Scanning single-mode fiber optic catheter–endoscope for optical coherence tomography

G. J. Tearney, S. A. Boppart, and B. E. Bouma

Department of Electrical Engineering and Computer Science, Research Laboratory of Electronics, Massachusetts Institute of Technology, Cambridge, Massachusetts 02139

M. E. Brezinski, N. J. Weissman, and J. F. Southern

Massachusetts General Hospital, Boston, Massachusetts 02114

J. G. Fujimoto

Department of Electrical Engineering and Computer Science, Research Laboratory of Electronics, Massachusetts Institute of Technology, Cambridge, Massachusetts 02139

Received October 30, 1995

We describe a new optical coherence tomography catheter–endoscope for micrometer-scale, cross-sectional imaging in internal organ systems. The catheter–endoscope uses single-mode fiber optics with a novel transverse scanning design. The distal end of the catheter–endoscope uses a gradient-index lens with a microprism to emit and collect a single spatial-mode optical beam with specific focusing characteristics. The beam is scanned in a circumferential pattern and can image transverse cross sections through the structure into which it is inserted. A device with a diameter as small as 1.1 mm has been achieved, and imaging of *in vitro* human venous morphology is demonstrated. © 1996 Optical Society of America

A technique for performing an optical biopsy (high-resolution micrometer-scale cross-sectional optical imaging) of tissue architectural morphology, without the need to excise tissue specimens, would have a powerful effect on the diagnosis and clinical management of many diseases. Optical coherence tomography (OCT) is a new optical imaging technique that uses low-coherence interferometry to perform high-resolution cross-sectional imaging in biological systems.[1] OCT has been extensively applied in ophthalmology to image transparent structures tomographically in the eye.[2,3] Clinical studies show that OCT provides tomographic images of the retina with unprecedented resolution and can be used to diagnose a wide range of retinal macular diseases.[4] The problem of OCT imaging in other human tissues is more challenging because of optical scattering.[5–7] However, *in vitro* studies have demonstrated that OCT can image the architectural morphology in tissues that are highly optically scattering.[8–11] One important and challenging application of OCT is imaging the vascular system for the diagnosis of atherosclerotic lesions that are prone to rupture.[11] Intravascular ultrasound is an existing catheter-based technique for obtaining cross-sectional images of human vasculature.[12] However, optical imaging methods, such as OCT, have the capability to image vascular lesions with much higher resolution. Moreover, additional research has shown that OCT may be clinically useful for performing high-resolution imaging of other organ systems, such as the gastrointestinal tract.[9]

OCT performs high-resolution cross-sectional imaging by illuminating tissue with low-coherence light and measuring the backscattered light as a function of time delay or range at different transverse positions.[1] Optical ranging is performed by low-coherence interferometry with a fiber-optic Michelson-type interferometer. Precision ranging is possible because interference is observed only when the optical path lengths of the sample and the reference arms match to within the coherence length of the source.[1,13,14] Tissue reflectance is obtained axially by varying the reference arm length and digitizing the magnitude of the demodulated interference envelope.[13,14]

A cross-sectional image is produced by recording axial reflectance profiles while the beam is scanned across the sample. The OCT system in this study is implemented by use of a superluminescent diode with a center wavelength of 1300 nm and a spectral bandwidth of 50 nm coupled into a single-mode fiber-optic Michelson interferometer.[1] The OCT system has a free-space axial resolution of 20 μm, a sample arm power of 150 μW, and a signal-to-noise ratio of 110 dB. The signal-to-noise ratio is determined by comparing the signal from a 100% reflector with the variance of the signal with the sample arm blocked. The total scan time depends on the image size and ranges from 5 to 60 s.

A key technology that is necessary for application of OCT for imaging of internal organ systems is a catheter–endoscope that is capable of delivering, focusing, scanning, and collecting a single spatial-mode optical beam. In addition, the catheter must be flexible and have a small diameter to facilitate its entry into internal channels such as coronary arteries that have inner diameters of ~1 mm. Here we describe the design and performance of a prototype single-mode fiber-optic scanning OCT catheter. This device is an enabling technology for developing a wide range of catheters and

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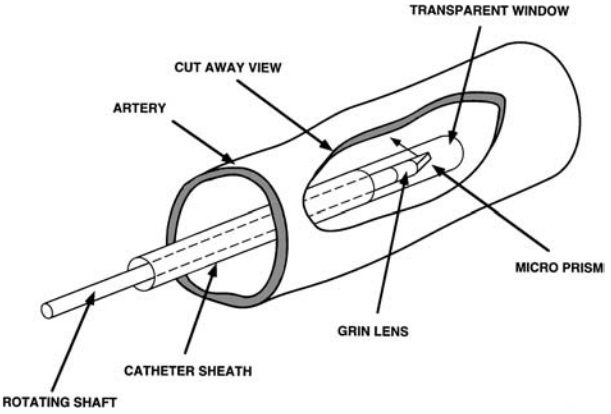

Fig. 1. General use of the OCT catheter–endoscope. The catheter–endoscope is inserted in the center of the lumen of a vessel and scans in a circumferential pattern perpendicular to the axis of the catheter.

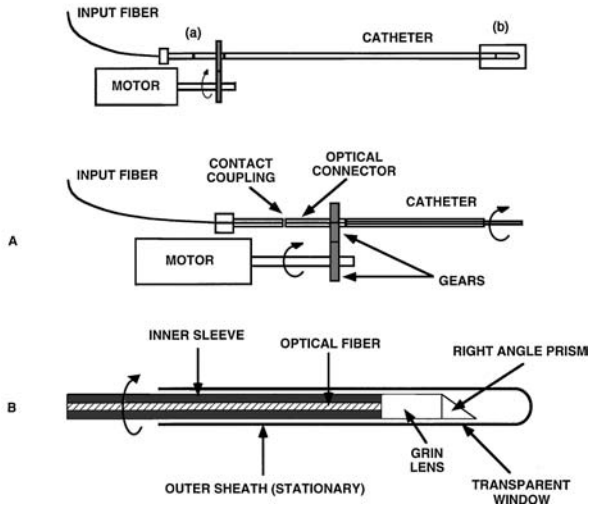

Fig. 2. Schematic of catheter–endoscope: A, Proximal end; B, body and distal end.

endoscopes and will permit the OCT imaging of many tissue and organ systems that were not previously accessible.

The OCT catheter consists of an optical coupling element at its proximal end, a single-mode fiber running the length of the catheter, and optical focusing and beam directing elements at the distal end. The catheter is designed to scan the beam in a circumferential pattern to image cross sectionally through the vessel (or other biological structure) into which it is inserted (Fig. 1).

Beginning at the proximal end of the device, incident light from a fixed single-mode optical fiber is coupled across a small air gap into a second single-mode fiber that can rotate (Fig. 2A). The drive assembly of the catheter, located at the proximal end, uses an optical fiber connector. A gear is attached to the connector and a shaft assembly consisting of the connector, the fiber in the catheter, and the distal focusing elements (Fig. 2A). A dc motor is used to drive the shaft assembly through a gear mechanism.

The catheter scans by rotating the optical fiber and the distal optics. We ensure precise alignment of the optical coupling between the fixed and rotating fibers by mounting the fibers in a precision mechanical ferrule assembly. This is accomplished by use of two male fiber connectors (AT&T-type connectors) and a male-to-male optical fiber coupler typically used for rigidly connecting two fibers. The input fiber connector is fixed to the coupler; the catheter fiber connector is free to rotate. The precise tolerances of the coupler ensure alignment and coupling of light from the fixed to the rotating fiber as the drive shaft rotates. This approach is superior to, and cheaper than, optically coupling across the rotating junction with a pair of lenses and free space. Reimaging the fiber ends onto each other is highly alignment sensitive because it maps angular variations in the beam direction onto transverse spatial displacements of the focused spot position as the shaft assembly rotates.

The body of the catheter is composed of a flexible, rotating inner sleeve that fits loosely inside a stationary outer sheath (Fig. 2B). The body is flexible to allow for bending during passage through the contours of internal channels, such as blood vessels. The outer sleeve allows the inner sleeve to rotate freely and provides a smooth exterior for passage through internal organ systems. The inner sleeve utilizes a hollow flexible cable similar to one found in a speedometer. The optical fiber is fixed within the center of the inner sleeve, which is rotated as a unit with the optical connector. Rotational torque exerted at the proximal end by the drive mechanism rotates the shaft assembly consisting of the fiber and cable within the outer sleeve.

The distal end of the catheter is composed of miniature beam focusing and directing optics. The single-mode fiber is attached to a gradient-index (GRIN) lens (Fig. 2B). The pitch of the lens is chosen to yield the required Gaussian beam parameters. The confocal parameter and the spot size are chosen so that the spot size (or transverse resolution) is comparable with the axial OCT resolution while maintaining an appropriate working distance (~3 mm). A microprism is mounted at the distal surface of the GRIN lens to direct the beam perpendicular to the axis of the catheter (Fig. 2B). The distal end of the catheter is enclosed by a transparent sleeve, which is continuous with the sleeve housing the body of the catheter (Fig. 2B). The microprism, the GRIN lens, the inner sleeve, the optical fiber, and the proximal rotating optical connector are all attached with ultraviolet curing optical cement to form a single unit, the rotating shaft portion of the catheter.

During image acquisition the catheter is inserted into the tissue structure being imaged (the artery or other internal tissue channel), and, as the drive motor turns, the shaft of the catheter and the distal optics circumferentially scan the focused beam perpendicular to the axis of the catheter (Fig. 1). An OCT image is acquired as the beam angle of rotation is varied over some range (usually 360 deg). The speed of imaging depends on the speed of the rotation and the OCT unit acquisition speed. Real-time imaging speeds of greater than 30 frames/s can be achieved by increasing both the rate of rotation of the drive motor and the power of the low coherence source.

A prototype OCT catheter has been designed, constructed, and analyzed. At the proximal end, the catheter uses a 1-rpm dc motor mechanically coupled to a modified AT&T connector through a custom 1:1

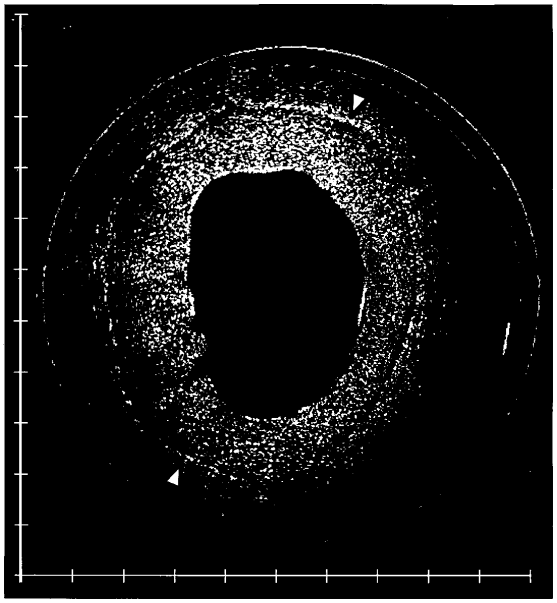

Fig. 3. OCT image of *in vitro* human saphenous vein acquired with the catheter–endoscope. Arrows indicate the border between layers in the vessel wall. Tick marks on the horizontal and vertical axes are spaced 500 μm apart.

ratio gear box. A single-mode optical fiber with a core diameter of 9μm guides the light through the body of the catheter. The distal end comprises a 0.7-mm-diameter, 0.2725-pitch GRIN lens and a 0.5-mm right-angle microprism. The catheter outer diameter is 1.1 mm at the transparent window of the catheter. We experimentally measured the confocal parameter and the focused beam diameter by measuring the reflectance from a mirror as a function of distance perpendicular to the catheter axis. The confocal parameter was 1.74 mm and the spot size was 38 μm at a wavelength of 1300 nm. The focal distance from the center of the catheter was ~3.0 mm.

Variations in optical coupling at the proximal end produced loss as the fiber was rotated. We determined losses that were due to optical coupling variations by measuring the change in reflectance from the fiber–GRIN interface as a function of angle. The loss as a function of angle was found to be $\pm$3 dB. These angular variations can be reduced by use of higher mechanical tolerances or can be normalized out of the image by simple processing techniques. Power loss caused by index mismatches within the catheter was 5 dB. Thus the overall signal-to-noise ratio of the OCT system coupled into the catheter was 105 $\pm$ 3 dB.

To demonstrate imaging with the catheter, we performed OCT on an *in vitro* human saphenous vein. The intact specimen was taken postmortem and imaged with the catheter in the center of the lumen of the vessel. We displayed the rectangular raw data image array in polar coordinates, using a linear interpolation algorithm. The image was also processed to correct for the depth-dependent exponential power loss caused by the optical attenuation of the tissue. The image represents a micrometer-scale tomographic image taken transluminally through the vessel and, to our knowledge, is the first OCT image of this type. The processed image shows the presence of the border between layers in the vessel wall (Fig. 3), demonstrating the ability of this device to image architectural morphology within the lumen of internal organ systems.

To conclude, we have designed and demonstrated a single-mode optical fiber scanning catheter–endoscope for OCT imaging of tissue architectural morphology within human internal organ systems. The catheter is compact and inexpensive and can be readily engineered into a clinically useful form. This device is an enabling technology that will permit the development of OCT-based optical biopsy techniques for a wide range of diagnostic imaging applications in tissues such as the vascular system, the gastrointenstinal tract, the urinary tract, and the respiratory tract.

This research is supported in part by National Institutes of Health contract NIH-9-RO1-EY11289-10 and by the Medical Free Electron Laser Program, U.S. Office of Naval Research contract N00014-94-1-0717. We gratefully acknowledge scientific discussions with E. A. Swanson.

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