Chapter 22
What is OCT?

Krzysztof Izdebski & Brian J.F. Wong

Abstract

Optical Coherence Tomography (OCT) is an optical based technology that uses light to generate cross-sectional images of turbid media such as living tissue. In medicine and biology, OCT can provide images to a depth of about 1 mm, and does not expose patients to the risks of ionizing radiation. There are several key organs where detailed micro-anatomic information of surface structure is extremely important and this includes the retina, coronary vasculature, and of course the delicate mucosa of the upper aerodigestive tract, in particular the vocal folds.

Keywords: *OCT, mucosa, vocal folds, malignant changes*

With conventional white light (WL) imaging (e.g., laryngoscopy, stroboscopy), we only see the outside surface of a structure. NBI® provides observation of vascular components located "deeper," but to see the internal anatomic structure in patients, conventional imaging technologies are typically used. These include computer tomography (CT), magnetic resonance (MRI), or ultrasound. But there is always a tradeoff between the depth of imaging and resolution. Fortunately, critical vocal fold (VF) structures are relatively superficial, which makes OCT an ideal technology to visualize subsurface organization. To look *in vivo* inside the VF we need to employ a specific technology. This can now be accomplished using Optical Coherence Tomography (OCT) [1].

OCT is analogous to ultrasound in that it relies upon the backscattering of light incident in tissue to generate images. In OCT, differences in tissue optical properties provide contrast. In MRI, CT, and ultrasound imaging, contrast is provided by proton density, electron density, and tissue acoustic impedance, respectively. An OCT system consists of a low coherence light source—known as Michaelson interferometer, which separates light into two optical pathways—and a detector.

During OCT imaging, light is focused at a target, and a portion of this light is reflected from sub-surface. The light or beam generated by backscattered light from different depths is collected simultaneously. Before leaving the instrument, part of the light from the source is diverted away from the optical pathway by a beam splitter. This split light travels down a second pathway and is incident on a reflective surface. The light reflects and re-enters the beam splitter where it is recombined with the light which is backscattered from the target tissue surface. This arrangement of a beam splitter, a light source, and two pathways is known as Michealson interferometer (the heart of all OCT systems).

Constructive interference occurs when photons have traveled the exact same distance along the two different optical pathways (toward the tissue and toward the reflector), and a robust signal is detected. This signal depends upon the local optical properties in the region, and it is this quantity (optical scattering) that is the physical basis for contrast in an OCT image. By adjusting the position of the reflective surface to and fro, depth scans in tissue can be constructed, forming what are called "A-line scans," terminology borrowed from ultrasonography. A group of A-line scans are generated by physi-

cally moving the device or beam across the tissue surface. In this manner a 2D image (or B-scan) is generated. This approached described above is used in Time Domain OCT and is often referred to as raster scanning.

Time Domain OCT has largely been supplanted, as the optoelectronics has become more sophisticated using either laser light sources that "sweep" across a range of wavelengths or diffraction gratings that separate wavelengths across different pathways allowing collection of data from multiple depths within the tissue using a detector array. The details of these approaches (frequency domain OCT) are beyond the scope of this discussion, but frequency domain systems are used in virtually all commercial and research devices presently.

Resolution in OCT systems is somewhat complex to describe, in that axial resolution differs from lateral resolution. Axial refers to the direction of light propagation into the tissue. Lateral resolution refers to the direction perpendicular to the incident light. Currently, the axial resolution of most OCT systems is limited to about 10 µm in tissue and is limited by the coherence length of the light sources. Both LEDs and lasers are used as sources for OCT systems. Lateral resolution is determined by the optical design of the system (optical components and their arrangement) and is diffraction-limited as in conventional light microscopy or endoscopy. Typically, lateral resolution is on the order of magnitude of axial resolution.

Depth of penetration is determined by the propagation of light into the tissue and the backscattering of photons to the detector. This depends upon the optical properties of tissue (absorption coefficient, scattering coefficient, and anisotropy), which are wavelength dependent. In general, OCT systems have migrated toward the use of infrared light as it is less highly absorbed by tissue than visible wavelengths.

Depending on the tissue optical properties for the illuminating wavelength, micrometer resolution can be achieved allowing visualization of the tissue at reliable depths of about 1 mm. This process is not unlike looking through a low-power microscope, but the difference being that images can be obtained at a depth and *in vivo*, and can provide visualization of tissue architecture well in excess of ultrasound or MRI.

Currently, OCT penetrates at the depth of 1 mm in most biological tissue, without radiation or a need for invasive maneuvers. Therefore OCT can be used *in vivo* safely. OCT has been used in imaging the VF for quite some time [2-7], though commercial office-based systems are unavailable.

References

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