

Research article

A photoacoustic sensing probe using optical fiber acoustic delay line

Arif Kivanc Ustun*, Jun Zou

Department of Electrical and Computer Engineering, Texas A&M University, College Station, TX, 77843-3128, USA

ARTICLE INFO

Keywords:

Photoacoustic sensing
Medical probe design
Optical fiber
Acoustic delay line

ABSTRACT

In this paper, we report a new photoacoustic sensing probe design consisting of two optical fibers. One optical fiber is used for delivering the excitation light pulses. The other one serves as an acoustic delay line to relay the generated PA signal from the target to an outside ultrasound transducer. With the addition of suitable time delay, the original PA signal can be easily separated from the interference signals. To demonstrate this new design, a prototype probe was designed, fabricated and tested. The PA sensing performance was characterized with different concentration of black and red dye solutions. The testing results show that the PA sensing probe can provide good sensitivity and maintain high linearity over a wide range of concentrations. The detection of bovine blood embedded into chicken breast tissue was also conducted to demonstrate its potential usefulness for *in-vivo* applications.

1. Introduction

For biomedical applications, photoacoustics (PA) has become a useful technique that combines both rich optical absorption contrast and good acoustic penetration depth beyond optical diffraction limit [1,2]. Specific tissues in human body have greater optical absorption coefficients than tissues nearby, such as blood vessels, tumors, and lesions [3]. Capitalizing upon the optical absorption differences between close tissues, PA sensing and imaging can deal with many discriminative analyses among tissues [4], features such as melanin or hemoglobin [5], and even analytes in different states, gases and liquids [6]. Although better than conventional optical methods, the penetration depth of PA sensing and imaging in tissues is typically limited to a few centimeters by the maximal allowable laser fluence and the optical absorption and acoustic attenuation in tissues. As a result, it is still difficult to detect a target buried deeply inside human body with a good contrast.

In recent years, new PA sensing probe [7,8] or guided biopsy needles [9,10] have been developed to conduct localized measurements. Because the light delivery and acoustic detection are conducted near the target, both optical and acoustic divergence and attenuation inside the tissue can be minimized, which results in higher sensitivity. However, with the ultrasound transducer placed close to the target, the PA signal will arrive at the transducer within a very short period of time. Therefore, it could be easily buried under the much stronger interference signals occurring at each PA excitation. This situation makes the extraction, processing and interpretation of the PA data rather difficult.

For example, when the pulsed laser fires, the strong electrical pulse generated by the driver circuit could be easily coupled into the transducer-amplifier loop. In addition, after reflection and backscattering, part of the incident light could directly illuminate the optically-absorbing transducer surface and induce large PA response directly on the transducer.

To address this issue, we report a new PA sensing probe design using two optical fibers. Specifically, one optical fiber serves as the optical waveguide for delivering excitation laser pulses onto the target. The second optical fiber functions as an acoustic delay line to detect and transmit the generated PA signal from the target to an outside ultrasound transducer, while creating a desirable amount of acoustic time delay. With the added acoustic time delay, the PA signal will arrive at the transducer after the interference signals diminish and therefore can be easily distinguished in the time domain. Due to its high optical transparency, the direct PA response at the tip of the acoustic delay fiber can be minimized. As a result, a clean PA signal originating from the target can be obtained. Furthermore, with the transducer located outside, the PA probe consists of only two optical fibers placed closely to each other. A small probe diameter can be achieved to minimize its invasiveness for *in-vivo* use. For demonstration, a prototype probe was designed, fabricated, and tested with different concentration of dye solutions and in the biological tissues such as chicken breast tissues with bovine blood. The testing results show that the PA sensing probe can provide good sensitivity and maintain high linearity over a wide range of dye concentrations. To demonstrate its potential usefulness for *in-vivo* sensing applications, the detection of bovine blood embedded

* Corresponding author.

E-mail addresses: akustun87@tamu.edu (A.K. Ustun), junzou@tamu.edu (J. Zou).<https://doi.org/10.1016/j.pacs.2018.11.001>

Received 31 May 2018; Received in revised form 22 October 2018; Accepted 5 November 2018

Available online 11 November 2018

2213-5979/ © 2018 The Authors. Published by Elsevier GmbH. This is an open access article under the CC BY-NC-ND license (<http://creativecommons.org/licenses/by-nc-nd/4.0/>).

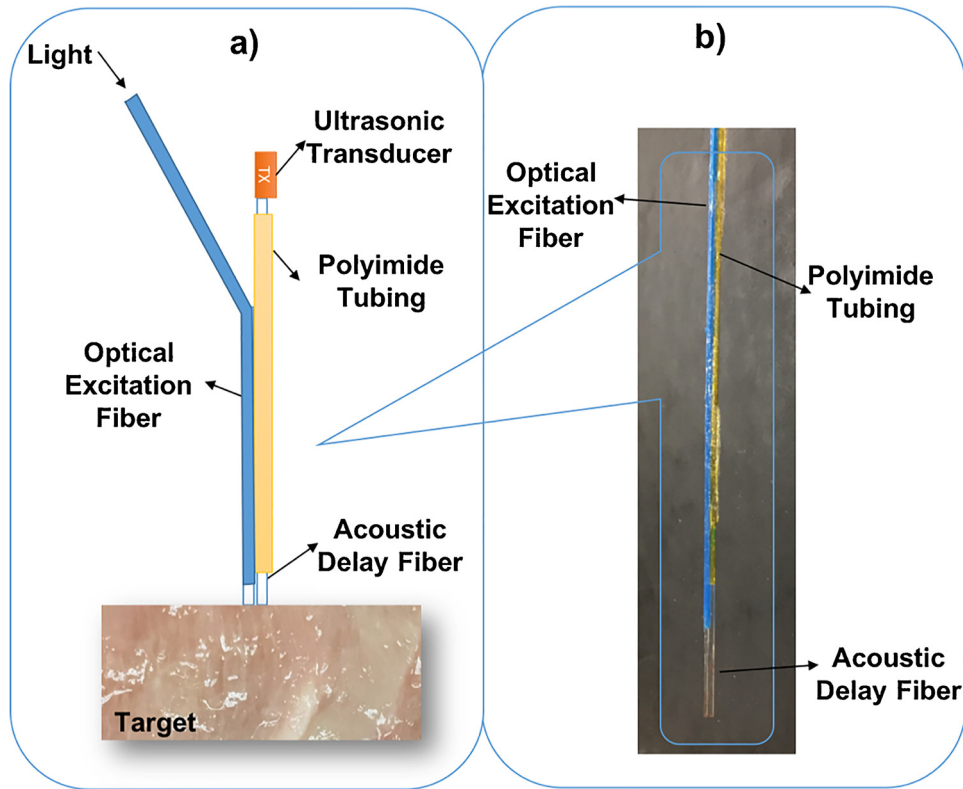


Fig. 1. a) Schematic of PA Sensing Probe Design and b) Picture of the Probe Shank with Glued Optical Excitation and Acoustic Delay Fibers.

into chicken breast tissue was also conducted. In addition to single-point sensing, the PA sensing probe can be used to conduct photoacoustic tomography without transducer arrays. When multiple PA sensing probes with different delay times are bundled into an array, the PA signals will reach the (single-element) transducer at different times and therefore can be received unambiguously in a time-serial manner. 2D or 3D image reconstruction can be performed based on the received PA signals [11].

2. Experimental procedure

2.1. Probe design and construction

Fig. 1a shows the schematic design of the PA sensing probe. Two optical fibers are laid side by side along the probe shank. The optical excitation fiber is used to send excitation laser pulses to the target. The acoustic delay fiber is used to receive and transmit the generated PA signal from the target to the ultrasound transducer with certain time delay, which is determined by the length and the sound velocity of the acoustic delay fiber. The acoustic delay should be made longer than the duration of the electrical pulse during excitation, such that the real PA signal from the target can be completely distinguished. Both the optical excitation and acoustic delay fibers are made of fused silica multimode fibers (FT400UMT, THORLABS, Newton, NJ) with a core diameter of 400 μm , a clad layer of 12.5 μm , and an overall diameter of 730 μm . The 400- μm core diameter allows the transmission of $\mu\text{J} \sim \text{mJ}$ laser pulses without damaging the fiber tip.

To reduce the acoustic attenuation, the polymer jacket layer of the acoustic delay fiber was removed. The fiber core together with the cladding layer was placed inside a polyimide tubing (Microlumen, Oldsmar, FL) with an inner diameter slightly larger than that of the acoustic delay fiber. The polyimide tubing provides good structural protection and also acoustic insulation for the acoustic delay fiber. With a snug fitting, it does not cause extra acoustic attenuation for the PA signal transmission. During the operation of the probe, the acoustic

delay fiber can easily contact with surrounding tissue or media. Without the acoustic isolation, the transmitted PA signals can be easily damped out. With the use of a polyimide tubing to encapsulate the acoustic delay fiber, the optical delivery and acoustic delay fibers can be directly glued together without affecting the performance (Fig. 1b). In this way, the PA sensing probe can be very compact, easy to build, and therefore suitable for in-vivo use. In addition, the optical excitation and acoustic detection areas can be largely overlapped to provide an optimal sensing condition.

The acoustic delay fiber is interfaced with a homemade PZT (APC 851, APC International, Ltd., Mackeyville, PA) transducer with a thickness of 1.1 mm. For robust interfacing contact between the acoustic delay fiber and the transducer, an acrylic spacer and a holder are used. Acrylic spacers and holders were cut from a 1.5-mm-thick acrylic sheet with a laser cutter (Universal Laser Systems, PLS6.75). The resonance frequency of PZT transducer's thickness mode is ~ 1.85 MHz. The size of the transducer ($4.0 \times 2.0 \text{ mm}^2$) is kept as small as possible to reduce the parasitic capacitance to the transducer. Since both PZT and silica have high acoustic impedance, no matching layer was applied between them. A thin layer of epoxy was applied to directly bond the transducer onto the tip of the fiber. The transducer frequency is chosen based on the non-dispersive single-mode transmission frequency limit of the optical fiber, which is usually around $0.1 \sim 0.2 c/d$, where c and d are the acoustic velocity and diameter of the fiber core, respectively [12,13]. Suppose the acoustic velocity is 5500 m/sec and the fiber core diameter is 400 μm , this gives $1.25 \sim 2.5$ MHz, which covers the peak frequency components of PA signals generated under unfocused illumination conditions. Distortion will occur for ultrasound transmission at frequencies much higher than this value due to dispersion and excitation of multiple modes. To increase the acoustic detection frequency range, the single large-core acoustic delay fiber could be replaced with a bundle of optical fibers with a smaller core diameter. However, because the ultrasound transmission along the optical fibers is very sensitive to the surface and contact conditions, careful measures have to be taken to ensure good acoustic isolation for every fiber in the bundle,

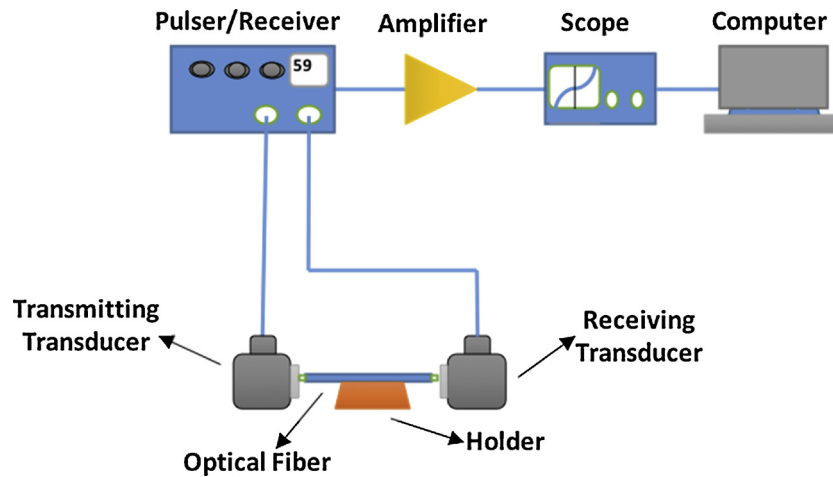


Fig. 2. A Two-Port Ultrasound Testing Setup for the Acoustic Delay Fiber.

which otherwise would be implemented in practice.

2.2. Ultrasound transmission of acoustic delay fiber

To identify optimal acoustic transmission conditions for the PA signals, two-port ultrasound testing was conducted to study the ultrasound transmission in the acoustic delay fiber under three conditions: 1) with the jacket layer, 2) with the jacket layer removed, and 3) placed in polyimide tubing (and with the jacket layer removed). To determine the actual acoustic velocity and attenuation of the acoustic delay fiber (without jacket layer and tubing), six fiber samples with different lengths ranging from 8.4 cm to 54 cm were also prepared.

Two 2.25 MHz flat contact transducers (V104 and V125, Olympus NDT, Waltham, MA, USA) were used to transmit and receive the ultrasound signals (Fig. 2). The two ends of the acoustic delay fiber were polished and contacted onto the transmitting and receiving transducers, respectively. Mineral oil was applied onto the contacts between the ends of acoustic delay fiber and the surfaces of transmitting & receiving transducers to enhance coupling efficiency and minimize unwanted reverberation. The pulser/receiver (5072PR, Olympus NDT, Waltham, MA) was set to the transmission mode, which sent a driving voltage pulse to the transmitting transducer to generate ultrasound signals and also amplified signals detected by a receiving transducer. Each received ultrasound signal was averaged 16 times and recorded on a digital oscilloscope (TDS2002C, Tektronix Inc., Beaverton, OR, USA) to determine their peak amplitude and time delay.

2.3. Verification of interference signal separation

The time delay separating the PA and interference signals was verified with a piece of black tape as the target. The black tape is a good optical absorber, which can generate strong PA signals. In addition, a highly-reflectively mirror surface was also used as the target to test the PA response of the tip of the acoustic delay fiber, which should be minimized as it will mix with the real PA signal from the sensing target. With a high reflectivity, the mirror surface itself is a weak PA generator. It reflects much of the incident light back onto the acoustic delay fiber, which provides a good scenario for observing the PA response at its tip.

2.4. Dye characterization

Fig. 3 shows the test setup of the dye characterization experiment. This setup was also used for the verification of interference signal separation (except that a black tape and a mirror were used as the targets). The excitation light pulse was delivered from a 532 nm Q-switched Nd:YAG laser (SPOT-10-200-532, Elforlight Ltd, Northants, UK)

with a 1.75 ns pulse width and a 1 kHz pulse repetition rate. The output laser energy is 20 $\mu\text{J}/\text{pulse}$. The output laser beam was focused down to ~ 0.7 mm in diameter and incident onto the input end of the optical excitation fiber. Since the core diameter of the optical excitation fiber ($\sim 400 \mu\text{m}$) is smaller than the focused beam diameter, only part of the incident light was coupled. The pulse energy measured at the output end of the optical excitation fiber was $\sim 1.31 \mu\text{J}/\text{pulse}$. Based on the core diameter of the optical excitation fiber, the maximal optical fluence at the output end of the of the optical excitation fiber was estimated to be $1.04 \text{ mJ}/\text{cm}^2$, which is well below the ANSI safety limits for biological tissues of $20 \text{ mJ}/\text{cm}^2$ [14]. The generated PA signals were detected and delayed by the acoustic delay fiber, and serially received by the built-in single-element PZT transducer. The received PA signals were amplified with a homemade PCB pre-amplifier with a gain of 40 dB. The amplified PA signals were displayed on the oscilloscope and stored on the computer.

To prepare the dye solutions as the sensing target, black and red dye powders (Rit® Dye, Phoenix Brands, Stamford, CT) were dissolved in DI water with an initial concentration of 0.1 g/mL or 10% (wt/vol %). The stock solution was further diluted to lower concentrations by adding suitable amount of water. The dye solution to be measured was transferred into an acrylic container. The PA sensing probe was first mounted onto a Z-stage and gradually lowered till the ends of both the optical excitation and acoustic delay fibers just touched the surface of the dye solution (Fig. 3). For each concentration, The PA measurement is repeated five times. The captured PA voltages were averaged to determine the overall PA response.

2.5. Detection of bovine blood in chicken breast

To demonstrate its potential usefulness for *in-vivo* sensing in biological tissues, the PA sensing probe was used to detect bovine blood (Quad Five, Ryegate, MT, USA) embedded inside a chicken breast tissue. The bovine blood was injected into a small plastic pouch made of clear food wrap film and placed inside a piece of chicken breast tissue. The tip of the PA sensing probe was contacted onto the chicken breast tissue (see the Fig. 8 for its diagram). The thickness of the chicken breast tissue separating the blood pouch and the probe tip is varied between 0 mm and 3 mm. Under the same testing conditions for dye characterization, the PA measurement was repeated 6 times. The captured PA voltages were averaged to determine the overall PA response in the bovine blood detection test.

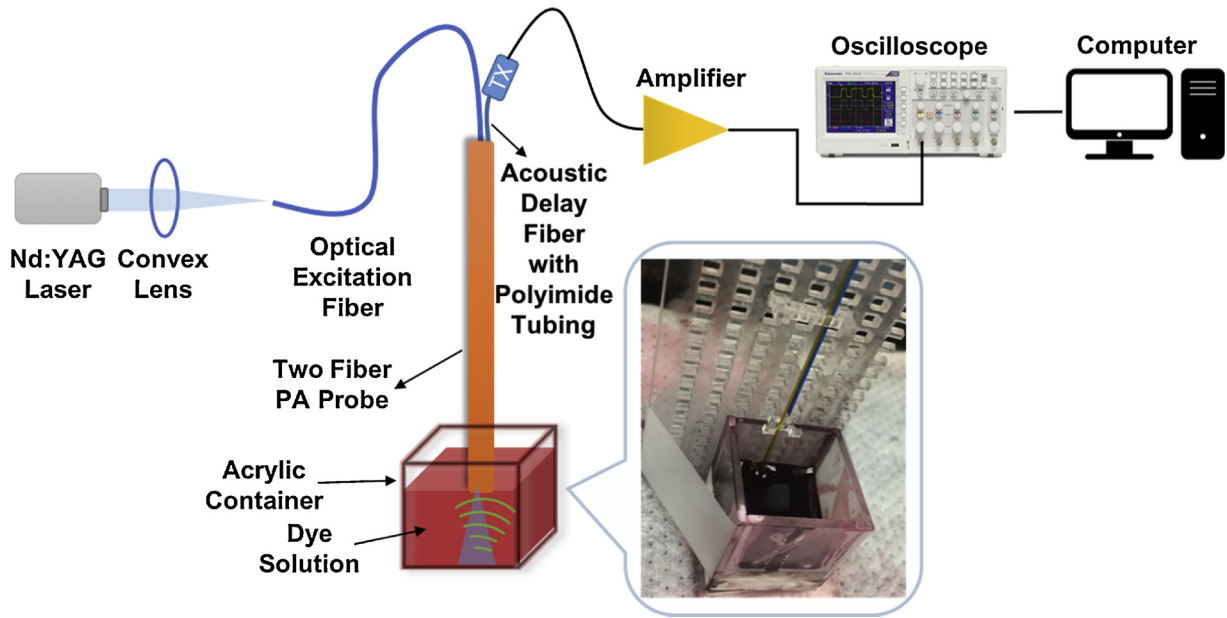


Fig. 3. Schematic of Dye Testing Setup.

3. Results and Discussion

3.1. Ultrasound transmission of acoustic delay fiber

Fig. 4a shows the received ultrasound signal after travelling through the acoustic delay fiber (with the jacket layer) with a length of 100 mm. The ultrasound signal arrived at the receiving transducer with a time delay of $\sim 18.4 \mu\text{s}$ while the source signal can be seen at the zero point in time domain. However, with the plastic jacket layer, the acoustic attenuation is high. For comparison, the ultrasound transmission testing was repeated after the jacket layer was removed and also after the optical fiber was placed inside the polyimide tubing. As shown in Fig. 4b, the acoustic attenuation is significantly reduced after the removal of the jacket layer. Similar result was obtained after the optical fiber (without the jacket layer) was placed inside the polyimide tubing (Fig. 4c), which shows that the polyimide tubing does not significantly affect the ultrasound transmission through the acoustic delay fiber.

The acoustic delay time increases almost linearly with the length of the fiber. Fig. 5 represents that linear relationship between the acoustic

time delay and the length of the fiber. An average acoustic velocity of the fiber is determined to be $\sim 5434 \text{ m/s}$, which is close to the typical sound velocity of fused silica [13,15]. Fig. 5b shows the peak-to-peak voltage amplitude of the received ultrasound signals through the tested fibers. Using the peak-to-peak voltage amplitude of the shortest fiber as the reference, the acoustic loss of each fiber is calculated as $20\log(V_{\text{out}_a} / V_{\text{out}_b})$, where “ V_{out_a} ” refers to the peak-to-peak voltage amplitude of that fiber and “ V_{out_b} ” refers to that of the shortest fiber. The acoustic attenuation (per unit length) of each fiber is determined by dividing the acoustic loss by the length difference (Δl) between the two fibers. Due to the possible different fiber/transducer contact conditions during the testing, the acoustic attenuation (@ 2.25 MHz) of the fibers ranges from 0.041 to 0.266 dB/cm (due to different contact condition in each measurement) with an average value of 0.152 dB/cm.

Although the ultrasound transmission testing is able to reveal the attenuation in the acoustic delay fibers, the potential PA signal reduction caused by the use of the acoustic delay fiber between the target and transducer (compared with direct target-transducer contact) also need to be studied. In our experimental setup, direct contacting transducer

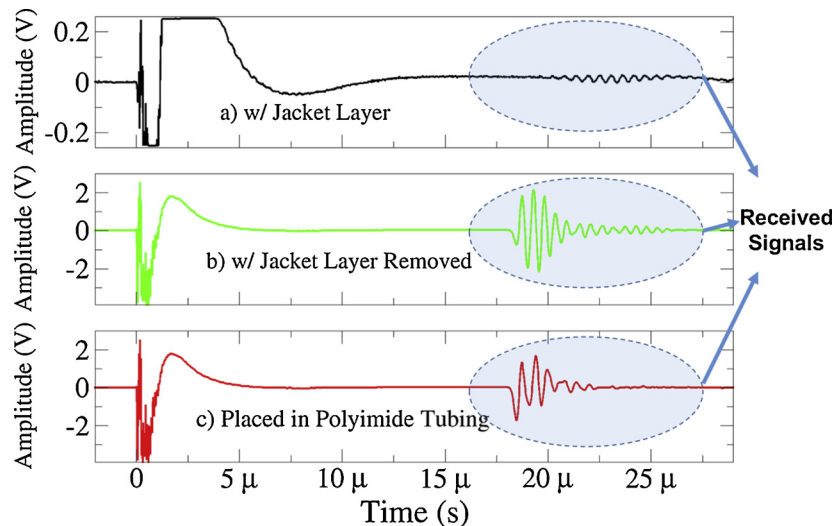


Fig. 4. Received Ultrasound Signals after Travelling through the Acoustic Delay Fiber for Different Fiber Conditions: a) with the Jacket Layer, b) with the Jacket Layer Removed, and c) Placed in Polyimide Tubing (with the Jacket Layer Removed).

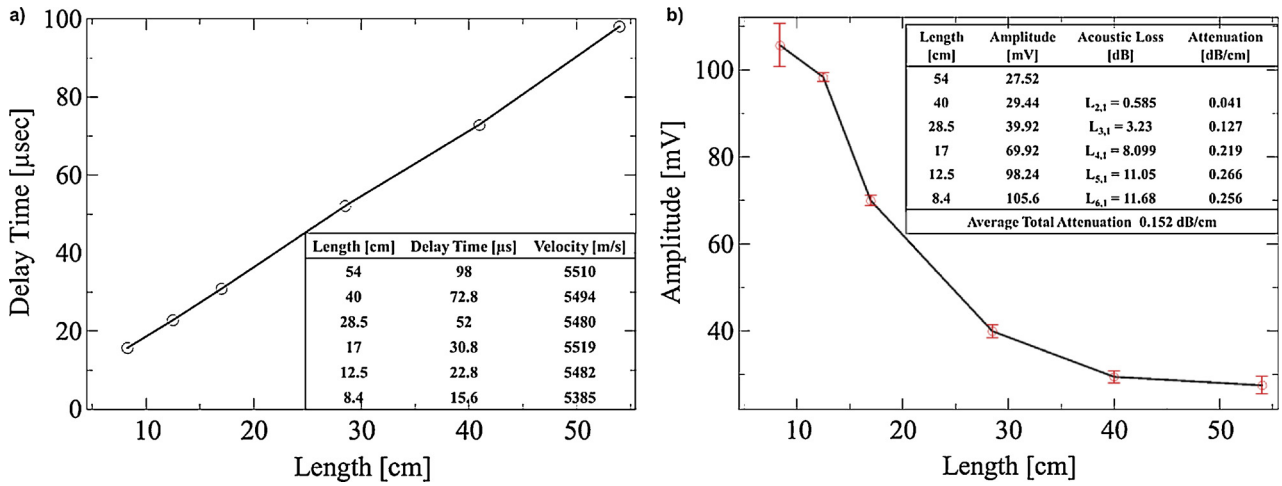


Fig. 5. Two Port Ultrasound Test Through Optical Fibers with Different Lengths for a) Acoustic Time Delay, Velocity and b) Acoustic Loss, Attenuation.

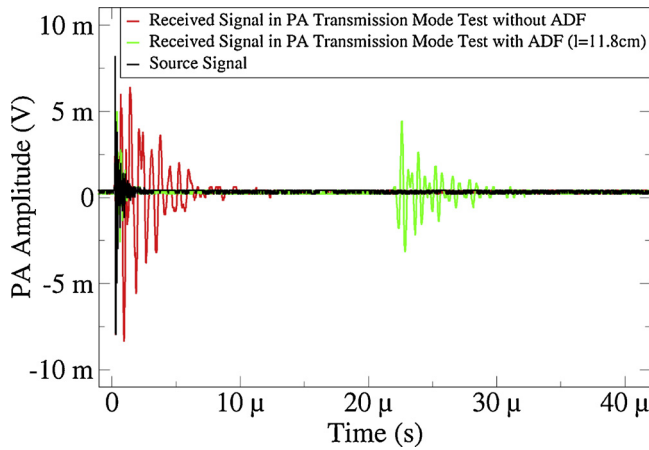


Fig. 6. Received PA Signals in Transmission Mode with and without Acoustic Delay Fiber (ADF).

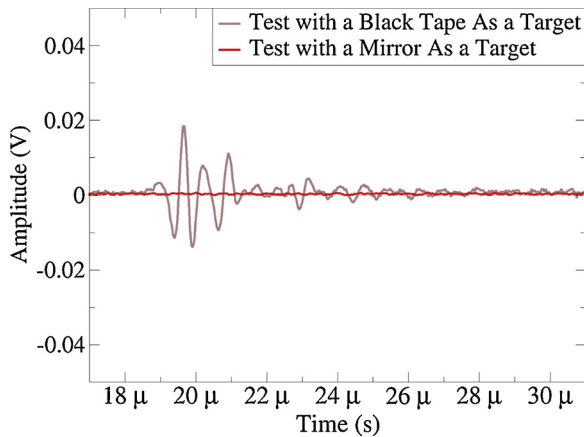


Fig. 7. Received PA Signal after Travelling through the Acoustic Delay Fiber from Black Tape and Mirror as the Target.

onto the target will completely block the excitation light, which makes the reflection-mode testing not feasible. Therefore, a transmission-mode testing was performed on a black tape target to compare the difference in the PA detection performance with and without the acoustic delay fiber. As shown in Fig. 6, the peak-to-peak PA voltage without the acoustic delay fiber is around 16.2 mV. The peak-to-peak PA voltage with the acoustic delay fiber is around 8.2 mV, which is due

to a smaller contact area and also the acoustic attenuation in the acoustic delay fiber. With a delay length of 11.8 cm, the difference in the arrival time between the two PA signals is around 21.60 μs, which approximately matches with its delay time and typical sound velocity of fused silica.

3.2. Verification of interference signal separation

Fig. 7 shows the comparison of received signals when a piece of black tape and a highly-reflective mirror surface were used as the target. The data acquisition was triggered by the firing of the pulsed laser. The PA signal from the black tape target arrived at the transducer after a delay of ~18 μs, which matches with the delay time determined in the ultrasound transmission testing of acoustic delay fiber. Although the strong interference signal still existed at the moment of the firing of the pulse laser, there was no noticeable PA signal with using mirror as the target. This shows that the PA response of the tip of the acoustic delay fiber is negligible, which can be explained by its high optical transparency and low absorption from the mirror.

3.3. Dye characterization and bovine blood detection

Fig. 8 shows representative PA signals received from a 0.04 g/mL red dye solution and a 0.04 g/mL black dye solution. The change in the average PA voltage amplitude of the first peaks as a function of the black dye concentration is shown in Fig. 8b. The peak-to-peak PA voltage increases almost linearly with the dye concentration with a significant correlation ($R^2 = 0.98291$ for black dye and $R^2 = 0.88371$ for red dye). When the black dye concentration was reduced down from 0.1 g/mL to 0.002 g/mL, the signal-to-noise ratio (SNR) dropped from 20.6 dB to 2.67 dB, the received PA waveform started to bury under the noise level. This indicates a detection limit of about 0.002 g/mL. When the red dye concentration was reduced down from 0.1 g/mL to 0.005 g/mL, the signal-to-noise ratio (SNR) dropped from 24.7 dB to 3.37 dB. This indicates a detection limit of about 0.005 g/mL for red dye solution in the PA experiment.

Fig. 9 shows the test setup and recorded PA signals from the bovine blood depending on the thickness of the chicken breast tissue. The PA signal amplitude drops when the thickness of the chicken breast increases. The total time delay of the PA signal is determined by the arrival time of the PA signal. The time delay in the chicken breast tissue between the probe tip and the blood is calculated by subtracting the time delay of the acoustic delay fiber (~21.83 μs) from the total time delay (~23.87 μs with 3 mm thick chicken breast piece). Assuming the acoustic velocity of chicken breast is ~1540 m/s, the distance between the probe tip and the bovine blood is estimated based on the calculate

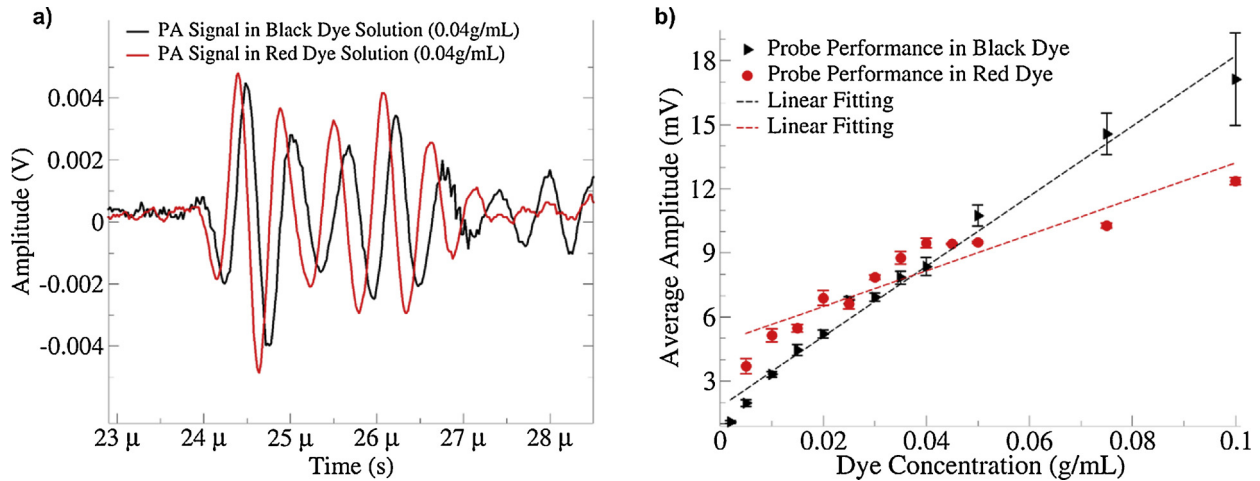


Fig. 8. PA Testing Results of Black and Red Dye Solutions a) PA signals from Black and Red Dye Solution of 0.04 g/mL Concentration and b) Average Amplitudes of PA signals from Black and Red Dye Solutions with Different Concentrations.

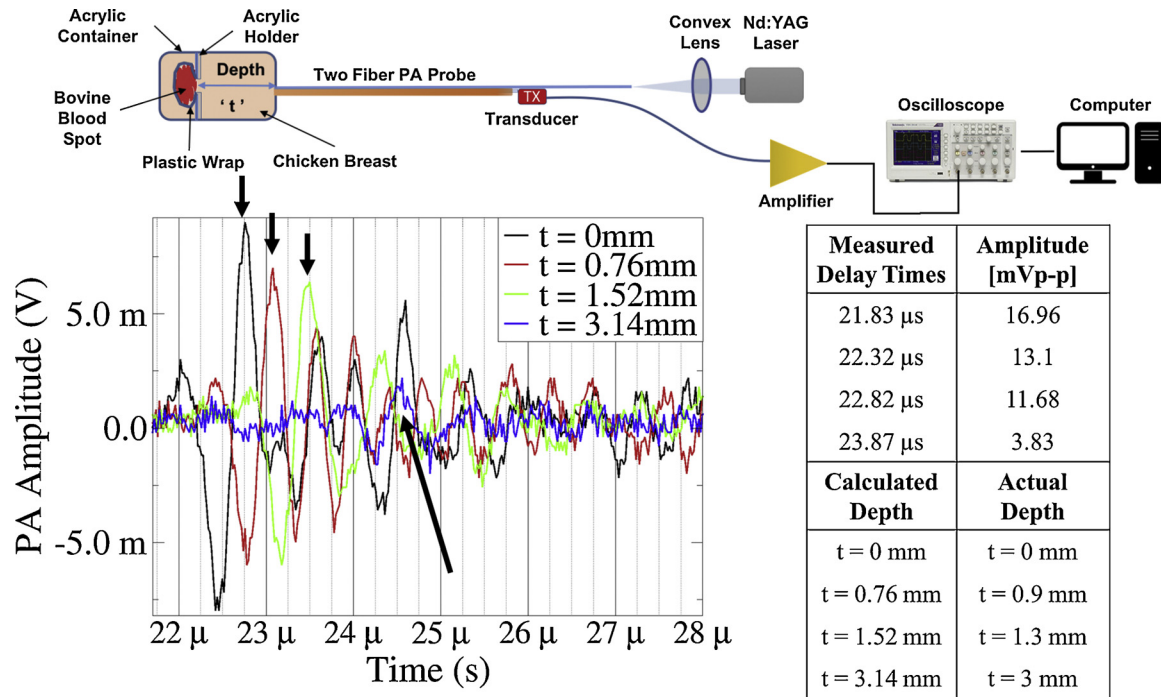


Fig. 9. Detection of Bovine Blood in Chicken Breast Tissue.

time delay in the chicken breast tissue, which matches with the actual thickness of the breast tissue. First positive peaks also provide good comparisons for different thicknesses in PA sensing probe design. This test shows that the PA sensing probe can detect not only the existence of the blood target, but also determine its distance based on the acoustic travel time.

4. Conclusion

In this work, we have demonstrated a new PA sensing probe with light delivery and ultrasound detection in a compact package. Different from conventional approaches, the use of optical fiber as acoustic delay line addresses two challenging issues in PA sensing probe design. First, with a suitable time delay, the PA signals will arrive later at the receiving transducer and not mixing with the much stronger interference signals. This allows PA measurements to be performed on targets close to or even contacted with the probe. Second, the receiving ultrasound

transducer can be located outside the probe shank. As a result, the diameter of the probe shank can be significantly reduced to minimize its invasiveness for *in-vivo* use. In the future, we plan to optimize the fabrication and assembly of the new PA sensing probe and investigate its *in-vivo* applications.

Conflict of interest

The authors declare that there are no conflicts of interest.

References

- [1] A. Oraevsky, A. Karabutov, *Optoacoustic tomography*, Biomedical Photonics Handbook vol. 34, (2003), pp. 1–34.
- [2] L.V. Wang, *Photoacoustic Imaging and Spectroscopy*, CRC press, 2009.
- [3] X. Wang, Y. Xu, M. Xu, S. Yokoo, E.S. Fry, L.V. Wang, *Photoacoustic tomography of biological tissues with high cross-section resolution: reconstruction and experiment*, Med. Phys. 29 (12) (2002) 2799–2805.
- [4] J. Xia, J. Yao, L.V. Wang, *Photoacoustic tomography: principles and advances*,

- Electromagn. Waves Camb. (Camb) 147 (2014) 1.
- [5] T. Lister, P.A. Wright, P.H. Chappell, Optical properties of human skin, *J. Biomed. Opt.* 17 (9) (2012) pp. 0909011-09090115.
 - [6] M. Kinnunen, R. Myllylä, Frequency spectrum analysis of pulsed photoacoustic signals in Intralipid, *Adv. Laser Technol.* 2007 7022 (2008) 702215 International Society for Optics and Photonics.
 - [7] Y. Cho, et al., Handheld photoacoustic tomography probe built using optical-fiber parallel acoustic delay lines, *J. Biomed. Opt.* 19 (8) (2014) pp. 086007-086007.
 - [8] C. Kim, T.N. Erpelding, L. Jankovic, L.V. Wang, Performance benchmarks of an array-based hand-held photoacoustic probe adapted from a clinical ultrasound system for non-invasive sentinel lymph node imaging, *Philos. Trans. Math. Phys. Eng. Sci.* 369 (1955) (2011) 4644–4650.
 - [9] D. Piras, C. Grijnsen, P. Schütte, W. Steenbergen, S. Manohar, Photoacoustic needle: minimally invasive guidance to biopsy, *J. Biomed. Opt.* 18 (7) (2013) pp. 070502-070502.
 - [10] Et al., "Dual-modality photoacoustic and ultrasound imaging system for noninvasive sentinel lymph node detection in patients with breast cancer, *Sci. Rep.* 5 (2015).
 - [11] M.K. Yapici, et al., Parallel acoustic delay lines for photoacoustic tomography, *J. Biomed. Opt.* 17 (11) (2012) 116019.
 - [12] J. May, Wire-type dispersive ultrasonic delay lines, *IRE Trans. Ultrason. Eng.* 7 (2) (1960) 44–52.
 - [13] G. Boyd, L. Coldren, R. Thurston, Acoustic clad fiber delay lines, *IEEE Trans. Sonics Ultrason.* 24 (4) (1977) 246–252.
 - [14] L. ANSfSUo, "American national standard for the safe use of lasers," American National Standards Institute. ANSI Z136. New York, NY: American National Standards Institute, 2007.
 - [15] I. Gelles, Optical-fiber ultrasonic delay lines, *J. Acoust. Soc. Am.* 39 (6) (1966) 1111–1119.



Arif Kivanc Ustun received Bachelor's degree in Electrical and Electronics Engineering from Anadolu University, Eskisehir, Turkey in 2009. He has also M.Eng degree in electrical engineering in 2014, and currently pursuing Ph.D. degree in electrical engineering from Texas A&M University. He worked as a research assistant in Bilecik University in 2009, then was awarded the scholarship of Turkish Ministry of Education since 2011. Currently, he is a teaching assistant at Texas A&M University from 2017. His current research interests include Microelectromechanical systems (MEMS), acoustic/ultrasonic delay lines, novel photoacoustic/acoustic medical probe systems, and micro and nanofabrication methods. Arif Kivanc Ustun is a

member of the International Society for Optics and Photonics (SPIE).



Jun Zou received his Ph.D. degree in electrical engineering from the University of Illinois at Urbana-Champaign in 2002. In 2004, he joined in the department of electrical and computer engineering at Texas A&M University, where he is currently a full professor. His current research interests lie in the development of micro and nano opto-electro-mechanical devices and systems for biomedical imaging and sensing applications. He is a Senior Member of IEEE and SPIE, and a Member of OSA.