \(<5\) mm. The corresponding fiber throughput also remains high (\( >94\% \)) when \( L \) of the different etched fibers is \(<5\) mm. Also of note in Fig. 4 is a dip in the power coupled to the adjacent fibers. The dip for SMF A occurs near 2.7 mm and for SMF B near 3.9 mm. This dip results from power oscillating between the launch and adjacent fibers.

Beam coupling between fibers is also computed as a function of fiber center spacing with fixed interaction lengths of 2 and 5 mm for the two types of SMF. The results are shown in Fig. 5. For this analysis when the separation is increased the cladding thickness is also increased so that the fibers remain in contact. If \(-35\) dB is used as a value for the maximum acceptable coupling and a 5 mm reduced fiber length is needed for packaging, then a 12 \( \mu \)m separation is the minimum separation that can be used for SMF 28 fiber, while a separation of \(<11\) \( \mu \)m is acceptable for SMF B. Smaller fiber separation dis-

**Table 1. Fiber Parameters Used in the Array and Fraction of Collected Backscatter \( P_o \)**

<table>
<thead>
<tr>
<th>Fiber Type</th>
<th>NA</th>
<th>( d_c ) (( \mu )m)</th>
<th>( n_{co} )</th>
<th>( d_s ) (( \mu )m)</th>
<th>( P_o )</th>
</tr>
</thead>
<tbody>
<tr>
<td>SMF A</td>
<td>0.12</td>
<td>8.2</td>
<td>1.4517</td>
<td>14</td>
<td>0.00014393</td>
</tr>
<tr>
<td>SMF B</td>
<td>0.16</td>
<td>6.0</td>
<td>1.4578</td>
<td>14</td>
<td>0.0001744</td>
</tr>
<tr>
<td>MMF A</td>
<td>0.55</td>
<td>9.0</td>
<td>1.5800</td>
<td>12</td>
<td>0.0013212</td>
</tr>
<tr>
<td>MMF B</td>
<td>0.25</td>
<td>8.0</td>
<td>1.4680</td>
<td>12</td>
<td>0.0004165</td>
</tr>
<tr>
<td>Fiber C</td>
<td>0.16</td>
<td>8.2</td>
<td>1.4517</td>
<td>14</td>
<td>0.00021408</td>
</tr>
</tbody>
</table>

![Fig. 3. Rayleigh range from a fiber (a) with \( 2L_o \) and (b) without a lens \( L_o \). \( 2\omega \) is the mode field diameter of the fiber.](image)

![Fig. 4. Coupling between fibers as a function of interaction length \( L \). MMF A, 9.0 \( \mu \)m core diameter, 0.55 NA, 12 \( \mu \)m fiber separation. SMF A, 8.2 \( \mu \)m core diameter, 0.12 NA, 15 \( \mu \)m fiber separation. SMF B, 6.0 \( \mu \)m core diameter, 0.16 NA, 15 \( \mu \)m fiber separation. MMF B, 8.0 \( \mu \)m core diameter, 0.25 NA, 12 \( \mu \)m fiber separation.](image)
tances are possible if the required interaction length is 2 mm. The oscillation of power between the launch and the adjacent fibers is very noticeable in this case. The oscillation tends to decrease as the fiber separation increases as expected owing to less interaction of the mode field with adjacent fibers.

The MC simulation of backscattered light collected by different types of fiber as a function of separation from a launch fiber is shown in Fig. 6. The simulated cross talk from a fiber of interest to adjacent fibers is computed using the following relation:

$$P_{XT-S}(dB) = 10 \log \left( \frac{P_m}{P_0} \right), \quad m = 1, 2, \ldots, 10, \quad (2)$$

where $P_m$ is the fraction of the incident light that is backscattered and collected by an adjacent fiber, and $P_0$ is the fraction of incident power that is backscattered and collected by the fiber of interest (the signal).

The specific values for $P_0$ for different types of fiber are given in Table 1. The ratio of backscattered signal power $P_0$ to backscattered cross-talk power $P_{XT}$ is plotted for each type of fiber in Fig. 7. The results indicate that a trade-off exists between NA and core diameter for optimizing $P_0/P_{XT}$. Increasing the NA will in general increase both $P_0$ and $P_{XT}$, but decreasing the fiber core diameter reduces $P_{XT}$ and increases the $P_0/P_{XT}$ ratio. This suggests that fibers with larger NAs and small fiber cores (SMF B) will provide good scattered signal detection in array configurations. This finding indicates that single-mode fibers may be suitable since the core diameter can be reduced to keep the V number at $\leq 2.405$ for single-mode operation and still have good signal collection efficiency.

### 4. Experimental Fiber Array Measurements

#### A. Fiber Cross-Talk Measurements

An experimental fiber array was fabricated as a test system for comparing measured data to the calculated results for cross talk. The fiber array consists of SMF 28 fibers that have been etched to a core diameter of $\sim 14 \mu m$ diameters using a standard buffered oxide etch (BOE) solution and mounted in a silicon V-groove array with $15 \mu m$ groove spacing. The diameter of the fiber is reduced to $14 \mu m$ for a length of $\sim 2 mm$ and gradually increases to the standard cladding diameter of $125 \mu m$. The taper is produced by gradually withdrawing the fiber from the etch solution. The reduced fiber sections are cemented in place with an epoxy (Angstrom Bond OG134) with a refractive index of 1.435 that is lower than the core index of the fibers (1.4517) and helps confine the mode profile. In addition, this cement can be polished to provide a good optical interface with a tissue sample. The refractive index of the cement was chosen by performing a beam propagation analysis of an array with the parameters for the SMF 28 fiber surrounded by a material with a range of refractive index values. The OG 134 cement refractive index matches one of the optimum values near 1.435 and was therefore used for the experimental array.

The experimental system for measuring light scattered from the tissue phantom solution and coupled...
back into different fibers in the array is illustrated in Fig. 8. The light from a 1310 nm superluminescent light-emitting diode (SLD) is coupled to an input fiber of a 3 dB fiber coupler, and one output fiber from the coupler is connected to a fiber of interest in the array. The other output fiber from the coupler is connected to detector 3 to monitor the input power level of the source. The second input fiber to the coupler is used to measure the backscattered signal returning to the fiber of interest. The backscattered light that is coupled to the adjacent fibers in the array is measured by successively coupling detector 2 to each of the adjacent fibers. Backscattered signals were measured from an undiluted Intralipid-10% solution and normalized to measurements from a 0.9% saline reference solution to remove contributions from interface reflections and background noise. It was necessary to use a concentrated Intralipid-10% solution to provide measurable cross-talk signals. While Intralipid-10% has a higher reduced scattering coefficient than most tissues, it is useful in obtaining robust signals for comparison with simulation results.\(^22\)

Figure 9 shows the experimental backscattered cross-talk values obtained with the fiber array as a function of radial distance from the center of the launch fiber. Data samples were obtained to a radial distance equivalent to ten fiber diameters (150 \(\mu\)m). The cross-talk values \(P_{\text{EXT}}\) of the scattering experiment are determined by using the relation

\[
P_{\text{EXT}}(dB) = 10 \log \left( \frac{(B_i - B_S)}{(A_i - A_S)} \left( \frac{T_{\text{adjacent}}}{T_{\text{launch}}} \right) \right),
\]

where \(B_S\) and \(B_i\) are the optical power values measured by detector 2 with the 0.9% saline and Intralipid-10% solutions. \(A_S\) and \(A_i\) are the optical power values measured by detector 1 with the 0.9% saline solution and Intralipid-10% solution. The parameter \(T_{\text{adjacent}}\) represents the transmission efficiency \(I_{\text{out}}/I_{\text{in}}\) of the adjacent fiber in the fiber array, \(T_{\text{launch}}\) represents the transmission efficiency of the launch fiber in the fiber array, and \(R_{3\,dB}\) represents the splitting ratio of the coupler for the returned signal through the fiber connected to detector 1. These transmission and splitting ratio values were measured prior to the experiment.

B. Coherent Depth Measurement with Reduced Fiber Diameter Array

The reduced diameter SMF array was used in a coherent distance measurement system. The experimental system is illustrated in Fig. 10 and consists of a fiber interferometer attached to one of the fibers in the array. The source has a center emission wavelength of 1310 nm and a spectral bandwidth of 40 nm. The free-space coherence length is

\[
l_c = 0.44 \frac{\lambda_0^2}{\Delta \lambda} = 19 \, \mu\text{m}.
\]

The SMF 28 fiber has a \(V\) number of 2.36 and a mode field diameter \([2\omega\text{ from Eq. (1)}]\) of 9.129 \(\mu\)m. The resulting Rayleigh range is

\[
z_R = \frac{\pi (2\omega)^2}{4\lambda} = 50 \, \mu\text{m}.
\]
The system uses a mirror (M2) on a galvo system to scan the coherence region over a distance of 1.18 mm at a rate of 0.5 Hz. The output from the signal fiber is reflected from a mirror (M1) that is mounted on a micrometer stage for accurate positioning relative to the tip of the fiber. Figure 11 shows the interference fringes formed within the coherence length of the source with the mirror M1 located 50, 150, and 250 μm from the edge of the fiber array.

Fig. 11. Interference fringes formed within the coherence length of the source with the mirror M1 located 50, 150, and 250 μm from the edge of the fiber array.
patterns acquired using a LabView control program at distances of 50, 150, and 250 μm from the end of the fiber. Although the amplitude of the fringe pattern decreases with distance, the signal is sufficient at 250 μm to measure the presence of the surface. This experiment indicates that reduced fiber diameters that are packaged in an array can be used to measure surfaces without a focusing lens that are considerably farther from the fiber tip than the Rayleigh range. Although more work is necessary, this concept can greatly simplify the design of endoscopes for limited subsurface measurement applications.

5. Conclusions
We have investigated the characteristics of fiber arrays formed with different types of reduced diameter optical fibers. It is assumed that each fiber in the array transmits and collects an optical signal. The motivation was to determine values for fiber NA and core diameter that allow minimum spacing between single-mode fibers operating at 1310 nm. This wavelength is of particular interest for studies of biological tissue, and single-mode operation provides the potential for use in high-contrast coherent detection processes in an endoscope. Fiber parameters for commercially available single-mode, multimode, and fiber image guide fibers were evaluated for fiber-to-fiber coupling, return backscatter signal strength, and backscatter cross talk returning from adjacent fibers in the array. The reduced diameter fiber-to-fiber coupling for all fiber types as a function of the interaction length shows that coupling remains less than −35 dB provided the interaction length is less than 5 mm. Higher-NA SMF and MMF fibers have less coupling than lower NA fibers with comparable core diameters and fiber spacing. Coupling as a function of fiber spacing for SMF fibers and 5 mm interaction length shows that separation between fibers of 12 μm can be achieved with −35 dB of coupling. For single-mode fibers the coupling oscillates because of the power transfer between fibers.

A MC analysis of the backscattered signal cross talk for different types of SMF and MMF fibers was conducted. The results indicated that a trade-off exists between NA and core diameter for optimizing $P_0/P_{XT}$. Increasing the NA will in general increase both the signal and cross-talk backscatter levels, but decreasing the fiber core diameter reduces $P_{XT}$ and improves the $P_0/P_{XT}$ ratio. This suggests that fibers with large NA and small fiber cores will provide good scattered signal detection in array configurations.

An experimental fiber array was fabricated from etched Corning SMF 28 single-mode fibers and packaged in an array with 15 μm fiber spacing and bonded with an optical cement that minimized coupling. The experimental array was used to evaluate the levels of backscattered signal and cross talk returning from an Intralipid tissue phantom solution. The backscattered measurements are in good agreement with the MC simulation results and indicate that this model is suitable for analyzing reduced diameter fiber arrays.

The experimental array with reduced diameter fibers was also used in a coherent distance measuring demonstration system without a lens. High-contrast interference patterns were obtained with light reflected from a mirror positioned out to 250 μm from the tip of the fiber. This was five times the Rayleigh range corresponding to the beam emitted from the fiber.

Our custom-fabricated, etched-tip linear fiber array provides an acceptable solution to the problem initially posed of realizing a single-mode fiber array at 1310 nm with closely spaced fiber (~15 μm) for use in an endoscope. The array, with or without focusing optics, could be used in coherent, proximal scanning or parallel channel acquisition applications as evidenced from the high-contrast interference patterns obtained in Fig. 11. Further studies are warranted with fibers having more complex fiber index profiles, different lens systems, and different data acquisition techniques to fully take advantage of these arrays for medical detection and imaging applications.

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