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A Concentric Three Element Radial Scanning Optical Coherence Tomography Endoscope

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Abstract

We have developed a 2.1 mm outer diameter optical coherence tomography endoscope that provides radial scans of luminal structures. The endoscope consists of three elements: (1) a stationary central core containing the fibers and focusing elements, (2) a rotating intermediate tube with a distal rod prism, and (3) a stationary sterilized glass cover. This design enables radial and spiral scanning and allows adjustment of the axial focal distance. Additionally, this design is capable of focusing light from multiple fibers into tissue. The performance of the endoscope was demonstrated in a study of tissue engineered blood vessels imaged at various time points during development.

Optical coherence tomography (OCT) is a technology that uses near infrared light to provide depth-resolved, cross-sectional images of tissue. Using low coherence interferometry and optical heterodyning, OCT provides significant improvements in both resolution and dynamic range compared to ultrasound. Axial and lateral resolutions typically range from 5 to 30 μ m depending on the light source and sample arm optics. These performance improvements come at the expense of penetration depth -OCT is typically limited to 1 -2 mm in highly scattering tissues.

OCT systems can be developed using standard fiber optic components. The low cost and small diameter of single mode fiber facilitates the construction of miniature OCT endoscopes. Researchers have developed a variety of probes to image tissues such as colon, esophagus, and blood vessels.[1] OCT endoscopes can be classified by how the light is directed out of the endoscope. End-firing designs image tissue at the tip of the endoscope. Side-firing designs direct the beam out of the side of the endoscope to image adjacent tissue. The side-firing endoscopes can be divided into longitudinal and radial scanning designs. The first radial side-firing OCT probe used a gradient index lens and prism to focus and direct light from a single mode fiber into tissue. These elements were rotated using a proximal motor and a rotary fiber coupler.[2] Other researchers have developed radial scanning endoscopes that use micromotors located in the distal end of the endoscope. This motor rotates a prism while the fiber and gradient index lens remain fixed.[3,4]

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We have developed another form of radial scanning, side-firing OCT endoscope. The aims for this design were to develop a miniature endoscope that minimizes time-varying fiber birefringence, eliminates wire shadows, and provides complete 360° radial or spiral scans of the surrounding tissue.

As shown in Figure 1, the endoscope design contained three concentric elements: a stationary central core containing the focusing optics, a rotating intermediate tube containing the beam-diverting rod prism, and a stationary sterile glass cover. An 18-gauge hypodermic stainless steel tube with an inner diameter of 1.07 mm was used for the central core. The optics were designed for 1300 nm center wavelength operation, and consisted of a single mode fiber (SMF-28), a ferrule, and a gradient index lens. The 0.975 mm diameter ferrule was used to radially center the fiber on the gradient index lens. The 3.348 mm long, 1.0 mm diameter gradient index lens (SLW-1.0, NSG America) placed the focus 2.35 mm from the distal end of the lens. These components were secured with standard fiber optic epoxies (NOA63, Norland, New Jersey and F120, Thorlabs, New Jersey).

The intermediate tube consisted of 16-gauge hypodermic steel (1.35 mm inner diameter) with a 1.0 mm aluminum coated rod prism affixed at the distal end with epoxy to deflect the beam out the side of the endoscope. A 180° window 1.7 mm in length was machined 4.5 mm from the distal end of the intermediate tube to allow the light to exit. The rod prism was tilted with a 100 mm thick spacer so that the reflecting surface was oriented at 48° with respect to the lens optical axis. This tilt caused the beam to exit at a slight angle to the endoscope radial axis, minimizing reflections from the outer tube.

The outer glass cover consisted of a custom glass tube, sealed at the distal end, with an inner diameter of 1.8 mm, a wall thickness of 150 μm , and a length of 20 cm. A steel ring adapter was epoxied to the proximal end of the tube to facilitate mounting to the endoscope motor mount (F120, Thorlabs, New Jersey). The glass cover was sterilized using standard steam autoclave procedure. When the endoscope was assembled, the central core and intermediate tube were positioned longitudinally so that the rod prism was displaced 900 mm axially from the distal end of the GRIN lens. In this configuration, the focus was placed 400 μm from the outer dimension of the endoscope. However, the focal plane position could be modified by manual translation of the central core within the intermediate tubes. A set screw in the mechanical mount is used to lock the central core in place. This adjustment allowed the focal plane to be positioned at the surface of the endoscope to 1.3 mm outside the endoscope. The optics provided a theoretical spot size with $1/e^2$ diameters of 26 and 22 mm at the focal plane. The astigmatism was created by the beam's path through the cylindrical glass cover. There was negligible change in spot size or shape with change in focal plane position.

To achieve radial scanning, a direct current motor with a planetary gearhead and optical encoder were used. A motion controller enabled precision rotation of the endoscope optics using a serial port connection with the OCT system computer. A custom aluminum mount was designed to house this motor assembly and transfer the rotary motion to the intermediate tubing of the endoscope using identical gears attached to the motor shaft and intermediate tubing. For longitudinal motion, a linear actuator positioned the central core and intermediate tubes within the glass cover. Simultaneous engagement of rotational and longitudinal actuators led to spiral scanning. At the distal tip of the endoscope, the air space separating the central core and intermediate tubes could vary less than 80 μm (between 150 and 230 μm). This tight tolerance prevented the optics from experiencing significant wobble during image acquisition for radial scanning speeds up to 1 rotation per second. A diagram of the endoscope optics and motion control is shown in Figure 1.

To demonstrate the performance of the endoscope, we imaged ten tissue engineered blood vessel mimics (BVMs). The endoscope was connected to a time-domain OCT system that has been described previously.[5] Briefly, a superluminescent diode light source with a central wavelength of 1300 nm and bandwidth of 100 nm provided a measured axial resolution of 10 μm in air (7.5 μm in water). Reference arm scanning was provided by a Fourier domain optical delay line operating at 200 Hz. The tissue engineered constructs consisted of two primary components: (1) a biocompatible polymeric scaffold and (2) a thin cellular lining. Three types of 4.0 mm inner diameter tubular scaffolds were used – three expanded polytetrafluoroethylene (ePTFE), three polyurethane, and four Dacron. Microvessel endothelial cells from human adipose tissue were seeded at a density of 1.25×10^6 cells/ cm^2 on to the inner surface of these scaffolds using identical protocols. Each blood vessel was developed in a bioreactor, which has been described previously.[6] The endoscope could be inserted through a three-way tye port into the lumen of the bioreactor. The glass cover was sterilized prior to each imaging session to prevent contamination of the bioreactors. We imaged each BVM at 1, 4, 7, 14, and 21 days post-seeding. Radial cross-sectional images consisting of 2000 a-scans with 625 points per a-scan were obtained every 1.0 mm for the central 10 mm of the BVM. While the endoscope was capable of three-dimensional, spiral scanning, the slow speed of our OCT system meant that individual two-dimensional images were more practical to obtain on the living BVMs.

Both ePTFE and Dacron material appeared relatively hyperintense. Polyurethane appeared relatively hypointense and additionally large media-filled, non-scattering pores were visualized. Figures 2 and 3 illustrate the development of a confluent cellular lining within an ePTFE mimic. Figure 2 shows the entire radial scan and in figure 3, a portion of the image is enlarged to show the cellular lining. A very thin cellular lining with clumps of cells was seen within the mimic one day after seeding (figure 2a, 3a). By day 4 (figure 2b, 3b), the lining had developed into a thick, hypointense confluent cellular layer. At days 14 and 21 (figure 2c, 3c and figure 2d, 3d, respectively) the lining was visualized to generally become thicker and more hyperintense. Islands of cellular material were visualized on the polyurethane grafts. No cells were visualized on the Dacron material, indicating that alternative protocols are required for successful development of BVMs using polyurethane and Dacron material. Decreases in the OCT signal intensity for deep regions of the mimic were recorded. This signal degradation is due to out of focus imaging, and could be mitigated in future experiments by adjusting the focal distance of the endoscope. This experiment supports the use of endoscopic OCT as a tool for process control and development of tissue engineered constructs.

The concentric three element endoscope design offers advantages in comparison to previously-reported radial scanning designs. There are several merits of keeping the fiber optics, ferrule, and gradient index lens stationary during imaging. First, an optical rotary coupling and rotating fiber are not necessary. This eliminates time varying birefringence that is introduced by stressing the fiber as it is being rotated. Second, the fixed positions of the fiber and focusing optics allow for the option of multiple imaging channels within the endoscope, e.g. for simultaneous acquisition of fluorescence spectra. Commercially available multiple channel rotary couplers have unacceptable losses and are limited in wavelengths of operation. Third, the design allows the operator to adjust the working distance (location of focal plane from endoscope window) while keeping the numerical aperture fixed. This can be performed by adjusting the position of the stationary center core within the rotating intermediate tube.

A disadvantage of distal micromotor-mounted prism designs is the presence of electrical wires obscuring a portion of the radial cross-section. Our endoscope provides complete 360° radial cross-sections of the surrounding tissue without these obscurations. Micromotors can

also have problems with excessive temperature, because they are contained within an insulating material like glass or plastic, and the size of the micromotor places a limit on how small the probe can be made.[1] Although the current concentric three element design was developed to image within vessels with a luminal diameter greater than 2.1 mm, the endoscope can be reduced in size using smaller gauge hypodermic tubing and optical focusing elements. Since the motor in this design is housed outside of the endoscope, higher quality motors with optical encoder feedback can be used for accurate positioning, which is important for image registration when collecting volumetric datasets.

The major limitation of the current design is that applications are limited to tissues that can be accessed with a semi-rigid endoscope. However, rigid endoscopes are widely used for cystoscopy, laparoscopy, thoroscopy, rhinoscopy, and arthroscopy. Examples of tubular structures that could be imaged with this rigid endoscope include the urethra, vagina, airway, and distal colon. A flexible version could be developed but the low torsional stiffness of flexible materials may lead to inaccurate rotational scanning, e.g. non-uniform rotational friction introduced by a flexible rotating intermediate tube may lead to non-uniform rotational speed during image acquisition. Further design and analysis is needed to determine if a flexible concentric three element endoscopic system could be developed.

In future implementations, we plan to incorporate several multimode fibers into the stationary central core that will enable simultaneous endoscopic acquisition of OCT and laser-induced fluorescence spectroscopy of the tissue. Similar to previous longitudinal-scanning designs,[7] additional multimode fibers illuminate tissue with the excitation light and collect the resulting fluorescence. This endoscope would enable rapid spiral scanning acquisition of potentially synergistic structural information from optical coherence tomography and biochemical information from laser-induced fluorescence.

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Figure 1.

The endoscope design provides radial scanning using three concentric tubes. Rotational motion from a DC motor (M) is transferred to the intermediate tube (IT) using two identical gears (GR) within the motor mount (MT). A stationary central core (CC) contains the single mode fiber, ferrule (F), and GRIN lens (G). A glass cover (GC) provides a transparent and sterile interface between the tissue and endoscope. The inner tubes (IT and CC) can be moved in and out of the glass cover to provide longitudinal positioning.

Figure 2.

OCT time series of an ePTFE tissue engineered blood vessel mimic. (A) The 1 day image shows a thin confluent lining. (B) The 4 day image shows an approximately 100 μ m thick, hypointense cellular lining. The 1st and 2nd reflections are from the outer glass window and multiple reflections occurring within the GRIN lens, respectively. (C) The 14 day image shows a more hyperintense, approximately 85 μ m thick lining with similar intensity as the ePTFE scaffold. (D) The 20 day image indicates an approximately 150 μ m thick cellular lining. Scale bar = 1.0 mm.

Figure 3. Enlarged portions of the images shown in figure 2. (A) Day 1, arrow points to clumps of cells, (B, C, D) Day 4, 14, and 21, respectively, arrows show bounds of cellular lining. Scale bar = 0.5 mm.