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Research article

Quantitative comparison of PZT and CMUT probes for photoacoustic imaging: Experimental validation

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ABSTRACT

Photoacoustic (PA) signals are short ultrasound (US) pulses typically characterized by a single-cycle shape, often referred to as N-shape. The spectral content of such wideband signals ranges from a few hundred kilohertz to several tens of megahertz. Typical reception frequency responses of classical piezoelectric US imaging transducers, based on PZT technology, are not sufficiently broadband to fully preserve the entire information contained in PA signals, which are then filtered, thus limiting PA imaging performance. Capacitive micromachined ultrasonic transducers (CMUT) are rapidly emerging as a valid alternative to conventional PZT transducers in several medical ultrasound imaging applications. As compared to PZT transducers, CMUTs exhibit both higher sensitivity and significantly broader frequency response in reception, making their use attractive in PA imaging applications. This paper explores the advantages of the CMUT larger bandwidth in PA imaging by carrying out an experimental comparative study using various CMUT and PZT probes from different research laboratories and manufacturers. PA acquisitions are performed on a suture wire and on several home-made bimodal phantoms with both PZT and CMUT probes. Three criteria, based on the evaluation of pure receive impulse response, signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR) respectively, have been used for a quantitative comparison of imaging results. The measured fractional bandwidths of the CMUT arrays are larger compared to PZT probes. Moreover, both SNR and CNR are enhanced by at least 6 dB with CMUT technology. This work highlights the potential of CMUT technology for PA imaging through qualitative and quantitative parameters.
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1. Introduction

Photoacoustic imaging (PAI) has been proven to be a promising imaging technique due to its ability to provide high resolution images at enhanced contrast related to the optical absorption [1,2]. In addition, PAI does not cause harmful effects to the patient. Photoacoustic (PA) waves are generated from a tissue when it is subjected to a pulsed laser irradiation. The energy carried by each laser pulse causes a local increase of temperature of the tissue, related to its optical absorption, leading to a thermal expansion, which generate an acoustic perturbation in the ultrasound frequency range [3,4]. The resulting ultrasound (US) waves propagate through the tissue to the body surface where they can be detected [4].

The advantages of PAI rely on its hybrid nature and the combination of the two imaging methods: optical imaging and ultrasound imaging. By coupling them, this modality overcomes some of their limitations. More precisely, PAI features resolution of ultrasound imaging while its contrast derives from the optical absorption. This imaging modality is particularly interesting for vascular imaging as blood is a strong optical absorber in the near infrared and presents a good contrast with the surrounding medium [5].

The theoretical PA signal generated by a spherical absorber surrounded by a lossless medium is a short pulse, characterized by a single-cycle shape, which is often referred to as N-shaped. This

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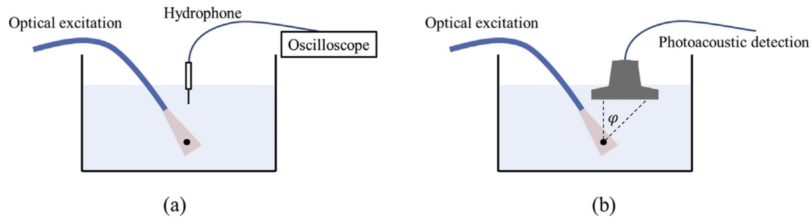

Fig. 1. Experimental set-up for the pure receive impulse response of the US probes of the study. The computation is made in two steps: (a) receiving the signal on the hydrophone, (b) receiving the same signal on the US probes. The φ angle allows evaluating the acceptance angle of the various probes.

shape results from the summation of a diverging compressive wave coming from the absorber and a converging compressive wave coming from the center of the absorber and reaching the detector with a delay, as a rarefaction wave [6]. The frequency bandwidth of a PA signal increases as the absorber gets smaller. Consequently, the spectral content of a PA signal generated by biological tissues may range from 1 MHz up to 100 MHz [4,6].

PA signals can be recorded using an US probe coupled with an US scanner. However, classical piezoelectric US probes, using PZT technology, have a limited bandwidth in both transmission and reception. An emerging alternative technology, CMUT (capacitive micromachined ultrasound transducer, [7]), can overcome this limitation. As compared to conventional PZT transducers, CMUTs may offer higher sensitivity and wider bandwidth [8]. As specifically regards reception (RX) operation in ideal electrical loading conditions, while the voltage frequency response of PZT transducers has a band-pass characteristic, CMUTs exhibit low-pass voltage [9] and charge [10] RX frequency responses theoretically reaching a 200% fractional bandwidth in reception. Such peculiar broadband characteristic, together with the higher RX sensitivity, motivated the first investigations on the use of CMUTs in PAI [11,12]. Several research groups have researched on the potential of CMUTs for PAI. In [13], *in vitro* three-dimensional PAI results using 2D CMUT arrays were successfully obtained. In [14], a particular CMUT technology was established to fabricate an optical-acoustic integrated PA imager, consisting of an infrared-transparent US array backed by an optical source. Furthermore, the technological advantages of CMUTs were exploited in the realization of miniaturized arrays for both two-dimensional [15] and three-dimensional [16] endoscopic PAI. However, the current literature lacks information on the potential performance increase achievable by using CMUT technology in place of classical PZT technology. Only recently, a RX-mode operation performance comparison between a PZT linear probe and an equivalent CMUT probe was carried out [17,18], and the benefits achievable by the higher sensitivity and the wider acceptance angle of the CMUT probe were discussed in a PA imaging application context.

This paper explores the potential advantages achievable by using CMUT technology for PA applications through a qualitative and quantitative imaging assessment. A preliminary study was conducted in [19] and is extended hereafter. A comparative study is carried out by conducting *in vitro* imaging experiments on a suture wire and on bimodal phantoms using different PZT and CMUT probes. Two CMUT linear probes, developed and manufactured by different research laboratories, were used in conjunction with open US scanners to acquire PA signals and generate PA images of the phantoms. The same experiments were carried out using two linear PZT commercial probes from different manufacturers. Qualitative and quantitative criteria, based on the computation of pure receive impulse response, signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR), were used to assess PA image quality.

2. Materials

2.1. Photoacoustic experimental set-up

The PA imaging set-up employed for this study consists of a Q-switched Nd:YAG laser at 1064 nm (Quanta-ray INDI, Spectra-Physics) delivering 5-ns laser pulses at a repetition rate of 10 Hz. The beam diameter is 8 mm. The spatial impulse response was measured using a 100 μm black absorbing suture wire (Ethilon 5-0, Polyamid 6, Ethicon) placed in a water tank. The suture wire was illuminated through the Nd:Yag source coupled with a 10-mm fiber bundle (CeramOptec GmbH, Bonn, Germany) composed of 431 individual optical fibers, each with 0.3 mm silica core diameter and 0.22 numerical aperture. In all measurements, the illumination position was fixed with respect to the suture wire position. The photoacoustic signal detection was made first with a needle hydrophone (Preamplifier W235052, Precision Acoustics), vertically placed 25 mm above the suture wire, using an oscilloscope. The frequency band of the hydrophone is wide enough in order to acquire the complete spectral content of the PA signal generated by the suture wire. The measured signal is averaged over 100 acquisitions. Then, the detection of the same PA signal is conducted using the different US probes used in this study. In all measurements, the probes were positioned in a way that the distance between the suture wire and the closest probe element, i.e. the border element of the active aperture, is set to the same distance of 25 mm. The PA signal was then acquired using all the elements of the active aperture in order to evaluate the angular acceptance of the probes, i.e. the evolution of the reception frequency response as a function of the arrival angle of the PA signal on the probe. The experimental setup is highlighted in Fig. 1.

For the phantom acquisitions, the laser beam was directed towards the phantoms while the US probe was positioned perpendicularly to the laser excitation, as shown in Fig. 2. The signal reception was optimized by improving the US coupling between the probe and the phantom using US gel. The described set-up allows recording simultaneously both PA and US images.

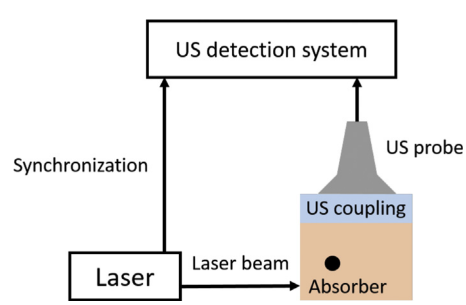

Fig. 2. Experimental photoacoustic set-up. An optical pulse illuminates the medium and the absorbers. A photoacoustic signal is generated and received by the US probe and post-processed by the scanner.

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