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Research article

Characterization of lens based photoacoustic imaging system☆

Kalloor Joseph Francis[a,1,*], Bhargava Chinni[b], Sumohana S. Channappayya[a], Rajalakshmi Pachamuthu[a], Vikram S. Dogra[b], Navalgund Rao[c,**]

[a] Department of Electrical Engineering, Indian Institute of Technology Hyderabad, 502285, India
[b] Department of Imaging Sciences, University of Rochester Medical Center, 601 Elmwood Ave, Rochester, NY 14642, USA
[c] Chester F. Carlson Center for Imaging Science, Rochester Institute of Technology, 54 Lomb Memorial Drive, Rochester, NY 14623, USA

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ABSTRACT

Some of the challenges in translating photoacoustic (PA) imaging to clinical applications includes limited view of the target tissue, low signal to noise ratio and the high cost of developing real-time systems. Acoustic lens based PA imaging systems, also known as PA cameras are a potential alternative to conventional imaging systems in these scenarios. The 3D focusing action of lens enables real-time C-scan imaging with a 2D transducer array. In this paper, we model the underlying physics in a PA camera in the mathematical framework of an imaging system and derive a closed form expression for the point spread function (PSF). Experimental verification follows including the details on how to design and fabricate the lens inexpensively. The system PSF is evaluated over a 3D volume that can be imaged by this PA camera. Its utility is demonstrated by imaging phantom and an *ex vivo* human prostate tissue sample.

1. Introduction

The photoacoustic (PA) phenomenon employs pulsed laser light to induce thermoelastic pressure increase in the tissue absorbers, which in turn leads to the generation of ultrasound (US) waves. PA imaging techniques focus on efficient ways to measure these US waves and form an image representative of the optical absorption profile of the tissue. This unique combination of light and sound brings together the high contrast capability of optical imaging and the high resolution of US imaging [1]. The intrinsic optical contrast of tissue molecule at specific wavelengths in the near infrared window enables PA imaging to be a potential modality in clinical applications like the early cancer diagnosis, metabolism imaging, etc. [2].

Conventional reconstruction algorithms use triangulation on multiple sensor observations to localize tissue absorbers [3]. Typically, given the US propagation model and measurement from multiple sensors, the profile of the initial pressure can be reconstructed. These methods require measurements on a closed surface surrounding the target volume and for this reason are effective in imaging small animals and *ex vivo* tissue samples. However, in clinical studies, US measurement on such a closed surface is nearly impossible. For example in thyroid and prostate imaging, a 360° view of the target tissue is hindered by other body parts resulting in a limited set of views. Robust reconstruction of tissue absorption profile from these limited measurements is an ongoing challenge for PA imaging [4–6]. Additionally, these reconstruction algorithms are computationally complex, and real-time imaging requires expensive and dedicated hardware [7]. In this paper, we present the design, fabrication and use of an acoustic lens based imaging system which we call a PA camera, as a possible alternative to the digital reconstruction based methods mentioned above. The key difference is that in the latter, spatial and temporal sampling of the PA signal occurs before the reconstruction while in the former, reconstruction, or more correctly focusing, occurs in the continuous space-time domain and the PA signal is sampled subsequently. Like in an optical camera, an acoustic lens is used to simultaneously focus PA signal from different points in a 3D volume. The lens performs the major task of focusing the pressure profile from an object plane to the corresponding imaging plane, thus eliminating the need for reconstruction algorithms. A PA camera is ideally suited for real-time C-scan imaging with the availability of a 2D sensor array in

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* Corresponding author.
** Principal corresponding author.
E-mail addresses: ee14resch12001@iith.ac.in (K.J. Francis), narpci@cis.rit.edu (N. Rao).
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the imaging plane. However, a B-scan image can also be formed using a linear US transducer array, without the need for reconstruction algorithms.

Early works on sound focusing using acoustic lens can be found in [8,9], where the authors studied focal length and gain with an acoustic lens. A more extensive theoretical and experimental study on the pressure gain with a biconcave lens can be found in [10]. With the wide use of acoustic lens attached to single element US transducers, the sound field and focusing action have become a well understood technique. However, time consuming point by point scanning is required with such a set-up to acquire C-scan or B-scan image data [11]. The imaging system we describe in this paper is different in that the lens is placed in between the object plane and the imaging plane, and a multi-element US sensor array acquires image data simultaneously at multiple pixel locations. In 2006, He et al. [12] used an acoustic lens for the first time in PA imaging. Other works from the group also include the peak holding circuit for real-time PA imaging [13] and the introduction of 4*F* imaging system [14–16]. A low-cost method using 3D printing technology to manufacture acoustic lenses and a preliminary characterization was conducted by Rao et al. in 2008 [17]. Along with the use of 4*F* imaging system, the group developed a scanning probe, known as PA camera [18]. This technique has proved to be a cost-effective alternative to the conventional PA imaging system with several advancements on the clinical side, including *ex vivo* studies [19] and system designs for *in vivo* imaging [20]. All these systems are designed to time-gate the acoustic signal to image an object plane at 2*F* distance from the lens. An attempt to image multiple depths is presented in [21] with limited success in phantom studies. Several important aspects lacking in the literature include a rigorous system characterization of such a PA imaging camera, resolution analysis of the system and the identification of limiting factors. It is also not clear how to specify the design parameters of an acoustic lens and a transducer to obtain a required resolution. In this work, we intend to bridge the gap between PA camera design and applications and also to open up possibilities of post-processing.

A theoretical model for the PA camera is presented in Section 2. We analyze wave propagation through a thin acoustic lens and present an expression for the pressure detected by the transducers. The proposed model is very flexible in that it allows for the computation of theoretical PSF for any camera design. We also propose a new PA signal model mimicking Gabor wavelets for a finite size source in this section. In Section 3, we present a PA camera design and a detailed specification of the PSF and tissue imaging experiments. In Section 4, a comparison of the theoretical and experimental PSFs is presented along with a study of changes in the PSF at off-axis and on-axis locations. We also demonstrate *ex vivo* prostate tissue imaging as an application of this system. We discuss the advantages and limitations of the proposed theoretical model and the system in Section 5.

2. Theory

In this section we derive the PSF of the acoustic lens, combining the wave propagation with the thin lens model. A separable theoretical axial and lateral PSF for the lens-based system is presented.

2.1. Acoustic lens-based imaging system

In acoustics, a biconcave surface is used as a converging lens if the index of refraction of the lens material is higher than the surrounding medium. The design of a spherical biconcave lens with focal length F and diameter 2ρ is considered here. To achieve a unit magnification we consider the object plane and imaging plane at a distance of 2*F* on either side of the lens as in Fig. 1. The unit magnification was chosen to eliminate the need for scaling the obtained image in this study. However, the lens allows for a different magnification as well. Laser exposure (not shown) excites a short US pressure profile at the object plane P_0. US waves from the object plane propagate in water to the anterior plane at P_1^-. The lens introduces a phase change to the wavefront, focusing and forming an image at the image plane P_2.

2.2. Acoustic lens action

Consider an US point source $\delta(x, y, z)$ at the origin. Let us consider the wave generated by the point source having an envelop signal $a(t)$ modulated with a sinusoid $e^{-i\omega_0 t}$. The source at origin can be defined as,

$$P_0(x,y,z,t) = \delta(x,y,z)a(t)e^{-i\omega_0 t}, \tag{1}$$

where t is time, $\omega_0 = 2\pi f_0$, and f_0 is the modulation frequency. US waves propagating from the source in a homogeneous medium

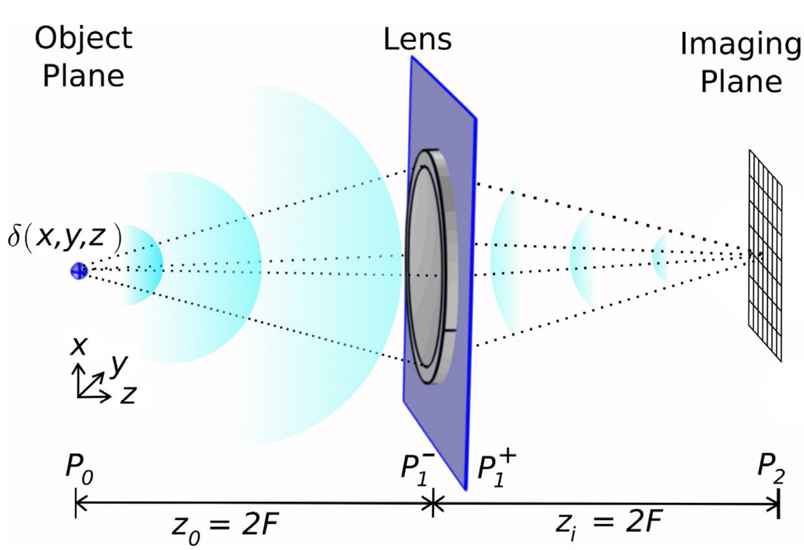

Fig. 1. Acoustic lens system with 4*F* geometry. z_o is the distance between object plane and lens, z_i is the distance between lens and imaging plane. P_0 is the object plane, P_1^- plane anterior to the lens, P_1^+ plane posterior to the lens and P_2 is the imaging plane. F is the focal length of acoustic lens.