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Low-noise broadband light generation from optical fibers for use in high-resolution optical coherence tomography

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Broadband light generation from a single-mode optical fiber was developed for high-resolution optical coherence tomography (OCT). No noise amplification was observed for light broadened by self-phase modulation. The investigation showed that the intensity noise of light broadened by self-phase modulation in a single-mode optical fiber was much lower than that of continuum light from a microstructure fiber (MSF). The spectral width of a femtosecond input laser pulse was successfully broadened by a factor of 11, and a coherence length of 3.7 μm was achieved with this source. The application of light broadened by a single-mode optical fiber and MSF was compared for use in OCT imaging. The results showed that a single-mode fiber with a small core diameter is a useful way to generate low-noise, broadband light for high-resolution OCT imaging. © 2005 Optical Society of America

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1. INTRODUCTION

Optical coherence tomography (OCT) can be used to perform *in vivo*, high-resolution cross-sectional imaging of microstructures in transparent as well as turbid biological tissues.^{1,2} Longitudinal resolution, governed by the source coherence length, is inversely proportional to the light source bandwidth. Superluminescent diodes (SLDs) are often used for OCT imaging and typically have 10–15 μm longitudinal resolution.³ However, achieving fast imaging speed with a high signal-to-noise ratio (SNR) requires more than the milliwatt-level power typically available from SLDs. Additionally, the limited optical bandwidth of SLDs precludes imaging on a cellular level.

A Kerr-lens mode-locked Ti:sapphire laser, optimized for a short coherence length, was demonstrated to achieve sub-2- μm longitudinal resolution.⁴ Although cellular resolution imaging was obtained, this source is not widely available due to its inherent complexity. Moreover, the wavelength-dependent penetration depth and group-velocity dispersion (GVD) in biological tissues require OCT light sources at wavelengths, such as 1.3⁵ and 1.0 μm ,⁶ different from that of the Ti:sapphire laser.

Recently, there has been great interest in broadband light generation from nonlinear optical fibers. With a highly nonlinear material-doped optical fiber, high-resolution OCT was achieved by broadband light generation.^{7,8} With stimulated Raman scattering (SRS), broadband light was generated at a wavelength of 1.55 μm using erbium-doped fibers.⁹ However, the broadband SRS output had a white amplitude noise. Highly nonlinear microstructure fibers (MSFs) can generate an

extremely broadband continuum light spectrum from the visible (800 nm) to the near infrared (1.6 μm) by using low-energy femtosecond (fs) pulses.¹⁰ This is achieved because of the waveguide dispersion characteristics of the fibers, which shift the zero-dispersion wavelength to shorter wavelengths, and the small core diameters, which provide tight mode confinement. An OCT system with a longitudinal resolution of 2.5 μm using a MSF source has been reported.¹¹ However, continuum light from a MSF exhibits a highly complicated structure that is extremely sensitive to input pulse energy, which results in a poor SNR.^{12,13} This sensitivity to input fluctuations could limit the potential utility of MSFs for imaging applications such as OCT.

In a low-coherence interferometer, reflection sensitivity is limited by the intensity noise associated with low-coherence sources. In OCT, the SNR can be expressed as¹⁴

$$\text{SNR} = \frac{2R^2 P_{\text{ref}} P_{\text{dut}}}{4kT\Delta f/R_{\text{eff}} + 2qRP_{\text{ref}}\Delta f + (\text{RIN})R^2 P_{\text{ref}}^2 \Delta f}, \quad (1)$$

where R_{eff} and Δf are the effective noise resistance and detection bandwidth, respectively, of the receiver. P_{dut} and P_{ref} are the detected powers from the sample and reference arms, and R is the responsivity of the photodetector. The three terms in the denominator represent the receiver noise, shot noise, and relative intensity noise (RIN) contributions. Equation (1) shows that the SNR of OCT is inversely proportional to the source RIN and the detection bandwidth Δf . In OCT, the detected signal bandwidth is proportional to the image acquisition rate and source bandwidth. Improved OCT imaging speed and resolution

will increase Δf and, conversely, decrease the system SNR. Therefore, to obtain high-resolution, fast-speed OCT, a low-RIN light source is required to achieve a high SNR. Herein, we present our investigation of broadband light generation for high-resolution OCT by using self-phase modulation (SPM) in a single-mode optical fiber. Our investigation shows that the intensity noise level of light broadened by SPM in fibers is much lower than that of the light from a MSF. A longitudinal resolution of $3.7 \mu\text{m}$ is achieved with this source.

2. ANALYSIS

The evolution of an optical pulse in a single-mode fiber can be described by the dimensionless nonlinear Schrödinger equation,¹⁵ which includes the effects of SPM and GVD:

$$i \frac{\partial V}{\partial (z/z_0)} = \frac{\pi}{4} \left[\frac{\partial^2 V}{\partial (t/t_0)^2} + 2|V|^2 V \right], \quad (2)$$

where the complex variable V is pulse amplitude. High-order nonlinear terms, such as SRS, are not considered in Eq. (2). The time variable t is a retarded time and is defined such that, for any distance z along the fiber, the center of the pulses is at $t=0$. The normalized length z_0 is defined by

$$z_0 = 0.322 \frac{\pi^2 c^2 \tau_0^2}{|D(\lambda)|\lambda}, \quad (3)$$

where τ_0 is the intensity FWHM of the input pulse, $D(\lambda)$ is the GVD in dimensionless units, c is the velocity of light, and λ is the vacuum wavelength. Numerical solutions of Eq. (2) are obtained and shown in Fig. 1 for the pulse output spectrum after light propagation through a single-mode optical fiber. The input pulse duration is 100 fs. After the output laser pulse passes through an 80 cm long optical fiber, the pulse duration is broadened to 5 ps.

As the pulse propagates through the fiber, the fiber refractive index n is related to light intensity I owing to the Kerr effect, which can be expressed as $n = n_0 + n_2 I$. The nonlinear refractive index n_2 will introduce an additional phase shift $\Delta\phi$ to the input light, which can be expressed as $\Delta\phi = 2\pi n_2 I l / \lambda_0$, where l is the fiber length. Thus, for ul-

trashort pulses passing through a nonlinear optical fiber, the rapid intensity variation dI/dt will introduce a rapid phase shift:

$$\frac{\partial(\Delta\phi)}{\partial t} = \frac{2\pi}{\lambda_0} n_2 l \frac{\partial I}{\partial t}, \quad (4)$$

which is equivalent to a frequency shift with the expression $\delta\omega = -\partial(\Delta\phi)/\partial t$. Thus, the steep leading and trailing edges of the laser pulse generate additional frequencies through the nonlinear index of the fiber.¹⁶ The resulting broadening of the spectral width can be seen in Fig. 1, where the solid and dashed curves show the output and input spectra, respectively. Because the new frequencies are primarily generated at the leading and trailing edges of the laser pulse, the output spectrum is symmetric. It can be seen that the spectrum broadened by SPM was smooth compared with the continuum light generated from MSF as shown in Fig. 3 in Ref. 13.

Noise in the wideband light from an optical fiber can be characterized by the RIN measured on a photodetector, which is related to the fluctuations of the input pulse energy. To the first order, the RIN of light from an optical fiber can be written as the product of the laser RIN and a nonlinear amplification factor.¹² The smooth spectrum structure in Fig. 1 indicates that the output spectrum broadened by SPM has low spectrum fluctuations and that the intensity noise level of the input laser would not be significantly amplified.

3. EXPERIMENT

In our experiment, a single-mode optical fiber (F-SPV, Newport) was employed for the investigation of broadband light generation. The fiber length was 70 cm with a mode field diameter of $3.2 \mu\text{m}$. With a small core diameter, light can be confined in a small area for high light intensity, which increases nonlinear broadening. The pumping source for continuum generation was a self-mode-locked Ti:sapphire laser. The total output power of the laser was greater than 700 mW, with a pulse duration of 110 fs and repetition rate of 76 MHz. The laser output wavelength was 803 nm. The laser beam was then coupled into the single-mode fiber after it passed through a Faraday isolator to prevent interference of backreflected light with the mode locking, as shown in Fig. 2. The focus length of the coupling lens L_1 was 6 mm, and the beam diameter was 1.0 mm before L_1 .

The spectrum of light was broadened when the light propagated through the fiber because of SPM. The broadband output of light was collimated by a 4.5 mm focal-length lens L_2 . The output spectrum was broadened gradually as the laser intensity increased. In Fig. 3(a) the output spectrum width varied as the pump power increased. It can be seen that increasing the pump power enhances broadband light generation. When the input power was 700 mW, the output light spectrum was broadened to a FWHM of 90 nm. Broadband light generation from 730 to 880 nm was observed from the fiber, as shown in Fig. 3(b) by the solid line. The central wavelength was 809 nm. The dashed curve in Fig. 3(b) shows the input spectrum with a FWHM of 8 nm. Note that the output

Fig. 1. Input and output spectra of an ultrashort laser pulse broadened by SPM. The laser wavelength is 800 nm, pulse duration is 100 fs, and pulse energy is 5 nJ. The fiber length is 80 cm with a mode field diameter of $3.3 \mu\text{m}$. The fiber GVD is $30 \text{ fs}^2/\text{mm}$.

Fig. 2. Diagram of broadband light generation through optical fiber. WP, waveplate; PD, photodetector; OSA, optical spectrum analyzer; ESA, electrical spectrum analyzer; AC, autocorrelator; OF, optical fiber; AT, attenuator; L_1 , L_2 , optical lenses; M_1 , M_2 , mirrors.

Fig. 3. (a) Output spectrum width with increasing of pump power. (b) Input-output spectra; the dashed and solid curves, the input and output spectra, respectively.

spectrum was successfully broadened by a factor of 11. The total output power from the optical fiber can be as high as 350 mW. The influence of fiber length on broadband light generation was also investigated. In this experiment, a different type of small-core optical fiber was used (FS-PM-4611). The output spectra for different fiber lengths were recorded, as shown in Fig. 4. When the fiber length was shorter than 53 cm, the output spectrum width was gradually broadened with increasing fiber length. When the fiber length was increased from 53 to 200 cm, the spectrum width did not increase further. However, the spectrum became smooth.

The intensity noise of the broadband light was measured by an electrical spectrum analyzer (HP 8560E), as

shown in Fig. 5(a) by the solid curve. The dashed curve in Fig. 5(a) shows the intensity noise of the pump laser. A neutral-density attenuator was used to keep the same 20 mW light power onto the photodetector without focusing for both measurements. The bandwidth of our detector was 200 kHz (New Focus, 2001). Note that the noise level of light from the optical fiber is identical to that from the laser. The measured intensity noises of the laser and broadened light from the fiber were -112 and -114 dBm/Hz at 100 kHz, respectively. No noise amplification was observed. This is in contrast to the supercontinuum generation from MSF, which is extremely sensitive to initial pulse energy.^{13,17} An increase of 0.1% in the pulse energy can dramatically change the fine structure of the output spectrum from a MSF. In our previous experiment, we obtained continuum light generation from 400 to 1400 nm using a MSF.¹⁸ In order to compare the noise between a single-mode fiber and MSF, we also measured the noise level of continuum light from a MSF as shown in Fig. 5(b). The solid curve shows the intensity noise from the MSF. The dashed curve shows the intensity noise from the pump laser. An attenuator was used to keep the same 20 mW light power from both the MSF and the pump laser for the measurements. Note that there is white amplitude noise at a level about 10 dB higher than that of the pump laser. The measured intensity noise of light from the MSF was -102 dBm/Hz at 100 kHz. This increase resulted from nonlinear amplification of the intensity noise of the input light by the MSF¹² due to complex nonlinear procedures that included SRS, four-wave mixing, and SPM. As light power onto the photodetector increased, we observed an increase in the intensity noise. Thus, increasing the light power from the MSF will not increase the SNR of the OCT system. However, for light broadened by the SPM in a single-mode fiber, no noise amplification was observed with different pump powers.

To demonstrate the application of broadband light from a single-mode fiber for high-resolution OCT, we measured the coherence function of the broadened spectrum shown

Fig. 4. Output spectra for different fiber lengths. (a) 23 cm, (b) 33 cm, (c) 53 cm, (d) 200 cm. This fiber is different from the small-core fiber for Fig. 3.

Fig. 5. Intensity noise of light from a (a) single-mode fiber, (b) MSF.

in Fig. 3(b). To overcome the bandpass limitation in fiber-based OCT systems, we designed and constructed an open-air, high-resolution OCT system as shown in Fig. 6. A pair of prisms made of fused silica glass was chosen to balance the GVD of the two arms of the interferometer. A neutral-density filter was inserted into the reference arm to decrease the intensity and reduce background noise. Both an objective lens and a rectangular prism were mounted on a voice coil translation stage to track dynamic focusing during longitudinal scanning to keep a zero path-length difference in the focus region between the reference and sampling arms.¹⁹ This made it possible to increase imaging depth with constant high lateral resolution. The baseplate was mounted on a stage for lateral scanning. In the experiment, the high-resolution OCT system was optimized to support a $3.7\text{ }\mu\text{m}$ longitudinal resolution with broadband light from optical fiber, as shown in Fig. 7(a). Considering the refractive index of tissue (1.35), the corresponding resolution is $2.7\text{ }\mu\text{m}$ in tissue. The FWHM of the coherent function of the input light with a spectrum shown as the dashed curve in Fig.

3(b) was about $35\text{ }\mu\text{m}$, as shown in Fig. 7(b). Thus, OCT resolution was enhanced.

With a fast Fourier transform (FFT) of the interference fringe, the dynamic range of the OCT system can be analyzed. In Fig. 8, the dynamic range of the interference fringe in Fig. 7(a) was analyzed by FFT. The result showed that the dynamic range of the OCT system with broadband light broadened by SPM was about 50 dB. In Fig. 5, we showed that the light from a MSF has a high intensity noise. To compare the influence of noise from a SPM light source with a MSF source on OCT, continuum light from a MSF was generated. The pumping source for the MSF was the same Ti:sapphire laser used in the first experiment. The total output laser power was about 700 mW, with a pulse duration of 110 fs. To get continuum light generation from a MSF fiber, the pump laser wavelength must be set near the zero-dispersion wavelength of the MSF. Thus, the laser output wavelength was adjusted to 780 nm, which was near the zero-dispersion wavelength of our MSF. The laser beam was then coupled into the MSF after it passed through the Faraday isolator

and $\lambda/2$ wave plate. The focal length of the coupling lens L_1 was 6.0 mm. A half-wave plate after the Faraday isolator was used to adjust the polarization state of the light input to the fiber to optimize the spectrum. Continuum light generation from 400 to 1400 nm was observed from

Fig. 6. Schematic of the ultrahigh-resolution OCT system: M_3 , wideband cube beam splitter; M_1 , M_2 , M_4 , and M_5 , silver mirrors; F_1 , long-pass filter; L_1 , coupling lens; L_2 , collimated lens; L_3 , objective lens; P , rectangular prism; D , InGaAs detector; F_2 , neutral-density filter; OF , optical fiber; FS , fused-silica prisms.

Fig. 7. (a) Interference fringe of light broadened by SPM. (b) Interference fringe of the input laser.

Fig. 8. FFT of the interference fringe in Fig. 7(a).

Fig. 9. Broadband light generation from a MSF after a long-pass filter.

the fiber. A long-pass filter F_1 was used to block all light with wavelengths shorter than 800 nm. The output spectrum ranging from 800 to 1430 nm from the fiber is shown in Fig. 9 after a long-pass filter. The total output power from the MSF could be as high as 100 mW, and the remaining power was approximately 50 mW after the filter. In the experiment, the ultrahigh-resolution OCT system was optimized to support a $1.8 \mu\text{m}$ longitudinal resolution in free space at a center wavelength of $1.1 \mu\text{m}$. Figure 10(a) shows the interference fringe with a silver mirror as the sample. Considering the refractive index of tissue, the corresponding resolution is $1.3 \mu\text{m}$ in tissue. Because the spectrum from the MSF is not perfectly Gaussian shaped, sidelobes in the interferometric autocorrelation are visible. The dynamic range of the interference fringe is analyzed by the FFT and shown in Fig. 10(b). It can be seen that the dynamic range is about 35 dB, which is 15 dB lower than the SPM source dynamic range as shown in Fig. 8. Therefore, with a MSF light source, the OCT system dynamic range is decreased greatly compared with that from a SPM light source because of its high intensity noise, although it has higher axial resolution.

The lateral resolution of our OCT system was $4 \mu\text{m}$, which was determined by the achromatic lens. The feasibility of high-resolution imaging was demonstrated in biological tissues by using a New Zealand white rabbit eye.

Figure 11(a) shows an *in vitro* OCT image of the rabbit cornea with the SPM light source shown in Fig. 3(b). The incident power of the system was 20 mW. The voice coil stage was scanned at a speed of 20 mm/s at a repetition rate of 5 Hz. The image size is $1.3 \text{ mm} \times 1.0 \text{ mm}$ at $2.7 \mu\text{m} \times 4 \mu\text{m}$ (longitudinal $\times$ lateral) resolution. The calibrated thickness of the epithelium is $47 \mu\text{m}$. Figure 11(b) shows an *in vitro* OCT image of the rabbit cornea with the MSF light source shown in Fig. 9. The voice coil stage was scanned at a speed of 4 mm/s at a repetition rate of 1 Hz. The image size is $1.3 \text{ mm} \times 1.0 \text{ mm}$ at $1.3 \mu\text{m} \times 4 \mu\text{m}$ (longitudinal $\times$ lateral) resolution. Because the rabbit eye was kept in liquid over night, the corneal stroma was thicker than that shown in Fig. 11(a) as a result of tissue hydration. It can be seen from the image in Fig. 11(b) that there is a background noise distribution that is much higher than that of Fig. 11(a). This high background noise is due to the intensity noise of the MSF light source. With faster imaging speed, image quality was decreased. Therefore, with a MSF light source, OCT imaging speed is limited, and the system dynamic range is low even at a slower imaging speed.

The imaging abilities of SPM and MSF sources for highly scattering biological tissue were compared in Fig. 12. In Fig. 12(a), a human nail bed was imaged *in vivo* with broadband light from the single-mode fiber. The image size was $1.5 \text{ mm} \times 1.5 \text{ mm}$. The tissue structure can be seen clearly. In Fig. 12(b), the same sample was imaged by the continuum light from a MSF. The image size was $1.5 \text{ mm} \times 1.5 \text{ mm}$. Compared with Fig. 12(a), it can be seen that the area marked A in Fig. 12(b) is not as clear as that area in Fig. 12(a). It is hard to see the nail plate and nail matrix under the proximal nail fold. At the same time, the imaging depth of Fig. 12(b) is not as deep as in Fig. 12(a). In Fig. 12(b), the center wavelength of the light source was at $1.1 \mu\text{m}$. According to a previous report,⁵

Fig. 10. (a) Interference signal at the detector with a MSF light source. (b) FFT of the interference fringe as shown in (a).

Fig. 11. High-resolution OCT imaging of the rabbit cornea. Ep, epithelium; En, endothelium; CS, corneal stroma. (a) Imaging by a SPM light source. (b) Imaging by a MSF light source.

Fig. 12. High-resolution OCT imaging of a human nail bed. NP, nail plate; NM, nail matrix; PNF, proximal nail fold. (a) Imaging by a SPM light source. (b) Imaging by a MSF light source.

near-infrared light should penetrate more deeply into highly scattering tissue than 800 nm light. Thus, this experiment shows that a MSF source is not suitable for OCT imaging of highly scattering tissue because of its high intensity noise.

Researchers routinely use dual-balanced detection to reduce the influence of the source intensity noise to get high system SNR. This technique has also been used for the high-resolution OCT system with a MSF source. With balance detection, the reported OCT system sensitivity was 94 dB with longitudinal resolution of 2 μm with a MSF source.¹¹ However, with a broadband self-mode-locked laser as the light source, the OCT system sensitivity can achieve 110 dB with resolution of $\sim 1 \mu\text{m}$ with balance detection.⁴ Both of these results were reported by the same group at MIT with the same OCT configuration. It can be seen that with a MSF and a fs laser source, there is a system SNR difference of 16 dB, which coincides with the result measured in this paper. The dynamic range shown in Fig. 8 with a SPM source is 15 dB higher than that with a MSF source as shown in Fig. 10(b). In our experiment, we observed that the intensity noise of a SPM source is the same as that of the fs laser. Therefore, with balance detection, the decrease in SNR resulting from the high intensity noise from MSF cannot be reversed.

To get continuum light generation from a MSF fiber, the pump laser wavelength must be set close to the zero-dispersion wavelength of the MSF. At the same time, the output spectrum is very sensitive to the input polarization state. The stability of the MSF source is limited because of the special requirement of the pump wavelength and polarization state. However, with a SPM source, there is no special requirement regarding the pump wavelength and polarization state. It is easy to generate a broadband light with a stable spectrum for OCT application. Moreover, with a different fs pump laser, SPM has the flexibility to produce broadband light sources at different wavelengths. For example, with a fs 1.3 μm laser pulse to pump a piece of dispersion-shifted fiber, a wide-band light spectrum with a FWHM of 120 nm was generated at 1.3 μm by SPM several years ago.²⁰ The system resolution was improved to 6 μm , and the measured SNR of the OCT system was 115 dB with this SPM source.

4. SUMMARY

In summary, broadband light generation from a single-mode optical fiber was evaluated. No noise amplification was observed for light broadened by SPM in a single-mode optical fiber. The investigation showed that the intensity noise of light broadened by SPM in a single-mode fiber was much lower than that of light from a MSF. The coherence length of a fs laser was successfully shortened from 35 to 3.7 μm after propagation through a small-core-diameter optical fiber. High-resolution OCT imaging was demonstrated. The imaging ability of light broadened by a single-mode fiber and a MSF were compared. The results show that the application of a MSF is limited for highly scattering biological tissue. It is also not suitable for high-speed imaging applications because of its high intensity noise, although it has high axial resolution. A comparative study shows that a low-intensity-noise light

source is very important for fast-speed, high-resolution OCT. Our investigation showed that a single-mode fiber with a small core diameter is a useful way to significantly broaden relatively low-bandwidth pulses from a fs laser to get low-noise, reliable broadband light for high-resolution OCT imaging.

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