

UC Irvine

UC Irvine Previously Published Works

Title

High-resolution optical coherence tomography using self-adaptive FFT and array detection

Permalink

<https://escholarship.org/uc/item/4183847m>

Authors

Zhao, Yonghua
Chen, Zhongping
Xiang, Shaohua
[et al.](#)

Publication Date

2001-05-23

DOI

10.1117/12.427904

Copyright Information

This work is made available under the terms of a Creative Commons Attribution License, available at <https://creativecommons.org/licenses/by/4.0/>

Peer reviewed

High-resolution optical coherence tomography using self-adaptive FFT and array-detection

Yonghua Zhao*, Zhongping Chen, Shaohua Xiang, Zhihua Ding,

Hongwu Ren, and J. Stuart Nelson

Beckman Laser Institute and Medical Clinic, University of California, Irvine, CA 92612

Jinendra K. Ranka, Robert S. Windeler, and Andrew J. Stentz

Bell Laboratories, Lucent Technologies

ABSTRACT

We developed a novel optical coherence tomographic (OCT) system which utilized broadband continuum generation for high axial resolution and a high numeric-aperture (N.A.) objective for high lateral resolution ($<5 \mu\text{m}$). The optimal focusing point was dynamically compensated during axial scanning so that it can be kept at the same position as the point that has an equal optical path length as that in the reference arm. This gives us uniform focusing size ($< 5 \mu\text{m}$) at different depths. A new self-adaptive fast Fourier transform (FFT) algorithm was developed to digitally demodulate the interference fringes. The system employed a four-channel detector array for speckle reduction that significantly improved the image's signal-to-noise ratio.

Keywords: Optical coherence tomography (OCT), Continuum generation, FFT, Speckle

1. INTRODUCTION

Optical coherence tomography (OCT) has found several applications [1-4] in biomedical engineering because of its ability to perform high-resolution cross-sectional imaging of tissue *in vivo* and *in situ*. OCT is based on a low-coherence Michelson interferometer with sample and reference mirrors in the two arms of the interferometer. Interference only occurs when the reference and sampling arms are matched within the coherence length of the source. Axial scanning is achieved by moving the reference mirror, and the intensity of interference fringes at different time windows during the scan can be used to represent the reflection (or scattering) information at different depths in a sample. The coherence length of the light source determines the system's capability to distinguish axially neighboring scattering points, i.e., the axial resolution. Light sources with a broader spectrum-bandwidth have a shorter coherence length and, therefore better axial resolution. Alternatively, lateral resolution is dependent on the beam focusing size, which is relative to the transverse mode quality, or spatial coherence of the light source. Better mode quality, which leads to better spatial coherence, makes it easier to focus onto a smaller size with better lateral resolution. A single-mode fiber is often used as a spatial filter to obtain high transverse mode quality. Therefore, broad spectrum-bandwidth and high coupling efficiency into a single-mode fiber are two of the requirements for an OCT light source.

Currently, superluminescent diodes (SLDs) are widely used in OCT, with a center wavelength of 850 or 1300 nm. The axial resolution for a typical SLD is approximately 10 to 20 μm with a single-mode power of 1 to 10 mW. A specially-designed, mode-locked Ti:sapphire laser would seem to a very good choice at 850 nm for high-resolution OCT imaging because it can provide an ultra-broadband spectrum (350 nm bandwidth which corresponds to about 1 μm in resolution) and high power ($> 100 \text{ mW}$) [5] in a single transverse-mode. However, the complicated mode-locking technique and high cost of such a laser system

* Correspondence: Email: yzhao@bli.uci.edu or zchen@bli.uci.edu; WWW: <http://www.bli.uci.edu/>; Telephone: 949-824-4713; Fax: 949-824-8413

makes it impractical for clinical applications. Furthermore, the penetration depth of OCT in scattering media (for example, human skin) at 850 nm is much shorter than at 1300 nm, which limits its imaging potential in cases where information from deep tissue structure is important.

Recently, researchers at Bell Laboratory developed an air-silica microstructure fiber with anomalous dispersion at 800 nm [6]. By injecting pulses of 100-fs duration, 800-pJ energy at a center wavelength of 790 nm into a 75-cm section of fiber, it was possible to generate an ultra-broadband continuum extending from 390 to 1600 nm. The nonlinear effects in such a fiber, such as self-phase modulation and Raman scattering, results in a broad, flat spectrum. Undoubtedly, this could be an ideal source for high-resolution OCT in any wavelength region spanning from visible to near IR.

Here we demonstrate a high-resolution OCT system that uses an air-silica microstructure fiber to produce the spectrum continuum light as a source. We also developed a novel self-adaptive algorithm to remove the side effects caused by the instability of the scanning device. A four-channel array detection system is used to reduce the speckle noise in OCT image acquisition.

2. SYSTEM SETUP

Most OCT systems utilize a 2x2 fiber coupler (50/50) to construct a Michelson interferometer. The advantages of such a fiber-based system are its compact size, easy alignment, and flexible probe design in the sampling arm. However, some limitations are noted when such a fiber-based OCT system is used for high-resolution imaging. First, a 2x2 coupler with a bandwidth more than 200 nm is difficult to make. Second, a lens system such as fiber collimators or focusers will also limit the efficient bandwidth because of chromatic aberration. Third, the fiber would be multi-mode if the short-wavelength limit of the total spectrum were smaller than the cut-off wavelength of the fiber. This will introduce ghost-lines in the OCT image. Fourth, polarization mode dispersion in a single-mode fiber may also limit the axial resolution [4]. Due to these limitations, we built a high resolution OCT as an open system, shown in Figure 1.

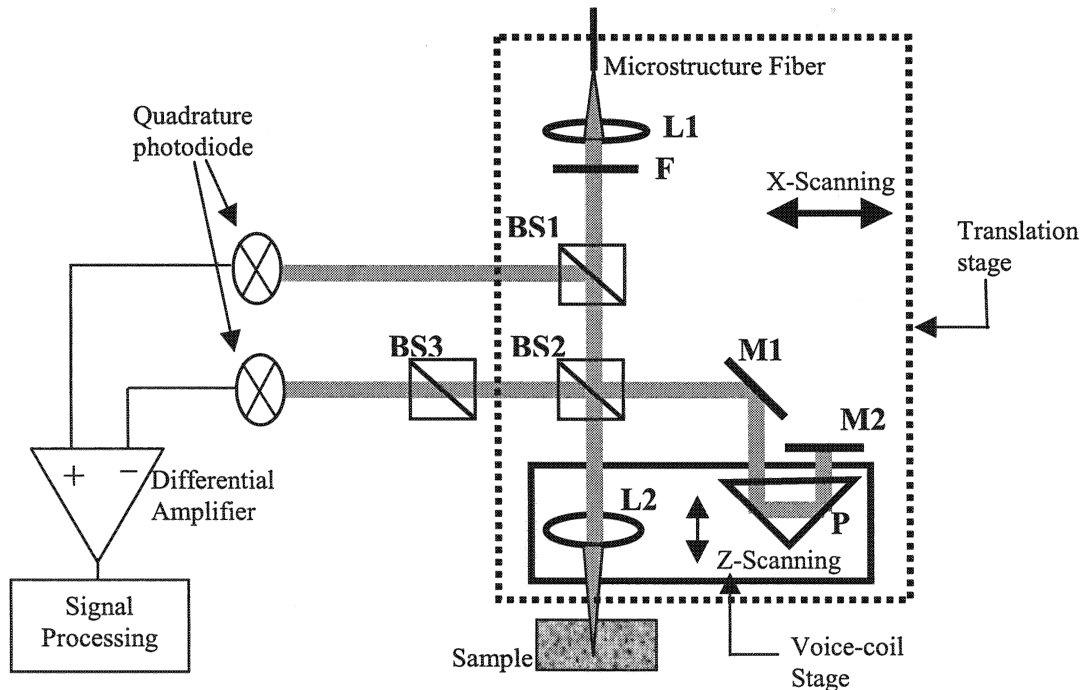


Fig.1: High-resolution OCT optical system. F: filter; BS1, BS2, BS3: 50% beam splitter; L1: collimator lens; L2: objective lens; M1, M2: high-reflection mirror; P: 180-degree prism; L2 and P are both mounted on a voice-coil translation stage.

The light source for this interferometer is continuum generation from an air-silica microstructure fiber. The pumping source for continuum generation is a self-mode-locked Ti:sapphire laser. The total output power of the laser is greater than 300 mW, with a pulse-duration of 45 fs and a repetition-rate of 80 MHz. The laser beam is then coupled into the microstructure fiber after an optical isolator, as shown in Fig. 2A. The spectrum of light is broadened when propagating inside the fiber because of self-phase modulation and Raman scattering. The color of the fiber varied from red, to yellow, to blue, along the fiber, as shown in Fig. 2B. The total output power from the microstructure can be as high as 70 mW and the remaining power is about 32 mW after a long-pass filter, which blocks all light with a wavelength shorter than 850 nm.

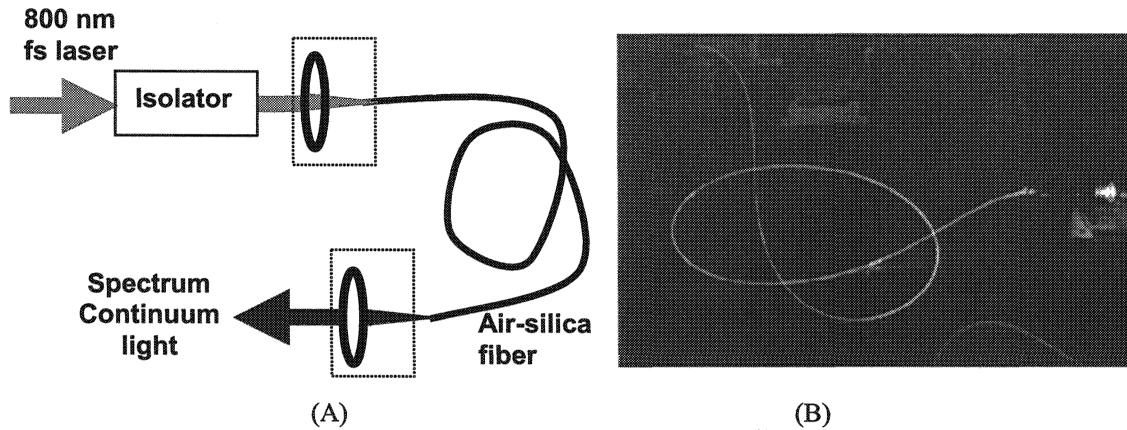


Fig. 2: Continuum generation in an air-silica microstructure fiber. A: Schematic of the system. B: picture of continuum generation from the air-silica microstructure fiber

To achieve high lateral resolution, it is necessary to use an objective lens with a high numerical aperture (NA) in the sampling arm to obtain a small focusing size. However, a small focusing size also means a small confocal parameter, which represents the axial imaging range with a similar beam size. For a Gaussian beam, the relationship between confocal parameter (Z_0) and beam waist radius (w) is given by

$$Z_0 = \frac{\pi w^2}{n\lambda} \quad (1)$$

where n is the sample refractive index and λ is the center wavelength of source. If the beam size is 4 μm , for example, the confocal parameter is 30 μm . In other words, within an imaging depth of 30 μm , the lateral resolution is 4 μm , all other regions have a lateral resolution much greater than 4 μm .

Several methods were developed to resolve this problem. One example is using C-mode scanning to reconstruct tomographic images, but this increases image acquisition time. In our system (Fig. 1), we use a dynamic focusing tracking method [7] to keep a zero path length difference between the reference and sampling beam in the focus region during axial scanning based on the following. Assuming the moving speed of the voice-coil stage (for axial scanning) is V , the moving speed of the focusing point inside sample will be nV because the objective lens is mounted on the voice-coil stage. Consequently, the moving speed of the optical path length for the focal point in the sampling arm is n^2V . Alternatively, the scanning speed of the optical path length in reference arm will be $2V$ because the 180-degree prism is mounted on the same voice-coil stage. If $n = \sqrt{2} = 1.414$, the two scanning speeds would match exactly. For a biological tissue sample ($n = 1.33$ to 1.45), this method can dynamically track the focus well enough to keep the lateral resolution within a few micrometers.

To reduce speckle noise in the OCT images, we used a quadrature detector to receive the interference signal. Speckle is formed because of the multiple scattering of photons in tissue and the limited spatial-frequency bandwidth of partial coherence sources. Wavefront distortion due to multiple scattering and

tissue inhomogeneity results in a speckle pattern. Signal loss occurs whenever different portions of the returning wave add destructively. Speckle noise reduces image contrast and makes boundaries between highly scattering structures in tissue difficult to resolve. If we use a multiple detector array, wave front distortions due to multiple scattering and tissue inhomogeneity create a different speckle pattern at each detector. Incoherent addition of the imaging from the multiple detectors increases the signal to noise ratio by a factor of square root of N , where N is the number of detectors. Figure 3 shows an example using this technique. The sample is from nail part of a human finger. Fig. 3A is the OCT image reconstructed from one detector, and Fig. 3B is the image after averaging from a array of four detectors.

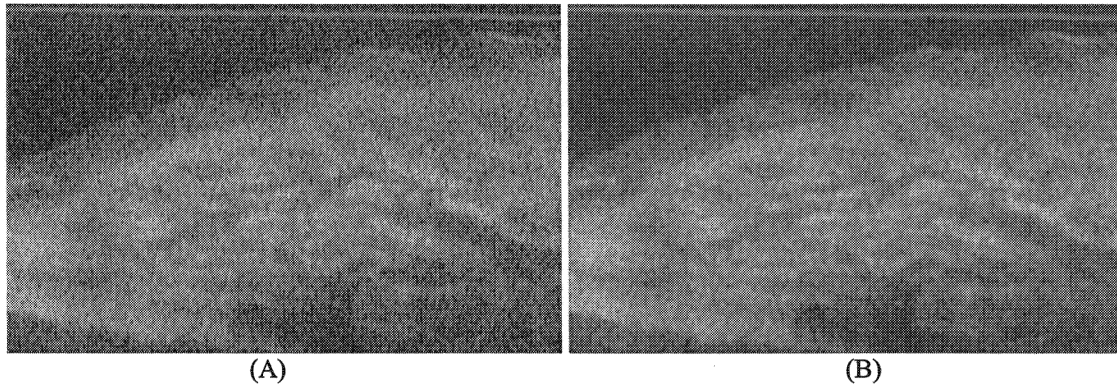


Fig. 3 Speckle reduction by array detection technique.

The demodulation quality is largely dependent on knowledge of the carrier frequency. Unfortunately, the velocity of the voice-coil translation stage is not constant because of the inherent shortage in its control loop, which causes the carrier frequency change during axial scanning. To demodulate the signal correctly, we developed a self-adaptive FFT algorithm to track the carrier frequency. A diagram of this algorithm is shown in Fig. 4. Our result shows that the dynamic range can exceed 120 dB using this in conjunction with a SLD as a light source.

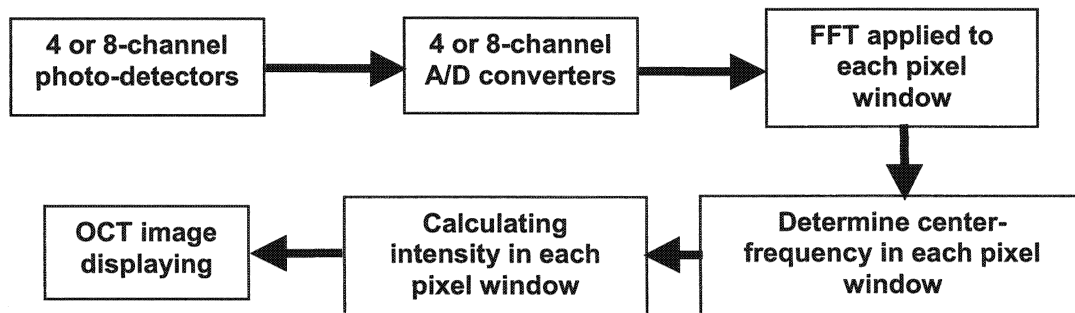


Fig. 4 Self-adaptive fast Fourier transform used in high-resolution OCT.

Lateral scanning is completed by moving the whole interferometer so that the scanning will not affect the performance of the objective lens. The whole numerical aperture of the lens can be utilized to obtain the smallest focusing size with this design.

3. RESULTS

Figure 5 shows two preliminary results using this novel high-resolution OCT system. The sample used in the first experiment is onion and its OCT image is shown is Fig. 5A. The image size is 1 mm × 1 mm.

Single cell structures can be identified in this image. Fig. 5B is an OCT image taken from human skin and the image size is also 1 mm \times 1 mm. The image has much better contrast when compared to the images obtained when using a SLD as the light source. We are currently working on characterizing and optimizing the system performance in terms of spatial resolution, contrast, and penetration depth.

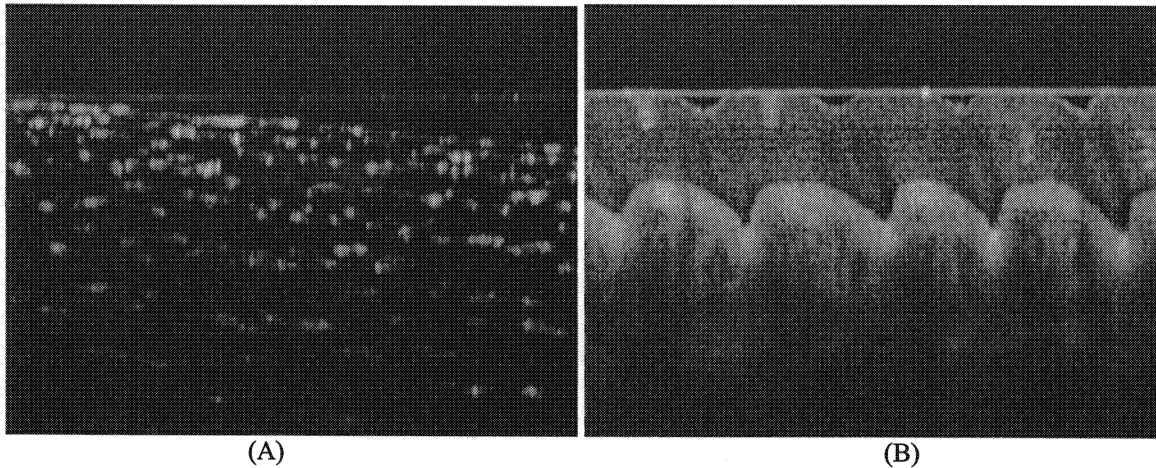


Fig. 5 OCT images of onion (A) and human skin (B).

4. CONCLUSIONS

In summary, we have demonstrated a high-resolution OCT system that uses continuum generation from a special microstructure fiber as a light source. The system is capable of dynamic focusing compensation to obtain constant lateral resolution and has a four-channel array detection to reduce speckle noise. A self-adaptive FFT algorithm was developed to enhance system performance to demodulate the fringe signal that has an unstable carrier frequency.

5. ACKNOWLEDGMENTS

This work was supported by research grants awarded from the Whitaker Foundation (23281, ZC), National Science Foundation (BES-86924), and National Institutes of Health (HL-64218, RR-01192 and GM-58785). Institutional support from Air Force Office of Scientific Research (N00014-94-1-0874), Department of Energy (DE-FG03-91ER61227), and the Beckman Laser Institute and Medical Clinic Endowment is also gratefully acknowledged.

REFERENCES

1. D. Huang, E. Swanson, C. P. Lin, J. S. Schuman, W. G. Stinson, W. Chang, M. R. Hee, T. Flotte, K. Gregory, C. A. Puliafito, and J. G. Fujimoto, *Science* 254, 1178(1991)
2. J. M. Schmitt, S. L. Lee, and K. M. Yung, *Opt. Commun.* 142, 203(1997)
3. Y. Zhao, Z. Chen, C. Saxer, S. Xiang, J. F. de Boer, J. S. Nelson, *Opt. Lett.* 25, 114(2000)
4. C. E. Saxer, J. F. de Boer, B. H. Park, Y. Zhao, Z. Chen, and J. S. Nelson, *Opt. Lett.* 25, 1355(2000)
5. W. Drexler, U. Morgner, F. X. Kartner, C. Pitris, S. A. Boppart, X. D. Li, E. P. Ippen, and J. G. Fujimoto, *Opt. Lett.*, 24, 1221(1999)
6. J. K. Ranka, R. S. Windeler, and A. J. Stentz, *Opt. Lett.* 25, 25(2000)
7. Z. Chen, T. E. Milner, D. Dave and J. S. Nelson, *Opt. Lett.* 22, 64 (1997)