Development of All-Optical Quantitative Ultrasound Imaging System

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DEVELOPMENT OF ALL-OPTICAL QUANTITATIVE ULTRASOUND IMAGING SYSTEM

A dissertation submitted in partial fulfillment of the requirements for the degree of

DOCTOR OF PHILOSOPHY

in

BIOMEDICAL ENGINEERING

by

Mohamed A. Almadi

2021
To: Dean John L. Volakis  
College of Engineering and Computing  

This dissertation, written by Mohamed A. Almadi, and entitled Development of All-Optical Quantitative Ultrasound Imaging System, having been approved in respect to style and intellectual content, is referred to you for judgment.

We have read this dissertation and recommend that it be approved.

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Florida International University 2021
DEDICATION

I would like to dedicate my work to my parents Abdulrahman and Refa, my wife Rawan, and my children Yara and Sami, who have always been there for me.
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First and foremost, I would like to express my gratitude and appreciation to my supervisor Dr. Wei-Chiang Lin for his continued support throughout this journey. Without his profound knowledge, guidance, and encouragement, this thesis would not have been achievable. I consider myself lucky to have been advised by someone who is truly all there. Secondly, I would like to thank my committee members Professor Shuliang Jiao, Dr. Jessica Ramella-Roman, Dr. Armando Barreto, and Professor Richard Bone, for their time reviewing my work. Their insightful suggestions, discussion, support, and advice will ever be appreciated.

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Finally, to my children Yara and Sami, the words cannot describe the love I have for you. My appreciation extends to my wonderful family members.
ABSTRACT OF THE DISSERTATION

DEVELOPMENT OF ALL-OPTICAL QUANTITATIVE ULTRASOUND IMAGING SYSTEM

by

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Florida International University, 2021

Miami, Florida

Professor Wei-Chiang Lin, Major Professor

Ultrasound (US) is a well-established deep-tissue imaging modality in biomedicine. It distinguishes different tissue types based on their echogenicity, but this approach provides limited diagnostic sensitivity and accuracy. The majority of the US transducers nowadays rely on lead zirconate titanate (PZT) ceramic elements to transmit and receive ultrasound. Unfortunately, significant limitations arise from these transducers due to their frequency characteristics and complex fabrication process. A recently introduced technique, Quantitative Ultrasound (QUS) Measurement, shows a great promise to improve US-based tissue diagnosis, but it requires a transducer with a large spectrum bandwidth, which is a feature not available in PZT transducers.

Recent research has shown that optical methods for ultrasound generation and detection may overcome these limitations. The significant advantages provided by the optical approaches are the tunable center frequency, large spectrum bandwidth, high sensitivity, and miniaturization friendly. More importantly, they are exceptionally suitable for quantitative measurements.
The primary objective of this Ph.D. research is to develop an all-optical ultrasound transducer (AOUT) for quantitative ultrasound measurements and imaging and, eventually, for tissue characterization. In order to accomplish this goal, an optical ultrasound transmitter (OUT) and a sensitive optical refractometry ultrasound detector (RUD) were developed. For ultrasound generation, an OUT was designed and built based on the photoacoustic (PA) effect. The OUT developed consists of two-layer Candle soot-PDMS coated on a glass substrate and can generate a maximum pressure ($P_{\text{max}}$) of 0.16 MPa, a center frequency ($f_0$) of 35 MHz, and a spectrum bandwidth ($f_{BW}$) of over 38 MHz. For ultrasound detection, a RUD was designed and built based on the probe beam deflection technique (PBD). The RUD developed in this study has a sensitivity of 55 Pa and an axial resolution of 0.27 mm.

By combining OUT and RUD, an AOUT was designed and fabricated. The imaging and the quantitative measurement capabilities of the AOUT were validated using three ultrasound phantoms. The first phantom contains 106-125 µm diameter soda-lime microspheres, the second phantom contains 70-90 µm diameter microspheres, and the last phantom contains 38-45 µm diameter microspheres. The reconstructed images show the profiles of the phantoms with reasonable accuracy, and the frequency spectra of the backscattering signals acquired from these phantoms were distinctively different, which confirms the feasibility of using AOUT to perform quantitative ultrasound measurements.
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<td>AC</td>
<td>Alternating Current</td>
</tr>
<tr>
<td>Al</td>
<td>Aluminum</td>
</tr>
<tr>
<td>AG</td>
<td>Agarose</td>
</tr>
<tr>
<td>AOQUIS</td>
<td>All-Optical Quantitative Ultrasound Imaging System</td>
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<tr>
<td>AOUT</td>
<td>All-Optical Ultrasound Transducer</td>
</tr>
<tr>
<td>A-mode</td>
<td>Amplitude Mode</td>
</tr>
<tr>
<td>DEHA</td>
<td>And Di(2-Ethylhexyl) Adipate</td>
</tr>
<tr>
<td>BSC</td>
<td>Backscattering Coefficient</td>
</tr>
<tr>
<td>BBP</td>
<td>Benzyl-Butyl Phlathate</td>
</tr>
<tr>
<td>B-mode</td>
<td>Brightness Mode</td>
</tr>
<tr>
<td>CS</td>
<td>Candle Soot</td>
</tr>
<tr>
<td>CMUT</td>
<td>Capacitive Micromachined Ultrasonic Transducer</td>
</tr>
<tr>
<td>CB</td>
<td>Carbon Black</td>
</tr>
<tr>
<td>CNF</td>
<td>Carbon Nanofibers</td>
</tr>
<tr>
<td>CNT</td>
<td>Carbon Nanotube</td>
</tr>
<tr>
<td>CW</td>
<td>Continuous Wave Laser</td>
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<tr>
<td>Cr</td>
<td>Copper</td>
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<tr>
<td>EAC</td>
<td>Effective Acoustic Concentration</td>
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<tr>
<td>ESD</td>
<td>Effective Scatterer Diameter</td>
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<td>EMI</td>
<td>Electromagnetic Interferences</td>
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<td>FPI</td>
<td>Fabry–Pérot Interferometer</td>
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<tr>
<td>Fiber Bragg Grating</td>
<td>FBG</td>
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<tr>
<td>Gas-Coupled Laser Acoustic Detection</td>
<td>GCLAD</td>
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<td>Lead Zirconate Titanate</td>
<td>PZT</td>
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<td>Mach-Zehnder Interferometer</td>
<td>MZI</td>
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<td>Michelson Interferometer</td>
<td>MI</td>
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<td>Micro Ring Resonator</td>
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<td>NIR</td>
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<td>Quantitative Ultrasound</td>
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<td>Radio Frequency</td>
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<td>Term</td>
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<td>Refractometric Ultrasound Detector</td>
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<td>SNR</td>
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<td>Time Gain Compensation</td>
<td>TGC</td>
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<td>Time-Of-Flight</td>
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<td>Ultrasound</td>
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<td>Ultraviolet</td>
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<td>X-Ray Computed Tomography</td>
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Chapter 1: Introduction

1. Background

Biomedical imaging technologies including X-ray Computed Tomography (CT), Magnetic Resonance Imaging (MRI), Optical Imaging, and Ultrasound (US) imaging play critical roles in disease and injury diagnosis and treatment [1]. Each imaging modality has its unique operating principle, which gives it certain advantages over other modalities but, unfortunately, also leads to some disadvantages. Clinical MRI and CT systems can create very detailed 3D images of the whole body; however, their high operating and maintenance costs, safety concerns, and bulkiness make them available only in modern clinical settings [2]. Optical imaging, including optical coherence tomography (OCT), can overcome these shortcomings, but its imaging depth is significantly limited by the high scattering and/or absorption properties of biological tissues [3]. Compared to the medical imaging modalities mentioned before, ultrasound imaging is much more versatile. First of all, it does not use ionizing radiation and hence is not a health hazard. Secondly, ultrasound imaging's depth and resolution can be easily tuned by changing the transducer's center frequency [4]. Finally, recent advances in electronic technologies have led to the development of small ultrasound imaging devices that are incredibly portable, inexpensive, and user-friendly [5]. Due to these unique characteristics, ultrasound imaging has re-established itself as a standard diagnostic tool in the medical field.

1.1. Conventional Ultrasound

Since its discovery in 1794, ultrasound has become an essential tool in the industrial and medical fields [6]. Nowadays, medical ultrasound is routinely used in ophthalmology [7]–
Low power ultrasound is typically used to image the internal structures of soft tissue and blood flow in the vessels as well as in the heart. When high power ultrasound is focused, it produces significant local heating and causes irreversibly damage to the target tissue. Therefore, medical ultrasound is not only a diagnostic modality but also a therapeutic modality.

Conventional ultrasound imaging is built upon the detection of the ultrasound waves reflected by the internal structure of the tissue under investigation. The reflection locations are associated with the spatial inhomogeneities in tissue acoustic impedance ($Z$) [24]. For biological tissues, their acoustic impedance is controlled by their density ($\rho$), sound speed ($v$), and compressibility ($\kappa$). That is [25]:

$$Z = \rho v = \sqrt{\rho / \kappa}$$  \hspace{1cm} \text{Eq. 1.1}

Table 1.1 shows the density and the speed of sound of various materials and biological tissues and their corresponding acoustic impedance [26].

The amount of the reflected energy is governed by the magnitude of the acoustic impedance mismatch between the two media [24]. It is mathematically expressed as [27]:

$$\mathcal{R} = \frac{(Z_2 - Z_1)^2}{(Z_1 + Z_1)^2}$$  \hspace{1cm} \text{Eq. 1.2}

where $\mathcal{R}$ is ultrasound reflectance coefficient, and $Z_1$ and $Z_2$ are the acoustic impedance of media 1 and 2, respectively.
<table>
<thead>
<tr>
<th>Material</th>
<th>Density [Kg m$^{-3}$]</th>
<th>Speed of Sound [m/s]</th>
<th>Acoustic Impedance [MRayl]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air</td>
<td>1.183</td>
<td>346</td>
<td>$4.29 \times 10^{-4}$</td>
</tr>
<tr>
<td>Water</td>
<td>1000</td>
<td>1500</td>
<td>1.50</td>
</tr>
<tr>
<td>Blood</td>
<td>1060</td>
<td>1570</td>
<td>1.66</td>
</tr>
<tr>
<td>Fat</td>
<td>952</td>
<td>1450</td>
<td>1.34</td>
</tr>
<tr>
<td>Muscles</td>
<td>1075</td>
<td>1590</td>
<td>1.70</td>
</tr>
<tr>
<td>Bone</td>
<td>1900</td>
<td>4080</td>
<td>7.80</td>
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<tr>
<td>Brain</td>
<td>1030</td>
<td>1550</td>
<td>1.60</td>
</tr>
<tr>
<td>Liver</td>
<td>1060</td>
<td>1590</td>
<td>1.65</td>
</tr>
<tr>
<td>Kidney</td>
<td>1050</td>
<td>1570</td>
<td>1.63</td>
</tr>
</tbody>
</table>

Table 1.1: Typical acoustic properties of biological tissues [26].

In addition to acoustic impedance mismatch, ultrasound reflectance is also affected by the geometric characteristics of the interface. The first reflection type is called specular reflection, where ultrasound waves are reflected by a smooth and flat boundary. The reflected ultrasound waves travel in a particular direction; the majority of them are captured by the transducer. The second reflection type is called diffuse reflection. Here, ultrasound waves are reflected by a non-smooth boundary (i.e., rough interface). In this case, reflected waves propagate in various directions, and only a portion of them are captured by the transducer. The third type of reflection is called backscattering. This phenomenon happens when ultrasound waves interact with structures whose sizes are smaller than the width of the ultrasound wavelength, resulting in non-isotropic ultrasound scattering [27]. Fig.1.1 provides the graphic illustrations of these three different types of ultrasound reflections.
Fig. 1.1: Different types of ultrasound reflections. (a) Specular reflection occurs when the object is larger than the ultrasound wavelength. (b) Diffuse reflection occurs when the object's surface is not smooth. (c) Scattering occurs when the object (i.e., scatterer) is smaller than the ultrasound wavelength.
To form an image, conventional ultrasound imaging relies on the reflection of ultrasound waves at the interfaces of different tissue types/structures (i.e., discontinuation in acoustic impedances). In brightness mode (B-mode) ultrasonography, biological tissue boundaries appear as bright lines, which correspond to the actual anatomical borders. Moreover, the biological tissue appears to be filled with small scatterers, representing the echogenicity (i.e., speckles) [28]. Fig.1.2 shows a typical B-mode image of a phantom containing microscatterers. Moreover, as the ultrasound travels deep into the tissue, its energy is reduced because of reflection, scattering, and absorption [27]. Therefore, the reflected ultrasound waves from the deep part of the tissue are inherently weaker than those from the superficial part. For this reason, all modern ultrasound imaging systems use a process called Time Gain Compensation (TGC) to remove this effect and hence maintain the contrast throughout the entire image depth [27]. This process can cause an artifact known as a posterior acoustic enhancement, where the tissue beyond a water/tissue interface becomes hyperechoic [29], [30]. This artifact is associated with signal processing and the inaccurate assumption of the speed of sound or attenuation in the tissue under investigation [31].

1.2. Quantitative Ultrasound

Despite its widespread clinical applications, B-mode ultrasound imaging has limitations of its own. This limitation is attributed primarily to the echo envelope-detection and the speckle removal process in conventional ultrasound imaging [32]-[33]. In other words, the frequency characteristics of the ultrasound reflection, containing wealth information of the microstructures of the tissue, are removed and distorted. Therefore, the echogenicity in B-
mode ultrasound imaging is qualitative in nature, making the diagnostic information derived from it tend to be highly subjective and user experience-dependent [34].

Fig.1.2: B-Mode Image of a phantom with scatterers, which simulate microscopic inhomogeneity in biological tissue.

To remove this shortcoming and enhance the clinical values of ultrasound imaging, researchers have been developing various approaches that would enhance the diagnostic capability of ultrasound. One approach is to derive quantitative measurements related to tissue properties from the ultrasound reflection. This approach is referred to as quantitative ultrasound (QUS) [35]. To date, a wide variety of QUS techniques have been developed, including frequency domain analysis of backscattering signals [36], flow estimation through Doppler [37], tissue elastography [38], and envelope statistics [32]. Some of these techniques have already been adopted by clinical applications (e.g., Doppler and elastography), while some of these techniques are still under development [32]. Among all QUS techniques explored and implemented, the one analyzing the spectral characteristics of the backscattering signals is of particular interest to this dissertation. This technique typically yields the backscattering coefficient (BSC) of a tissue, which is related
to the tissue's intrinsic properties and hence greatly useful for tissue diagnosis [32]. This technique looks at the frequency component of the BS waves in the frequency domain, rather than just the intensity of waves in the time-domain. It has been shown in the literature that the BSC depends on the shape, size, spatial organization, and acoustic impedance of the scatterers [39]. This QUS method hypothesizes that different tissue types may scatter ultrasound differently due to their distinctive microscopic characteristics. These differences are embedded in the characteristics of backscattering waves. Moreover, if the tissue's microscopic characteristics changed over time due to disease, injury, or response for treatment, then by sensing these changes and expressing them quantitatively will significantly enhance the diagnostic capability of the B-mode ultrasound imaging. Over the past decade, these QUS techniques have been successfully applied to characterizing different diseases and tissue conditions in prostate [40]–[43], heart [44]–[50], eyes [51], breast [52]–[54], and liver [55]–[59].

Estimating the BSC with high accuracy is not straightforward, and several approximations and compensations need to be considered. Nevertheless, as a starting point, the BSC can be defined by the equation derived by Lizzi et al. [60]:

\[
\sigma_{BSC}(f) = \frac{R^2}{V} \frac{\langle I_{sc}(f) \rangle}{I_{inc}(f)} \propto W(f) \quad \text{Eq. 1.3}
\]

where \( R \) is the distance to the scattering volume of interest, \( V \) is the scattering volume which is defined by the beam-width and the gate length, \( I_{sc}(f) \) and \( I_{inc}(f) \) are the intensity of the scattered and incident fields respectively, and \( W(f) \) is the normalized power spectrum.
In the planar reference technique, the normalized power spectrum \( W(f) \) is given by [61]:

\[
W(f) = \frac{1}{N} \frac{\mathcal{R}^2}{4} A(f, L) \sum_{n=1}^{N} \frac{|S_{\text{sample}}(f)|^2}{|S_{\text{planar}}(f)|^2}
\]  
Eq. 1.4

where \( N \) is the number of scan lines, \( \mathcal{R} \) is the reflection coefficient of the planar reference material, \( A(f, L) \) is an attenuation compensation, \( S_{\text{sample}}(f) \) and \( S_{\text{planar}}(f) \) are the Fourier transform of the backscattered signal and the reference signal, respectively.

The first and simplest approach used to estimate the scattering properties is by fitting a line to a BSC spectrum \( \sigma_{b\text{sc}}(f) \). From this line, the slope, the intercept, and the mid-band are extracted. The slope is correlated with the scatterer size, while the intercept is correlated with the concentration [62].

The second approach relies on fitting \( \sigma_{b\text{sc}}(f) \) with a theoretical BSC using a scattering model such as the spherical Gaussian model or the fluid-fill sphere [63]. This approach can estimate two main parameters: the effective scatterer diameter (ESD) and effective acoustic concentration (EAC). Even though the model-based approach is widely used for QUS, applying a simple scattering model to the biological tissue would not yield accurate results [32].

While the above mentioned QUS technique provides valuable information about tissue microstructures, challenges have quickly emerged, limiting the adoption of this technique in clinical devices. First, the accuracy of this QUS technique, as mentioned before, is hinged on a deep understanding of the ultrasound scattering mechanism in biological tissues. However, this knowledge is still incomplete to this date [64], [65]; the available backscattering models often yield inaccurate and inconsistent BSC estimations and hence degrade its performance [66]. Therefore, improving the model parameters to correlate to
the anatomical properties will significantly improve the estimation accuracy. Second, to correctly estimate BSC, the effect of acoustic attenuation on the backscattering waves must be compensated for [67]. If the attenuation was not adequately compensated, the characteristics of the backscattering waves could be affected, which leads to inaccurate estimations. In the literature, attenuation compensation techniques are well-established; however, they require prior knowledge of the tissue attenuation coefficient. The challenge is that the biological tissue is heterogeneous; therefore, the attenuation coefficient is not uniform as a function of depth [32]. Third, the backscattering signals are not entirely defined by the tissue's microstructures but also influenced by the system-dependent factors such as the bandwidth, the focusing, and the mechanical and electrical properties of the transducer [32]. The correction for these factors is done through reference measurements (e.g., reflection from a flat surface or reference phantom). A study conducted by Madsen et al. found considerable differences in the BSC estimates from well-calibrated tissue-mimicking phantom among ten laboratories [68]. The differences were attributed to the different ultrasound systems, measurement techniques, and methods to estimate BSC utilized by each laboratory.

Kruger et al. developed a much simpler method to estimate the scatter size based on the backscattering waves [69]. In this method, instead of the derivation of BSC, the frequency components of the backscattering waves from the dominant regime are used to determine the intrinsic property of the scattering medium. Several assumptions related to scattering medium are used in this method, including (1) the acoustic inhomogeneities are induced by the spherical scatterers embedded, (2) both the background medium and the scatterers are isotropic, and (3) the scattering events are limited in the Rayleigh scattering regime.
(i.e., the wavelength is much larger than the particle size). The dependence of the scatterer size \((a)\) with the frequency of maximum amplitude \((\omega_m)\) can be expressed as:

\[
a = (2n \kappa_s \bar{x})^{-1/6} \omega_m^{-2/3} \quad \text{Eq. 1.5}
\]

where \((n)\) is the concentration of the particles, \(\kappa_s\) is defined by \((\kappa_s = 4\pi g/9v^4)\), \(g\) is a factor that depends on the elastic properties of the scatterers, \(v\) is the speed of sound, \(\bar{x}\) is the window length.

It should be noted that, since the experimental systems have limited bandwidth, a system correction must be applied to the estimation. The transducer transfer function \((B(\omega))\) is modeled as a simple Gaussian function with center frequency \((\eta)\) and FWHM bandwidth \((\sigma)\) as:

\[
B(\omega) = \frac{B_0}{\sqrt{2\pi \sigma}} e^{-\frac{(\omega-\eta)^2}{\sigma^2}} \quad \text{Eq. 1.6}
\]

where \(B_0\) is a scale factor. Inserting Eq. 1.6 into Eq. 1.5 yields the expression of \((a)\) as a function \((\omega_m)\) for such a Gaussian band-limited system:

\[
a = \left(1 + \frac{\omega_m(\omega_m - \eta)}{2n \kappa_s \bar{x} \omega_m^4}\right)^{1/6} \quad \text{Eq. 1.7}
\]

The main drawback of Kruger's method is the number of assumptions related to the scattering properties of the medium. Furthermore, this method needs a fair and reasonable estimation of the concentration in order to yield precise estimations of the scatterer size. Finally, its accuracy is positively related to the bandwidth of the ultrasound system; the larger the bandwidth, the higher the scatterer size estimation accuracy in general.
1.3. Ultrasound Generation and Detection

As far as ultrasound imaging is concerned, the frequency of the ultrasound waves controls the imaging depth [27]. To image deep into the body, low-frequency ultrasound waves, typically between 2 and 5 MHz, are used. On the other hand, high-frequency ultrasound waves, typically between 5 and 15 MHz, would be needed to image superficial tissue structure [24]. Apart from imaging depth, the resolution is another parameter that depends on the ultrasound frequency; high-frequency ultrasound waves yield a high axial resolution and vice versa. Therefore, the selection of the ultrasound frequency and hence the transducer strongly depends on the needs of an imaging application.

Over the years, an array of methods has been developed for transmitting and/or detecting ultrasound waves. Based on the methodology employed, modern ultrasound transducers can be classified into three major types: (1) piezoelectric-based transducers, (2) capacitive micromachined ultrasonic transducers, and (3) optical-based ultrasound transducers. The following section presents an overview of these transmitter types with their advantages and limitations:

1.3.1. Piezoelectric Ultrasound Transducers

The piezoelectric effect has been the most commonly used mechanism for ultrasound generation and receiving since the dawn of medical ultrasound imaging [70]. In the transmission mode, a high voltage pulse is applied to the piezoelectric element, which generates vibration and hence pulsed ultrasound waves. In the receiving mode, the acoustic pressure is applied to the piezoelectric element, which generates electric charges [71]. The cross-section of a typical piezoelectric ultrasound transducer is shown in Fig.1.3.
Fig. 1.3: Cross-section of a piezoelectric ultrasound transducer. Positive and negative electrodes allow for electrical connections. The damping block absorbs the ultrasound wave propagating backward and prevents excessive vibration of the piezoelectric element. The matching layer creates an interface between the piezoelectric element and the surrounding medium.

The piezoelectric materials found in ultrasound transducers include piezoelectric crystals (e.g., quartz), piezoceramics (e.g., lead zirconate titanate (PZT)), and piezoelectric polymers (e.g., polyvinylidene fluoride (PVDF)) [72]. Due to its native characteristics, the fundamental resonant frequency \( f_r \) of a piezoelectric element can be determined by its physical dimension (i.e., thickness) and the speed of sound in it \( v_{piezo} \) [73]. That is:

\[
f_r = \frac{v_{piezo}}{2l}
\]

Eq. 1.8

When the frequency of the applied voltage matches with its fundamental resonance frequency, the vibration amplitude of a piezoelectric element will be at its highest. Therefore, low-frequency ultrasound transducers use thicker piezoelectric elements than high-frequency ones.
Also, due to the nature of the piezoelectric element and the transducer's design, the piezoelectric elements will always continue vibrating beyond the termination of the electrical pulse. This is known as the ringing effect [74]. Therefore, a backing material is often placed directly behind the active elements to dissipate and prevent excessive vibrations (i.e., ringing effect). A highly damped material with an acoustic impedance close to the piezoelectric element can efficiently shorten the length of the generated ultrasound wave, thus improving the axial resolution and widening the spectrum bandwidth of the transducer [75].

Moreover, because of the acoustic impedance mismatch between the piezoelectric material and the material under investigation (e.g., water or soft tissue), a matching layer between the two media must be introduced to facilitate the transmission of ultrasound pulses. The thickness of the matching layer is a quarter of the wavelength of the produced ultrasound [76].

One of the main limitations of piezoelectric-based ultrasound transducers is the manufacturing difficulty and complexity. Fabricating a piezoelectric ultrasonic transducer with a center frequency above 30 MHz is still technically challenging due to micro-scale dicing and electrical connections [72]. Another limitation is the acoustic impedance mismatch between the piezoelectric layer (>30 MRayl for PZT) and the surrounding medium such as air (0.0004 MRayl), water (1.5 MRayl), and soft tissue (1.6 MRayl) [77]. This huge mismatch limits the ultrasound wave transmission from the piezoelectric-based transducer to the surrounding medium. Moreover, even with the presence of a matching layer, as seen in Fig.1.3, a coupling medium between the transducer and the imaging target still has to be used to efficiently transmit ultrasound waves from the transducer to the target.
Another major disadvantage of piezoelectric transducers is the requirement of a high voltage driver, which raises serious electrical safety concerns in environments such as the operating field [24]. Finally, because of the electronic components in the piezoelectric transducers, they are sensitive to electromagnetic interferences (EMI), affecting the quality of the ultrasonic signals. In addition to fabrication drawbacks, piezoelectric-based transducers have limitations in frequency characteristics as the thickness of individual piezoelectric elements determines the operating frequency [73], and the generated bandwidth is determined by the acoustic impedance of the backing layer (damping block) [74].

1.3.2. Capacitive Micromachined Ultrasonic Transducers (CMUTs)

CMUTs are considered to be the next generation of ultrasound transducers [78]. Unlike piezoelectric-based transmitters, CMUTs rely on electrostatic principles for both ultrasound emission and detection [79]. This type of transducers is constructed using the micromachining technique, where a cavity is formed by two electrodes, as seen in Fig.1.4. The working principle of CMUT is provided as follows. In the transmission mode, an AC signal is applied to the electrodes of a CMUT, which causes the membrane to vibrate and produces ultrasound pulses. In receiving mode, the incident ultrasound pulses deform the CMUT membrane, which varies the capacitance of the CMUT and hence generates a voltage signal [80].

Compared to the conventional PZT transducers, CMUTs offer advantages such as higher sensitivity, broader bandwidth, and reduce voltage requirements. However, the high performance of CMUTs can only be achieved when a large AC signal is applied across the
electrodes, increasing the risk of device failure [72]. Despite the outstanding performance of CMUTs, its output pressure is lower compared to conventional PZT in the immersion applications (i.e., medical imaging) [81], and the sensitivity is slightly lower than PZT transducers [82]. Moreover, when CMUTs are built into an array, they suffer from significant crosstalk between the elements [83].

Fig.1.4: The schematics of Capacitive Micromachined Ultrasound Transducer (CMUT). The AC signal is applied to the electrodes, which causes the membrane to vibrate and produces ultrasound pulses. The incident ultrasound pulses deform the CMUT membrane, which varies the capacitance of the CMUT and hence generates a voltage signal.

### 1.3.3. Optical Ultrasound Transducers

The limitations of the piezoelectric ultrasound transducers and CMUTs provide strong motivation for the investigation of a new ultrasound transduction mechanism. Over the past two decades, optical approaches for ultrasound generation and detection have been investigated, and high-quality results yielded. Today, they are considered an attractive base for developing ultrasound imaging techniques [84].

The main advantage of using optical methods to detect ultrasound waves is the independence between the sensitivity of the detector and its size. Unlike piezoelectric-based ultrasound transducers, where the sensitivity drops as the element size is reduced,
optical ultrasound transducers can preserve their sensitivity when miniaturized. Moreover, this transducer type is immune to EMI. Another attractive feature for optical ultrasound transducers is the tunability of the center frequency of the generated ultrasound pulses. In other words, the center frequency of an optical ultrasound transducer can be instantaneously adjusted in accordance with the imaging application needs by altering the temporal profile (i.e., pulse width) of the excitation laser [84]–[86]. In the following sections, the established methods for optical ultrasound generation and detection by other research groups will be briefly reviewed and compared.

1.3.3.1. Optical Ultrasound Generation

The most common optical process for ultrasound generation uses the photothermal mechanism to create a photoacoustic (PA) effect [84]. In this process, the optical energy is absorbed by an absorbing material, causing a transient increase in local temperature. This temperature elevation, in turn, causes the absorbing material to expand through the process called thermoelastic expansion. Finally, the expansion leads to the generation of an ultrasound wave. To efficiently generate ultrasound waves using the PA effect, the thermal and the stress confinement conditions are required. In other words, the excitation laser pulse width must be much shorter than the thermal relaxation time and the stress relaxation time [87]. In addition, the excitation pulse energy should be kept low enough so the acoustic waves generated do not introduce physical damage to the absorbing material.

The ultrasound generation via optical means was demonstrated for the first time in 1963 by White et al. [88]. Since then, interest in this method has grown steadily. Initially, the research work focused on the PA effect of ultrathin metal films. However, due to their
inherited low thermoelastic coefficient, the ultrasound generation efficiency of these films was poor [89]. The research then evolved towards the use of carbon-based absorbers, such as carbon black, mixed with elastomers or epoxy materials, such as PDMS, strongly enhanced the PA effect because of their high thermoelastic expansion coefficient [90]–[92]. To date, researchers have combined PDMS with different absorbers, such as graphite [93], carbon nanoparticles [94]–[99], and gold nanoparticles [100]–[103]. The schematics of different optical ultrasound transmitters are shown in Fig.1.5.

![Schematics of different optical ultrasound transmitters](image)

Fig.1.5: Schematics of (a) fiber-optic ultrasound transmitter, (b) planar ultrasound transmitter array, and (c) focused ultrasound transmitter.

In those investigations, improving the efficiency of the PA effect was the primary goal so that optical ultrasound transmitters can produce sufficiently strong ultrasound waves, comparable to PZT transmitters, for imaging purposes. The photoacoustic conversion efficiency is known to depend on parameters such as thermoelastic expansion, optical absorption, and thermal capacity of the absorbing film [104]. However, there is not yet a
comprehensive study that explores the interplays of these parameters and hence provides a clear guideline about generating a highly efficient optical ultrasound transmitter.

1.3.3.2. Optical Ultrasound Detection

Optical ultrasound detectors can be classified into two main categories: (1) Interferometry and (2) refractometry, based on the mechanisms employed [105].

1.3.3.2.1. Interferometric Methods

Using interferometry as the foundation, various interferometric techniques for ultrasound detection have been developed over several decades. These techniques include Fabry–Pérot interferometer (FPI) [106]–[114], Fiber Brag Grating (FBG) [115]–[118], Micro Ring Resonator (MRR) [119]–[122], Mach-Zehnder interferometer (MZI) [123]–[126], and Michelson interferometer (MI) [127]–[129]. Among them, FPI, FBG, and MRR are the most frequently used due to their high sensitivity and high-frequency response. In the following sections, the working principles of these techniques, their advantages, and their limitations will be discussed.

1.3.3.2.1.1. Fabry-Pérot Interferometers (FPI)

A typical fiberoptic FPI is composed of two mirrors separated by a transparent medium to form a cavity, as shown in Fig.1.6. The partially reflective mirror reflects a portion of the probe laser beam, which becomes the reference beam. The portion of the probe beam transmitting through the partially reflective mirror will be reflected by the total reflective mirror (the sample beam) and travel back to the photodetector. At the photodetector, the
reference beam and the sample beam are summed together. Because of the extra traveling distance in the cavity of the sample beam, its phase will be different from that of the reference beam. This phase difference affects the outcome of summing the reference and the sample beams (i.e., constructive and destructive inferences).

Fig. 1.6: Schematic of a typical fiberoptic Fabry–Pérot Interferometer (FPI). An incident ultrasound wave changes the thickness of the Fabry-Pérot cavity and hence the interference outcome between the reference and the sample beams at the detector side.

When an ultrasound wave strikes the front mirror of the FP cavity, the front mirror moves, and the overall length of the FP cavity changes. This change, in turn, alters the phase of the sample beam and hence the interference outcome between the reference and the sample beams at the detector side. The optical output of an FPI is typically proportional to the applied pressure [101]. The sensitivity of a fiberoptic FPI sensor for ultrasound may be tuned by selecting the total reflective mirror with different mechanical properties.

FPI is one of the more promising optical ultrasound detection methods due to its good signal-to-noise ratio (SNR), broad spectrum bandwidth, and high sensitivity. However, like any other technique, fiberoptic FPI has some limitations. The main limitation of fiberoptic FPI is that it is a point detection technique. In order to perform 2D or 3D ultrasound
imaging using a fiberoptic FPI, an array of fiberoptic FPIs may be used, which is complex to fabricate. Alternatively, a scanning approach may be used, which, unfortunately, would increase the imaging time. Another drawback of fiberoptic FPI is its sensitivity to temperature. When subject to temperature, the thickness of the FP cavity will change accordingly due to the thermal expansion properties of the FP cavity. This change will compromise the linearity of the sensor hence reduce its sensitivity [114].

1.3.3.2.1.2. Fiber Bragg Gratings (FBGs)

Fiber Bragg Gratings (FBGs) are optical fibers with a periodic refractive index modulation inside their cores, as shown in Fig.1.7. The periodic refractive index feature in the core of an FBG is created by exposing the fiber core to a patterned intense ultraviolet (UV) energy [130]. When a continuous wave (CW) probe laser transmits through an FBG, a small portion of it is reflected by each high reflective band (i.e., groove). All the reflected light signals then form one single reflection at a particular wavelength that satisfies the Bragg condition. When an ultrasound wave interacts with an FBGs, the refractive index of the FBGs is altered, leading to a shift in the FBGs reflection spectrum [105]. Several studies have shown that FBGs have higher sensitivity than conventional ultrasound transducers [85], [131]. Moreover, due to the fact that the fiber's diameter can be in the micrometers range, an FBG-based ultrasound detector can be miniaturized hence make it suitable for, for example, intravascular applications. However, since this technology depends on the refractive index modulation, FBGs are sensitive to background vibrations and temperature fluctuations [105].
Fig. 1.7: Schematic of a Fiber Bragg grating (FBG). The periodic refractive index grating reflects a small portion of the transmitted laser beam. When an ultrasound wave interacts with an FBGs, the refractive index of the FBGs is altered, leading to a shift in the FBGs reflection spectrum.

1.3.3.2.1.3. Micro-Ring Resonators (MRRs)

A Micro-Ring Resonator (MRRs) is composed of a looped optical waveguide (ring) that is coupled with a bus waveguide, as shown in Fig. 1.8. The incoming ultrasound will deform the ring waveguide dimensions, which changes the refractive index of the ring via the elasto-optic effect [105]. The change in the ring’s refractive index modulates the round-trip phase, causing a phase modulation and an amplitude modulation of the output intensity [120]. Thus, by detecting the variation in the output optical intensity, the corresponding ultrasound waves can be recorded.

MRRs have several advantages, such as broad detection bandwidth, high acoustic sensitivity, and compact size [120]. However, the major limitation is the optical loss due to curvature. Moreover, like FBGs, MRRs are sensitive to temperature fluctuations and background noise.
Fig. 1.8: The schematic of Micro-Ring Resonator (MRR). The incoming ultrasound will deform the ring waveguide dimensions, which changes the ring's refractive index, hence modulating the round-trip phase, causing a phase modulation and an amplitude modulation of the output intensity.

1.3.3.2.2. Refractometric Methods

The interaction between the ultrasound waves and the medium alters the medium's refractive index via the photoelastic principle, which causes the probe beam to deflect from its original path [132]. Various ultrasound detection techniques have been developed based on this principle, such as probe beam deflection (PBD), phase-sensitive ultrasound detection with a Schlieren beam, and Intensity-sensitive detection of refractive index [133]–[136]. In this section, the discussion is going to focus on the PBD techniques.

PBD detects ultrasound waves in a transparent medium by sending a probe laser beam through the region where the ultrasound wave is present and monitoring the change in its propagation direction. A typical setup of PBD is shown in Fig. 1.9. The ultrasound wave, which consists of compression and rarefaction cycles, produces similar fluctuations in the index of refraction of the medium. The fluctuations deflect the probe beam from its original optical path. A high-speed photodetector can be used to detect the fluctuation in the probe
beam direction, producing a signal corresponding to the temporal profile of the ultrasound wave.

![Diagram of a typical Probe Beam Deflection (PBD) technique sensor](image)

**Fig.1.9:** A typical Probe Beam Deflection (PBD) technique sensor. When an ultrasound wave propagates through a transparent medium, it produces fluctuations in the index of refraction. The fluctuations deflect the probe beam from its original optical path. The deflection shown here is greatly exaggerated.

The deflection of a probe beam propagating in a sound field was reported first by Lucas and Biquard in 1932 [137]. Since then, this technique has been used in a wide range of applications, from investigating thermal effects [138] to measuring acoustic effects [139]. A quick search on the usage of this technique in the medical field shows that PBD has been used for ultrasound detection, photoacoustic measurements, and photoacoustic imaging [125]–[139].

There are several advantages associated with the PBD technique in terms of ultrasound detection. First of all, this unique technique is composed of only a few components and is easy to build, as seen in Fig.1.9. Second, PBD does not require direct contact with the sample to be imaged, making it suitable for remote sensing. For example, Gas-coupled Laser Acoustic Detection (GCLAD) has been utilized to sense ultrasound waves travel through the air in various studies [142], [144], [154]. Third, unlike other optical ultrasound
sensing techniques, PBD is insensitive to the background noise, minimizing the need for acoustic isolation [155]. Fourth, PBD can be easily implemented to perform one-dimensional (line) or two-dimensional (area) detection. Fifth, PBD is immune to excessive vibrations (e.g., the ringing effect associated with the PZT) since the probe beam is only deflected when an ultrasound wave propagates through it [132]. Because of these advantages, PBD can be used in a wide range of biomedical applications, especially when contact with the sample is prohibited. Examples include but are not limited to imaging of the burned tissue, eyes, and brain.

2. Motivation and Significance

The features and advantages of optical ultrasound generation and detection techniques provide a strong foundation for the development of an all-optical ultrasound imaging system. This system can be used in, for example, endoscopic ultrasound imaging and minimally invasive surgery, where conventional ultrasound imaging systems are not applicable.

For optical ultrasound generation, the current works show high pressure and high-frequency capabilities; however, their applications were mainly limited to subsurface and high-resolution imaging. For applications such as ophthalmology, dermatology, and intravascular imaging, a transmitter with a frequency >30 MHz is preferred [156]. However, abdominal imaging (i.e., liver, kidneys, pancreas) and cardiology require a low-frequency transmitter [157]. Therefore, developing an optical ultrasound transmitter that can generate a high-pressure ultrasound wave with a tunable center frequency can open new possibilities for biomedical applications.
For optical ultrasound detection, the PBD technique is very attractive due to its high sensitivity and wide spectrum bandwidth. While this technique has been utilized in photoacoustic imaging applications, integrating it with an optical ultrasound transmitter to form an all-optical ultrasound transducer has not been demonstrated yet.

Moreover, as mentioned before, the diagnostic capability of the conventional ultrasound imaging systems can be improved by introducing quantitative analyses to the raw radio frequency (RF) signals. All quantitative ultrasound studies found in the existing literature utilize piezoelectric-based transducers to perform their measurements. To the best of my knowledge, the feasibility of using an all-optical ultrasound system to perform quantitative ultrasound has not yet been demonstrated.

3. Aims and Objectives

This Ph.D. research aimed to design and develop an all-optical quantitative ultrasound imaging system (AOQUIS) for imaging and tissue diagnosis. It should have a tunable center frequency, leading to an adjustable imaging depth and resolution. In addition, it should provide broadband backscattering signals to facilitate quantitative ultrasound analyses. To achieve this goal, three major investigations have been carried out. They are:

Aim 1: Develop optical ultrasonic transmitters with a tunable center frequency:

- Design and develop a planar optical ultrasound transmitter using carbon-based absorbers and Polydimethylsiloxane (PDMS).
- Execute a set of experiments to characterize the performance of the transmitters built and hence identify the optimal transmitter design for an AOQUIS.
Aim 2: Develop a broadband optical ultrasound detector:

- Build an optical ultrasound detector based on the PBD technique
- Establish in-depth knowledge about the effects of the intrinsic ultrasound and probe beam characteristics on the performance of the PBD-based ultrasound detector.
- Use experiments to validate the performance of the PBD-based ultrasound detector and identify the pathway for optimization.

Aim 3: Design and develop an all-optical quantitative ultrasound system for 2D imaging:

- Design and develop a novel all-optical ultrasound transducer (AOUT) that consists of a photoacoustic-based ultrasound transmitter and a probe beam deflection-based ultrasound detector.
- Use a set of experiments to validate the performance of the AOUT in terms of imaging and quantitative ultrasound measurements.

4. Dissertation Organization

The organization of the remaining dissertation is based on the specific aims stated above. Chapter 2 will describe the development of a highly efficient optical ultrasound transmitter. The fabrication processes of a single and multilayer optical ultrasound transmitter will be described. The outcomes of the experimental studies evaluating the effects of various transmitter properties, such as absorber type, on transmitter's efficiency and their corresponding analyses will be reported. Finally, the optimal optical ultrasound transmitter design will be disclosed.
Chapter 3 will describe the development of a PBD based ultrasound detector. The process of building this detector will be disclosed. Next, the outcomes of various experimental studies characterizing the performance of the detector will be reported. Through the analysis of the experimental results, the optimization strategy for the detector will be identified. Finally, evidence confirming the quantitative ultrasound measurement capability of this detector will be presented.

Chapter 4 presents the design, fabrication, and evaluation of the novel all-optical ultrasound transducer (AOUT). This AOUT is an integration of the multilayer CS-PDMS ultrasound transmitter reported in Chapter 2 and the PBD based ultrasound detector disclosed in Chapter 3. Moreover, the outcomes from a series of experiments demonstrating the capabilities of the proposed system to produce a-mode signals, B-mode images, and quantitative ultrasound measurements will be presented.

Chapter 5 is the last chapter of this dissertation. It will summarize the findings of this research work, discuss the limitations of the first AOUT, and finally layout the scope for future research.
Chapter 2: Optical Ultrasound Transmitter

1. Introduction

In the majority of ultrasound applications, ultrasound generation relies on materials such as quartz, crystals, and lead zirconate titanate (PZT) [158]. These materials can convert applied electrical energy to mechanical energy and vice versa through a process referred to as the piezoelectric effect [159]. While popular, piezoelectric ultrasound transmitters process some critical limitations from the biomedicine point of view. The primary limitation is the center frequency and the bandwidth of each transmitter are fixed [70]. In addition, dicing piezoelectric material, building electrical connections, and embedding electronic circuitry introduce significant transmitter fabrication complexity [23], [24], [31], [80]. Moreover, piezoelectric ultrasound transmitters require a high driving voltage to generate sufficiently strong ultrasound waves for imaging purposes; they may cause an electrical hazard if not appropriately insulated.

Optical ultrasound transmitters (OUT) have been considered as a practical alternative to conventional piezoelectric transducers [84], [160]. This transmitter type generates ultrasound through the photoacoustic (PA) effect, which refers to the generation of acoustic waves by converting the optical energy into mechanical energy [161], [162].

Optical generation of ultrasound waves can be achieved through two main regimes, the ablation regime and the thermoelastic regime [163]. In the ablation regime, ultrasound is generated by using high laser energy. The high energy causes the temperature of the material to exceed its melting point. Continued heating causes the surface of the material to be removed through vaporization. This rapid phase change generates ultrasound waves [164]. While the ablation regime produces much stronger acoustic pressure, high laser
energy is destructive to the material exposed. In contrast, generating ultrasound waves in the thermoelastic regime is a non-destructive approach [165]. In the thermoelastic regime, the ultrasound is generated through the PA effect, where the light energy is converted into heat, leading to local thermal expansion in the material [85], [88], [166]. The latter is the preferred regime to build an OUT because of its non-destructiveness; it was the primary mechanism explored in this Ph.D. research.

The main advantage of OUTs compared to the traditional PZT transmitters is that the characteristics of the generated ultrasound waves can be controlled by changing the parameters of the excitation laser, such as pulse duration and laser spot diameter. Another advantage of this transmitter type is the absence of electronic components, which makes them immune to EMI and compatible with other imaging technologies such as MRI [84]. Finally, a linear or 2D OUT array can be easily produced by scanning the excitation laser over an OUT film; the element size and element density of this OUT array can be adjusted by changing the spot size and the scanning speed of the excitation laser used.

Another critical advantage of OUTs is that the temporal profile of its generated ultrasound waves follows the temporal profile of the excitation laser [84]. By using a laser pulse with a pulse width of a few nanoseconds, an ultrasound wave with a center frequency of tens of MHz can be generated and used for high-resolution imaging. By increasing the laser pulse width to more than a hundred nanoseconds, an ultrasound wave with a center frequency of a few MHz can be generated and used for depth tissue imaging [27]. This versatility is extremely attractive in biomedical applications such as cancer surgery.
2. Photoacoustic (PA) Theory

The PA generation is the underlying mechanism of an OUT [160]. The PA generation process starts with the conversion of optical energy to thermal energy by an absorbing material. The local thermal energy subsequently causes the surrounding material (elastomer, in most cases) to expand via thermal expansion. Finally, this expansion leads to the creation of ultrasonic acoustic waves. These steps are graphically illustrated in Fig. 2.1.

![Diagram of PA process](image)

Fig. 2.1: Process of the photoacoustic (PA) effect in which optical energy is converted to ultrasound waves through a series of energy conversions.

To maximize the PA generation efficiency, the following conditions have to be met. First, the laser pulse should be shorter than the thermal relaxation time of the surrounding material. Second, the laser pulse should be shorter than the acoustic relaxation time of the surrounding material. These two requirements are referred to as thermal and stress confinements, respectively [104]. Under these prerequisites, the generated PA pressure ($P_{PA}$) can be described by:

$$P_{PA} \equiv \left( \frac{\beta v^2}{C_p} \right) \frac{\mu_a F}{c \tau} = \frac{\mu_a F}{C_p} \frac{E_{th}}{c \tau}$$

Eq. 2.1

where $\beta$ denote the thermal expansion coefficient [$^\circ$C$^{-1}$], $v$ the sound speed [m/s], $C_p$ the specific heat capacity [J/(K kg)], $\mu_a$ the optical absorption coefficient [mm$^{-1}$], $F$ the light
fluence \((F = E_p/r^2)\) where \(E_p\) is the laser pulse energy [nJ] and \(r^2\) effective focal spot area [mm\(^2\)]. \(\Gamma\) is referred to as the Grüneisen coefficient \((\Gamma = \beta v^2/C_p)\) expressed by the thermal and acoustic properties of the surrounding material, \(\tau\) is the laser pulse duration [ns], and \(E_{th}\) is the thermal energy [nJ/mm\(^3\)] \([95]\). Eq. 2.1 shows how \(P_{PA}\) is influenced by the absorbing/surrounding material properties as well as the excitation laser parameters. These insights provide meaningful guidance to the material selection process of developing a highly efficient OUT. Theoretically, to generate high-pressure ultrasound waves, the OUT material should have a high optical absorption coefficient, a high thermal conductivity, a large thermal expansion coefficient, a high speed of sound, and a low heat capacity. In addition, high laser fluence and short pulse duration are also preferred. These critical parameters are listed in Table 2.1.

3. Structure of the Optical Ultrasound Transmitter

In the early research about PA ultrasound generation, researchers focused only on materials with high optical absorption (absorbers). However, these absorbers typically have a low thermal expansion coefficient and consequently yield low PA efficiency. It was later found a significant improvement in PA efficiency could be obtained by embedding the absorbers in elastomers with a high thermal expansion coefficient \([167]\).

In the following section, two main absorbing materials (i.e., metal-based and carbon-based materials) that are widely used in OUT developments will be discussed. In addition, elastomeric materials available and suitable to create highly efficient OUT will be reviewed.
<table>
<thead>
<tr>
<th>Parameter</th>
<th>Description</th>
<th>Unit</th>
<th>Belong to</th>
</tr>
</thead>
<tbody>
<tr>
<td>( \beta )</td>
<td>Thermal expansion coefficient</td>
<td>°C(^{-1})</td>
<td>OUT material</td>
</tr>
<tr>
<td>( v )</td>
<td>Speed of sound</td>
<td>m/s</td>
<td>OUT material</td>
</tr>
<tr>
<td>( C_p )</td>
<td>Specific heat capacity</td>
<td>J/(K kg)</td>
<td>OUT material</td>
</tr>
<tr>
<td>( \mu_a )</td>
<td>Optical absorption coefficient</td>
<td>mm(^{-1})</td>
<td>OUT material</td>
</tr>
<tr>
<td>( F )</td>
<td>Laser fluence</td>
<td>nJ/mm(^2)</td>
<td>Excitation laser</td>
</tr>
<tr>
<td>( E_p )</td>
<td>Laser pulse energy</td>
<td>nJ</td>
<td>Excitation laser</td>
</tr>
<tr>
<td>( r^2 )</td>
<td>Effective focal spot area</td>
<td>mm(^2)</td>
<td>Excitation laser</td>
</tr>
<tr>
<td>( \tau )</td>
<td>Laser pulse duration</td>
<td>ns</td>
<td>Excitation laser</td>
</tr>
</tbody>
</table>

Table 2.1 Laser and material parameters affecting OUT efficiency.

### 3.1. Absorbing Materials

A good absorbing material for OUTs should have high optical absorption, low light reflection, and high thermal conductivity [168]. Low reflection and high absorption allow the majority of the excitation laser energy to enter into and then be absorbed by the material, which favors the thermal energy buildup in the material. Owing to their low thermal expansion coefficient, embedding the absorbers in elastomers with a high thermal expansion coefficient can significantly improve the efficiency of the OUT. In such a case, the thermal conductivity of the absorbing material is another critical consideration when it comes to building an OUT [86]. To further understand the absorbers' thermal conductivity, let’s consider an OUT consisting of absorbing and elastomer particles, as shown in Fig. 2.1. The absorbing particles absorb the excitation laser, which is subsequently converted into heat. Some of that heat is transferred into the surrounding elastomer particles. The output
acoustic wave is determined by combining the acoustic wave produced by both particles. In the case where the absorber's thermal expansion coefficient is low (e.g., metal-based and carbon-based materials), the contribution of the absorber to the acoustic wave is negligible, and the thermal energy in the absorber is wasted [168]. Thus, it is critical to reduce the absorber's thermal energy ($\gamma_A$) and increase the elastomer's thermal energy ($\gamma_E$). By solving the heat conduction equation, the fraction of the thermal energy ($\gamma_E$) can be estimated as [169]:

$$\gamma_E = \frac{1}{\frac{(\rho C_p)_A}{(\rho C_p)_E} d_A + 1}$$

Eq. 2.2

where $\rho$ is the density, $C_p$ is the specific heat capacity at constant pressure, $d_A$ is the thickness of the absorber, $l_E$ is the heat penetration depth of the elastomer. Eq. 2.2 shows that the ratio of the specific heat capacities is critical to increasing $\gamma_E$. Moreover, Eq. 2.2 states that decreasing the absorber thickness $d_A$ consequently decreases the absorber's specific heat capacity, thus increasing $\gamma_E$. From the thermal conduction perspective, nanoscale absorbers are desirable for their low specific heat capacity and fast thermal conduction rate [168]. Therefore, in order to achieve a high PA efficiency in OUTs, absorbing materials with high thermal conductivity are preferred [170].

3.1.1. Metal-based Absorbing Materials

Metal films have been used in PA applications since 1963 [171]. However, due to the low thermoelastic coefficient, their PA efficiency was extremely low [172]. Moreover, due to the difference in the acoustic impedance between the metallic film and the surrounding
medium, most of the generated ultrasound waves cannot propagate out of the metal film. However, mixing metallic particles with PDMS would significantly enhance the overall PA efficiency [100]–[103], [173].

3.1.2. Carbon-based Absorbing Materials

Another type of absorber commonly used in the OUT development is carbon-based materials. In 1880, Alexander Graham Bell used a fully modulated beam of sunlight to transmit the sound signal (i.e., speech) and lampblack particles to act as a receiver. His invention, which is known as the photophone, was the first introduction to the photoacoustic effect [88]. Since then, carbon-based materials have been widely adopted in the PA research community. Highly absorbing carbon materials that have been used in PA applications include carbon black (CB), carbon nanotube (CNT), carbon nanofibers (CNF), and candle soot (CS) nanoparticles [32]–[44].

Among these carbon-based nanomaterials, CS nanoparticles have demonstrated the most superior performance, in terms of PA efficiency, due to their 3D nanoscale profile and branch-like structure [92]. These distinctive features allow CS nanoparticles to have fast heat transfer to the surrounding thermal expansion medium (PDMS), thus improving the efficiency of generating PA waves. Moreover, due to its nanoscale size, CS nanoparticles have the largest surface-to-volume ratio than other carbon-based nanoparticles [95].

To date, CS nanoparticles have been used in various applications. For example, they are used to build Na⁺-storage anodes for dual-ion batteries [182], optical sensors for infrared (IR) light [183], and fluorescent tracers for imaging [184]. Several research groups have shown interest in using CS nanoparticles to produce OUTs due to their simple synthesis.
process, high absorption, rapid heat transfer, and nanoscale size [90], [95], [99], [178], [181], [185], [186]. Recently, the CS nanoparticles and PDMS composite show a significantly improved PA efficiency compared to other absorbers. Chang et al. fabricated an OUT by placing a glass slide with uncured PDMS onto another glass slide with CS nanoparticles to produce a planar OUT [181]. The fabrication process was improved later by directly spin-coating low-viscosity PDMS on a glass slide covered by a CS nanoparticle layer [92]. Moreover, CS nanoparticles have also been coated on optical fibers to fabricate a miniature focused OUT [185].

3.2. Elastomeric Material

Elastomeric materials in an OUT are used to convert the thermal energy, produced by the absorbing material, to mechanical energy through a process known as the thermoelastic effect [160]. They are also referred to as 'the surrounding material' in OUTs. The volume expansion of a material can be described by the thermoelastic equation [187]. That is:

$$\Delta V = \beta \theta V$$

Eq. 2.3

where $\beta$ is the thermal expansion coefficient, $\theta$ change in the temperature, and $V$ is the initial volume. PDMS is one of the elastomeric materials processing a very high thermal expansion coefficient [188], which is $300 \times 10^{-6} \, ^{\circ}\text{C}^{-1}$. In comparison, the thermal expansion coefficients of metals such as aluminum (Al) and copper (Cr) (i.e., $24 \times 10^{-6} \, ^{\circ}\text{C}^{-1}$ and $17 \times 10^{-6} \, ^{\circ}\text{C}^{-1}$, respectively) are significantly lower. Having a high thermal expansion coefficient means that, under a constant thermal energy input, PDMS can expand more than other materials, thus generate higher acoustic pressure. This characteristic explains why PDMS is a popular choice in an OUT development. Besides, PDMS is the preferred elastomeric
material in building OUTs for biomedical applications because of its biocompatibility [189].

4. Research Objective

While various groups have demonstrated the feasibility of producing OUTs for imaging and therapeutic purposes, a systematic evaluation of the interplay between the design parameters of an OUT and its PA efficiency has not yet been carried out. Therefore, the design process of OUTs remains highly empirical, and the published results are difficult to reproduce. More importantly, a clear guide to optimizing the performance of an OUT is lacking. The research work presented in this chapter aims to bridge this knowledge gap. Specifically, a series of experimental studies were carried out to quantitatively assess the effects of (1) the absorber type, (2) the structure of the transmitter, (3) the absorption coefficients and the thickness of the absorbing material, (4) the thickness of the elastomeric material, and (5) the thermomechanical properties of the elastomeric material on the PA efficiency of an OUT. To limit the scope of this exploration, the absorbing materials used to build OUTs were CB and CS nanoparticles; the elastomeric material was PDMS.

5. Materials and Methods

5.1. Experimental Setup

The experimental setup used to quantify the ultrasound waves generated by an OUT is shown in Fig.2.2. All the measurements were conducted with both the OUT and the hydrophone immersed in water at room temperature (22°C). The optical irradiation source was a 532 nm pulsed laser (SPOT-10-200-532, ELFORLIGHT) with a pulse width of 1 ns
and a repetition rate of 10 kHz. The laser beam was focused onto the OUT through the glass water container using a 30-mm focal length aspherical lens. A calibrated high-frequency hydrophone (HGL-0200, Onda), with an aperture size of 200 µm and a center frequency of 40 MHz, was used to record the generated ultrasound waves. The hydrophone was attached to a precision XYZ linear translation stage to provide accurate scanning and positioning of the hydrophone during the experiments. The distance between the hydrophone and the OUT was maintained at 15 mm in all experiments. The hydrophone signals were amplified by a preamplifier (AG–2010, Onda) and then recorded by a digital oscilloscope (TEKTRONIX TDS 2014B) with a sampling rate of 1 GS/s.

Fig.2.2: The experimental setup for the acoustic characterization of OUTs.

5.2. OUT Fabrication Processes

Two OUT fabrication methods employed in this study are illustrated in Fig.2.3. To fabricate a single-layer OUT, the absorber-PDMS mixture was spin-coated on a glass
substrate. To fabricate a two-layer OUT, the absorbers and PDMS were coated sequentially onto the glass substrate.

Fig. 2.3: The structures of the optical ultrasound transmitters (OUT) fabricated in this study. (a) Single-layer OUT, and (b) two-layer OUT.

5.2.1. Fabrication of Single-Layer OUT

For the fabrication of a single-layer OUT, two carbon-based absorbers were used. The first one was a commercially available CB Nanopowder (US1067, US Research Nanomaterials Inc.), with a particle diameter of around 150 nm. The second one was synthesized CS nanoparticles with an average particle diameter of around 30 nm. These CS nanoparticles were collected from a paraffin candle using a flame synthesis method [95].

In this study, PDMS was used as the surrounding material of the fabricated OUTs. It was prepared using the 10:1 ratio of the elastomer base and the curing agent (Sylgard 184, Dow Corning, Midland, MI, USA). CB particles were mixed with the liquid PDMS in a 1:5 w/w ratio to form the CB-PDMS composite. The CS nanoparticles collected on the metal plate were then carefully removed and mixed with liquid PDMS in a 1:5 w/w ratio to form the CS-PDMS composite.
The two composites were then degassed for 30 min in a vacuum chamber to remove air bubbles. Next, the mixtures were spin-coated directly on a microscope glass slide using a spinner (SCK-300P Spin Coater, Instras Scientific, USA) to form a thin homogeneous layer. The rotation rate and the spin duration were 2700 RPM and 15 minutes, respectively. Finally, the samples were placed in an oven to cure for 30 minutes at 125°C and form the CB-PDMS OUT (T1) and CS-PDMS OUT (T2). The complete fabrication process of single-layer OUTs is graphically illustrated in Fig.2.4.

Fig.2.4: The fabrication process of single-layer OUTs, where the absorber-PDMS mixture is spin-coated on a glass substrate creating a homogeneous thin film.

5.2.2. Fabrication of Two-layer OUT

To fabricate two-layer OUTs, a sequential coating approach was employed, as illustrated in Fig.2.5. The fabrication process started with a flame synthesis method using a paraffin wax candle [95]. A glass slide was placed within the candle flame, at a predetermined distance from the wick, to collect CS nanoparticles for a fixed duration, which yielded an absorption layer. In the meantime, the PDMS was prepared by mixing the elastomer base and the curing agent at a specific w/w ratio and then degassed for 30 min in a vacuum chamber. The liquid PDMS was slowly poured onto the glass slides with CS nanoparticles; these slides were left alone for about 30 min to allow PDMS to permeate the CS nanoparticle layer. The glass slides were then spanned to remove the excessive PDMS as
well as produce an even-thickness PDMS layer. Finally, the slides were placed in a 125°C oven for 30 minutes to cure PDMS, which yielded two-layer OUTs for the experimental studies.

![Fig.2.5: The fabrication process of a two-layer OUT, where the absorber is coated first, and then the PDMS mixture is spin-coated on a glass substrate creating two individual layers.](image)

To find the optimal transmitter design, the effects of the optical, material, and geometric properties of an OUT had to be individually evaluated. For this reason, six groups of OUTs were designed and fabricated. Each group provided an opportunity to investigate the contribution of one specific property (e.g., absorber type) to the characteristics of the ultrasound waves generated by an OUT. Details about these six groups of samples are provided as follows:

- **G1**: OUTs with different optical absorber types. Two single-layer OUTs: CB-PDMS OUT \((T_1)\) and CS-PDMS OUT \((T_2)\), were fabricated using the exact fabrication process, base materials, and curing conditions. This OUT group was used to investigate the efficiency of CB and CS nanoparticles with respect to the PA generation.

- **G2**: OUTs with different structural characteristics. Here a single-layer CS-PDMS OUT \((T_2)\) and a two-layer CS-PDMS OUT \((T_3)\) were fabricated. Both OUTs had a similar optical absorption spectrum and thickness. The PDMS composition of both OUTs was also maintained at 10:1 (elastomer base curing agent ratio). The curing condition was
kept constant for both transmitters. The characteristics of the ultrasound waves generated by these two transmitters were compared to determine their advantages and disadvantages.

- **G3**: OUTs with CS nanoparticles obtained at different collection sites. It has been shown that the physical and chemical properties of CS nanoparticles vary in accordance with the deposition location in a flame synthesis process [190]. To test the effects of CS nanoparticle properties on the PA generation, two sets of OUTs were fabricated. In the first set, the CS nanoparticles were collected from the top of the candle flame (T4). In the second set, the CS nanoparticles were collected from the middle of the candle flame (T5). Other fabrication parameters and processes, such as the collection time, the PDMS composition, and the curing process, were kept the same for both transmitter groups.

- **G4**: OUTs with different PDMS layer thickness. Five OUTs were fabricated following the two-layer fabrication process depicted in Section 5.2.2. The collection time and the deposition location of CS nanoparticles, the PDMS composition, and the curing condition were kept the same. Five different PDMS layer thicknesses were obtained by setting the spin-coating speed at 1000, 2000, 2400, 2700, and 3000 RPM (T6, T7, T8, T9, and T3). The thickness of each transmitter was determined by measuring its cross-section thickness under an optical microscope (T610C-IPL-14M3, AmScope).

- **G5**: OUTs with different PDMS mechanical properties. It has been shown that changing the elastomer base to curing agent ratio of a PDMS matrix would modify its mechanical properties [191]. To understand how this characteristic would impact the PA generation, three OUTs were fabricated following the same two-layer fabrication
process. The elastomer base to curing agent ratio of PDMS used in these transmitters was varied from 5:1 (T10), 10:1 (T3), to 30:1 (T11).

- **G6**: OUTs with varying CS nanoparticle collection time. It has been shown that the thickness of the CS nanoparticle layer could be controlled by varying the collection time [95]. To understand the effects of the thickness of the CS nanoparticle layer on the PA generation, five OUTs were prepared using the same two-layer fabrication process. The collection time was varied from 10, 20, 30, 40, and 100 sec among those five samples (T12, T13, T14, T15, and T16).

The physical characteristics of all the fabricated OUTs used in this dissertation are summarized in Table 2.2.

### 5.3. Study Protocols

In each comparison study, the performance of each OUT was measured using the setup depicted in Fig.2.2 and quantitatively analyzed. To achieve a high signal to noise ratio (SNR), the ultrasound waves produced by each OUT were recorded repeatedly for 128 times and then averaged to produce a representative ultrasound wave. From the representative ultrasound wave, the peak amplitude ($P_{max}$), the peak frequency ($f_0$), and the spectrum bandwidth ($f_{BW}$) were obtained. The definitions of $P_{max}$, $f_0$, and $f_{BW}$ are illustrated in Fig.2.6. These parameters were compared among different OUT types using either parametric or non-parametric statistical methods.
<table>
<thead>
<tr>
<th>Variable Examined</th>
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<th>Structure</th>
<th>PDMS Composition</th>
<th>Thickness (µm)</th>
<th>Collecting Location</th>
<th>Concentration/Collection Time</th>
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<td>CB</td>
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<td>10:1</td>
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<tr>
<td></td>
<td>T2</td>
<td>CS</td>
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<td>15.0±0.3</td>
<td>-</td>
<td>20%</td>
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<tr>
<td>Structure</td>
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<tr>
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<td>CS</td>
<td>Two-layer</td>
<td>10:1</td>
<td>68.0±2.0</td>
<td>Top</td>
<td>10 Sec</td>
</tr>
<tr>
<td></td>
<td>T5</td>
<td>CS</td>
<td>Two-layer</td>
<td>10:1</td>
<td>35.0±5.0</td>
<td>Middle</td>
<td>10 Sec</td>
</tr>
<tr>
<td>PDMS Thickness</td>
<td>T6</td>
<td>CS</td>
<td>Two-layer</td>
<td>10:1</td>
<td>150±0.7</td>
<td>Middle</td>
<td>10 Sec</td>
</tr>
<tr>
<td></td>
<td>T7</td>
<td>CS</td>
<td>Two-layer</td>
<td>10:1</td>
<td>37.5±0.4</td>
<td>Middle</td>
<td>10 Sec</td>
</tr>
<tr>
<td></td>
<td>T8</td>
<td>CS</td>
<td>Two-layer</td>
<td>10:1</td>
<td>23.1±0.7</td>
<td>Middle</td>
<td>10 Sec</td>
</tr>
<tr>
<td></td>
<td>T9</td>
<td></td>
<td></td>
<td></td>
<td>15.1±0.8</td>
<td>Middle</td>
<td>10 Sec</td>
</tr>
<tr>
<td></td>
<td>T3</td>
<td></td>
<td></td>
<td></td>
<td>12.8±0.3</td>
<td>Middle</td>
<td>10 Sec</td>
</tr>
<tr>
<td>PDMS Composition</td>
<td>T10</td>
<td>CS</td>
<td>Two-layer</td>
<td>5:1</td>
<td>11.0±0.7</td>
<td>Middle</td>
<td>10 sec</td>
</tr>
<tr>
<td></td>
<td>T3</td>
<td>CS</td>
<td>Two-layer</td>
<td>10:1</td>
<td>12.8±0.3</td>
<td>Middle</td>
<td>10 sec</td>
</tr>
<tr>
<td></td>
<td>T11</td>
<td>CS</td>
<td>Two-layer</td>
<td>30:1</td>
<td>10.1±1.5</td>
<td>Middle</td>
<td>10 sec</td>
</tr>
<tr>
<td>CS Concentration</td>
<td>T12</td>
<td>CS</td>
<td>Two-layer</td>
<td>30:1</td>
<td>12.8±0.3</td>
<td>Middle</td>
<td>10 Sec</td>
</tr>
<tr>
<td></td>
<td>T13</td>
<td>CS</td>
<td>Two-layer</td>
<td>30:1</td>
<td>17.5±0.3</td>
<td>Middle</td>
<td>20 Sec</td>
</tr>
<tr>
<td></td>
<td>T14</td>
<td>CS</td>
<td>Two-layer</td>
<td>30:1</td>
<td>22.1±0.5</td>
<td>Middle</td>
<td>30 Sec</td>
</tr>
<tr>
<td></td>
<td>T15</td>
<td>CS</td>
<td>Two-layer</td>
<td>30:1</td>
<td>25.8±0.4</td>
<td>Middle</td>
<td>40 Sec</td>
</tr>
<tr>
<td></td>
<td>T16</td>
<td>CS</td>
<td>Two-layer</td>
<td>30:1</td>
<td>73.3±0.5</td>
<td>Middle</td>
<td>100 Sec</td>
</tr>
<tr>
<td>Reproducibility</td>
<td>T17</td>
<td>CS</td>
<td>Two-layer</td>
<td>30:1</td>
<td>18.6±0.6</td>
<td>Middle</td>
<td>20 Sec</td>
</tr>
<tr>
<td></td>
<td>T18</td>
<td>CS</td>
<td>Two-layer</td>
<td>30:1</td>
<td>17.5±0.3</td>
<td>Middle</td>
<td>20 Sec</td>
</tr>
<tr>
<td></td>
<td>T19</td>
<td>CS</td>
<td>Two-layer</td>
<td>30:1</td>
<td>20.1±0.2</td>
<td>Middle</td>
<td>20 Sec</td>
</tr>
<tr>
<td></td>
<td>T20</td>
<td>CS</td>
<td>Two-layer</td>
<td>30:1</td>
<td>22.0±0.3</td>
<td>Middle</td>
<td>20 Sec</td>
</tr>
</tbody>
</table>

Table 2.2: The physical characteristics of all the fabricated transmitters used in this dissertation.
6. Results and Discussion

The experiments depicted in this chapter were carefully designed to answer two main questions:

1) Can the characteristics of the ultrasound waves generated by an OUT be controlled by varying the excitation laser parameters?

2) Can the characteristics of the ultrasound waves generated by an OUT be controlled by varying the OUT properties and configurations?

In the experiments designed to answer the first question, the explored laser parameters included pulse width, pulse energy, and fluence. The outputs of these experiments are summarized in Section 6.1. In the experiments designed to answer the second question, the explored properties of OUT included: absorber type, CS nanoparticle deposition location, PDMS layer thickness, PDMS composition, CS collection time, and backing material. The outputs of these experiments are summarized in Section 6.2.

6.1. Effect of Laser Properties

In this section, the effects of laser parameters, including pulse energy, fluence, and pulse duration, on the characteristics of ultrasound waves generated by an OUT will be presented. It should be noted that all the results presented here were obtained using the OUT T1 (see Table 2.1).
Fig. 2.6: (a) Photograph of the PA measurement setup used in this study. The distance between the hydrophone and the OUT was maintained at 15 mm throughout the study. (b) A typical ultrasound wave detected by a broadband hydrophone. The peak-to-peak amplitude was used to estimate the corresponding maximum pressure $P_{\text{max}}$. (c) A typical frequency spectrum of the ultrasound wave generated by an OUT. The peak of this frequency spectrum $f_0$ is located at 32 MHz, and its bandwidth $f_{\text{BW}}$ is represented by the distance between the points $f_H$ and $f_L$ on the frequency spectrum where the amplitude is equal to half of the maximum amplitude (i.e., FWHM).
6.1.1. Pulse Energy

Fig. 2.7 shows the maximum pressure $P_{\text{max}}$, the center frequency $f_0$, and the spectrum bandwidth $f_{\text{BW}}$ of the ultrasound waves generated by the OUT T1 as a function of laser pulse energy. It was observed that $P_{\text{max}}$ increased from 0.009 ± 0.001 to 0.041 ± 0.002 MPa when the laser pulse energy increased from 0.94 to 2.5 µJ. This positive correlation appears to be linear, which is supported by the PA theory detailed in Section 2. According to Eq. 2.1, increasing the laser energy ($E_p$) will result in an increase in the amplitude of the PA pressure ($P_{PA}$), as shown in Fig. 2.7 (a). Similar linearity between the amplitude of the ultrasound wave, generate through the PA effect, and the incident laser energy has been reported by various groups [105], [181]. This linear relation is attributed to the amount of energy delivered by the pulse. It should be noted that, while the laser energy is positively correlated to the PA pressure, it should not cause photoablation to the OUT material [95]. $f_0$ and $f_{\text{BW}}$, on the other hand, did not change significantly with the increase or decrease of laser pulse energy. This observation is expected since all these results were produced from the same transmitter (i.e., T1).
Fig. 2.7: Effects of laser pulse energy on (a) the maximum pressure $P_{\text{max}}$, (b) the center frequency $f_0$, and (c) the spectrum bandwidth $f_{\text{BW}}$ of the ultrasound waves generated by a single-layer CB-PDMS OUT (T1).

6.1.2. Laser Fluence

Fig. 2.8 shows the maximum pressure $P_{\text{max}}$, the center frequency $f_0$, and the spectrum bandwidth $f_{\text{BW}}$ of the ultrasound waves generated by the OUT T1 as a function of laser fluence. In this experiment, the laser fluence was elevated from 0.8 to 3.1 nJ/ mm$^2$ by altering the laser spot size on the transmitter. Here, the diameter of the laser spot was determined by the full width half maximum (FWHM) of the laser intensity profile; the laser fluence was calculated by the ratio of the laser pulse energy (J) and the area of the laser spot (mm$^2$). It was observed that $P_{\text{max}}$ was 0.014 ± 0.001 MPa when the laser fluence was
0.8 nJ/mm². When the laser was focused, the laser fluence was increased by four-folds, but $P_{\text{max}}$ was only increased by three-folds (i.e., $0.043 \pm 0.003$ MPa). Similar nonlinearity between the amplitude of the ultrasound wave, generate through the PA effect, and the incident laser fluence has been reported by various groups [192]–[195]. The nonlinearity is likely attributed to the changes of the thermophysical parameters (i.e., thermal expansion coefficient, speed of sound, density, and heat capacity) due to significant temperature rise [193]. However, the exact interaction mechanism between the ultrasound amplitude and laser fluence is not fully understood, and further studies are needed. Moreover, it was also observed that the nonlinear increase in the ultrasound amplitude was accompanied by a change in the temporal profile of the ultrasound wave and, consequently, a change in its frequency characteristics. The 0.8 nJ/mm² fluence yielded ultrasound waves with a center frequency $f_0$ of $18.411 \pm 1.041$ MHz. Increasing the laser fluence to 3.1 nJ/mm² would shift $f_0$ to $23.639 \pm 0.886$ MHz. The variation in $f_0$ and $f_{\text{BW}}$ with the incident laser fluence has been reported as well [181], [196].
b) Fig. 2.8: Effects of laser fluence on (a) the maximum pressure $P_{\text{max}}$, (b) the center frequency $f_0$, and (c) the spectrum bandwidth $f_{\text{BW}}$ of the ultrasound waves generated by a single-layer CB-PDMS OUT ($T_1$). * indicates statistically significant difference (p < 0.05).

6.1.3. Laser Pulse Duration

Two laser systems were used in this experiment; a 905 nm nanosecond laser (L-CUBE-9-40/200-30/100) with a pulse width tunable between 30 ns and 116 ns and a 532 nm pulsed laser (SPOT-10-200-532, ELFORLIGHT) with a fixed pulse width of 1 ns. The selection of these laser systems enabled the investigation of the effects of pulse duration (1 ns to 116 ns) on the characteristics of the ultrasound waves generated. The same experimental setup
depicted in Fig.2.2 was used for both laser systems. Four pulse durations, 1, 27, 79, and 116 ns, were evaluated altogether.

The effects of laser pulse duration on the characteristics of the ultrasound waves generated by the OUT T1 are shown in Fig.2.9. The 116 ns laser pulses yielded ultrasound waves with a center frequency \( f_0 \) of \( 4.212 \pm 0.409 \) MHz. Reducing the pulse duration to 30 ns would elevate \( f_0 \) to \( 8.319 \pm 0.813 \) MHz. Further, reducing the pulse duration to 1 ns would increase the center frequency \( f_0 \) to \( 23.852 \pm 2.161 \) MHz. This phenomenon was compared to the one prediction by the PA simulation found in [197], and a good agreement was observed between them. The results demonstrate that the laser pulse duration can be used to tune the PA temporal profile hence its corresponding frequency. The spectrum bandwidth of the ultrasound waves \( f_{BW} \) was also negatively affected by the laser pulse width. By varying the laser pulse width from 1 to 116 ns, \( f_{BW} \) decreased from \( 42.504 \pm 2.445 \) to \( 7.092 \pm 0.816 \) MHz.

Because the absorption coefficients of CB nanoparticles at the two laser wavelengths are different and the output energy of the two lasers could not be adjusted to the same level, the effect of the pulse width on \( P_{max} \) could not be objectively analyzed and hence are not presented. As shown in Fig.2.7, laser pulse energy does not affect \( f_0 \) and \( f_{BW} \). Therefore, it is safe to assume that the variations in \( f_0 \) and \( f_{BW} \) observed in Fig.2.9 can be directly attributed to pulse width differences.

6.2. Effects of OUT Properties

According to the PA theory, both laser and OUT properties would impact the characteristics of the ultrasound waves generated. The experiments investigating the effects
of laser parameters were carried out with constant OUT properties; their results were presented in the previous section. In this section, the results of the investigations on the effects of OUT properties are presented. These investigations examined several OUT properties, such as absorber type and the transmitter structure; they were carried out with fixed laser parameters (i.e., pulse width, pulse energy, and fluence were 1 ns, 2.5 µJ and 3.16 µJ /mm², respectively).

Fig.2.9: Effects of laser pulse duration on (a) center frequency $f_0$, and (b) spectrum bandwidth $f_{BW}$ of the ultrasound waves generated by a single-layer CB-PDMS OUT (T1).
6.2.1. Absorber Type

The OUT T1 and T2 were used in this experiment, and the outcomes are summarized in Fig.2.10. The maximum pressure $P_{\text{max}}$ measured from the OUT T2 is almost twice as much as that from the OUT T1. According to their absorption spectra, the absorption coefficient of the OUT T1 is actually higher than that of the OUT T2. This phenomenon suggests that CS nanoparticles are more superior to CB nanoparticles in terms of PA generation. This finding may be attributed to the smaller size of CS nanoparticles: the nano-scale CS particles shorten the thermal pathway and hence fasten the heat dissipation to the surrounding PDMS medium [181].

Ultrasound waves generated by the OUT T1 and T2 share a similar frequency profile, although these transmitters have different absorbers and thicknesses. The center frequencies $f_0$ of the ultrasound waves generated by the OUT T1 and T2 were 18.541 ± 2.732 MHz and 20.522 ± 0.315 MHz, respectively; the spectrum bandwidth $f_{\text{BW}}$ was 44.406 ± 5.643 and 45.057 ± 3.110 MHz, respectively. These properties were not statistically significantly different. These findings match with the reported experimental and theoretical results from other research groups [181], [198]. They also suggested that the frequency spectral characteristics of the ultrasound waves generated by an OUT are predominantly determined by the temporal profile of the laser pulse, as long as the OUT is not too thick.

6.2.2. Transmitter Structure

In the previous section, it was found that CS nanoparticles are far more superior to CB nanoparticles in terms of PA generation. In the subsequent study, the effects of the transmitter structure, that is single-layer vs. two-layer, on ultrasound wave characteristics
were investigated. The results of this investigation are summarized in Fig. 2.11. In general, ultrasound waves generated by the two-layer OUT (T3) possess more desirable qualities than those from the single-layer OUT (T2). The maximum pressure of the generated ultrasound waves was $0.071 \pm 0.002$ for the OUT T2 and $0.117 \pm 0.004$ MPa for the OUT T3 (Fig.2.11 (b)). In a single-layer OUT, the presence of CS nanoparticles within the PDMS medium may change the network of PDMS elastomer chains and hence affect its thermal expansion coefficient, which, in turn, reduces the PA pressure [199]. Mixing the CS nanoparticles with PDMS breaks their branch-like structure; however, the destruction of this structural characteristic does not seem to affect the absorption spectrum, as seen in Fig.2.11 (a). Moreover, it is challenging to obtain uniform mixing of carbon particles in PDMS since carbon particles tend to agglomerate within the PDMS matrix. This non-homogenous characteristic would lead to uneven local light absorption and hence local heating throughout a one-layer transmitter and degrade the PA generation.

![Absorbance vs Wavelength graph](image)
Fig. 2.10: Effects of absorber type on ultrasound waves generated by single-layer OUT T1 and T2. (a) The absorption spectra, (b) the maximum pressure $P_{\text{max}}$, (c) the center frequency $f_0$, and (d) the spectrum bandwidth $f_{\text{BW}}$ obtained from OUT T1 and T2. * indicates statistically significant difference (p < 0.05).
Changing the structural characteristic (i.e., one-layer vs. two-layer) of a transmitter also significantly impacted the center frequency of the generated ultrasound waves $f_0$. In this study, $f_0$ was $20.466 \pm 0.313$ MHz for the OUT T2 and $32.112 \pm 2.127$ MHz for the OUT T3; this difference is statistically significant. A statistically significant difference was also found in the spectrum bandwidth of the two OUTs. The OUT T2 bandwidth is much narrower ($43.349 \pm 1.040$ MHz) than the OUT T3 ($53.094 \pm 3.530$ MHz). The superior performance of the two-layer OUT compared to single-layer OUT may have been arising from the nonuniform mixing of the CS nanoparticles in the PDMS matrix and not from the acoustic attenuation as the total thickness of each OUT is similar. The generated ultrasound shape depends on the optical penetration depth, which depends on the optical attenuation coefficient. As the optical attenuation coefficient increase, the width of the generated ultrasound wave become narrow. However, for a low optical attenuation coefficient, the penetration depth increase, which generates a broader ultrasound signal [181].

### 6.2.3. CS Nanoparticles Collection Location

Based on the temperature, a candle flame can be divided into three distinctive regions: the top, the middle, and the inner portions. Interestingly, CS nanoparticles collected from each region possess unique intrinsic properties such as the amount of wax contained and electrical conductivity. CS nanoparticles collected from the tip of the candle flame contain the least amount of wax, thus exhibit higher conductivity and hydrophilic properties. On the contrary, CS nanoparticles collected from the inner (bottom) portion of the flame contain more wax materials; thus, they are non-conductive and superhydrophobic [200]. To understand the effects of the collection location-dependent intrinsic properties of CS
nanoparticles on the PA generation, two two-layer OUTs, \textbf{T4} and \textbf{T5}, were prepared using CS nanoparticles collected from the tip and the middle of the candle flame as seen in Fig.2.12. In the fabrication process of these two transmitters, the collection time, PDMS composition, and spin-coating parameters were kept the same.
Fig. 2.11: Effects of the transmitter structure on ultrasound waves generated by the OUT T2 and T3. (a) The absorption spectra, (b) the maximum pressure $P_{\text{max}}$, (c) the center frequency $f_0$, and (d) the spectrum bandwidth $f_{\text{BW}}$ obtained from the OUT T2 and T3. * indicates statistically significant difference ($p < 0.05$).

Fig. 2.12: CS nanoparticles collecting by placing the glass substrate (a) on the top of the candle flame and (b) in the middle of the candle flame. The thickness of the CS nanoparticles layer was 55 µm for T4 and 22 µm for T5.
As seen in Fig. 2.13, the maximum pressure $P_{max}$ of the ultrasound waves generated by OUT T4 and T5 was $0.153 \pm 0.007$ and $0.154 \pm 0.004$ MPa, respectively. The frequency spectra of the ultrasound waves from both transmitters share similar characteristics; their center frequency $f_0$ and bandwidth $f_{BW}$ were comparable (i.e., no statistically significant difference). These findings suggest that the site of CS nanoparticles collection does not influence the efficiency of multiple layers OUT. In other words, the alterations in electromechanical properties of CS nanoparticles do not negatively impact the process and the outcomes of ultrasound wave generation by a two-layer OUT.
6.2.4. PDMS Thickness

The spin-coating method employed in the transmitter fabrication process allows control of the thickness of the PDMS layer in a two-layer OUT. By keeping the elastomer base and curing agent ratio (10:1) and the spinning time (15 minutes) constant, the PDMS layer thickness of a two-layer OUT was varied from 150.213 ± 0.744 to 12.835 ± 0.319 μm by increasing the spinning speed from 1000 to 3000 RPM correspondingly. Based on this methodology, five two-layer OUTs, denoted as T6, T7, T8, T9, and T3, were fabricated. Their corresponding total thickness were 150.208 ± 0.771, 37.524 ± 0.409, 23.159 ± 0.766, 15.137 ± 0.844, and 12.807 ± 0.322 μm respectively. The CS absorbing layers of these transmitters were created by collecting CS nanoparticles in the middle region of the candle flame for 10 sec. These samples were used to investigate the effects of PDMS layer thickness of a two-layer OUT on PA wave generation.
The outcomes of the investigation of the PDMS layer thickness effects are summarized in Fig. 2.14. The absorption spectra of the transmitters, as shown in Fig. 2.14 (a), were very similar, which suggested the intrinsic properties of the absorption layers of these transmitters were well controlled in the fabrication process and would not be responsible for any characteristic changes in the ultrasound waves observed in this study. The maximum pressure $P_{max}$ of the ultrasound waves increased from $0.071 \pm 0.013$ to $0.107 \pm 0.011$ MPa when the thickness of the transmitter decreased from $150.206 \pm 0.711$ (T6) to $12.832 \pm 0.343 \mu m$ (T3). Moreover, the alteration in the PDMS layer and hence the overall transmitter thickness significantly affected the center frequency $f_0$ of the ultrasound waves generated; it shifted from 30 MHz to ~ 40 MHz when the thickness of the transmitter decreased from $150 \mu m$ (T6) to ~13 $\mu m$ (T3) as shown in Fig. 2.14 (b). Finally, the bandwidth of the ultrasound waves generated by a transmitter increased from $51 \pm 2.164$ to $61 \pm 1.072$ MHz when the PDMS layer thickness decreased. As the thickness of the PDMS layer increases, the maximum pressure, the center frequency, and the bandwidth of the ultrasound waves generated would decrease. This behavior may be explained by the attenuation effect introduced by the PDMS layer; acoustic attenuation of PDMS affects the high-frequency components of ultrasound waves more strongly than the low-frequency ones [201]. This hypothesis is supported by the publication of Hou et al., in which the authors concluded that the transmitter medium beyond the absorption depth (i.e., $1/\mu_a$) would not contribute to the PA generation but only attenuate acoustic wave generated [103].
Fig. 2.14: Effects of PDMS layer thickness on ultrasound waves generated by a two-layer OUT. (a) The absorption spectra, (b) the maximum pressure \( P_{\text{max}} \), (c) the center frequency \( f_0 \), and (d) the spectrum bandwidth \( f_{\text{BW}} \) obtained from OUT T6, T7, T8, T9, and T3. * indicates statistically significant difference (p < 0.05).

6.2.5. PDMS Composition

According to the photoacoustic theory, the PA pressure generated \( P_{\text{PA}} \) is positively related to the surrounding material's thermal expansion coefficient (\( \beta \)). Therefore, it should be possible to improve the PA efficiency of OUTs mentioned in the previous section by elevating the thermal expansion coefficient of PDMS. One way to modify \( \beta \) of PDMS is to change its elastic modulus \( E \). This approach was derived from the definition of thermal expansion coefficient \( \beta \) [191]. That is:

\[
\beta = \frac{\Gamma \rho C_p}{3E} \quad \text{Eq. 2.4}
\]

where \( \Gamma \) is Grüneisen coefficient, \( \rho \) is density, and \( C_p \) the specific heat capacity. Moreover, \( E \) of PDMS varies linearly with the elastomer base and curing agent ratio \( n \) [191]:

\[
E = \frac{20 \text{ MPa}}{n} \quad \text{Eq. 2.5}
\]
Based on Eq. 2.4 and 2.5, it is apparent that $\beta$ of PDMS can be increased by increasing the elastomer base and curing agent ratio ($n$). Using this insight, three two-layer OUTs were fabricated using PDMS with $n = 5:1$, 10:1, and 30:1 (i.e., $T_{10}$, $T_3$, and $T_{11}$). The other properties of these transmitters, such as the absorption layer thickness, were kept the same. Their efficiency in generating ultrasound waves was evaluated using the setup depicted in Fig. 2.2. The outcomes of this investigation are presented as follows.

The effects of PDMS composition on the characteristics of the ultrasound waves generated by a two-layer OUT are shown in Fig. 2.15. When $n$ was increased from 10:1 to 30:1, the maximum pressure of the ultrasound waves $P_{max}$ increased from $0.087 \pm 0.006$ to $0.104 \pm 0.012$ MPa. On the other hand, $P_{max}$ decreased to $0.075 \pm 0.002$ MPa when $n$ was reduced from 10:1 to 5:1. These changes were found to be statistically significant. This observation validated the hypothesis derived from Eq. 2.4 and 2.5: an increase in $n$ of PDMS would lead to an increase in its $\beta$ and hence improve the PA efficiency of a PDMS based transmitter. Since the absorption spectra and the thicknesses of all these transmitters are comparable, as shown in Fig. 2.15 (a), the increase or decrease in $P_{max}$ observed in this study would definitely be attributed to the alterations of PDMS elastic modulus. Furthermore, no significant changes in the center frequency and the bandwidth of the ultrasound waves were found among the three samples in this investigation.
6.2.6. CS Nanoparticles Collection Time

To gauge the effects of CS nanoparticle collection time on PA generation of a two-layer OUT, five transmitters \( T_{12}, T_{13}, T_{14}, T_{15}, \) and \( T_{16} \) were fabricated using the standard procedure depicted in Section 5.2.2. The CS nanoparticle collection time was varied from 10, 20, 30, 40, and 100 sec, while the remaining fabrication parameters were kept constant. The absorption spectra of these five transmitters are shown in Fig. 2.16 (a). It clearly shows that CS nanoparticles are strong absorbers between 380 nm and 880 nm. Also, as expected, the absorbance of the transmitter increased continuously with the increase in collection time. Moreover, the thickness of the deposited CS nanoparticle (i.e., the absorbing layer) exhibited a positive linear relationship with the collection time, as shown in Fig. 2.16 (b). Using curve fitting, it was determined that the slope of this linear relationship was 0.7 µm/s.
Fig. 2.16: (a) Absorption spectra of the OUTs with five different collection times of CS nanoparticles. (b) The thickness of the CS nanoparticle layer as a function of collection time.

The effects of CS nanoparticle collection time on ultrasound wave characteristics are summarized in Fig. 2.17. According to the PA theory, higher optical absorption should lead to stronger ultrasound waves. The maximum pressure $P_{\text{max}}$ generated by the OUT T12 was $0.092 \pm 0.001$ MPa. When the collection time was doubled (i.e., 20 sec), the light absorption of the transmitter increased by $\sim 40\%$ and $P_{\text{max}}$ reached $0.109 \pm 0.001$ MPa. However, increasing the collection time beyond 20 sec actually led to a steady decline in $P_{\text{max}}$, as shown in Fig. 2.17 (a). This trend reversal may be explained by the fact that the collection time impacts both the thickness and the density of the CS nanoparticle layer [95].
When the density of CS nanoparticles in the absorbing layer becomes high, the diffusion of PDMS into the absorbing layer would be obstructed. In turn, this effect causes a reduction in heat dissipation from CS nanoparticles to PDMS and hence weakens the PA efficiency of the transmitter. Moreover, dense CS nanoparticles would also reduce the adhesion between particle and substrate as well as particle and PDMS, which, again, diminishes the PA efficiency of the transmitter [181]. A similar finding and conclusion have been reported in the study about the impact of carbon nanotube density on photoacoustic effects [177]. While altering $P_{\text{max}}$, CS nanoparticle collection time does not influence the center frequency, and the bandwidth of the ultrasound wave generated. Unlike PDMS, this behavior suggests that the CS absorbing layer does not attenuate acoustic waves [92].

**6.2.7. Backing Effect**

To test the backing effects, a two-layer OUT was fabricated following the procedure described in Section 5.2.2. Removing the CS-PDMS composition from the glass substrate while reserve its original shape was extremely difficult for thin OUTs; therefore, a thick transmitter (790 µm) was fabricated to facilitate the investigation. The same fabrication process for two-layer OUTs described in 5.2.2 was followed. In order to produce a thick sample, the rotation speed of the spin coating was set to 100 RPM.
Fig. 2.7: Effects of CS nanoparticle collection time on (a) the maximum pressure $P_{\text{max}}$, (b) the center frequency $f_0$, and (c) the spectrum bandwidth $f_{\text{BW}}$ of the ultrasound waves generated by a two-layer OUT from the OUT T12, T13, T14, T15, and T16. * indicates statistically significant difference ($p < 0.05$).
The effects of backing material on the characteristics of ultrasound waves generated by a two-layer OUT are shown in Fig. 2.18. The maximum pressure of the ultrasound waves generated by the OUT with a glass backing (hard-backing) was twice as high as the one with a water backing (soft-backing). Moreover, the average center frequency and the bandwidth of the ultrasound waves generated by the hard-backing OUT are slightly lower than those from the soft-backing transmitter. However, these differences are not statistically significant. These findings are in agreement with those reported by Lee et al. [202].

This enhancement introduced by the hard backing can be attributed to the constructive interference effect of acoustic waves. When a PA wave is generated in the absorbing layer of a transmitter, it travels in two directions: one toward the distal surface of the transmitter $\text{PAW}_f$ and the other toward the glass substrate $\text{PAW}_b$ (Fig. 2.19). $\text{PAW}_b$ is mostly reflected at the transmitter and glass interface because of the significant mismatch in acoustic impedance. Therefore, the measured PA wave from a transmitter is actually the summation of $\text{PAW}_b$ and $\text{PAW}_f$. Depends on the delay time between the two signals, construction or destruction interference could occur between two waves.

![Graph showing pressure comparison between Glass and Water](image-url)
Fig. 2.18: Effects of backing material on (a) the maximum pressure $P_{\text{max}}$, (b) the center frequency $f_0$, and (c) the spectrum bandwidth $f_{BW}$ of the ultrasound waves generated by a two-layer OUT. * indicates statistically significant difference ($p < 0.05$).
6.3. PA Wave Field Distribution

In addition to quantitative analyses of ultrasound waves generated, the acoustic fields of the fabricated OUTs were also characterized. The experimental setup used here was similar to the one depicted in Fig.2.2. The high-frequency bullet hydrophone was mounted on an XYZ linear translation stage, so the magnitude of the ultrasound waves on the YZ plane could be scanned, as shown in Fig.2.20 (a). When the hydrophone was placed at a specific scanning location on the YZ plane, the pulse laser was fired continuously, and the generated ultrasound waves were recorded 124 times and averaged. The acoustic field was measured over a rectangular area of 3 mm × 50 mm in the y and z-direction, as shown in Fig.2.20 (a). A representative result from this acoustic field characterization study is shown in Figure 2.20 (b). The measured PA pressure is the highest at the surface of the transmitter, which is about 1 MPa in this case. The PA pressure decreased continuously along the z-axis due to acoustic attenuation. The acoustic field is confined within a specific rectangular
region with a width (y-direction) of 1 mm, which matches the diameter of the excitation laser.

![Diagram of excitation laser and hydrophone](image)

**Fig. 2.20:** (a) The scanning process of the acoustic field of an OUT. (b) A representative acoustic field is measured from a single-layer transmitter.

### 6.4. Reproducibility

Upon optimizing the fabrication process and the configuration parameters of the two-layer OUT, the reproducibility of the fabrication process was tested. Four samples with identical configuration parameters were produced (i.e., OUT T17, T18, T19, and T20), and the
ultrasound waves generated by these OUTs were quantitatively analyzed and compared. The findings of this study are summarized in Fig. 2.21. The absorption spectra from the four transmitters were similar, suggesting that the CS nanoparticle collection procedure produced a consistent absorbing layer. The maximum pressure $P_{\text{max}}$ generated was $0.098 \pm 0.061$, $0.098 \pm 0.044$, $0.101 \pm 0.061$, and $0.088 \pm 0.012$ MPa for OUT $\text{T17}$, $\text{T18}$, $\text{T19}$, and $\text{T20}$, respectively. No statistically significant difference was detected in $P_{\text{max}}$ from the four OUTs. The same analysis conclusion was found in the comparison of the center frequency as well as the bandwidth of these four transmitters. For these reasons, the reproducibility of the fabrication process is successfully verified.
c)

Fig.2.21: Reproducibility of the fabrication process optimized for the fabrication of two-layer OUTs. (a) The absorption spectra, (b) the maximum pressure $P_{\text{max}}$, (c) the center frequency $f_0$, and (d) the spectrum bandwidth $f_{\text{BW}}$ obtained from the OUTs evaluated.

6.5. Stability

In the final characterization study, the output stability of the OUTs was evaluated. An OUT (T19) was excited continuously by the pulse laser for an hour, and the ultrasound waves generated were periodically recorded. The characteristics of these acquired ultrasound waves were collectively compared, and the outputs of this investigation are summarized in Fig.2.22. The maximum ultrasound wave pressure generated by OUT T19 was stable around $0.103 \pm 0.001$ MPa over the one-hour evaluation period. The center frequency and the spectrum bandwidth of this OUT remained around $30.392 \pm 0.641$ and $52.803 \pm 2.622$.
MHz, respectively, over the same period. No significant difference was found in these characteristics.

Fig. 2.22: The stability of the outputs of the two-layer OUTs (T19) fabricated in this study. (a) The maximum pressure $P_{\text{max}}$, (b) the center frequency $f_0$, and (c) the spectrum bandwidth $f_{\text{BW}}$ of the ultrasound waves generated by a two-layer OUT (T19) as a function of time.
7. Conclusion
A systematic investigation was carried out to understand how the mechanical, optical, and geometric properties of an OUT would influence the properties of generated ultrasound waves, including maximum pressure \( P_{\text{max}} \), center frequency \( f_0 \), and spectrum bandwidth \( f_{\text{BW}} \). In addition, the effects of laser parameters, including pulse width, pulse energy, and local fluence, on the properties of generated ultrasound waves were examined. The significant findings of these investigations are:

1. CS nanoparticles are more efficient than CB nanoparticles in terms of PA generation.
2. Two-layer OUTs produce more desirable ultrasound waves (i.e., high maximum pressure, high center frequency, and broad-spectrum bandwidth) than the single-layer ones. Moreover, they are much easy to manufacture.
3. The efficiency of the two-layer OUT in terms of maximum PA pressure would decrease when the thickness of the absorbing layer exceeds \( \sim 20 \) um and its absorbance greater than 2.5.
4. The efficiency of a two-layer OUT, in terms of maximum PA pressure, would increase with PDMS with lower elastic modulus is used.
5. The variations in the intrinsic properties of CS nanoparticles do not influence their efficiency in terms of PA wave generation.
6. A high acoustic impedance backing significantly improves the output of a two-layer transmitter.
7. The laser pulse width only influences the center frequency and the bandwidth of a transmitter.
These findings form a scientific guideline, based on which highly efficient OUTs with a
tunable center frequency and bandwidth can be designed and fabricated. Moreover, a
simple, cost-efficient, and reliable fabrication process was also developed in this study.
These two pieces of knowledge hold great potential for the future development of all-
optical ultrasound imaging.
Chapter 3: Optical Ultrasound Detector

1. Introduction

In ultrasound imaging, a piezoelectric material, such as piezoelectric crystals (e.g., quartz) and piezoceramics (e.g., PZT), is commonly used for detecting acoustic waves by converting pressure waves to measurable electrical signals (voltage). However, piezoelectric ultrasound transducers suffer from several limitations. From a quantitative ultrasound imaging point of view, one major limitation is their narrow spectrum bandwidth [203]. Another limitation is the manufacturing difficulties, especially for those very high-frequency transducers [72].

As an alternative to piezoelectric ultrasound detectors, optical-based ultrasound detectors hold promise to overcome these limitations. Optical ultrasound detection can be classified into two categories based on the mechanisms: interferometry and refractometry [105]. Interferometry methods detect the change in the optical interference patterns induced by ultrasound waves, where ultrasound waves change the mean free path or the optical wavelength of the probe beam and hence alter the interference results between the probe and the reference beams. On the other hand, refractometry methods rely on the change in the medium's refractive index induced by mechanical pressure, also known as the photoelastic principle [87]. In this detection mechanism, a probe beam is used to interrogate the local disturbance in the refractive index of a medium induced by ultrasound waves; the intensity, the deflection angle, or the phase alterations of the probe beam indirectly provides ultrasound wave information [105].

Several studies have shown that optical methods of ultrasound detection have several advantages over the traditional piezoelectric detectors, including broader spectrum
bandwidth, similar acoustic impedance to biological tissues, and insensitive to EMI. Therefore, they can be easily integrated into other diagnostic imaging modalities such as photoacoustic microscopy (PAM), photoacoustic endoscopy (PAE), and optical coherence tomography (OCT) [204]–[215]. Moreover, in contrast to the piezoelectric detectors whose sensitivity depends on the element size, the sensitivity of an optical-based ultrasound detector remains the same despite the reduction of the detector size [216]. Optical-based ultrasound detectors also offer the feasibility of noncontact implementation [217], [218]. Finally, several groups have demonstrated that the sensitivity of optical-based ultrasound transmitters is comparable or even better than piezoelectric transmitters [219]–[225].

Currently, the most representative optical-based ultrasound detectors are Fabry–Pérot Interferometer (FPI) [106]–[108], [111], [113], [114], [226]–[228], Micro-Ring Resonators (MRRs) [122], [229], and Fiber Bragg Gratings (FBGs) [230]–[234]. Nevertheless, these optical-based ultrasound detectors require a complex optical setup. Moreover, they are sensitive to background vibration and temperature fluctuations [235]–[237]. Besides, these optical-based ultrasound detectors are considered a point detection device. In order to perform a line or an area detection, either a detector array has to be built, which is complicated to fabricate, or a mechanical scanning approach has to be employed, which is time-consuming. Therefore, it would be beneficial to develop a simple and straightforward optical ultrasound detector that does not process these limitations.

Unlike the optical ultrasound detection mechanisms mentioned above, the probe beam deflection (PBD) technique shows its capabilities to detect the acoustic wave remotely, has a simple and straightforward optical setup, and is insensitive to the background and environmental vibrations [155]. The PBD technique is a line detector; it detects ultrasound
anywhere along the probe laser beam path. For this reason, the BPD technique is a good foundation for developing a line or 2D optical ultrasound detector. Finally, the PBD technique is immune to the ringing effect because the measurement occurs from the deflection of the probe laser beam following Snell's law, rather than vibrating the detector's membrane/construct [132], [142], [238]. To date, PBD technique has been employed to detect thermal effects [239], [240], shock wave-front [152], high-intensity focused ultrasound (HIFU) [145], and photoacoustic effects [139]–[141], [147], [153].

2. **Probe Beam Deflection Technique (PBD) Theory**

Acoustic waves are pressure waves that consist of crests and troughs. As an acoustic wave propagates through a medium, the medium's molecules are compressed at the crests and rarefied at the troughs [24]. In turn, these density changes cause an increase and decrease in the refractive index of the medium, known as the photoelastic effect [153] (Fig.3.1).

![Illustration of the photoelastic effect. The high-pressure region (compression) is colored in red, and the low-pressure region (rarefaction) is colored in green.](image)

Fig.3.1: Illustration of the photoelastic effect. The high-pressure region (compression) is colored in red, and the low-pressure region (rarefaction) is colored in green.
The working principle of the PBD technique is presented in Fig. 3.2. The front part of the acoustic wave, the crest, would produce a positive gradient index in the local medium, which behaves like a positive lens. Therefore, the rays passing through this medium region will be deflected toward the optical axis. On the other hand, the back part of the acoustic wave, the trough, would produce a negative gradient index in the local medium, which behaves like a negative lens. Hence, the rays passing through this medium region would be deflected away from the optical axis. The ray deflection induced by acoustic waves would affect the number of probe rays reaching the optical detector, which is usually placed on the optical axis of the probe beam, and hence its output.

The change in refractive index (Δn) of a medium induced by ultrasound pressure (Δp(x, y, z, t)) may be described by the Lorentz formula [145]. That is:

$$\Delta n = n_0 + K \Delta p(x, y, z, t)$$  \hspace{1cm} \text{Eq. 3.1}$$

where $n_0$ is the refractive index of the undistributed medium, and $K$ is the photoelastic coefficient ($\partial n / \partial p$). When $K$ and $\Delta p(x, y, z, t)$ are known, the degree of the probe ray deflection may be estimated using Fermat's principle [241].

The study presented in the chapter aims to design and develop a prototype optical ultrasound detector based on the PBD technique because it provides a more straightforward pathway to 2D, 3D, and 4D all-optical quantitative ultrasound imaging systems envisioned.

To achieve this goal, three specific tasks have been identified and have to be completed:

- Quantify the sensitivity of the optical ultrasound detector developed,
- Quantify the resolution of the optical ultrasound detector developed, and
- Confirm the capability of the optical ultrasound detector in terms of performing quantitative ultrasound measurements.
Fig.3.2: Graphic illustration of the working principle of the PBD technique. (a) Before the interaction between the acoustic wave and the probe beam rays, the rays travel parallel to each other through the medium. (b) The front part of the acoustic wave produces a positive gradient index, which deflects the rays towards the optical axis forming the positive lobe of the detector signal. (c) The acoustic wave's tail part produces a negative gradient index, which deflects the rays away from the optical axis forming the negative lobe of the detector signal. (d) As the acoustic wave propagates beyond the interrogation region, the probe beam rays return to their original paths.

3. Refractometric Ultrasound Detector (RUD) Design Concept

An ultrasound detector built upon the PBD technique, denoted as refractometric ultrasound detector (RUD), would comprise the following components:

1. A continuous wave (CW) light source that provides a collimated probe beam,
2. A coupling medium in which the probe beam interacts with the ultrasound waves, and
3. A photodetector that quantitatively detects the magnitude of the probe beam deflection.

The intrinsic characteristics of a RUD, including its sensitivity, would be strongly influenced by the properties of these elements. For the probe beam, the beam power, the beam diameter, and the beam profile were the primary considerations. For the
photodetector, the response time and sensitivity were essential. In the first RUD prototype, a diode laser and a high-speed photodetector, readily available in the lab, were used. The selection of a coupling medium was essential to the RUD building because the degree of probe beam deflection heavily depends on the properties of the coupling medium. In general, a good coupling medium should have the following characteristics: low absorption at the probe beam wavelength, high photoelastic coefficient, and high acoustic velocity. The other properties of the coupling media important to this research work were acoustic attenuation and acoustic impedance. Since this RUD was developed for quantitative ultrasound imaging (i.e., measuring the weak backscattering acoustic signals), its acoustic attenuation had to be low over a broad frequency spectrum. Finally, the coupling medium selected should have an acoustic impedance similar to that of biological tissues, allowing the reflected ultrasound waves, originated inside the tissue, to pass through the tissue-coupling media interface easily. Among all available materials for the coupling medium, water met every design criterion stated above. First of all, water has minimal absorption in the visible wavelength region [242]. Moreover, water's photoelastic coefficient is similar to other polymers (1.5×10⁻⁴ MPa⁻¹) [243]. Finally, the acoustic attenuation of water (0.0022 dB cm⁻¹ MHz⁻¹) is very low compared to other polymers (e.g., polymethyl methacrylate (PMMA), polystyrene, and PDMS) [244], [245], and its acoustic impedance is close to the average acoustic impedance of the soft tissue (1.48 MRayl) [26]. For these reasons, water was selected as the coupling medium in the prototype RUD.
4. Experimental Evaluation of The Prototype RUD

A benchtop test station, as depicted in Fig.3.3, was built to evaluate the functionality and sensitivity of the prototype RUD. As mentioned previously, the RUD consisted of three main components: a probe laser beam, a coupling medium, and a fast photodiode. A collimated diode laser (STR-640-A-HR-L01-30-S Coherent) was utilized as a probing beam; its wavelength was 640 nm and output beam diameter 4 mm. In order to alter the probe beam diameter, a pinhole was placed in front of the laser output. This approach would reduce the diameter of the probe laser beam to 2 mm without introducing diffraction. The laser was mounted on the side of a custom-made glass container with dimensions of 75 mm × 25 mm × 25 mm and a wall thickness of 1 mm; the probe beam was transmitted through the center of the container. A high-speed photodetector (PDA10A2 Thorlabs) with a spectrum bandwidth of 150 MHz was centered on the probe beam axis and measure the intensity profile fluctuation of the probe beam transmitting through the coupling medium. To control the portion of the probe beam delivered to the photodetector and to reduce the ambient light and specular reflection influences, a second pinhole was used. The diameter of this pinhole was kept the same as that in front of the laser. The output of the photodetector was displayed on a digital oscilloscope (TEKTRONIX TDS 2014B) and recorded for further processing and analysis.

To generate ultrasound waves, a 10 MHz unfocused PZT transducer (V312-SM Olympus) with a 6 mm diameter element was used. The transducer was carefully aligned such that its outputs were perpendicular to the probe beam axis. The distance between the PZT transducer tip and the probe beam was maintained at 15 mm. A pulser/receiver (STEVAL-IME013V1 STMicroelectronics) was used to drive the transducer with two cycles of 10
MHz square waves. By changing the voltage supplied to the transducer from 5 to 33 volts, the generated ultrasound pressure was elevated from 8.48 to 53.8 kPa, respectively. This voltage-to-pressure conversion was calibrated and verified using a hydrophone (HGL-0200, Onda).

Fig.3: The experimental setup used for characterizing and improving the prototype RUD. In this setup, the ultrasound waves were generated using a PZT transducer placed 15 mm above the probing beam. The PZT transducer was operated in a pulse-echo mode; the transducer generated an ultrasound pulse and then detected the ultrasound pulse reflected by the bottom of the glass container. This method allowed for a direct comparison between the PZT transducer measurements and those of the RUD.

A series of carefully designed experiments were systematically carried out to establish the critical know-how about building a RUD for an all-optical quantitative ultrasound imaging system. Specifically, these experiments aimed to uncover the pathways to improve the sensitivity of a RUD. In other words, they investigated the correlation between the RUD response and (1) the probe beam power, (2) the probe beam diameter, (3) the magnitude of the acoustic waves, (4) resolution, (5) the operation time (i.e., stability), and finally (6) quantitative backscattering measurements.
In the study of the effects of the probe beam power on the RUD response, the probe beam power was varied by changing the input voltage supplied to the diode laser from 0.6 mW to 6.4 mW. Other parameters, such as the ultrasound pulse characteristics and the probe beam diameter, were kept the same. The probe beam power effects on the RUD responses were quantitatively assessed in both the time and frequency domains.

In the study of the probe beam diameter effects, the probe beam diameter was varied from 4 mm to 2 mm by changing the size of a pinhole placed in front of the probe laser. Other parameters, such as the ultrasound pulse characteristics and the probe beam power, were kept the same. The probe beam diameter effects on the RUD responses were quantitatively assessed in both the time and frequency domains.

In the study of the RUD sensitivity, the driving voltage supplied to the PZT transducer was varied from 5 V to 33 V, and the corresponding RUD responses were recorded. The RUD responses were analyzed in both the time and frequency domains; the effects of the ultrasound wave magnitude on the RUD responses were quantitatively assessed. The sensitivity of the RUD was extracted from this analysis.

In the study of the RUD resolution, the pulser of the PZT transducer was programmed to produce two consecutive ultrasound pulses with a specific time delay between them. This acoustic wave template was used to simulate reflected ultrasound pulses from two separate objects. The time delay between the two pulses was varied from 0 ns to 1600 ns, which corresponded to 0 mm to 2.4 mm separation between the reflective objects. The RUD responses were analyzed by measuring the time-of-flight (TOF) between the two pulses in the time domain and then compared to the theoretical predictions.
In addition to the evaluation studies mentioned above, a phantom study was carried out to verify the RUD's capability in detecting backscattering signals (i.e., quantitative ultrasound). By placing the phantoms in the water container under the probe beam and the PZT transducer, as depicted in Fig. 3.4, the backscattering signals were able to be simultaneously detected by the RUD and the PZT transducer. The ultrasound phantoms used here were Polyvinyl Chloride Plastisol (PVCP) phantoms, which were fabricated following the procedure disclosed by Vogt et al. [246]. Briefly, a stock of PVCP solution was prepared by combining 8% v/v PVCP resin with a 90:10 volume ratio solution of Benzyl-butyl phthalate (BBP) and Di(2-Ethylhexyl) adipate (DEHA). The PVCP solution was then stirred for 30 minutes in a beaker through a magnetic stirrer and degassed in a desiccator chamber for 60 min. After degassing, different sizes of glass-lime spheres, ranging from 38 - 45 µm diameter to 70 – 90 diameter, were added and magnetically-stirred in for five minutes. The final solution was then poured into a round bottom flask and placed in a magnetically-stirred oil bath maintained at 190 °C. After 12-15 minutes. The solution was then poured into a lubricated aluminum mold and polymerized within 5 minutes after insertion into the mold.

In this chapter, the RUD responses were quantitatively compared to the PZT transducer in both the time domain (i.e., temporal profile) and the frequency domain (i.e., width and peak of the frequency response) to confirm that the RUD performed as well as PZT transducers in terms of ultrasound detection.
5. Results and Discussion

5.1. Effects of Probe Beam Parameters on RUD Responses

To understand the effects of probe beam parameters on the RUD responses, two parameters of the probe beam, power, and diameter, were separately investigated. The outputs of these investigations are presented in the next two sessions.

Fig.3.4: (a) A photograph of the experimental setup. (b) The schematic diagram of the experimental setup for the ultrasound backscattering measurements.
5.1.1. Probe Beam Power

To determine the effects of the probe beam power on the performance of the RUD, the response of RUD to the ultrasound waves, generated by the PZT transducer, were measured with the probe beam power varied from 0.6 mW to 6.4 mW. As the probe beam power increase from 0.6 to 6.4 mW, the maximum amplitude of the RUD signal increased linearly, as shown in Fig.3.5 (a). The temporal profile, the center frequency, and the spectrum bandwidth of the RUD signals, as shown in Fig.3.5 (b and c), remained constant. The results suggest (1) the photodetector of the RUD was operating in the linear region for those probe beam powers tested, (2) the probe beam profile did not change with the alterations in its output power. In addition, they also support the idea of improving the signal to noise ratio of RUD via increasing the probe beam power, which would be useful for weak ultrasound wave detection (i.e., quantitative ultrasound).

5.1.2. Probe Beam Diameter

The effects of the probe beam diameter on the RUD responses are summarized in Fig.3.6. With a 4 mm diameter probe beam, the time duration of the RUD signal, as shown in Fig.3.6 (a), was more than 2 µs. When the probe beam diameter was reduced to 2 mm, the time duration of the RUD signal was reduced to 1.25 µs. This phenomenon can be explained by the response duration of the RUD, which describes the time it takes for an ultrasound pulse to travel through the probe beam. Theoretically, the time required for two cycles of 10 MHz ultrasound waves to travel through a 4 mm window in water would be ~ 2.8 µs. However, the required time would drop to 1.4 µs when the window size is reduced to 2 mm. These theoretical predictions match the experimental results, which confirm the
response duration of a RUD is primarily determined by the probe beam diameter. Since the response duration of a RUD directly influences the axial resolution of an all-optical ultrasound imaging system, it is ideal to keep it close to a half duration, time-wise, of the ultrasound pulse generated by an optical ultrasound transmitter (OUT). In other words, the ideal probe beam diameter would be half of the ultrasound pulse width, distance-wise.

In the frequency domain, another effect of the probe beam diameter was revealed. The 4 mm probe beam yields a spectral bandwidth of 1.1 MHz and the 2 mm beam of 2.6 MHz, both of which were much smaller than the true spectral bandwidth of the acoustic waves generated by the PZT (10 MHz, measured by calibrated hydrophone). The primary reason behind this spectral bandwidth reduction is the extra cycles produced by the RUD, which is associated with the probe beam diameter [247]. Reducing the diameter further would further improve the bandwidth of the detected signal, which favors quantitative ultrasound measurement and imaging.

![Graph showing the relationship between RUD laser power and RUD amplitude](image)

\[ R^2 = 0.9977 \]
Fig. 3.5: (a) The effect of the probe beam power on the amplitude of the optical signal produced by the RUD, (b) the temporal profile of the RUD signals with various probe beam power levels, and (c) the corresponding frequency spectra of the RUD signals. The acoustic waves generated by the PZT had a 10 MHz center frequency, 4 MHz spectrum bandwidth, and a peak pressure of 45 kPa.
Fig. 3.6: The effects of the probe beam diameter on the ultrasound detection capability of the RUD. (a) The temporal profile of the RUD signals with a 4 mm and 2 mm probe beam diameter (b) the corresponding frequency spectra. The acoustic waves generated by the PZT had a 10 MHz center frequency, 4 MHz spectrum bandwidth, and a peak pressure of 45 kPa.

5.2. Acoustic Pressure VS RUD Responses

To determine the minimum acoustic pressure detectable by the prototype RUD, the peak acoustic pressure generated by the PZT transducer was adjusted by changing its driving voltage. The outcomes of this investigation are summarized in Fig. 3.7. As shown in Fig. 3.7 (a), there is a clear linear relationship between the driving voltage (0 to 33 V) and the peak acoustic pressure generated from the PZT transducer (0 to 45 kPa). Fig. 3.7 (b) shows the raw RUD signals induced by ultrasound waves with different peak acoustic pressures, which suggests the minimal detectable acoustic pressure of the RUD is as low as 8 kPa. The results are expected since the change in the refractive index is proportional to the amplitude of the ultrasound waves as described earlier. Fig. 3.7 (c) shows the corresponding frequency spectra of the raw RUD signals. The center frequency and the spectrum bandwidth of the RUD signals were apparently insensitive to the peak pressures of the ultrasound waves.
Fig. 3.7: (a) The relationship between the input voltage and the output pressure of the PZT. (b) The effects of the peak ultrasound pressure on the signals produced by RUD. The intensity of the RUD increases linearly with the ultrasound pressure. (c) The center frequency and (d) the spectrum bandwidth of the ultrasound waves detected by the RUD.

5.3. RUD Resolution

The axial resolution of the all-optical ultrasound imaging system built upon the RUD would be determined by the ability of the RUD to separate the echoes from two adjacent reflectors/back-scatterers. As described in the previous section, the RUD resolution is primarily influenced by its response duration, which is controlled by the probe beam diameter. This assertion was confirmed experimentally, as shown in Fig. 3.8. To simulate ultrasound reflection from two objects 2 mm apart along the z-axis, the pulser of the PZT transducer was programmed in such a way that two consecutive ultrasound pulses with a 1.33 µs delay were generated in each cycle (see Fig. 3.8 (a)). Here the distance between the reflective objects was calculated using the TOF difference between the two pulses multiplied by the speed of sound in water (1480 m/s). The responses of the RUD to these ultrasound pulses are presented in Fig. 3.8 (b and c). When the probe beam diameter was 4 mm, the RUD response did not show two distinct pulses (i.e., two separate echoes from the
objects). On the other hand, the RUD response with a 2 mm probe beam showed two distinguishable pulses (i.e., objects).

Fig. 3.8: The effect of probe beam diameter on the axial resolution of the RUD. (a) The PZT signal, (b) the RUD signal with a 4 mm diameter probe beam, and (c) the RUD signal with a 2 mm diameter probe beam.
To further evaluate the axial resolution of the RUD with a 2 mm diameter probe beam, the pulser was reprogrammed to produce two consecutive pulses with different delay times, and each pulse contains two cycles, as shown in Fig.3.9. The delay time was varied from 0 to 1600 ns, which was equivalent to 0 to 2.36 mm axial distance difference between the two pulses (i.e., reflective objects).

![Graph showing axial resolution for PZT transducer and RUD](image)

**Fig.3.9:** Axial resolution of (a) PZT transducer and (b) RUD.

The distance between the two pulses (i.e., objects) is determined based on the TOF method [248], where the distance between the two pulses ($S$) is determined from:
\[ S = v_{\text{water}} \times \Delta t \]  

Eq. 3.2

where \( v_{\text{water}} \) is ultrasound velocity in water (1480 m/s). The time delay (\( \Delta t \)) between the two pulses is determined by subtracting the time of the envelope peak (\( t_1 \)) from that of the envelope peak (\( t_2 \)), as illustrated in Fig. 3.10 (a). The RUD responses to the ultrasound pulses with different delay times are shown in Fig. 3.10 (b). The results suggest that the axial resolution of the RUD was not smaller than 0.5 mm. The axial resolution of the RUD could be further improved by decreasing the diameter of its probe beam.

Fig. 3.10: (a) TOF method used to determine the spatial distance between two pulses. (b) The delay and hence the distance between two ultrasound pulses are estimated by the RUD responses and the PZT response compared to the theoretical predictions.
5.4. Comparisons Between PZT Transducers and RUD

The ultrasound waves generated by the PZT transducer could be simultaneously detected by the two detection methods, as shown in Fig.3.3. This approach enabled a side-by-side comparison of the PZT transducer and the RUD. The signals detected by both detectors under the same test condition are compared in Fig.3.11 (a). The number of cycles in the signal from the PZT transducer was noticeably smaller than the one from the RUD. It was also noticed that, even with the probe beam diameter reduced to 2 mm, the temporal profile of the RUD signal still does not match that of the acoustic waves generated by the PZT transducer; the RUD signal contained many more cycles (5 cycles) than the actual acoustic waves (2 cycles). This can be easily explained by the long response duration of the RUD induced by the wide probe beam, as explained in the previous section.

The corresponding frequency spectra of both signals are shown in Fig.3.11 (b), where major differences between the PZT transducers and the RUD are identified. The center frequency of the RUD signal is 8.2 MHz; the center frequency of the PZT transducer signal, however, is 10 MHz (which matches the spec from the manufacturer). More importantly, there is a significant difference in the spectrum bandwidth between the two detectors. The spectrum bandwidth of the PZT transducer signal (4 MHz) is significantly wider than that of the RUD signal (2.6 MHz). This discrepancy is caused by the additional cycles introduced by the RUD, as explained before.
5.5. RUD Stability

The stability of the RUD was evaluated by analyzing its responses to the same ultrasound waves as a function of time. The outcomes of this investigation are summarized in Figure 12; the amplitude from the repeatedly measured RUD responses is shown in Fig.3.12 (a) and their corresponding center frequency and spectrum bandwidth in Fig.3.12 (b and c). The peak amplitude of the RUD signals did fluctuate slightly in a random fashion over the entire duration of evaluation (i.e., 1 hour). This behavior may be attributed to the instability of the probe laser beam or the PZT transducer. The center frequency and spectrum bandwidth of the RUD signals also remained the same during the entire duration of the
evaluation. These findings further confirmed the suitability of building an all-optical quantitative ultrasound imaging system based on the RUD proposed here.

Fig.3.12: The stability of the RUD responses over the duration of 1 hour. (a) The amplitude, (b) the center frequency, and (c) the spectrum bandwidth of the RUD signals acquired at different time points.
5.6. Backscattering Measurements

In this exploration, the backscattering signals from two scattering phantoms, one with 70-90 µm diameter soda-lime microspheres (Phantom 1) and the other with 38-45 µm diameter microspheres (Phantom 2), were simultaneously measured using the PZT transducer (reference) and the RUD. A representative backscattering signal detected by the RUD is shown in Fig.3.13 (a). The small oscillations in the signal after the echo from the front surface of the phantom were induced by the microspheres, as they did not exist in the phantoms without microspheres (Phantom 3) (Fig.3.13 (b)).

![Graph](image)

Fig.3.13: Typical backscattering signals from (a) the phantom with scattering microspheres and (b) the phantom without scattering microspheres.
Furthermore, the time-domain backscattering signals and their corresponding power spectra obtained from Phantom 1 and Phantom 2 are plotted in Fig. 3.14. Phantom 2 also showed a stronger location-dependency in its backscattering signals; the time domain signal from Location 3 is weaker than those from the other two locations, and its center frequency is also lower than the other two. This phenomenon may suggest the high inhomogeneity nature, in terms of scatterer distribution, of Phantom 2.

The peak frequencies of the backscattered signals from the phantoms were further compared with the theoretical predictions derived by Kruger et al. [249]. The results of this comparison are plotted in Fig. 3.15. The experimental results show that the increase in the scatterer size would decrease its corresponding peak frequency. The negative correlation was consistent with the theoretical predictions. However, due to the limited bandwidth of the RUD and the PZT transmitter in this experiment, the trend of the experimental results was not in complete agreement with that of the theoretical predictions.

![Amplitude vs Time Graph](image)
Fig. 3.14: (a) The backscattered signals detected from Phantom 1 and (b) their corresponding frequency spectra. (c) The backscattered signal detected from Phantom 2 and (d) their corresponding frequency spectra.
The correlations between the scatterer sizes and the peak frequency of the backscattering signals. The same negative correlation is found in both the theoretical predictions and the experimental results. However, a substantial discrepancy between the trends is found in the high peak frequency/small scatterer size region.

Next, to determine the minimum backscattering amplitude that could be detected by the RUD, the peak pressure of the incident acoustic waves generated by the PZT transducer was steadily decreased from 53.8 kPa to 0 kPa. The system setup is shown in Fig.3.16.

Fig.3.15: The correlations between the scatterer sizes and the peak frequency of the backscattering signals. The same negative correlation is found in both the theoretical predictions and the experimental results. However, a substantial discrepancy between the trends is found in the high peak frequency/small scatterer size region.

Fig.3.16: The experimental setup used to measure the minimum pressure that can be detected by the RUD.
The outcomes of this investigation are summarized in Fig.3.17. It is observed that the intensity of the backscattering signal decreased with the reduction of the incident ultrasound wave pressure. However, when the incident ultrasound wave pressure dropped below 17 kPa, the backscattering signal is no longer visible in the RUD response, as seen in Fig.3.17 (b).

The following procedure estimated the pressure of the backscattering waves. The amplitude ratio of the backscattered waves and the reference reflectance (i.e., water/glass interface) was 0.32%, considering the reflectance coefficient ≈ 94%. That means the microspheres backscatter 0.32% of the generated ultrasound wave, assuming the PZT response is linear in this pressure range. Thus, when the PZT transducer generates an ultrasound wave with a maximum pressure of 17 kPa, only 55.2 Pa of that would be backscattered. From that, it was estimated that the minimum pressure the RUD can detect is 55.2 Pa.

Fig.3.17 (c) shows the center frequency of the windowed backscattering signals from the RUD at different incident ultrasound wave pressures. The center frequency was not affected by the variations in the incident wave pressure. These results demonstrate the capability of this sensor to detect backscattering signals with sensitivity and specificity close to a PZT transducer. Again, these findings confirm the applicability of the depicted RUD to an all-optical quantitative ultrasound imaging system.
Fig. 3.17: (a) The effect of the incident ultrasound wave pressure on the backscattering signal detected by the RUD. (b) The SNR of the backscattering signals decreased linearly with the decrease in the incident ultrasound wave pressure. (c) The center frequency (i.e., ~11.2 MHz) of the corresponding RUD signals detected.
6. Conclusion

Most of the conventional ultrasound modalities use piezoelectric-based transducers to detect the reflected acoustic signals. Although these materials are widely used, they still have several drawbacks, such as narrow bandwidth, fabrication complexity, susceptibility to electromagnetic interference, and high acoustic impedance mismatch. Optical detection of ultrasound is a promising alternative to piezoelectric transducers.

In this work, a prototype optical ultrasound detector was developed and built to detect reflected and backscattered ultrasound from different ultrasound phantoms. The detector utilized the probe beam deflection (PBD) technique, where acoustic waves alter the regional refractive index and hence probe beam propagation. This detector is called a refractometric ultrasound detector (RUD). Unlike other optical-based ultrasound detectors, the RUD developed does not interfere with the acoustic waves, is simple to set up and insensitive to background vibrations and has the capability to perform 2D detection easily.

The functionality and performance of the prototype RUD were evaluated in a series of experimental studies. The outcomes of these experimental studies confirm the RUD is sensitive and reliable in terms of detecting ultrasound signals. The prototype RUD achieves an axial resolution below one millimeter and pressure sensitivity below 55 Pa. In a separate set of experimental studies, the feasibilities of (1) using the prototype RUD to detect the backscattered ultrasound waves from scattering phantoms and (2) using the backscattering signals detected by the RUD to differentiate phantoms with different scatterer sizes were successfully demonstrated. Finally, the performance stability of the prototype RUD was confirmed by producing consistent outputs over an hour.
In order to improve the performance of the RUD, further work should be focused on two main aspects; probe laser and coupling medium. Using a probe beam with a small diameter would broaden the detection bandwidth and improve the axial resolution. Also, replacing water with another material possessing a higher photoelastic coefficient would further improve the sensitivity of the RUD.
Chapter 4: All-Optical Quantitative Ultrasound Imaging System (AOQUIS)

1. Introduction

Commercially available ultrasound systems typically rely on PZT transducers to transmit and receive ultrasound waves [158]. However, significant limitations arise from these transducers due to their frequency characteristics and complexity fabrication process [70]. Ultrasound generation and detection by PZT transducers are most efficient at the resonance frequency (i.e., center frequency), which is determined by the shape/volume and the composition of the transducer elements [73]. Furthermore, the spectrum bandwidth of PZT transducers is limited by the continuous vibration of the PZT elements [250]. Once the element dimension and material are determined during the manufacturing process, the frequency characteristics of the PZT transducer are fixed [70]. For these reasons, clinical ultrasound imaging systems usually are equipped with a range of ultrasound transducers to cover different image needs [251].

The dice and fill method is considered the industry standard in terms of transducer fabrication. However, dicing a PZT plate into micron-scale elements for the purpose of producing high-frequency ultrasound transducers is still extremely difficult [252]. Fabricating an ultrasound array transducer poses additional challenges, as good acoustic matching and minimal cross-talk among the PZT elements are required. In addition, building a reliable electrical interconnection of the PZT elements is complex, especially when the element number is high [253]. An alternative to ultrasound array transducers is a synthetic array where a single transducer is mechanically scanned [254]. Multiple research groups were able to use such an approach to perform high-frequency B-mode ultrasonography, with a center frequency as high as 100 MHz [255], [256]; however, the
bulkiness of the scanning system makes it extremely difficult to miniaturize such a
transducer for, for example, endoscopic applications [74].
Optical approaches for ultrasound generation and detection have been considered viable
and attractive alternatives to those conventional piezoelectric ultrasound transducers [85].
Optical ultrasound generation is based on the PA effect, where light energy is converted to
acoustic energy by absorbing materials [160]. One advantage of optical ultrasound
generation is that the transmitter dimension is determined by the spot size of the excitation
light; therefore, miniaturizing an optical ultrasound transmitter down to micron-scale is
feasible [257]. Moreover, unlike conventional ultrasound transducers, optical ultrasound
transmitters can generate a high-frequency acoustic wave with a significantly broader
spectrum bandwidth [84]. Several optical ultrasound detection methods have been
developed to date, including Fabry–Pérot interferometer (FPI), polymer Microring
Resonators (MRRs), and fiber Bragg grating (FBGs) [72]. These detectors have the
advantage of miniaturized size, high sensitivity over broader spectrum bandwidth, non-
contact detection potential, and optical transparency [258].
Integrating both optical ultrasound transmitter and receiver into a single platform yield an
all-optical ultrasound transducer. In recent years, several studies have demonstrated that
the all-optical ultrasound transducer is a viable alternative to those conventional
piezoelectric ultrasound transducers in terms of imaging [91], [173], [176], [180], [213],
[225], [259], [260]. All-optical ultrasound transducers possess several unique advantages
over conventional ultrasound transducers. One significant advantage is their tunable
operating center frequency and broader spectrum bandwidth. In other words, the resolution
and imaging depth of an all-optical ultrasound transducer can be tuned easily by altering

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the temporal profile of the excitation laser [261]. Another attractive feature of all-optical ultrasound transducers is their efficiency does not degrade with the reduction of their sizes; therefore, they can be miniaturized easily [105]. Since all-optical ultrasound transducers do not contain any electrical components, they are immune to EMI as well as compatible with other imaging methods such as MRI [86].

While all-optical ultrasound transducers have been demonstrated in several previous studies, yet there is a need to improve their optical designs in order to make them versatile and applicable to various clinical applications. One significant improvement is to enhance the diagnostic capability of all-optical ultrasound transducers, which may be accomplished by incorporating the principle of quantitative ultrasound [262]. This concept has not yet been implemented to the best of my knowledge. Another valuable improvement would be to enhance the real-time 2D /3D imaging capability of all-optical ultrasound transducers. This, however, is rather difficult to achieve using a point transducer and will require an innovative optical redesign of an all-optical ultrasound transducer [263].

In this chapter, an all-optical ultrasound transducer (AOUT) was designed and developed based on the two developments reported in the previous chapter: (1) a two-layer optical ultrasound transmitter (OUT) based on candle soot nanoparticles (absorber) and PDMS (background medium) and (2) an optical ultrasound detector RUD based on probe beam deflection (PBD) technique. This prototype transducer demonstrated the following advantages in terms of ultrasound imaging: feasibility of miniaturization, operating field friendliness, and fast 3D imaging capability. More importantly, for the first time, the feasibility of using this system to perform quantitative ultrasound was demonstrated via imaging ultrasound phantoms.
2. Architecture, Mechanism, and Fabrication of AOUT

In Chapters 2 and 3 of this dissertation, the performance of optical ultrasound transmitters (OUT) built upon the CS-PDMS composition and optical ultrasound detectors based on refractometry (RUD) were quantitatively characterized. Here these two components were fabricated and then combined into a single component, named all-optical ultrasound transducer (AOUT), to perform quantitative ultrasound measurements and imaging. The design concept and the fabrication process of this AOUT are depicted as follows.

In the AOUT, ultrasound waves were optically generated from a two-layer CS-PDMS transmitter. The detailed fabrication process of this transmitter type can be found in Chapter 2. Briefly, CS nanoparticles were collected by placing a glass slide within the flame of a paraffin candle for 20 seconds, which is known as the flame synthesis method [95]. A transparent layer of degassed liquid PDMS was added on top of the CS nanoparticles via spin coating, creating a two-layer CS-PDMS film. Next, the CS-PDMS film was cured in a 125 °C oven for 30 minutes. This fabrication process is illustrated in Fig.4.1 (a).

In the AOUT, ultrasound detection was achieved by a refractometric ultrasound detector (RUD), developed in Chapter 3. To combine the RUD with the OUT, as shown in Fig.4.1 (b), a rectangular glass cube with an open top and bottom was first built using microscope slides. The OUT was then glued to the open bottom of the glass cube with the CS-PDMS layer facing inward, which formed a square container. The container was then filled with the coupling medium of the RUD, water in this case. Finally, a thin PDMS layer was glued to the top of the square glass container to prevent water leaks and, more importantly, create a contact interface between the transducer and tissue. In this final step of the transducer
assembly, a collimated diode laser was mounted on the side of the glass container; the laser beam traveled perpendicularly to the generated and reflected acoustic waves and probed the photoelastic effects in the coupling medium.

The AOUT design above produced a point ultrasound transducer, which is perfectly suitable for A-Mode ultrasound. Since the RUD can detect the ultrasound waves anywhere along the probe beam path, this transducer could perform B-mode ultrasound imaging by merely scanning the excitation laser location on the CS-PDMS film, as illustrated in Fig.4.2.

There are two essential features in this AOUT design. First, the coupling medium used for the optical ultrasound detector has a very low acoustic attenuation, facilitating the detection of weak backscattered waves. Table 4.1 shows the acoustic attenuation of various polymers...
and water [245], [264]–[266]. Second, it can acquire 2D (i.e., B-mode ultrasonography) and 3D images without employing sophisticated optical and mechanical components.

![Fig.4.2](image)

Fig. 4.2: The operation principles of the AOUT for (a) A-mode and (b) B-mode ultrasonography.

The working mechanism of the AOUT is illustrated in Fig.4.3. In this design, the excitation light illuminates the CS-PDMS film and generates an acoustic wave (W_{PA}) via the PA effect from the bottom of the transducer. W_{PA} propagates through the section of the
coupling medium where the probe beam situates and creates a disturbance in the local refractive index via the photoelastic effect, which alters the probe beam propagation direction. This event leads to the creation of the first RUD response post-laser excitation. As $W_{PA}$ reaches the coupling medium-PDMS interface, a small portion ($W_{PAR}$) will be reflected because of the acoustic impedance mismatch. $W_{PAR}$ travels through the probe laser beam and creates the second RUD response post-laser excitation. The remaining portion of $W_{PA}$ transmitting through the PDMS layer ($W_{PAT}$) will enter into a subject (e.g., tissue); the echoes created by the subject because of acoustic impedance mismatch or backscattering will reenter into the coupling medium and be detected by the RUD and used to form the A-mode information.

<table>
<thead>
<tr>
<th>Material</th>
<th>Material Formulation</th>
<th>Acoustic Attenuation (dB/cm/MHz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Polyacrylamide (PAA)</td>
<td>%5 Acrylamide, %0.1 Bis-acrylamide</td>
<td>0.17 ± 0.15</td>
</tr>
<tr>
<td>Agarose (AG)</td>
<td>%2 Agarose/water</td>
<td>0.18 ± 0.14</td>
</tr>
<tr>
<td>Polydimethylsiloxane (PDMS)</td>
<td>10:1 elastomer base/curing agent</td>
<td>2.35 ± 0.28</td>
</tr>
<tr>
<td>Polyvinyl chloride plastisol (PVCP)</td>
<td>%5 Softener/PVCP</td>
<td>0.55 ± 0.05</td>
</tr>
<tr>
<td>Water</td>
<td>Temperature 36°C</td>
<td>0.0022</td>
</tr>
</tbody>
</table>

Table 4.1 Acoustic attenuation of various polymers and water [245], [264]–[266].

3. Experimental setup

Based on the design concept as well as the fabrication process depicted in the previous section, a prototype AOUT was fabricated. The picture of this prototype AOUT is shown
in Fig.4.4; its dimension was 25 mm × 25 mm × 25 mm. The thickness of the glass wall was 1 mm (standard microscope slides).

The experimental setup to evaluate this prototype AOUT is shown in Fig.4.5. The excitation source was a 532 nm pulsed laser (SPOT-10-200-532, ELFORLIGHT) with a pulse width of 1 ns and a repetition rate of 10 kHz. The laser output was reflected by two
mirrors and then focused by a 30 mm focal length aspherical lens onto the CS-PDMS film of the OUT. The probe laser used in the RUD was a CW diode laser (STR-640-A-HR-L01-30-S Coherent) with a center wavelength of 640 nm, an output power of 6.4 mW, and a 4 mm collimated beam diameter. A pinhole was placed in front of the probe laser to reduce the probe beam diameter to 2 mm. A high-speed photodetector (PDA10A2 Thorlabs) with a spectrum bandwidth of 150 MHz was used to detect the probe beam transmitting through the transducer. The second pinhole was placed in front of the detector, which reduced the ambient light and specular reflection influences. The output of the detector was connected to a digital oscilloscope (TEKTRONIX TDS 2014B), where detector response was sampled, digitized, visualized, and then store the data for further processing.

3.1. OUT Characterization

The ultrasound generation capability of the prototype transmitter was quantitatively assessed using a calibrated high-frequency hydrophone (HGL-0200, Onda). The hydrophone was attached to a 2D translation stage so that the measurement position could be precisely controlled during the experiments. In this characterization study, the hydrophone was placed above the probing beam and 22 mm away from the OUT.

3.2. RUD Characterization

The ultrasound detection capability of the prototype detector was quantitatively assessed using a calibrated high-frequency hydrophone (HGL-0200, Onda). In this characterization study, the hydrophone was placed above the probing beam and 22 mm away from the OUT.
This method allowed for a direct comparison between the acoustic wave detected by the hydrophone and the RUD.

Fig. 4.5: (a) The schematics of the experimental setup for AOUT test and evaluation. (b) The photograph of this experimental setup.

The ability of the AOUT in terms of performing line (A-Mode) and area (B-Mode) ultrasound measurements were sequentially evaluated in the following experiments.
3.3. Pulse-Echo Experiment

A simple pulse-echo experiment was carried out to evaluate the echo measurement capability of the prototype AOUT. The procedure of this experiment was illustrated in Fig.4.6 (a). The target echo signal was the one reflected by the water/air at the distal end of the transducer. By adding/removing water from the glass container, the distance between the probe beam and the water surface (S) was varied from 1 mm to 7.5 mm. This change, in turn, altered the time delay (Δt) between the \( W_{PA} \) pulse (i.e., the ultrasound wave generated by the OUT) and the \( W_{PAR} \) pulse (i.e., the ultrasound wave reflected by the water/air interface). Using the TOF method, as shown in Fig.4.6 (b), the distance between the probe beam and the water surface (S) was estimated [248].

3.4. Phantom Study

3.4.1. Phantom Fabrication

Four ultrasound phantoms were used in this study. The fabrication process for this ultrasound phantom type was easy and straightforward. First, 10 g of gelatin (GELATIN #0, Humimic Medical) was added into a rectangular glass mold and heated in a 100°C oven for 15 min to melt the gelatin. When no visible air bubbles were observed in the melted gelatin, 100 mg of soda-lime glass microspheres were added. The microspheres were dispersed evenly within the gelatin by spinning the glass mold, which created a homogeneous phantom. Three sizes of soda-lime glass microspheres were used to generate distinctive backscattering properties for each phantom. 35-40 μm diameter microspheres were used in the first phantom (Phantom 1), 75-90 μm diameter microspheres in Phantom 2, and 106-125 μm diameter microspheres for Phantom 3. The last phantom (Phantom 4)
contained no microspheres and was used as the reference. The final dimension of the prepared phantoms was 17 mm × 17 mm × 5 mm.

![Diagram](image)

**Fig. 4.6:** (a) The pulse-echo experimental setup where the distance between the probe beam and the water surface was varied manually. (b) A representative pulse-echo signal used to estimate the distance between the probe beam and the water surface S. S can be calculated by the product of TOF (Dt3= (t3 -t0)/2) and the sound velocity in the coupling medium (1480 m/s).

### 3.4.2. Experimental Setup and Signal Processing

The ultrasound phantoms were mounted on the top of the AOUT to measure their backscattering characteristics, as depicted in Fig. 4.7. For each ultrasound wave generated
by the OUT, the reflected and backscattered ultrasound waves were recorded by the RUD in the AOUT. This process was repeated multiple times for each phantom; the recorded RUD responses were averaged to improve the signal to noise ratio (SNR) and then analyzed in both the time and the frequency domains. In the final investigation, the capability of the prototype AOUT in terms of performing B-Mode imaging was evaluated. Mirror II in the experiment setup was attached to a 1D precision linear translation stage. This approach enabled a line scan of the excitation laser on the CS-PDMS film, which was equivalent to introducing a linear transmitter array in the transducer (Fig 4.2. (b)).

![Fig.4.7: Setup for the ultrasound backscattering measurements using AOUT.](image)

4. Results and Discussion

4.1. System Characterization

4.1.1. Characteristics of OUT

A representative ultrasound wave acquired from the prototype AOUT is shown in Fig.4.8 (a), and its corresponding frequency spectrum is shown in Fig.4.8 (b). The ultrasound waves generated by the prototype transducer showed a bipolar characteristic in the time
domain, which is expected from the photoacoustic effect. The spectrum bandwidth ($f_{BW}$) of the generated ultrasound waves is $\sim 38.5$ MHz; the center frequency ($f_0$) $\sim 34$ MHz. The maximum acoustic pressure of the ultrasound waves generated was around 0.09 MPa. These results are consistent with the OUT performance reported in Chapter 2.

4.1.2. Characteristics of RUD

Placing the hydrophone above the probing beam of the prototype AOUT enabled a direct comparison between the hydrophone response (reference) and the RUD response. The ultrasound waves generated by the OUT and then measured by both detectors are shown

![Graph a) and b) showing characteristics of the ultrasound waves generated by the prototype AOUT in the time domain and frequency domain.](image)

Fig. 4.8: Characteristics of the ultrasound waves generated by the prototype AOUT in (a) the time domain and (b) the frequency domain.
in Fig.4.9 (a). The temporal profile of the hydrophone-measured signal, as expected, is a single bipolar pulse; the one produced by the RUD, however, contained multiple cycles. This time-domain discrepancy also led to a significant difference in their frequency spectral characteristics, as shown in Fig.4.9(b). The center frequency \( f_{0H} \) and the spectrum bandwidth \( f_{BW_H} \) of the calibrated hydrophone signal were 34 MHz and 38.5 MHz, respectively. However, the center frequency \( f_{0RUD} \) and the bandwidth \( f_{BW_{RUD}} \) of the RUD signal were only 19.7 MHz and 7 MHz. The reason behind this significant discrepancy between the two detectors is the probe beam diameter of the RUD. The probe beam width (2 mm) is much greater than the wavelength of the ultrasound wave (0.04 mm); therefore, the ultrasound wave would use 1.35 µs to travel through the probe beam. This, in turn, created additional cycles in the RUD response and hence artificially reduce its spectral bandwidth [247].

4.2. Pulse-Echo Experiment

The outcomes of the pulse-echo experiments are summarized in Fig.4.10. Clearly, a positive linear relationship was observed between \( \Delta t \) and \( S \). Assuming the speed of sound in the coupling medium is 1480 m/s, the distance between the probe beam and the water surface calculated via the TOF were 1 mm to 7.5 mm; which strongly agreed with the actual distances measured using a caliper. These findings suggest that the prototype AOUT can be used to perform A-mode measurements.
Fig. 4.9: (a) The OUT output recorded by the hydrophone (in blue) and the RUD (in orange). (b) Normalized frequency spectra of the time domain responses from the hydrophone (in blue) and the RUD (in orange).
Fig.4.10: (a) The representative signals produced by the AOUT with varying distance between the probe beam and the water surface $S$. The legend shows the actual $S$. (b) The relationship between $\Delta t$ and $S$. (c) The comparison between the calculated $S$ and the actual $S$.

4.3. Phantom Study Results

Representative responses of the prototype AOUT obtained from the phantom study are shown in Fig.4.11. Typically, there are three strong echoes in the AOUT response: the first one originated from the water and PDMS interface ($P_1$), the second one from the PDMS and phantom interface ($P_2$), and the third one from the back surface of the phantom ($P_3$). The fluctuations in the transducer response between $P_2$ and $P_3$ were only observed in
phantoms with microspheres; therefore, they were attributed to the acoustic wave scattering by the embedded microspheres.

![graph](image)

Fig.4.11: (a) Typical time-domain backscattering signals from Phantom with scatterers, and (b) zoom-in into the backscattering waves.

In order to accurately characterize the backscattering signals, random noises, as well as the pulses from the PDMS/phantom ($P_2$) and phantom/air ($P_3$) interfaces, had to be removed. Therefore, the following signal process routine was devised and applied to the raw AOUT responses.

It was found that the easiest way to reduce random noise in recorded transducer responses and improve the SNR was through averaging. By taking the average of repetitive
measurements, the SNR could be improved by the square root of the number of measurements [267]. In this phantom study, the final backscattered signal was obtained from the average of 128 times raw transducer responses. While signal averaging improves the SNR, it would require multiple data acquisitions simultaneously and hence increase the data acquisition time. This increase would negatively impact the imaging time of the transducer. At this point, the real-time imaging capability is not a required spec of the first AOUT prototype. According to the performance analyses of each component in the AOUT, the majority of the random noises detected were generated by the probe laser beam because of its temporal instability. Therefore, switching to a more stable probe laser would favor the detection of backscattering signals of the AOUT.

From an averaged transducer response, a windowing process was utilized to isolate the backscattering signal. The window's width was set to 1.1 us (1.6 mm distance-wise); it began immediately after the pulse from the PDMS/phantom (P2) interface reflection. The window length was determined by the SNR of the backscattering signals [268]–[270]. The averaged and windowed backscattered signals for Phantom 2 and Phantom 4 are shown in Fig. 4.12 (a and c), respectively. The frequency spectra for the windowed backscattering signals from both phantoms were compared to the frequency spectra of the signals from the PDMS layer (i.e., background noise). These backscattered signals from Phantom 2 show a clear difference, in their corresponding power spectra, from the background noise. However, the backscattering signals from Phantom 4 are similar to the background noise since the phantom does not contain scatterers. The results for both phantoms are shown in Fig. 4.12 (b and d).
Fig. 4.12: (a) Backscattering signals detected from Phantom 2 using the prototype AOUT and (b) the corresponding frequency spectra of the windowed backscattering signal and the PDMS layer. (c) Backscattering signals detected from Phantom 4 using the prototype AOUT and (d) the corresponding frequency spectra for the windowed backscattering signal and the PDMS layer.

To quantitatively assess the effects of the scatterer size on the backscattering signal, the backscattering signals were acquired from three distinct locations of each phantom. The frequency spectra of the averaged backscattering signals were compared among all four phantoms. As shown in Fig. 4.13 (a), the significant finding in this comparison study was the center frequency of the backscattering signal decreases as the scatterer size increases. This trend matches the predictions of the theoretical model of ultrasound scattering; the backscattering from larger scatterers would contain a more low-frequency component, and vice versa [249]. This finding confirms that it is feasible to use the frequency spectral characteristics of the backscattering signals to differentiate different scatterer sizes, which is the essence of the quantitative ultrasound. However, it should be noted that there is a substantial discrepancy between the center frequencies of the backscattering signals measured in this experiment and those predicted by the theoretical model, as shown in Fig. 4.13 (b). The negative correlation was consistent with the theoretical predictions. However,
due to the limited bandwidth of the AOUT in this experiment, the trend of the experimental results is not in complete agreement with that of the theoretical predictions. Unlike the PZT ultrasound transducers, where the frequency and bandwidth are fixed, there is still room to improve the bandwidth of the prototype AOUT by reducing the probe beam diameter, as demonstrated earlier.

Fig. 4.13: (a) The correlations between the scatterer sizes and the peak frequency of the backscattering signals. (b) A negative correlation was observed in both the theoretical predictions and the experimental results.
4.4. 2D Imaging

In this section, the imaging capabilities of the prototype AOUT was demonstrated. The same experimental setup described in Fig.4.5 was used in this section; however, Mirror II, which was attached to the 1D linear translation stage, was mechanically scanned along a line on the OUT. As mentioned before, the number of transmitter elements in this design was determined by the number of scanning points. The size of each transmitter element is proportional to the spot size of the focused excitation beam (d), and the distance between each element is defined by the step size (y) of scanning. This approach yielded the most straightforward and simplest 2D transmitter array, as shown in Fig.4.14.

![Fig.4.14](image)

Fig.4.14: The experimental procedure used to test the imaging capabilities of the prototype AOUS. The optical ultrasound transmitter array is defined by the size of the excitation probe (d) and the scanning interval (y), which determines the lateral resolution in the B-mode images produced.

In this imaging study, the ultrasound phantoms, introduced in the previous section, were used as the imaging targets. To produce a 2D image of each phantom, the following step-wise procedure was used [271]:

1. Align the probe beam with the phantom.
2. Scanning the mirror along the line on the OUT.
3. Recording the reflected data.
4. Repeating steps 1-3 for each scanning point.
5. Creating a 2D image from the recorded data.
1) At each position, the AOUT responses were averaged 124 times to reduce background noises.

2) Time Gain Compensation (TGC) technique was applied to the averaged AOUT responses to compensate for the acoustic energy loss resulting from attenuation.

3) The Hilbert transform was applied to the transducer responses to envelop the shape of the acoustic signal. In this step, the frequency-dependent information embedded in the backscattering signals is removed.

4) The logarithmic transformation was applied to the enveloped signals to reduce their dynamic range and achieve a balanced histogram.

5) The first four steps were repeated for each data acquisition location.

6) Finally, the processed data from all acquisition locations were combined to create an image (i.e., B-Mode Ultrasound Imaging).

The B-Mode images from the phantoms, produced by the prototype AOUT, are shown in Fig.4.15. In the B-mode images, the shape and the dimensions of the phantoms are clearly visible; however, the scattering property between Phantom 1, Phantom 2, and Phantom 3 cannot be distinguished using this method. Nevertheless, through the quantitative analysis of the backscattering signals, the differences among the phantoms can be easily detected, as shown in Fig.4.13.

5. Conclusion

In summary, a prototype all-optical ultrasound transducer (AOUT) was successfully produced by integrating a Candle soot nanoparticle–polydimethylsiloxane (CS-PDMS) ultrasound transmitter with a refractometric ultrasound sensor. The detailed fabrication
process of this prototype transducer was provided in this chapter. In addition, the intrinsic properties of the prototype transducer were characterized using a series of experimental studies. Finally, the capability of the AOUT in terms of performing quantitative ultrasound measurements and imaging was also confirmed experimentally. To the best of my knowledge, this exploration is the first to prove the viability of using an all-optical approach to build a 2D quantitative ultrasound imaging device. More importantly, the AOUT developed here is suited for imaging a large tissue area without using complex optical and mechanical components. It paved a solid foundation for the future development of all-optical quantitative ultrasound imaging systems (AOQUIS).

The ultrasound waves generated by the AOUT have a peak pressure of 0.09 MPa, measured at 22 mm from the CS-PDMS film. Furthermore, the transducer has a center frequency above 34 MHz and a spectrum bandwidth of at least 38 MHz; its active area, also known as the element size, is about 2 mm in diameter. On the other hand, the transducer has a detection bandwidth of 7 MHz and a noise equivalent pressure of about ~ 55.1 Pa. This transducer can perform 2D (B-Mode) imaging, achieved by mechanically scanning the excitation laser across the CSPDMS film.

The phantom study outcomes prove that the AOUT is capable of detecting acoustic backscattering signals from the scattering phantoms and uses their spectral characteristics to differentiate different phantom types. This observation, in turn, confirms the potential of the AOUS in tissue diagnosis. Finally, the imaging capability of the prototype AOUT was confirmed with the same scattering phantoms.
Fig. 4.15: 2D images acquired from the previous processing method. (a) for Phantom 1, (b) Phantom 2, (c) Phantom 3, and (d) Phantom 4.

The AOUT produced in this chapter provides many advantages over the conventional ultrasound systems; it is easy to fabricate and has a tunable center frequency, a broad spectrum bandwidth, a high sensitivity, and an immunity to electromagnetic interferences. The transducer's tunable center frequency allows users to condition the imaging depth and resolution based on their needs, which could be beneficial for a wide range of clinical applications.

One of the significant issues with the prototype AOUT is the limited detection bandwidth, resulting from the multi-cycle artifacts introduced by the RUD. This limitation is not insurmountable, fortunately. As described in the previous chapters, reducing the probe beam diameter can significantly improve the RUD's resolution and sensitivity. Furthermore, the sensitivity of the RUD can be improved by replacing the coupling medium with another one that has a higher photoelastic coefficient, high speed of sound, and lower acoustic impedance. These improvements will be implemented in the next generation of the AOUT.
Chapter 5: Conclusions, Limitations, and Future Works

1. Conclusions

The primary objective of this Ph.D. research is to develop an all-optical ultrasound system for quantitative ultrasound measurements and imaging and, eventually, for tissue characterization. In order to accomplish this goal, an efficient optical ultrasound transmitter (OUT) and a sensitive optical ultrasound detector (RUD) had to be developed first. Subsequently, the OUT and RUD were combined to form an all-optical ultrasound transducer (AOUT) capable of performing quantitative ultrasound measurements and imaging. This transducer becomes a critical foundation for the future development of an all-optical quantitative ultrasound imaging system for biomedicine.

For the development of an efficient OUT, a series of experiments were carried out to optimize the design and hence the PA efficiency of the OUT. In terms of absorption particles used in the OUT, it was found that CS nanoparticles are better than other carbon-based materials because of their high absorption coefficient in the UV-Vis-NIR wavelength region and, more importantly, their unique nanostructure allows for fast heat transfer to the surrounding media (e.g., PDMS) [181]. Another advantage of CS nanoparticles is that they can be quickly and easily produced using the flame synthesis method; the thickness of the CS nanoparticle layer is linearly proportional to the collection time.

By surrounding the CS nanoparticles with an elastomeric polymer with a high thermal expansion coefficient (i.e., 30:1 PDMS), the PA efficiency of the OUT was significantly improved. However, excessive PDMS would negatively impact the maximum pressure, the center frequency, and the spectrum bandwidth of the ultrasound pulses produced by the OUT due to the attenuation effect [168].
At the end of this optimization study, a two-layer CS-PDMS OUT was produced, and its fabrication process was developed. By coupling it with a high energy and short pulse laser, this OUT can generate ultrasound pulses with a pressure ($P_{\text{max}}$) of 0.1 MPa, a center frequency ($f_0$) of 35 MHz, and a spectrum bandwidth ($f_{\text{BW}}$) of over 38 MHz, which meet the needs of quantitative ultrasound imaging.

For ultrasound detection, a RUD was designed and built based on the probe beam deflection technique (PBD) [153]. This RUD consists of three main components; a probing beam, a coupling medium, and a sensing element; its sensitivity is governed primarily by the coupling medium and the probing beam. It was found that water is a practical choice of the coupling medium because of its low acoustic attenuation coefficient and adequate photoelastic coefficient [26], [243]. Moreover, its acoustic impedance of water (1.4 MRayl) is comparable to PDMS (1.5 MRayl) and biological tissue (1.64 MRayl). Other materials with a higher photoelastic coefficient, such as fused silica, Germanium, Arsenic trisulfide, and flint glass, were also considered, but their high acoustic attenuation coefficient and/or acoustic impedance made them less desirable [272]. The probing beam influences the performance of the RUD through its power and diameter. The experimental evaluation of the RUD showed that (1) probe beam power was linearly proportional to the amplitude of the RUD response, and (2) the probe beam diameter determines the axial resolution of the RUD in A-mode ultrasonography. According to the outcomes of the experimental evaluation, the RUD developed in this study has a sensitivity of 55 Pa and an axial resolution of 0.27 mm (2 mm diameter probe beam), which meets the needs of quantitative ultrasound imaging.
In the final stage of this dissertation research, the first all-optical ultrasound transducer (AOUT) was designed based on the OUT, and the RUD developed previously. It was then fabricated and evaluated in several experimental studies. While initially designed for point detection, the AOUT can efficiently perform B-Mode ultrasonography by scanning the excitation laser along the RUD probe beam axis on the OUT. This unique advantage is a direct result of the AOUT design and is not found in other optical ultrasound imaging systems published. The imaging capability of the AOUT was validated using scattering phantoms; the reconstructed images show the profiles of the phantoms with reasonable accuracy. Finally, to test the quantitative ultrasound capability of the AOUT, three scattering phantoms with different scatterer sizes were imaged. The frequency spectra of the backscattering signals acquired from these phantoms using the AOUT were distinctively different, which confirms the feasibility of using AOUT to perform quantitative ultrasound imaging.

2. Instrumentation limitations

Like all research projects, several limitations with the existing AOUT design were uncovered as the project progressed. The limitations are associated with both the OUT and RUD. The limitations of the developed OUT are related to its maximum photoacoustic pressure, center frequency tunability, and active area. As mentioned in Chapter 2, the photoacoustic pressure generated by the OUT is determined by the fluence of the excitation light. Therefore, the maximum output of the OUT (i.e., photoacoustic pressure) reported in this dissertation is actually limited by the laser available. Using a laser with higher pulse energy could alleviate this limitation, but it would increase of probability of photoablation.
The limitations of the developed RUD are its sensitivity and axial resolution. They are caused primarily by the CW probe laser beam. The instability of the probe laser beam reduces the signal to noise ratio (SNR) of the RUD and hence its sensitivity. The relatively large diameter of the probe laser beam introduces additional cycles in the RUD responses, which reduces the axial resolution of the RUD and artificially reduced the spectral bandwidth of the RUD responses. It should be noted that the RUD sensitivity is also hampered by the moderate photoelastic coefficient of the coupling medium (water, in this case). Finally, another drawback in the current RUD design is observed in several experimental results; multiple reflections of the ultrasound pulses occur within the RUD, as illustrated in Fig. 5.1. These reflections would cause interference among the echoes and hence introduce the reverberation artifact [273].

Despite the mentioned limitations, it is still possible to perform B-mode imaging and quantitative ultrasound using the AOUT developed. The success of using the backscattering signals acquired by the AOUT to differentiate different scattering phantom types is hugely encouraging and shows the great promise of this research project.

3. Future Works
All in all, the work described in this dissertation turns the concept of creating an all-optical quantitative ultrasound system into a reality by creating a working prototype. The prototype transducer, also known as AOUT, is capable of performing A-mode and B-mode ultrasonography and, more importantly, quantitative ultrasound measurements. While this accomplishment is satisfying, further research has to be carried out in order to improve AOUT’s performance and make it clinically relevant. Specifically, the transducer
sensitivity needs to be optimized, and the transducer needs to provide efficient 2D and 3D imaging capability. To achieve this ambitious goal, the required future research endeavors are described as follows.

Fig.5.1: (a) A schematic diagram showing the sources of reflection within the RUD; (b) A comparison between the PZT and RUD signals showing the multiple reflection effect in the RUD.
3.1. Optimizing the Optical Ultrasound Transmitter

One improvement needed for the OUT is its PA efficiency. The peak PA pressure generated by the best OUT produced in this dissertation is about 0.16 MPa (measurement distance 1.5 mm), which is still significantly lower than the acoustic pressure generated by the conventional ultrasound transducers (i.e., a few hundred kPa) [274]. One possible strategy to improve OUT's PA efficiency is to reduce the thickness of the CS-PDMS film. One additional benefit of this reduction is that it will reduce the attenuation effect of the film on high-frequency ultrasound components. Another possible strategy is to replace PDMS with a polymer possessing a greater thermoelastic property.

The maximum fluence of the excitation laser used in this dissertation research was 3.1 nJ/mm², which is significantly lower than the thermal damage threshold of those carbon-based materials [95]. Therefore, increasing the excitation laser fluence is a reasonable strategy to further enhance the photoacoustic pressure generated by the OUT, as long as it does not damage the photoacoustic film. Alternative optical excitation schemes may also be considered to deliver more optical energy to the photoacoustic film without exceeding its thermal damage threshold; these schemes include coded excitation and temporally modulated pulses [275].

3.2. Optimizing the Optical Ultrasound Detector

The outcomes of the RUD evaluation studies show that the current RUD suffers the spectrum bandwidth distortion and limited axial resolution. A simple solution to address these shortcomings would be reducing the probe laser beam diameter. Ideally, the width of the probe laser beam should be comparable to that of the ultrasound pulse generated by the
OUT. Furthermore, the next generation RUD may incorporate a probe laser sheet and linear photodetector array to provide the instantaneous line detection capability. Finally, the temporal and spatial stability of the probe laser beam has to be used in the next generation RUD because it would improve the signal-to-noise ratio of the RUD and hence its sensitivity. This improvement will be critical to quantitative ultrasound measurements (i.e., backscattering measurements). The RUD sensitivity can be further improved by increasing the photoelastic coefficient and the sound speed of the coupling medium. Hence, exploring other coupling media will be necessary for the development process of the next generation RUD. The idea coupling medium should process the following properties: higher photoelastic coefficient, higher speed of sound, lower acoustic attenuation, and similar acoustic impedance to biological tissue and PDMS. Finally, as shown in Fig.5.1, multiple reflections within the existing RUD cause imaging artifacts. Strategies to reduce the acoustic reflections from the transmitter will also have to be explored and implemented in the development process of the next generation RUD.

3.3. Optimizing the All-Optical Quantitative Ultrasound System

When the AOUT was used to perform B-mode ultrasonography in this dissertation work, the B-mode image quality was relatively low. The reason behind the low image quality is the scanning method employed; the excitation laser was manually scanned across the absorption film, and the translation stage limited both the number of scanning positions and the interval between two scanning positions. Several strategies may use to address this shortcoming. For example, a fast galvo mirror may be incorporated to improve the precision and stability of the excitation laser scanning process, as shown in Fig.5.2.
Another possible solution will be using an excitation laser line, instead of an excitation laser point, to drive OUT. One unique advantage of this solution is that, when it is coupled with a line RUD, instantaneous B-mode imaging can be achieved.

Fig.5.2: One proposed scanning mechanism to increase the number of transmitter elements and reduce the scanning time.
LIST OF REFERENCES


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