Toroidally focused sensor array for real-time laser-ultrasonic imaging: The first experimental study

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1. Introduction

Focused sensor arrays can be used in photoacoustic (PA) and laser-ultrasonic (LU) imaging to increase sensitivity and spatial resolution in a limited spatial region [1]. Focusing a linear array in the elevation direction allows forming the image plane for 2D cross-sectional (selective-plane) imaging [2]. Additional focusing of a linear array in the lateral direction can improve the lateral resolution, sacrificing the size of the sensitivity region [1,3–6]. Generally, the resulting shape of the array will be toroidal. Moreover, the larger the width of the spatial region with high sensitivity the lower the spatial resolution in the corresponding direction [3]. Focused arrays are of particular interest in hand-held PA imaging probes because of better lateral resolution, reduced imaging artifacts and an ability to image sloped surfaces [7]. In addition, only the sensitivity region should be illuminated, so the light delivery system can be optimized.

Combined PA and traditional ultrasound pulse-echo imaging can provide both functional and morphological information about the tissue [8]. In PA imaging, the laser radiation is absorbed in the object itself and the generated acoustic pulses are used to reconstruct the distribution and strength of optical absorbers [9,10]. In traditional immersion pulse-echo ultrasound, a short acoustic probe pulse is used to insonify the object and backscattered acoustic field is used to reconstruct the external boundary and internal acoustic inhomogeneities (scatterers). LU pulse-echo imaging is similar to the traditional ultrasound. However, instead of using piezoelectric transducers to both generate and receive a short acoustic probe pulse, it uses a short (typically < 10 ns) laser pulse to generate short (< 100 ns) and wideband (1–9 MHz) bipolar acoustic probe pulses by PA effect. In this case strongly damped piezoelectric transducers can be utilized to receive the backscattered acoustic field in a wide frequency band [4,12]. This combination allows broadband acoustic inspection of objects and effectively decouples generation and detection of acoustic signals to increase signal-to-noise ratio. Wideband acoustic generation and detection is the main reason why LU inspection has almost no “dead zone” and allows visualizing complex materials such as CFRP composites [13]. Furthermore, the transverse profile of the acoustic beam replicates the smooth transverse profile of the illuminating laser beam, thus practically eliminating the side-lobes. Therefore, combined PA and LU imaging can be advantageous [14,15].

The first multichannel combined PA and LU imaging system was cylindrically focused [14–16]. It had 16 PVDF piezofilm sensors, a plane-parallel optically absorbing plate (PA generator) and a cylindrical acoustic lens. The lens formed a 0.4 mm thick image plane, and the
longitudinal resolution in water was 0.1 mm. This setup was used for real-time imaging of a medical needle inside a model of a blood vessel [15] as well as the external and internal boundaries of duralumin and polymethylmethacrylate (PMMA) samples [17]. However, the lateral resolution of this system was limited by the width of a single transducer (1 mm). At the same time, the noise equivalent pressure of the transducer is inversely proportional to the square root of its surface area [4]. Thus, the width of a transducer also directly affects the sensitivity. The use of a toroidally focused array could improve the lateral resolution of 2D LU imaging maintaining the high sensitivity and could expand the potential applicability of the LU imaging technique. While the use of toroidally focused arrays in PA imaging is more or less understood, no experimental toroidally focused LU imaging systems have been presented. Thus, it is important to build and explore such systems before combining them with PA imaging.

Toroidally focused arrays require not only a complicated manufacturing process, but also a suitable image reconstruction algorithm capable of real-time operation. In PA imaging, the filtered back projection (FBP) algorithm was shown to provide good results with a moderate number of sensors in real time [3,16,18]. In LU imaging, the FBP algorithm can be used for small animal imaging [19], for visualization of the external [20] and internal [21] boundaries of various objects and approximating them with sub-PSF accuracy [17]. The possibility of treating a focused transducer as an effectively point-like virtual detector situated at the focus of the transducer has been studied in PA microscopy [22,23]. However, the toroidally focused arrays are expected to have the highest lateral resolution in the vicinity of the focus in the image plane [24], where the diffraction of the probe beam should be taken into account. In the case of a spatially focused array, all the virtual detectors will coincide, which will make imaging around them impossible without moving the array.

The model-based algorithms [25,26] were shown to be successful in PA imaging with arbitrary-shaped detectors [27]. Although some implementations of model-based algorithms were capable of operating in real time [28], such algorithms are much more computationally intensive than closed-form FBP algorithms [29]. Simulation of ultrasound scattering requires additional calculations of the probe beam field. Therefore, implementing model-based algorithms in real time oriented portable PA and LU systems can be challenging. Moreover, analytical inversion schemes can be used in theoretical analysis of spatial resolution [30] and imaging artifacts.

In this paper, we present the first real-time LU imaging system utilizing a toroidally focused sensor array. A modified FBP algorithm for real-time LU imaging of the region near the waist of the acoustic probe pulse is introduced. The achieved frame rate is up to 30 Hz. The spatial resolution of 30 μm × 0.32 mm in a region ~ 4 mm wide is demonstrated. The achieved spatial resolution of the toroidal array is significantly higher than the spatial resolution of the cylindrical array (0.1 mm × 1.1 mm) of the combined PA and LU imaging system presented earlier in [15]. However, the sensitivity region of the toroidal array is much smaller than that of the cylindrical array (~32 mm). The depth dependency of the sensitivity region width and lateral resolution are explored experimentally.

2. Material and methods

2.1. Experimental setup

Fig. 1 shows the schematic of the experimental setup and the photograph of the developed LU sensor array. Laser radiation from a Q-switched laser (1053 nm, 200 Hz pulse repetition rate, 500 μJ pulse energy, ~8 ns pulse duration, model TECH-1053, Laser-Compact, Russia) is delivered by the fiberoptic cable (600 μm core diameter, Optofiber, Russia) to the PA generator inside the sensor array, where it is absorbed. The PA generator is made of a black polymer material acoustically matched to the PMMA lens; the optical absorption coefficient is ~180 cm⁻¹ [14]. The short generated acoustic pulse travels through the toroidal PMMA acoustic lens and undergoes scattering by the object. The back-scattered acoustic waves propagate through the lens, and then they are registered by 16 wideband piezoelectric PVDF transducers (1 mm × 19 mm × 52 μm) with 2 mm pitch. Analog electrical signals from the detectors are pre-amplified and fed to the high-speed 32-channel data acquisition and processing system based on NI FlexRIO architecture (National Instruments, USA): NI 5752 ADC (12 bit, 50 MHz) and NI PXIe-7962R FPGA module. It digitizes, stores, averages the acquired data over several laser pulses, and then transmits the averaged digital signals via high speed communication lines (PCIe) to a personal computer (Intel® Core™ i7-4770 CPU @ 3.4 GHz, Intel, USA). The LU image was reconstructed in real time using the FBP algorithm modified for toroidally focused arrays. The algorithm was accelerated on a GPU (NVIDIA GeForce GTX 770, NVIDIA, USA) using NVIDIA CUDA computing platform.

2.2. Filtered back projection algorithm for laser-ultrasonic imaging with toroidally focused arrays

The LU FBP algorithm is a delay-and-sum type algorithm: the strength of a scatterer or the reflection coefficient of a boundary ε(r) at point r is considered to be proportional to the weighted sum of the signal values p(rₙ, t) recorded by the sensors located at points rₙ in the appropriate full travel times of the probe acoustic pulse:

ε(r) = \sum_{n=0}^{N_s} D(rₙ, r) F[p(rₙ, t = tₕ(r) + |rₙ - r|/c)] (1)

Here Nₛ is a number of sensors, D(rₙ, r) = G⁻¹(r)ᵣᵣ-associated ΔΩₛ/Ωₒ is a coefficient that accounts for 1/|rₙ - r| attenuation of the scattered waves and can account for the amplitude G(r) of the probe pulse at point r, ΔΩₛ is the solid angle of the m-th sensor as seen from the point r, Ωₒ is the total solid angle of the array. Tₕ(r) is the moment of the probe pulse to reach the point r from the PA generator, c is the speed of sound in the immersion fluid, F is the filtering operator. In PA imaging, F[p] is proportional to (−t)(t/Δt). In LU imaging, F[p] may be taken as (−dp/Δt) due to the bipolar waveform of the LU probe pulse. Here, in the experiments, it was assumed that G(r) = 1. It should be noted that the sensitivity region directly depends on the form of the correction factor G⁻¹(r), but the spatial resolution as a local characteristic of the system does not depend on G.

Both D(rₙ, r) and Tₕ(r) depend on the geometry of the PA generator and the transverse profile of the illuminating laser beam. After calculation at the preparation stage, they can be stored and used for image reconstruction until the parameters of the array or the positions of the image pixels are changed. However, the sensitivity region of a focused array as well as the object can be small. In experiments, it is often necessary to manipulate the image scale and position in order to find the object and get better results. In this scenario, it is advantageous to minimize the duration of the preparation stage. One of the possible ways is to approximate the probe beam with an astigmatic monochromatic Gaussian beam propagating along the x direction:

\[ U(X, Y, Z) = U₀ \left[ \frac{w₀Xw₀Y}{w_X(X)w_Y(Y)} \right]^{1/2} \exp \left\{ \frac{-Z²}{w_Z²(X)} \right\} \exp \left\{ \frac{-Y²}{w_Y²(Y)} \right\} \exp \left\{ -iκcT₀(X, Y, Z) \right\} \] (2)

\[ cT₀(X, Y, Z) = X + \frac{Z²}{2R₀(X)} + \frac{Y²}{2R₀(Y)} - \frac{1}{2κ} \ln \left\{ \frac{X - X₀Z}{X₀} + \frac{X₀ - X}{X₀Z} \right\} \] (3)

This beam has six parameters: wavenumber (κ), two Rayleigh lengths (X₀Z and X₀Y), two positions of the foci (X₀Z and X₀Y), and the
amplitude ($L_b$). The beam widths are $w_Z(X) = w_{0Z} \sqrt{1 + (X - X_{0Z})^2/X_{0Z}^2}$ and $w_Y(X) = w_{0Y} \sqrt{1 + (X - X_{0Y})^2/X_{0Y}^2}$, where $w_{0Z}$ and $w_{0Y}$ are the beam waists, which are connected to the Rayleigh lengths $X_{0Z} = kw_{0Z}^2/2$ and $X_{0Y} = kw_{0Y}^2/2$. The radii of curvature of the wavefront in the perpendicular planes are $R_Z(X) = (X - X_{0Z})(1 + X_{0Z}^2/(X - X_{0Z})^2)$ and $R_Y(X) = (X - X_{0Y})(1 + X_{0Y}^2/(X - X_{0Y})^2)$. The angular apertures are $2\theta_z = 4/kw_{0Z}$ and $2\theta_y = 4/kw_{0Y}$, correspondingly.

Consider a toroidally focused array, which has curvature radii $r_\nu$ and $(R + r_\nu)$ and angular apertures $2\theta_\nu$ and $2\Phi$. Let the center of curvature in the XY plane coincide with the origin (see Fig. 2). The foci of the corresponding Gaussian beam ($X_{0Z}$ and $X_{0Y}$) can be chosen so that they coincide with the two centers of curvature of the sensor array: $X_{0Z} = 0$ and $X_{0Y} = -R$. The Rayleigh lengths can be determined approximately from the angular apertures: $X_{0Z} = 2/k\Phi^2$ and $X_{0Y} = 2/k\Theta^2$. The effective wavenumber $k$ and Rayleigh lengths should be adjusted experimentally. If the right wavenumber is chosen, then the 2D LU image of a reflective plane will be a straight line and the Gouy phase shift will be compensated correctly, otherwise it will be curved and shifted.

The experimental array had $r_\nu = 40$ mm, $R = 4$ mm, $\Phi = 20^\circ$, $\Theta = 15^\circ$. The array shown in Fig. 2 has different parameters than the experimental one to illustrate the complex structure of the sensitivity region in the case of Gaussian astigmatic probe beam. The foci do not coincide and the overall sensitivity is less than for a confocal array.

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The spatial resolution of PA and LU imaging systems can be determined using the point-spread function (PSF) [1, 2, 9, 30]. A PSF is an image of a point-like object and it is determined by the parameters of the system. If the distance between the two point-like objects is less than the full widths at half-maximum (FWHM) of the PSF, the objects cannot be distinguished. That is why the FWHM of the PSF can be treated as spatial resolutions in the corresponding direction in PA and LU imaging. The three directions are considered in the case of a 2D transducer array: axial, lateral and elevation. In the case of a toroidal sensor array in Fig. 2 and a point-like object on the X axis the axial direction corresponds to the X axis, the lateral direction – to the Z axis, and the elevation direction – to the Y axis. If the object is off the X axis its image may be blurred and rotated. That is why in [3] the widths of the PSF were determined by fitting the ellipse to the binarized PSF.

The image processing procedure consisted of three steps. First, the LU image was reconstructed using the FBP algorithm (1) – (3). Second, the image was binarized – the values greater than the half of the maximum value were replaced by ones and the others were replaced by zeros. Third, the ellipse that had the same second-moments as the image was found using the MATLAB/Octave regionprops function. In addition, the bounds along the Z axis were found for a binarized image of a plane-parallel plate to determine the width of the sensitivity region.

3. Results and discussion

Fig. 3 shows the schematic of the experiment with a model of a point-like acoustic scatterer (an acupuncture needle 0.2 mm in diameter) and the reconstructed LU image. The needle was fixed in a duralumin holder that was mounted on an automated 3D translation stage (see Fig. 1). The LU image (1 mm × 1 mm, 400 × 400 pixels) was reconstructed using the developed FBP algorithm (1) – (3) in real-time (up to 30 Hz frame rate) in XZ plane, which was perpendicular to the needle. The needle was translated along the X axis, and the experimental signals were saved for post-processing.

It should be noted that the better representation of a point-like object would be a needle fixed parallel to the X axis. However, such a
setup would require a precise alignment of a needle and an array. A misalignment of 1° at a scan length of 20 mm would give \( \sim 0.35 \) mm shift off the image plane which would be close to the waist of the beam in the elevation direction (\( w_0 \approx 0.45 \) mm). This shift of the needle along Y axis should be compensated. The needle placed approximately parallel to the Y axis (as in [19]) allows to obtain the correct sensitivity depth profile without this compensation. The axial spatial resolution can be slightly worse due to the integration of the wavefront of the probe beam along the Y axis in this case.

Fig. 4 shows the dependencies of normalized sensitivity (black), axial (blue) and lateral (red) resolutions (FWHM of the image) on the position of the needle. Fig. 4 shows the reconstructed image and the binarized image of an acupuncture needle as well as the fitted ellipse.

The best lateral resolution achieved in the experiment was 0.31–0.33 mm in the region between the foci (–5 mm < X < 0 mm), and the best axial resolution is \( \sim 30 \) μm in the same region. The high sensitivity is achieved in the region around the foci: –9.6 mm < X < 1.8 mm.

The experimental determination of the size of the sensitivity region normally requires translating the needle along the Z axis. This procedure can be sped up if one uses reflective plane surface instead of a point scatterer. This plane surface can be considered as a superposition of infinite number of point-like scatterers. The size of the image of a reflective plane surface along the Z axis corresponds to the size of the sensitivity region. Thus, in the second experiment, we used a duralumin plate with a flat bottom surface mounted on the holder instead of the needle to model a reflective plane surface. Again, the plate was translated along the X axis, and the experimental signals were saved for post-processing.

The post-processing procedure was similar to that used for the needle images. The only difference was that we determined the left and right boundaries of the binarized image instead of calculating the second moments. The full width of the sensitivity region is shown in Fig. 5.

Fig. 5 clearly shows that the minimum width of the sensitivity region (magenta) is achieved at X \( \approx \) 0.7 mm. The minimum width of the sensitivity region is located near the center of curvature of the sensor array in the XZ plane (\( X_{0Z} \)) while the highest spatial resolution is achieved near the center of curvature in the XY plane (\( X_{0Y} \)). The sensitivity region (where high sensitivity is achieved) is between the two centers of curvature (\( X_{0Z} \) and \( X_{0Y} \)) (Fig. 2). Therefore, toroidal focusing of the LU sensor arrays can be used to expand its sensitivity region.

It can be clearly seen in Fig. 5 that the LU image of the plate at X \( \sim -5 \) mm resembles the negated version of the LU at X \( \sim 5 \) mm. This is the consequence of the Gouy phase shift (two terms with \( \pi \) in (3)). The total Gouy phase shift along the X axis is \( \pi \) which is equivalent to negating the probe pulse. This should be taken into account during the interpretation of the LU images because the polarity of the image of the

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**Fig. 3.** The schematic of the experiment (left). The LU image of the acupuncture needle 0.2 mm in diameter (right). The solid black line in the LU image corresponds to the half-maximum level of the image. The colorbar is linear scale.

**Fig. 4.** The dependencies of maximum amplitude (normalized sensitivity – black, top left), axial (\( \Delta X \), blue, middle left) and lateral (\( \Delta Z \), red, bottom left) FWHM of the image (spatial resolutions) on the position of the acupuncture needle 0.2 mm in diameter – the model of the point-like scatterer. The LU image of the needle reconstructed using the proposed algorithm (top right) and its binarized form with the fitted ellipse (dashed black line, bottom right). The solid black line in the LU image corresponds to the half-maximum level of the image. The lengths of the major and minor axes of this ellipse correspond to the widths of the image and spatial resolutions. The presented LU images correspond to the position of the needle X = 4.9 mm (near the focus).
boundary also depends on the reflection coefficient of the material. Since the glass plate in water is almost an acoustic hard boundary and the Guoy phase shift is taken into account in the calculation of $T_L(r)$ it is the positive part of the image that represents the correct position of the reflective boundary.

The high axial resolution (30–40 μm) can be attributed to the wide ultrasound frequency band 2–14 MHz at −6 dB. As it was noted in [16] the axial spatial resolution in LU imaging is $\sim 0.4 \frac{\lambda}{c} \approx 44 \mu m$ at 14 MHz. In [31] it is noted that the axial resolution for a spherical ultrasound radiator is $0.5c/BW \approx 63 \mu m$, where $BW = 12 \text{MHz} – \text{bandwidth of the transducer}$, the lateral resolution for the spherical radiator is $\lambda(f\text{-number}) = \lambda/(2\sin\Phi) \approx 292 \mu m$, where $\lambda \approx 0.2 \text{mm}$ is the average wavelength. It should be noted that the lateral resolution is actually less than 0.32 mm since the needle itself is 0.2 mm in diameter and its LU image is roughly the spatial convolution of the PSF and the cross-section of the needle. Therefore, the toroidal LU sensor array allows achieving the resolution that is close to the diffraction limit of the system near the focus in the image plane. If the diffraction of the probe beam is not taken into account by the modified algorithm, the images of the reflective plate will be curved, and imaging near the foci of the array will be impossible.

It should be noted that the acoustic lens had a thickness of $h = 2 \text{mm}$ in the image plane which corresponded to the round-trip time of $\Delta t = 2h/c_L \approx 1.8 \mu s$, $c_L$ – the speed of sound in the material of the lens. Therefore, there was a series of false artifact images of the external boundary of the object at distances $n_c\Delta t = n(2.6 \text{mm})$, where $n = 1,2,\ldots$ – the number of reflection above the object. Since the model of the point scatterer was 0.2 mm in diameter these reflections were outside the imaging region.

4. Conclusions

In this paper we presented a system for 2D LU imaging with improved lateral spatial resolution in the image plane due to the toroidal focusing of the sensor array. The width (1 mm) and number (16) of the transducers as well as the aperture (−45°) of the array were the same as we had before in the cylindrical sensor array (see [14,15]) to maintain high sensitivity and to preserve the single-sided access to the inspected object. A modified BP algorithm for real-time LU imaging of the region near the waist of the acoustic pulse is presented and tested experimentally. The developed toroidally focused LU imaging system allowed us to visualize a sample located at a depth of ≈ 4 cm in water with the lateral spatial resolution of 0.32 mm and axial spatial resolution of 30 μm in real-time. The achieved frame rate was up to 30 Hz. The minimum width of the sensitivity region is achieved near the focus of the sensor array in the image plane while the best spatial resolution is achieved near the focus in the plane perpendicular to the image plane. The high sensitivity is achieved between the foci of the array.

The developed array is intended to be used as a part of combined real-time PA and LU imaging system as in [14,15]. The manufacture of the toroidally focused imaging system is much more complicated than the manufacture of the cylindrically focused system. Therefore, it is important to explore experimentally the trade-off between the size of the sensitivity region and the lateral spatial resolution to optimize the setup for a particular tomographic application. The modified FBP algorithm might be improved and the advanced signal processing for combined real-time LU and PA imaging might be implemented in future studies. Current work was focused on the experimental study of the toroidally focused LU array. Thorough numerical analysis of the developed array will be the topic of our future work.

Declaration of Competing Interest

None.

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