Ultrahigh-resolution optical coherence tomography at 1.15 µm using photonic crystal fiber with no zero-dispersion wavelengths

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Abstract: We report a broad-band continuum light source with high power, low noise and a smooth spectrum centered at 1.15 µm for ultrahigh-resolution optical coherence tomography (OCT). The continuum is generated by self-phase modulation using a compact 1.059 µm femtosecond laser pumping a novel photonic crystal fiber, which has a convex dispersion profile with no zero dispersion wavelengths. The emission spectrum is red-shifted from the pump wavelength, ranges from 800 to 1300 nm and results in a measured axial resolution of ~2.8 µm in air. We demonstrate ultrahigh-resolution OCT imaging of biological tissue using this light source. The results suggest PCF with this type of dispersion profile is advantageous for generating SC as a light source for ultrahigh-resolution OCT.

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Reference
1. Introduction

In optical coherence tomography (OCT), the axial resolution is determined by the bandwidth of the light source. Continual efforts have been made to develop light sources with sufficient intensity, low noise and broad bandwidth for real-time, ultra-high resolution OCT (UHR-OCT) [1]. A smooth Gaussian-like spectrum is desired for suppression of side-lobes. In addition, a compact, convenient device is important in the clinical setting. In order to achieve better penetration depth in opaque tissue and cover the effective responsivity of InGaAs detectors, it is reasonable to choose the wavelength range of the light source to be above 1 μm [2]. It has been shown through numerical simulation that for light sources with center wavelengths above 1.2 μm, it is difficult to achieve axial resolution better than 2 μm to 3 μm at depth in tissue due to absorption and imperfect dispersion compensation [3]. Therefore, a broad-bandwidth light source for UHR-OCT centered between 1.1 μm and 1.2 μm could be advantageous for mitigating resolution degradation, covering the effective wavelength range of InGaAs detectors and maximizing imaging penetration depth.

Light sources based on supercontinuum (SC) generation have drawn much attention for UHR-OCT in the wavelength range above 1 μm because of their broad bandwidths [4-11]. SC is generated by pumping a highly nonlinear fiber, such as photonic crystal fiber (PCF), with femtosecond laser pulses either at around the zero dispersion wavelength (ZDW) [7, 9, 12] or in the normal dispersion area [6, 8, 10, 11] of the fiber. Pumping the fiber around a ZDW commonly used to generate SC with a very broad bandwidth. By pumping a PCF in the anomalous dispersion area with a Ti:Sapphire laser, broad band light sources centered at around 1.3 μm and 1.1 μm have been used to achieve 2.5 μm and 1.8 μm axial resolution (in air) with UHR-OCT [7, 12]. Further, by using a compact fiber laser at 1 μm to pump a PCF just below the ZDW, a SC covering from 800 nm to 1300 nm was also demonstrated for UHR-OCT [9]. However, when femtosecond laser pulses are launched near the ZDW of PCF, SC is generated by the excitation of unstable solitons [13]. Therefore, these light sources show large spectral modulations and tend to be noisy [14, 15]. 15 dB dynamic range reduction due to noise amplification has been reported compared with SC based on self-phase modulation (SPM) [16]. In addition, spectral filter was required for achieving desired spectral range, which reduced the output power and increase the complexity of the system [5, 12]. Therefore, SC generation in the normal dispersion area by SPM is preferred for UHR-OCT because this can achieve a smooth spectrum, low noise and high output power [16]. However, SC based on
SPM in conventional single-mode fiber generally results in a relatively narrow bandwidth and the output wavelength is limited to being around the pump wavelength. By pumping a UHNA fiber with a Nd:glass laser centered at 1.059 μm, a bandwidth of 139 nm has been demonstrated. The axial resolution in the air is around 5 μm[6]. Further extension requires higher peak power coupled into the fiber, which will increase the cost and size of the laser. Similar performance was also achieved with a Cr:fosterite fs laser and a dispersion shifted fiber[8]. However, compact and portable Cr:fosterite lasers are not currently commercially available. Recently, we and others have reported a dual-band light source with bands centered at 800 nm and 1300 nm by pumping a PCF with two closely spaced ZDWs[17, 18]. The SC generation of this light source is mainly attributed to SPM. However, spectrum shows appreciable spectral modulations which can only be avoided when ultra short pulses (<50 fs) are employed, and resolution is limited to about 6 μm at 1300 nm band. In order to achieve even broader-band SC based on SPM, a PCF was tapered down to 40 μm diameter, resulting in a constant, large normal dispersion profile at about 800 nm[10]. Although less than 2 μm axial resolution was demonstrated with this light source, the light was centered at 800 nm, not above 1μm. In addition, the tapered fiber is likely to be fragile and difficult to handle.

In this work, we demonstrate a new SC light source centered at 1.15 μm by pumping a novel PCF with a commercial compact femtosecond laser at 1.059 μm. The SC is attributable to SPM, so the spectrum is smooth and Gaussian-like with very low side-lobes, and low noise because there is no noise amplification. Also, because of the unique dispersion profile of this PCF, the SC is extremely broad, resulting in a clean, narrow point spread function (PSF) of only 2.8 μm width (in air). Moreover, the center wavelength of SC is shifted from the pump wavelength of 1.059 μm to 1.15 μm, which is desirable, as described above. This PCF transfers most of the power of the pump laser to a very broad band in the desired wavelength range but still maintains the benefits of SC generation by SPM and has sufficient power for real-time UHR-OCT.

2. Supercontinuum generation with PCF based on SPM

The fiber used for SC generation is a commercially available PCF (NL-1050-NEG-1, Crystal Fiber A/S). This fiber has a core size of about 2.3 μm and a convex dispersion profile without ZDWs. The convex dispersion profile results in a relatively flat dispersion area from 0.9 μm to 1.1 μm that is very close to zero dispersion (Fig. 1(a)). The dispersion profile is slightly modified from the manufacturer’s specification to best fit the simulation to the experimental result shown below. The cross section of the PCF is also shown in the inset of Fig. 1(a), where air diameter and pitch distance determines the dispersion profile of the PCF. The schematic of the experimental setup is shown in Fig. 1(b). 130 fs pulses were launched from a turn-key, compact Nd:glass laser (IC-150, High Q) into a 0.8 m length of PCF, which was then spliced to a 0.5 m length of conventional single-mode fiber (HI1060). HI 1060 is used to simplify coupling to the interferometer. The output of the pump laser centered at 1.059 μm was passed through a half-wave plate (HWP) and was focused to the PCF core by an aspheric lens (C230, Thorlabs). About 70% of the pump energy was coupled into the PCF, corresponding to an average power of 100 mW at 75MHZ repetition rate. Although the PCF was designed as non-polarization maintaining, we observed appreciable birefringence, presumably caused by the irregularity of the air holes and pitches of the PCF. The HWP adjusted the polarization of the light launched into the PCF to optimize the SC generation for OCT.
The SC spectra measured experimentally and generated by numerical simulation are compared in Fig. 2. After propagation in 0.8 m of PCF, the spectrum has been broadened to span from 0.8 µm to 1.3 µm with a full-width at half-maximum (FWHM) bandwidth of 240 nm centered around 1.15 µm (black solid line, Fig. 2(a)), when optimized with the polarization control. The spectrum of the pump pulse is also shown in Fig. 2(a) (dotted blue line). Fig. 2(b) presents the numerical simulations of the spectra generated using the nonlinear Schrödinger equation. The simulation agrees well with the experimental results. According to the simulation, the SC generation can be primarily attributed to SPM because SC is generated only in the normal dispersion regime. However, including stimulated Raman scattering in the simulation reproduces the small modulations observed in the experiments (green solid line). The low-dispersion peak of the PCF is at around 0.95 µm, and it is clear in Fig. 2 that the pump energy is distributed to bands on either side of 0.95µm. However, the majority of the energy is distributed to the longer-wavelength side. This is attributable to the asymmetric dispersion profile. Due to the flat and low dispersion area around 1µm, the spectrum first extends very quickly to both the short and long wavelength sides. However, the slope of the dispersion profile on the long wavelength side is less than that on short wavelength side. This permits the spectrum to be more easily extended to the longer wavelength side than to the shorter wavelength side. Therefore, most of the spectral energy concentrates on the long wavelength side and the center wavelength shifts from 1.059µm to about 1.15µm.
3. OCT imaging system and results

To demonstrate ultra-high resolution OCT imaging using this light source, the SC light output from the fiber was coupled into a free space time-domain OCT as shown in Fig. 1(b). The average power out of the fiber was about 40 mW due to 4 dB splicing loss between the PCF and the H1060. An achromatic lens with 45 mm focal length was used to focus the light onto the sample, resulting in approximately 7 μm lateral resolution. Scanning optical delay in the reference arm is realized by using a retroreflector mounted on a galvanometer. Dispersion compensation was achieved with two pairs of prisms made of SF11 and Lakn22. The axial PSF of the system was measured by recording the interferogram with a mirror as the sample and is plotted in Fig. 3(a). The measured axial resolution was about 2.8 μm in air, (corresponding to approximately 2.1 μm in the tissue). The inverse Fourier transform of the PSF is shown in Fig. 2(a) (red dashed line) to match very closely the spectrum of the SC input to the interferometer, showing that the interferometer did not significantly filter the SC light. The shorter wavelengths were cut off at around 0.95 μm due to the responsivity of the InGaAs photodetector which eliminates the requirement of a long pass spectral filter. The slight asymmetry of the PSF was due to residual unmatched higher order dispersion. As shown in the logarithmic plot shown in Fig. 3(b), the PSF is very clean down to -40 dB. Two small side lobes are observed 6 μm aside and -35dB down from the main peak, caused by small, low frequency modulations on the spectrum. This PSF performance represents a significant improvement over previously reported UHR-OCT light sources near this wavelength range using 1 μm femtosecond laser [9]. Fig. 3(b) was scaled to reflect 60 dB sample arm attenuation. By considering the 135 dBc/Hz relative intensity noise of the laser at carrier frequency of 300 KHz [6], 100 KHz signal detection bandwidth, 8 mW average light power...
on the sample and 0.1 mW power to the detector from the reference arm, the expected sensitivity was calculated to be around 103 dB with single detector. This is plotted as the dashed line in Fig. 3(b), and corresponds closely to the measured noise floor (solid line) confirming that there was no appreciable noise amplification during the SC generation based on SPM[6]. This is expected for SC based on SPM but not for SC generated by pumping around ZDWs with conventional PCF [7, 12]. The theoretical expected shot noise floor of 113 dB (dotted line) is also plotted in Fig. 3(b) for comparison. This performance could be approached by using balanced detection.

An in vivo ultrahigh-axial resolution OCT image of human skin and nail bed is demonstrated in Fig. 4(a). Stratum corneum, epidermis, nail pad, and blood vessels can be identified. The penetration depth is comparable to OCT imaging with 1300 nm light. An ultrahigh-resolution in vitro OCT image of de-epithelialized porcine cornea is demonstrated in Fig. 4(b), where Bowman’s layer, characterized by denser collagen lamellae, Descemet’s membrane and intrastromal morphology (for example, collagen lamellae) can be clearly identified. No obvious side lobes are observed in the images even at the specular reflection surface.
4. Discussion and conclusion

The SC spectrum demonstrated here shows an obvious shift of the center wavelength from the pump wavelength 1.059 μm toward 1.15 μm. This shift is not observed in previously-described SC based on SPM[6, 8, 10, 11] It is worth discussing if this shift might be advantageous or not. This is not comparative study, but previously published results [2][3] give evidences that a center wavelength at 1.15 μm may be a good choice for UHR_OCT. For ultra-high axial resolution OCT, such as we describe here, around 3 μm, the ideal axial resolution can not be maintained in tissue due to absorption, scattering and dispersion. In reference [3], Hillman et al numerically studied the effects of water dispersion and absorption in UHR-OCT by assuming that Gaussian broad-band light sources with various center wavelengths doubly pass through 1 mm of water. According to their results, at 3 μm axial resolution, if dispersion has been fully compensated, the axial resolution will not be affected by absorption for sources with center wavelengths as long as 1.2 μm. Even with no dispersion compensation, Hillman et al predict a flat area of least resolution degradation from 0.98 μm to
1.15μm. For opaque tissue, scattering also has to be considered. In reference [2], Sainter et al studied three representative opaque tissues and concluded that better light penetration depth can be achieved when the wavelength is above 1μm, except the area around 1.44μm due to strong absorption. From these results, we can expect that the optimal center wavelength range for UHR-OCT in opaque tissue is 1μm to 1.15μm. Choosing 1.15μm may provide other benefits. First, with longer wavelength, we probably can achieve deeper image penetration. Second, the effective responsivity characteristic of InGaAs detectors starts from 0.9μm. By shifting the center wavelength to 1.15μm, the light source can cover the range from 0.9μm to 1.4μm which overlaps with the effective range of the InGaAs detector, limited by the absorption peak at 1.44μm. Third, a light source centered at 1.15μm is probably appropriate for both ophthalmology and opaque tissue.

In conclusion, using a compact and commercially available 1.059μm femtosecond laser and a novel PCF, we have developed a new light source for UHR-OCT. Compared with previous SC based light sources at above 1μm wavelength range with conventional PCF, it has a smoother spectrum resulting in a very clean PSF with negligible sidelobes and no noise amplification. The axial resolution, about 2.8μm (in air), is almost two times improved over demonstrated results of SC generated with conventional single mode fiber. We have demonstrated OCT imaging with ~2.1μm axial resolution (in tissue). The SC generation is mainly attributed to SPM because the PCF has no negative dispersion area and the dispersion profile is close to zero dispersion at the pump wavelength. The PCF is commercially available and no tapering or further modification of the fiber is required. The spectral shift from the pump wavelength to 1.15μm provides an effective overlap with the responsivity of InGaAs detectors as well as good penetration depth in opaque tissue. The performance of this light source suggests that PCF with this type of convex, normal dispersion profile is more suitable for SC generation than conventional PCF or single mode fiber for UHR-OCT. Recently, femtosecond lasers emitting at around 1μm have been widely developed[19] including cheap, compact fiber lasers[20]. Availability of femtosecond fiber lasers such as these will make this light source very attractive for opaque-tissue imaging applications of UHR-OCT in the clinical setting where portability is important.

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