Optical modeling of an ultrathin scanning fiber endoscope, a preliminary study of confocal versus non-confocal detection

Erek S. Barhoum
Flight Deck R & D, Boeing Commercial Airplanes, Everett WA 98203
erek.s.barhoum@boeing.com

Richard S. Johnston and Eric J. Seibel
Department of Mechanical Engineering and Human Interface Technology Lab, University of Washington, Seattle WA 98195
johnston@hitl.washington.edu, eseibel@hitl.washington.edu

Abstract: An optical model of an ultrathin scanning fiber endoscope was constructed using a non-sequential ray tracing program and used to study the relationship between fiber deflection and collection efficiency from tissue. The problem of low collection efficiency of confocal detection through the scanned single-mode optical fiber was compared to non-confocal cladding detection. Collection efficiency is 40x greater in the non-confocal versus the confocal geometry due to the majority of rays incident on the core being outside the numerical aperture. Across scan angles of 0 to 30°, collection efficiency decreases from 14.4% to 6.3% for the non-confocal design compared to 0.34% to 0.10% for the confocal design. Non-confocality provides higher and more uniform collection efficiencies at larger scan angles while sacrificing the confocal spatial filter.

©2005 Optical Society of America

OCIS codes: (170.2150) Medical optics and biotechnology; (170.2150) Endoscopic imaging.

References and Links

1. Introduction

Minimally invasive medical procedures will benefit from flexible endoscopes that are extremely thin yet produce wide field of view and high resolution images. Current flexible endoscopes use fiber bundles or silicon image sensors placed in the distal tip to capture imagery while illumination is delivered via a separate fiber bundle. Using these technologies each pixel in the image is derived from a physical element in the distal tip, a fiber in the bundle or a sensor in the sensor-array (video camera). Typical image sensing elements require a pixel sensor of at least 3 to 5 microns in diameter, the outer diameter of the imaging fiber within the smallest flexible endoscopes being coherent fiber bundles. Therefore, total number of imaging pixels and distal tip diameter are directly related, increasing both the image field of view and resolution requires increasing the distal tip diameter.

Using the same optical fibers for both illumination and image collection allows further reduction in overall diameter of the standard flexible endoscope by eliminating the separate illumination fibers. The dual focus from and back to the same optical fiber is referred to as confocal endoscopy. Recent development of confocal flexible endoscopes rely on illumination and collection through the same small-core optical fibers and an optical scanner at either the proximal or distal end. Using a thin and flexible coherent fiberoptic bundle, the proximal end can be scanned using large mirrors from outside the body while in vivo images are acquired from the small distal tip [1, 2]. The recent application of micro-optical-electromechanical systems (MOEMS) has produced small mirror systems with a (8 mm long by 3.3 mm in diameter) rigid tip of a flexible endoscope [3]. Instead of a mirror scan system, the point source from a single optical fiber can be scanned using a small metal cantilever driven in vibratory mechanical resonance [4]. A more compact two-dimensional (2D) scanner is driving the optical fiber itself as a quartz mechanical resonator or micro-optical scanner.

At the University of Washington, a flexible scanning fiber endoscope (SFE) is being developed to provide high resolution images in which the resolution is less dependent on the distal tip diameter [5]. The SFE uses one singlemode optical fiber that is held at the distal end by a piezoelectric 2D actuator. A 2D resonant spiral scanning pattern is generated by driving the cantilevered optical fiber at its single mechanical resonance in an amplitude-modulated circular pattern or spiral pattern. Images can be generated either by a confocal geometry or simply by collecting backscattered light from sensors placed laterally to the fiber scanner [6]. By detecting scanned laser illumination in a non-confocal geometry, increased image quality and collection efficiency have been reported over the more standard confocal geometry [7]. Initial prototype SFEs have been made in the non-confocal geometry and their performance has been demonstrated to have 500 lines of resolution across a 1.5 mm diameter field with better than 10 micron resolution [8].

In this preliminary study, the expected performance of using SFEs to image tissue is modeled and the confocal arrangement is compared to non-confocal geometries to provide guidance to future laser-scanning endoscope designs. In an ultrathin endoscope, the confocal geometry has the advantage of simplicity by using the same optical fiber for illumination and collection, although the spatial light filtering of the confocal geometry significantly lowers the light collection efficiency. The objective of modeling the SFE geometries with adequate fidelity and accuracy is to analyze the tradeoffs in optical collection efficiencies between the two geometries using reflectance properties of human tissue. Scattering off the tissue sample
was modeled using results from prior studies of reflectance properties of human tissue [9] [10]. This data is essentially equivalent to Bi-Directional Reflectance Distribution Function (BRDF) data [11]. Based on this data, a function was built which provides the appropriate reflectance value as a function of the scatter direction, including both specular and diffuse behavior, as exhibited by human tissue.

1.1 Scanning Fiber Endoscope

1.1.1 System Description

The heart of the SFE is the fiber microscanner as sketched in Fig. 1 with outer diameter of 1.0 mm. In this device the piezo tube is driven by two amplitude modulated sine waves with 90 degrees phase difference at the first mode resonant frequency of the scanning fiber. The distal tip of the scanning fiber follows an expanding spiral pattern.

![Piezo tube](image1.png)

![Fiber attachment](image2.png)

![Scanner housing](image3.png)

![Piezo to housing](image4.png)

![Patterned piezo](image5.png)

![Scanning piezo](image6.png)

The Gaussian beam exiting the single mode scanning fiber is focused to the image plane using a device-specific optical system. Various applications may require different fields-of-view, working distances, depth-of-foci, or target resolutions. By changing the optical system, identical fiber microscanners can be used to create endoscopes for a variety of applications.

One monochrome red (635 nm Thor Labs laser diode source) SFE has been configured to image a 1.5 mm diameter area at a 2 mm working distance (40 degree field-of-view) [8]. Using a 5 KHz resonant frequency scanning fiber (4.3 cantilever length of Nufern 460HP singlemode optical fiber), images are created with 250 expanding spirals (500 pixel diameter circular images) at a frame rate of 16 Hertz. Fig. 2, is an image of a 1951 mylar USAF test target which shows resolution of the 9.8 micron bar and space pattern (group 5, element 5) demonstrating Gaussian performance of the optical system. In this prototype SFE the detector used was a separate silicon PIN photodiode placed adjacent to the distal tip of the endoscope in a non-confocal geometry (Hamamatsu) [5]. The lens system was two plano-convex lenses edged down in diameter (Edmund Optics 43-394 and 43-397). Future prototypes will integrate the optical detection system within the endoscope such as custom photodiodes in an annular ring or multimode optical fibers to collect light at the distal tip either outside or inside the SFE microscanner housing with sensors at the proximal end. In combination, a dual waveguiding optical fiber such as a dual clad singlemode-core optical fiber can be used to compare directly the collection efficiencies of the ultrathin SFE designs. Location and geometry of the light collection system will be determined through simulations (such as the one described herein) to be optimal for the specific application.
1.1.2 Software Description

A non-sequential ray tracing software package was used to model the scanning laser illumination and tissue backscatter (ASAP 8.0.3 Breault Research Organization, Tucson, AZ). ASAP is open-ended, which allows the user to introduce scatter models in a variety of ways. The two most prominent methods involve fitting a mathematical function to:

- Raw BRDF data
- Known reflectance behavior

ASAP allows coherent optical sources to be modeled and has the speed and accuracy needed for performing the required analysis. Presented in this paper are bitmap printouts from the ASAP program which is highly interactive, allowing the user to click anywhere on the image to show the value of the energy density at that location.

2. Development of the Model

2.1 Spatial model and dynamics

The model included geometry of the entire distal tip of the endoscope and a sample placed a distance of 2 mm from the optical system. A modified Henyey-Greenstein (H-G) function was assigned to the sample, as described below. The model considered rotations of the optical fiber in the range of $\theta = 0$ to 3 degrees (0-0.225 mm lateral displacement at the free end), centered about the point where it is held by the piezo. As the fiber is rotated, the rayset is also rotated. The source was modeled as a 635 nm coherent Gaussian Beam. The numerical aperture (NA) of the fiber is 0.13. The first waist of the beam is assumed to be at the tip of the fiber and its radius is 0.00175 mm, which is obtained by matching the beam width to the diameter of the fiber core. For scalability, the total power assigned to the source was normalized and set equal to 1 (one) Watt. The lens was modeled as an ideal lens with a focal length of 0.4 mm. This was determined using the Thin Lens Formula ($1/f = 1/u + 1/v$; where $f$ is the focal length, $u$ is the image distance and $v$ is the object distance) using an image distance of 2 mm.

2.2 Scatter model

A modified H-G function was assigned to the sample, allowing the direction and intensity of scattered rays to be determined. All scattered rays in ASAP are incoherent. Since ASAP apportions flux differently to incoherent and coherent rays, attempts to scatter coherent beams will result in energy conservation violations. ASAP implements Gaussian Beam Decomposition (GBD) by creating ‘coherent beams’ from a base ray and several parabasal
rays. The parabasal rays track the beam width and divergence. As with all Gaussian beams, the smaller the beam, the greater its divergence. Beams on the order of a few wavelengths or smaller violate the paraxial approximation implicit in the GBD method. As a result, roughness inherent to a surface cannot be sampled using GBD.

The Henyey-Greenstein function is a commonly-used formulation that depicts reflective behavior of light incident on a given surface, see equation (1). A Henyey-Greenstein function with a “g-factor” of 0.8 [10] is appropriate for modeling the reflective behavior of organic (human) tissue. The detection angle, $\phi$, is defined with respect to the surface normal. The Henyey-Greenstein function (H-G) is defined as follows:

$$H - G = \frac{1}{4\pi} \cdot \frac{1 - g^2}{(1 - g^2 - 2g \cos \phi)^{3/2}}$$

where $g$ is an anisotropy factor and $\phi$ is the detection angle.

In employing this function to accommodate both diffuse and specular behaviors, some factors and careful choices of angle arguments must be determined. The diffuse component was defined about the surface normal, resulting in a Lambertian element centered at $\phi = 0$. At the specular angle, $\phi_0$, the data indicates behavior that is much more directional in nature. To capture this specular dependence, we introduce the modified Henyey-Greenstein function with the cosine argument $\phi - \phi_0$, see equation (2). The $\phi - \phi_0$ term is known as the scatter angle and represents simply the angular departure of the flux from the specular angle. A normalization factor, $a$, was added as well as a proportioning factor, $\beta$, which determines the percent of energy scattered diffusely. This resulted in the H-G' function of (eqn. 2), and Fig. 3 shows a plot of this H-G’ function with a 25-degree specular angle, $\phi_0$:

$$H - G' = a \cdot \beta \cdot \cos \phi \cdot \left( \frac{1}{a} \right) + (1 - \beta) \cdot \frac{1 - g^2}{(1 - g^2 - 2g \cos(\phi - \phi_0))^{3/2}}$$

where $a$ is a normalization factor, $\phi$ is the detection angle, $\phi_0$ is the specular angle, and $\beta$ is the proportioning factor.
3. Simulations and Results

Rays were traced and statistics were generated to determine where the backscatter flux was located internally and able to be captured by the endoscope, as well as the general direction and intensity of rays external to the housing. The number of rays scattering off the sample in each raytrace was 33,808,716. The pictures displaying the results are shown in windows divided into pixels, as set by the user. It is important to trace enough rays as to “fill” each pixel. As a general rule, the number of rays used in each simulation was chosen so as to maintain a noise-to-signal ratio between 10% and 20%. Specifically, the number of pixels should be set equal to the square root of the number of rays on the desired object multiplied by the desired noise-to-signal ratio. For example, if the detector has 100,000 rays, for a 10% noise-to-signal ratio the number of pixels would be: \((100,000 \times 0.5) \times 0.1 = 31.62 \approx 32\) pixels. The noise: signal ratio essentially tracks the number of rays in each pixel used in the energy density calculation. If this number becomes large, the accuracy of the calculation diminishes.

3.1 Ray paths inside and outside of the SFE housing

Figure 4 shows a schematic of the system, and indicates the general relationships between fiber rotation, \(\theta\), and:

- fiber deflection
- deflection of the spot on the sample
- specular angle of the rays scattering off of the sample.

Fig. 3. A plot of the scatter model function in ASAP 8.0.3, based on the above H-G' function

(2) with \(\theta_0 = 25\) degrees, \(a = 17.25^\circ\pi\), and \(\beta = 0.58[9]\).
The deflection of the fiber is directly proportional to the angle of rotation; specifically $\delta = 4.3 \tan \theta$, where the cantilever length is 4.3 mm. However, in reality, the fiber does not remain rigid as it scans, but this simplification enables the use of trigonometry to determine the location of the fiber tip. Analytically, this does not affect the calculation, since ultimately only the position and direction of the emission affects the raytracing. The specular angle, $\phi_0$ is approximately $10^\circ \theta$ from an optical scan of ± 30 degrees at the tissue, as shown in Fig. 4. The illumination spot moves in the vertical and/or horizontal directions as: $\delta = 17.2^\circ \theta$ mm, with theta in radians.

Fig. 4. (above). A schematic showing the overall system and the relationships between various parameters. The fiber tip displacement to inner edge of housing is exaggerated for clearly illustrating scan angle. Rays shown are representations only, and do not represent energy traveling along actual ray paths. (below). The optical fiber end face is shown with illustrative cones of acceptance angles (NA) denoting light collection in confocal and non-confocal. The vertex of the two cones is located at the effective point source of the single-mode optical fiber, slightly within the endface surface along the optical axis.
In practice, ray totals are increased until very little change in result is obtained. Fig. 5 shows several plots of rays being traced as the fiber sweeps through a vertical angle. The plots show the fiber at angle $\theta=0,1,2,3$ respectively. The maximum deflection of the fiber is 0.225 mm. Not all rays traced were plotted, and note that at $\theta = 0$, most of the rays scatter directly back into the lens, as expected. This is evidenced by the black region between the center of the sample and lens, see top-left of Fig. 5. This conical region of backscatter appears darker than in the other plots in the figure due to the higher density of rays in that region. The pictures at the right of each plot in Fig. 5 show the movement of the focused spot on the sample for each case of $\theta$.

![Fig. 5](image1)

![Fig. 5](image2)

![Fig. 5](image3)

![Fig. 5](image4)

Fig. 5. The plots above show a profile of the system tracing rays at theta values of 0, 1, 2 and 3, and the corresponding location of the energy distribution on the sample. The scanning optical fiber is shown in Green. The vertical arrow indicates the location of the (ideal) lens.

Notice that at angles of theta as low as 1 degree, the specular emission misses the lens and endoscope tip completely, see arrow in second SFE graphic from the top of Fig. 5. Avoiding this specular reflection is desirable since reflection from a laser source can result in bright spots on the image. Multiple reflections were not included in this model.
Figure 6 shows the Gaussian beam incident on the tissue sample before it is reflected. For a fiber rotation of 2° (and corresponding fiber deflection of 0.15mm), the centroid of the spot is located 0.6 mm from the SFE optical axis or 17.2*θ, which agrees with the geometric relationship in Fig. 4. These plots are useful in determining the illumination spot size at the sample surface, directly relating to the image resolution in the non-confocal geometry. In Fig. 6 and in all the remaining figures, the direction of rotation used pertains to a rotation about the vertical axis instead of a rotation about the horizontal axis, as used previously. This change in fiber rotation is chosen arbitrarily and simply demonstrates the symmetry of the system.

![Figure 6](image)

*Fig. 6. (a) (Left) shows the spot from a global perspective. (b) (Right) shows the same spot centered in the picture window. The majority of the sampled rays are distributed within a 2 micron spot size.*

Figure 7 shows the intensity of the rays coming off of the sample in angle space. The beam is incident at 20 degrees horizontal. The reason that each direction is different is because the rotation is purely about one axis only (vertical). So, in one direction the reflection is based on the H-G’ function and in the other direction there is no angular function resulting in a pure Lambertian profile. If there was no rotation, the intensity scatter profile would be an axially symmetric diffuse spot. Consider each profile plot taken across the reflective scatter function in Fig. 7. At or near the specular angle, there is a sharply peaked profile in the direction of rotation, and a smooth Lambertian profile in the other direction. The shape of the scatter from the sample is identical to that which is plotted in Fig. 3 at θ=2, verifying that the scatter model is behaving in the desired manner. The specular direction indicated by the profile plot is approximately 20 degrees with a fiber rotation angle of θ=2. This figure validates the earlier assertion that φ₀ = 10*θ, where φ₀ is the specular angle and θ is the angle of fiber rotation.
Fig. 7. Intensity of the rays scattering off of the sample at $\theta=2$ degrees plotted ± 90 degrees on both axes.

Figure 8 shows the positions of scattered rays on an over-sized round detector (6 mm diameter) located at the front of the housing at a fiber rotation of 2°. Note the shape of the distribution, indicating that at $\theta=2$, the majority of the energy misses the housing. In fact, this occurs for values of theta as low as 0.9625°.

Fig. 8. Energy incident on an oversized 6-mm diameter detector at a reflection angle of 20 degrees ($\theta=2$) that slightly overfills the square window. Note the specular component on the left region of the detector, and the “hole” in the detector is where the microscanner housing and lens interface the surface of the much larger detector.
3.2 Confocal versus Non-confocal Detection

The most straightforward analysis of confocality in an ultrathin SFE design is comparing the amount of backscattered light re-entering the scanned illumination fiber that can be collected confocally versus non-confocally. Much of the energy scattered off the sample surface will land incident on the lens along a direction vector pointing primarily in the negative direction of the optical axis. These rays will then be re-focused by the lens back towards their origin, the tip of the optical fiber. It remains to separate the flux incident on the fiber tip as that belonging to one of four categories:

A. Flux incident on the fiber core inside of the fiber NA
B. Flux incident on the fiber core outside of the fiber NA.
C. Flux incident on the fiber cladding.
D. Combined flux from cladding and energy on core outside of the core NA.

Category A is important for creating a confocal system, while categories B and C are combined to provide the energy needed for image detection in a non-confocal system, indicated by category D. The plot in Fig. 9 shows the flux incident on the two elements at the tip of the optical fiber, namely the cladding region and the core region of the fiber tip. As illustrated in Fig. 9, the cladding region (CAT C) becomes a more prominent ray collector at higher rotation angles, while at small rotation angles, more energy falls directly onto the core region (CAT A). Notice that at very small values of theta, there is little energy on the tip of the fiber cladding. In fact, at $\theta=0$, there is no flux at all incident on the fiber cladding, and at $\theta=0.9625$ degrees, there is as little as 0.003 Watts (0.3%) incident on the cladding surface. However, most of the light striking the core is incident greater than the NA of the core. As clearly shown, the small diameter and low NA of the fiber core in single-mode optical fibers limits the collection of light compared to a non-confocal collection from the cladding at the proximal end (CAT D). While the rays initially exit the fiber tip completely within the core NA, they will not necessarily arrive there again within the same NA to be guided within the core to the proximal end.

![Flux vs. Fiber Rotation](image)

Fig. 9. Flux on internal components as a function of rotation angle. Note that the dashed lines combine to give the solid Cyan colored line.
The images listed in table 1 show the distribution of rays falling on the end face of the optical fiber in a non-confocal (column 1) versus confocal geometry (column 2) for the four scan angles of theta (images are plotted in angle space). All images shown in the table depict the rays striking the optical fiber core only. Since the effective point of focus for the majority of the collected rays is slightly behind the end face of the optical fiber (refer to Fig. 4, side view), the spatial distribution of rays roughly matches the angular distribution of rays striking the end face. For the confocal case, the percentage of rays entering the core ranges from 0.34% to 0.10% with theta values from 0 to 3 degrees as listed in Table 1. For the non-confocal case, rays incident on the core that are outside the NA of the fiber core (listed in Table 1), and rays incident on the cladding are summed as indicated in Fig. 9 (CAT D), and the total percentage of rays in the four angles of theta are listed in Table 1 (non-confocal system). The collection efficiency for the non-confocal system drops from 14.4% to 0.41%. Although both optical designs have lower collection efficiency with increasing scan angle, in the confocal case the percentage of rays that are collected by the core is always less than 0.4%. The significant difference in collection efficiency between optical designs is also shown graphically in Fig. 9 (CAT A versus CAT D).
Table 1. Images of energy density on non-confocal and confocal detection systems

<table>
<thead>
<tr>
<th>θ, deg</th>
<th>Non-confocal System</th>
<th>Confocal System</th>
<th>Non-Confocal System</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Core Only (NA&gt;0.13)</td>
<td>Core Only (NA&lt;0.13)</td>
<td>Core + Cladding</td>
</tr>
<tr>
<td>0</td>
<td><img src="image1.png" alt="Image" /></td>
<td><img src="image2.png" alt="Image" /></td>
<td>14.4%</td>
</tr>
<tr>
<td></td>
<td>1.44E-01 Watts (14.4%)</td>
<td>3.36E-03 Watts (0.34%)</td>
<td></td>
</tr>
<tr>
<td>1</td>
<td><img src="image3.png" alt="Image" /></td>
<td><img src="image4.png" alt="Image" /></td>
<td>10.3%</td>
</tr>
<tr>
<td></td>
<td>1.00E-01 Watts (10.0%)</td>
<td>1.92E-03 Watts (0.19%)</td>
<td></td>
</tr>
<tr>
<td>2</td>
<td><img src="image5.png" alt="Image" /></td>
<td><img src="image6.png" alt="Image" /></td>
<td>8.0%</td>
</tr>
<tr>
<td></td>
<td>2.7E-02 Watts (2.7%)</td>
<td>1.39E-03 Watts (0.14%)</td>
<td></td>
</tr>
<tr>
<td>3</td>
<td><img src="image7.png" alt="Image" /></td>
<td><img src="image8.png" alt="Image" /></td>
<td>6.3%</td>
</tr>
<tr>
<td></td>
<td>4.11E-03 Watts (0.41%)</td>
<td>9.56E-04 Watts (0.096%)</td>
<td></td>
</tr>
</tbody>
</table>

Images showing energy density collected on the fiber core tip (end face), from rays incident outside the core NA (column 1), and confocally collected within the core NA (column 2). Note for θ=1 the specular dependence of the incident beam causes a greater degree of anti-symmetry than for θ=0, and also for θ=2 and θ=3, as the specular rays “miss” the lens and hence do not land incident on the core tip. Since the majority of rays are being focused behind the fiber end face, the spatial distribution rays collected in the non-confocal optical system (column 1) form an annulus while the rays collected in the confocal system (column 2) have a more normal distribution. The image...
shown in the non-confocal system column for \( \theta = 3 \) shows a yellow-green ring around the outside of all the rest of the energy; this best corresponds to the spatial periphery of the fiber core. Since all figures use the same dimensional scaling, the 3.5 micron fiber core size scales each figure in Table 1.

4. Discussion

The optical modeling allowed a direct comparison between the confocal and non-confocal geometries of the scanning fiber endoscope under development. Since the smallest illumination spot is desired at the tissue, single mode operation is required. Therefore, the small core diameter and low NA restricts the collection efficiency in the confocal detection to about 50 times less than the non-confocal design. This large difference in collection efficiency is shown graphically in Fig. 9 and numerically in Table 1. Even in the case when there was no deflection angle (\( \theta = 0 \)) and all the collected light was incident on the core of the optical fiber, more than 40 times more light was incident outside the NA of the optical fiber core than inside. In all cases, a perfect lens was assumed which resulted in the light collection for the confocal design to be maximized, especially for the case of no deflection angle which had the highest collection efficiency (0.34%). While the confocal design was limited to the collection of rays within the NA of 0.13, the non-confocal design collected all rays incident on the cladding (NA<0.84) and the core at NA>0.13. Although high NA multimode fibers are typically in the range of 0.3 to 0.6 NA, higher collection NAs are possible for bare optical fibers by making the outer surface of the cladding highly reflective.

Confocal optical designs are the most common type of scanned imaging system under development for new flexible endoscopes. The main reason for the popularity of the confocal design is the rejection of optical noise and the axial spatial discrimination afforded by the spatial filter of the small core optical fiber and the simplicity of the optical design for the distal (in vivo) end. Furthermore, a confocal daughter scope introduced through the working channel or integrated within a standard flexible endoscope adds new functionality, such as imaging deeper within tissue and the future potential for optical biopsy [12]. Typically these confocal scopes are very low in field of view and depth of focus, thus referred to as microscopes, and their use is designed to be in conjunction with and always within the field of view of the mother scope.

As a design tool for the SFE development, the modeling of confocality helps to identify the major advantages for the non-confocal geometry, which are the high collection efficiency across wider fields of view, and much greater depths of focus. Although the collection efficiency dropped from a maximum at zero scan angle to a minimum at the largest scan angle modeled (\( \theta = 3 \)), the rate of reduction is 2x greater from \( \theta = 2 \) to 3 degrees in the confocal design compared to the non-confocal design (see Table 1). In the case of non-confocal imaging, light collection at wider fields of view can be increased by having additional sensors, possibly on the exterior of the SFE housing. In future analysis, these optical models will be used to identify any additional features necessary to acquire higher contrast and high resolution images. In one example, a high-resolution SFE could have the feature of confocal optical detection for the highest resolution imaging in three-dimensions at only the smaller fields of view, while the non-confocal detection is used for the wider fields of view. In another example, a long depth-of-focus SFE can be used as a stand-alone ultrathin flexible endoscope, viewing tissues that were previously inaccessible to standard flexible endoscopy, such as the biliary duct, main pancreatic duct, and the more peripheral lung.

Further enhancements to the SFE optical modeling may take into consideration such effects as polarization. As polarized light interacts with human tissue, some of it is reflected (specularly) by the top layer, while some of the energy is transmitted to sub-surface layers of various bi-refringent properties and scattered laterally, subsequently randomizing the polarization [13]. While the modified H-G' function used in this analysis does an adequate job of capturing illuminated tissue properties in terms of intensity and direction of rays, it does not take into consideration the optical polarization and deeper tissue layers. Future
models may include such things as polarized Gaussian sources, layered tissue, and polarizing filters for measuring the degree of polarization at the detectors in order to perform analyses on how the inclusion of such design elements may enhance the performance of the endoscope device, such as rejecting specular reflection. In confocal geometries, the rejection of specular reflection may be more critical since the tissue surface can vary within the scanned illumination field, and since multiple non-confocal signal collectors cannot be used as a filter to mitigate the specular reflections.

Acknowledgments

We would like to thanks Steven L. Jacques, Professor, Oregon Graduate Institute of Science and Technology, Portland, OR for helpful discussions, and grants from the NIH/NCI (R21 CA094303 and R21 CA110184) for partial support of this research, and Breault Research Organization, Inc., for the 2004-2005 academic license for ASAP 8.0.3.