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Photoacoustic wavefront shaping with a long coherence length laser

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ABSTRACT

Photoacoustic (PA) wavefront shaping (WS; PAWS) could allow focusing light deep in biological tissue. This could enable increasing the penetration depth of biomedical optical techniques including PA imaging. However, focussing at depth requires a light source of long coherence length (CL), presenting a challenge because the CLs of typical PA excitation lasers are short. To address this challenge, we developed a PAWS system based on an externally modulated external cavity laser with a long CL. The system was demonstrated by focussing light through rigid scattering media using both PAWS and optical WS. PAWS enabled focussing through diffusers with 8 × enhancements, while all-optical WS enabled focussing through various scattering media including a 5.8 mm thick tissue phantom. By enabling PAWS with increased coherence, the system could facilitate exploring the practical depth limits of PAWS, paving the way to focussing light deep in tissue.

Keywords: Photoacoustic, imaging, wavefront shaping

1. INTRODUCTION

The scattering of light in living tissue limits the capabilities of biomedical optics techniques. By destroying spatial coherence, scattering limits the penetration depth of optical microscopy. By attenuating the fluence, scattering also limits the penetration depth of techniques based on diffuse light including PA tomography. Overcoming scattering could increase the penetration depth and resolution of these techniques, significantly broadening their applicability.

One method that could compensate for the deleterious effects of scattering is optical wavefront shaping (WS) [1]. The principle of WS is that, by spatially patterning the light incident on a scattering medium, one can control the interference patterns produced inside the medium. In principle, this allows focussing light deep in tissue, increasing the fluence at depth. The challenge is discovering the optimal incident pattern. This typically requires making measurements of the light intensity at the target location [2], providing a so-called "guidestar" [2]. Focussing can then be achieved by tuning the structure of the incident light so as to maximize the guidestar signal [3]. In PAWS, the guidestar is a PA signal generated at the target location; the light field is optimized so as to maximize the amplitude of that PA signal.

PAWS studies have demonstrated the feasibility of focussing light through thin scattering media [4], [5]. However, focusing deep in tissue presents a number of additional challenges including those related to: (i) exerting enough control over the light field to significantly change the fluence; (ii) obtaining enough sensitivity to detect how the fluence changes; and (iii) shaping the light field quickly enough to avoid losing the focus to decorrelation. Perhaps most importantly though, the PAWS system must also produce light that can be controlled. For this to be the case, the light must be sufficiently temporally coherent to produce stable, high-contrast interference (speckle) patterns in the region of interest. Because scattering takes light along various (extended) paths, retaining temporal coherence at depth in scattering media requires a CL longer than the penetration depth. For example, assuming relatively weakly scattering tissue, to enable focussing at multi-cm penetration depths, a ~1 m scale CL would be needed [6], [7]. For focussing at a 10 cm depth, a >10 m CL would be needed [6]. This presents a challenge because conventional PA excitation lasers typically have CLs of a few millimetres. To address this challenge, we developed a PAWS system based on an amplified external cavity laser with a much longer CL. The system was demonstrated by focussing light through scattering media using PAWS and conventional optical WS.

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2. PAWS SYSTEM

Figure 1 shows a simplified diagram of the PAWS system. A laser beam is provided by a wavelength tunable 1440-1640 nm continuous wave (CW) laser (Tunics T100S-HP, Yenista). This laser can provide a linewidth as low as 400 kHz[1], indicating a CL in excess of 200 m. The output of the laser is temporally modulated by an acousto-optic modulator (AOM) driven by a pulse generator. This provides pulses with lengths down to tens of nanoseconds, limited by the AOM response time. The modulated light is amplified by a 27 dBm erbium doped fibre amplified (EDFA; FA-27-IO-CP, Pritel). With a CW input, this EDFA provides output optical powers in the range 0-500 mW. With a pulsed input, the amplified pulse energy depends on factors including the pulse length and the pulse repetition frequency (PRF). For example, for a pulse length of 75 ns and a PRF of 10 kHz, 1.8 µJ pulses can be produced. After amplification, the beam is spatially structured by a spatial light modulator (SLM; HSP1920-600-1300, Meadowlark), then transmitted through an optical diffuser (DG120, Thorlabs). A portion of the forward scattered light is transmitted via a beam-splitter to a light absorbing layer. This excites ultrasound waves, which are detected by a 5 MHz focussing piezoelectric ultrasonic transducer (UST; V310-N-SU, Panametrics NDT). The detected waves are amplified and digitized by a data acquisition (DAQ) system comprising a pulser-receiver (5072PR, Olympus) and an oscilloscope connected to a PC. Images of the field in the plane of the absorber are recorded by an infrared camera (C10633, Hamamatsu) positioned co-planar with the absorbing layer.

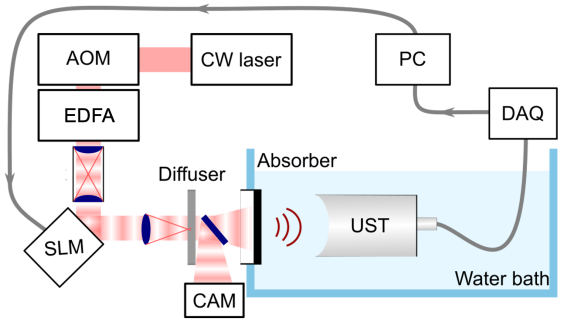

Figure 1. Diagram of the PAWS system. To simplify the diagram, passive optics are omitted including: a polarizer and half-wave plate prior to the SLM; lenses after the diffuser; neutral density filters on the camera.

3. OPTICAL WS

To investigate the focussing of light through scattering media, optical WS experiments similar to those in previous studies [1] were performed. In one experiment, the camera was re-positioned directly behind the diffuser. A random phase pattern was projected by the SLM so the field on the diffuser and the speckle pattern on the camera were both effectively random. The SLM pattern was then optimized via WS so as to maximize the intensity of a pixel of interest (POI) in the camera image. The POI overlapped with a single speckle grain. As such, the intensity of that speckle grain was optimized. The optimization was done using a genetic algorithm (GA) [8]. During the optimization, the intensity of the POI increased approximately linearly with increasing iterations. The POI then became saturated and the GA was terminated (after 55 iterations, each involving 30 measurements). The experiment lasted approximately 7 minutes. Images were recorded prior to and during WS.

The image acquired prior to WS was plotted in figure 2a. As expected, the image shows a seemingly random speckle pattern. The image obtained after WS is shown in figure 2b. In this image, the intensity of the POI has increased to produce a focus. The light intensity enhancement defined as the ratio between the optimized POI intensity and the average intensity before optimization [1], was estimated at $\eta \approx 30$. By comparison, the theoretical maximum was [3] $\eta_{th} = \alpha\, N/M \approx 800$, where: $\alpha = \pi/4$ is a constant accounting for the phase-only nature of the light field control [3], $N \approx 1000$ was the number of independent input modes (illuminated SLM super-pixels) and $M = 1$ was the number of independent output modes (individually resolvable speckles). The measured enhancement was lower than the theoretical maximum as expected given that the WS experiment was terminated due to saturation (not convergence). The experiment was repeated in different conditions involving focussing through different scattering media, including a 5.8 mm thick tissue phantom (not shown).

[1] The full width half maximum (FWHM) typical linewidth from the manufacturer's data sheet.

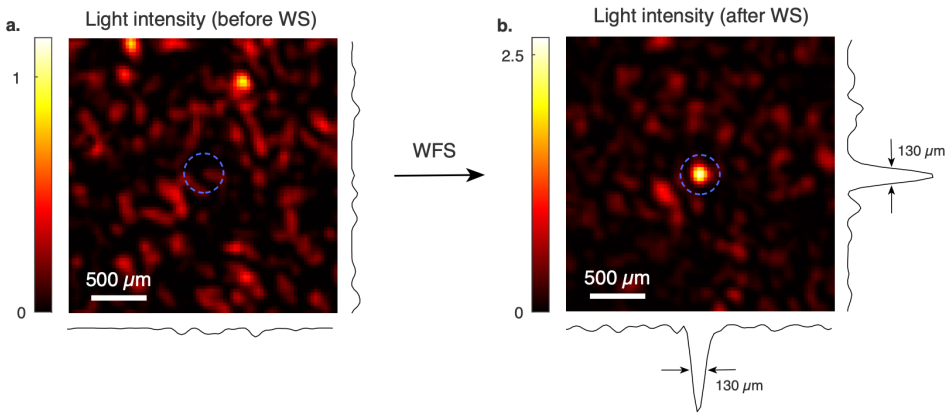

Figure 2. Images before and after optical WS. The POI is in the centre of the dashed blue circle. The plots below/next to the images are horizontal/vertical profiles through the POI. The images share a common arbitrary unit, selected to show the relative intensity (note the independent colorbars). The speckle size was ≈100 μm, while the pixel size was 30 μm, allowing individual speckles to be resolved. For this experiment, the laser was replaced with a different laser (ITLA, Pirelli). The system was operated in CW mode. The SLM elements were binned 32 by 32 times providing ≈ 2,000 super-pixels, half of which were illuminated.

4. PAWS

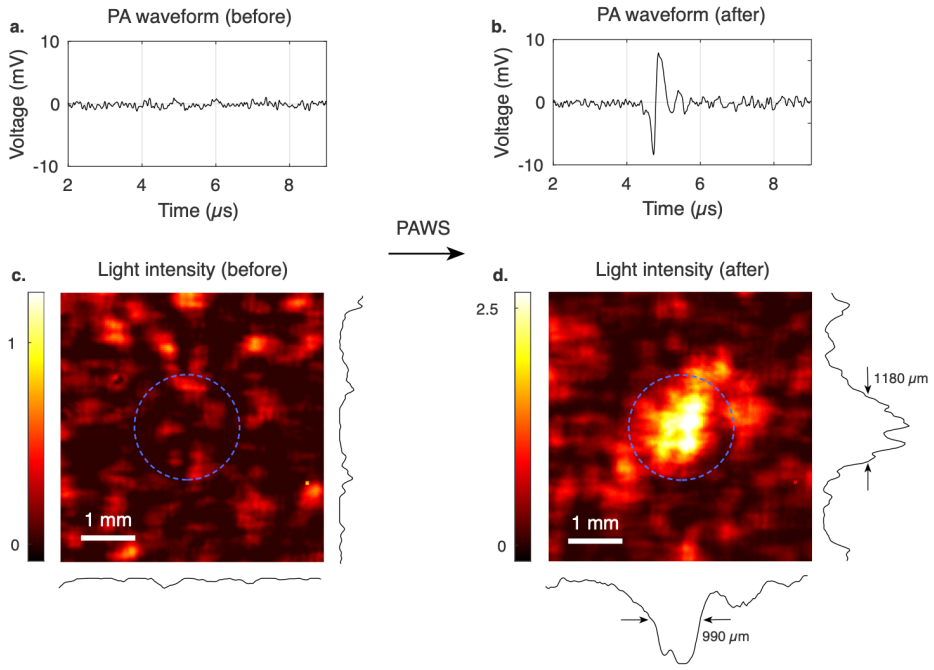

Figure 3. PA waveforms and optical images obtained before and after PAWS. The plots below/next to the images are horizontal/vertical profiles through the centre of the enhanced region. The speckle size was ~500 μm, approximately equal to the size of the ultrasonic focus. For this experiment, the light was pulsed, with a pulse length of 60 ns and a PRF of 10 kHz. The wavelength was 1540 nm. The SLM was set up as described in the caption of figure 2.

To investigate the focussing of light via PAWS, experiments similar to previous linear PAWS experiments [4], [9]–[11] were conducted using the system shown in figure 1. In one experiment, as above, a random pattern was projected by the SLM, such that the fields incident on the diffuser and absorber were effectively random. The SLM pattern was then optimized using a GA. However, unlike in the optical WS study, this time the GA sought to maximize the peak-to-peak amplitude of the detected PA waveforms. During the experiment, the peak-to-peak of the PA waveform increased steadily then reached an apparent plateau after 1000 iterations, whereupon the algorithm was terminated. The experiment lasted approximately 2 hours. Waveforms and images were recorded prior to and during PAWS.

The waveform obtained prior to PAWS was plotted in figure 3a. It shows no obvious PA signal above the noise floor. The root mean squared (RMS) of the waveform was ≈0.4 mV and the peak-to-peak, V_0, was 2.2 mV. The waveform obtained after PAWS was plotted in figure 3b. This waveform exhibits a bipolar signal with a peak-to-peak amplitude, V_1, of approximately 16 mV. The PA amplitude enhancement was thus estimated to be at least $\eta_{PA} = V_1/V_0 \approx 8$. This enhancement is significantly lower than the theoretical maximum enhancement, which was on the order of several hundred times as above. To see if the PA enhancement reflected an increased light intensity in the vicinity of the focus of the transducer, images of the light incident on the absorbing layer prior to and after PAWS were plotted in figures 3c-d. As expected, the image obtained prior to PAWS shows random speckle, while the image obtained after PAWS shows a region of enhanced intensity, presumed to coincide with the focal zone of the transducer.

5. DISCUSSION

A PAWS system was developed and demonstrated to enable focussing light through scattering media via PAWS and optical WS. Unlike previous PAWS systems, the system used an externally modulated external cavity laser. This laser has a long CL, offering the prospect of producing stable interference (speckle) patterns at significant depths in turbid media. Among other things, this could benefit experiments exploring the penetration depth limits of PAWS.

Based on the laser linewidth, the CL of the system could be as high as 240 m; long enough to produce speckle patterns through several tens of cm of tissue like media [6]. However, due to fundamental limits, the external pulsing of the laser limits the practically achievable CL to $<ct_p$, where c is the speed of light and t_p is the pulse length [7]. For 60 ns long pulses for example, the CL cannot exceed ~18 m, limiting the applicable depth in a tissue like media to around 10 cm [6]. It should also be noted that the coherence length of the whole system, as influenced by all its optical components, has yet to be experimentally measured. While no effects of limited coherence have been observed, thus far the system has only been tested in media of sub centimetre thickness.

Table 1. Reported PA enhancements in previous reported (linear) PAWS experiments.

Short reference (first author, year, journal, [ref])	**Maximum PA amplitude enhancement, η_{PA}**
Chaigne. 2014. Nature Photonics [5]	6
Chaigne. 2014. Optics Letters [9]	6
Dean-Ben. 2015. Optics Letters [12]	6
Zhao. 2021. Optics Letters [10]	7
Tzang. 2016. Optics Express [13]	9
Kong. 2011. Optics Letters [11]	10
Caravaca-Aguirre. 2013. Optics Express [14]	10
Chaigne. 2014. Optics Letters [15]	12
Tay. 2014. Optics Letters [16]	14
Lai. 2015. Nature Photonics [4]	60

The provided demonstrations showed the basic capability of the system to focus light through scattering media using WS and PAWS. The reported enhancements were relatively low. However, this was not unexpected. In the optical WS experiment, the enhancement was about 30 times. While lower than the theoretical maximum, this was limited by the finite dynamic range of the camera. To overcome this limit, the dynamic range of the measurements could be extended in a number of ways. For example, the laser power could be adapted throughout the measurement so as to keep the maximum pixel intensity just below saturation. In the PAWS experiment, the PA amplitude enhancement was estimated to be at least 8 times. Although seemingly low, this figure is comparable to the PA enhancements reported in a majority of previously reported linear PAWS experiments. For context, the PA amplitude enhancements reported in a number of PAWS studies

are summarized in table 1. All but one of these values are in the range 6-14, close the observed value. In this study, it is thought that the main factor limiting the enhancement was the sensitivity of the PAWS system. This might be increased by increasing the sensitivity of the ultrasonic detector or by increasing the pulse energy by using a higher power EDFA or varying the pulse parameters.

Although a useful system for PAWS research, the system is not suitable for experiments involving living tissue because the available wavelengths are too strongly attenuated by tissue. The system is thus suited to experiments involving other media whose optical properties can be tailored to these wavelengths. The system could be adapted however, by swapping the laser and amplifier for similar systems providing other wavelengths that are more tissue-compatible.

6. CONCLUSION

PAWS could allow focusing light deep in tissue by using PA signals to guide the structuring of coherent light. This could lead to fluence enhancements, enabling increasing the penetration depth of PA imaging and other techniques. However, focusing deep in scattering media – thus advancing the PAWS technique - requires a CL much longer than the CL of traditional PA light sources. To address this challenge, we developed a PAWS system based on an externally modulated external cavity laser with a very long CL. The system was demonstrated by focussing light through rigid scattering media using both PAWS and optical WS. PAWS enabled focussing through diffusers with 8 × enhancements, while optical WS enabled focussing through various media including a 5.8 mm thick tissue phantom. The system could be adapted to make it tissue-compatible by changing the working wavelengths. Higher enhancements might be unlocked by increasing the sensitivity. By enabling PAWS with practically unlimited coherence, the system could facilitate exploring the practical depth limits and other aspects of PAWS techniques, paving the way to focussing light deep in living tissue.

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