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Passive Acoustic Mapping for ultrasound therapy monitoring

Abstract: Passive Acoustic Mapping (PAM) is an ultrasound-based imaging method developed for monitoring therapeutic ultrasound. By using diagnostic transducers to passively record the acoustic signals that are emitted by cavitation bubbles, the origin of the bubbles can be reconstructed and displayed as intensity maps. In this study, two matrix arrays with different aperture sizes were used for the volumetric reconstruction of simulated and experimental data. In a second step, the number of elements being used for the reconstruction was reduced by more than the factor of eight in order to assess the influence on the imaging quality. In the numerical part of the study, the image quality was greatly improved by increasing the aperture size, while a high number of elements used for the reconstruction merely offers minor improvements. The experimentally obtained results were able to confirm the numerical findings regarding the achievable reconstruction quality.

Keywords: Passive Acoustic Mapping, ultrasound therapy monitoring, cavitation localization.

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1 Introduction

The emergence of therapeutic ultrasound as a promising treatment method with a rising number of possible applications comes in hand with the need of a reliable monitoring method. The application of therapeutic ultrasound relies on the induction of either thermal or mechanical effects in the desired tissue region [1]. The mechanical effects mostly relate to cavitation bubbles [1]. Due to complex mechanisms, the spatial and temporal occurrence of cavitation bubbles is

hard to predict [2]. As a consequence, whether cavitation is desired or not in the specific application, it is necessary to establish a spatial and temporal cavitation monitoring in order to prevent undesired tissue damage.

For thermal based applications, magnetic resonance imaging (MRI) thermometry is a precise method for monitoring the temperature in the treated area, but its high initial and operational costs as well as the availability limit the widespread use [3]. Ultrasound thermometry is able to overcome these limitations by taking advantage of the speed of sound's temperature dependence [4]. For mechanical based applications on the other hand, an ultrasound-based method generally referred to as Passive Acoustic Mapping (PAM) offers a more cost-efficient alternative approach with higher spatial and temporal resolution that could easily be implemented in the clinical environment [2]. Oscillating and collapsing cavitation bubbles emit a characteristic broadband signal that can be recorded by separate diagnostic arrays that are only passively receiving. By applying passive beamforming algorithms, intensity maps showing the location of the cavitation bubbles can be obtained.

Volumetric PAM using matrix arrays as receivers offers the possibility of monitoring a whole volume surrounding the focal area of the therapy transducer without moving the diagnostic transducer. Consequently, the occurrence of cavitation bubbles can be monitored not only in the treatment area, but also in the surrounding tissue, making the application of therapeutic ultrasound potentially safer. However, the high number of elements of a typical matrix transducer combined with the need to reconstruct several imaging planes leads to very time-consuming reconstructions and is demanding with respect to the used receive electronics [5]. As this limits the possibility to volumetrically monitor cavitation in real-time, a reduction of the elements used for reconstruction would greatly decrease the needed time. Therefore, the aim of this study is to show the influence of the aperture size and a decreased number of elements on the reconstruction image quality.

Under a number of reconstruction algorithms adapted for PAM, the Delay, Multiply, Sum and Integrate (DMSAI) algorithm has been chosen for this study. By an artificial widening of the aperture and a further amplification of

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stronger signals, DMSAI offers a superior image quality compared to other non-adaptive algorithms without being as numerically complex as adaptive algorithms [6].

2 Material and Methods

2.1 PAM Algorithm

The DMSAI algorithm was implemented as described by Lu et al. [6]. For completeness, the algorithm is described in briefness. After defining a voxel grid in the region of interest, the signals of each element are delayed individually, depending on the distance between the element and the voxel. The delayed signal for one element i at time t can be written as

$$s_i(\mathbf{x}, t) = d_i(\mathbf{x})p_i(t + d_i(\mathbf{x})/c), \quad (1)$$

where $d_i(\mathbf{x})$ is the distance between element i and the voxel at position $\mathbf{x}$, c is the speed of sound and p_i is the original signal recorded by the sensor elements.

Afterwards, the delayed and multiplied signals are summed over all channels N , forming the source strength q . With some steps of simplification that are not shown in detail, the source strength can be written as

$$q(\mathbf{x}, t) = [S_1^2(\mathbf{x}, t) - S_2(\mathbf{x}, t)]/2, \quad (2)$$

with $S_1(\mathbf{x}, t) = \sum_{i=1}^N \bar{s}_i(\mathbf{x}, t)$, $S_2(\mathbf{x}, t) = \sum_{i=1}^N \bar{s}_i^2(\mathbf{x}, t)$ and $\bar{s}_i(\mathbf{x}, t) = \text{sign}[s_i(\mathbf{x}, t)]\sqrt{|s_i(\mathbf{x}, t)|}$.

The source strength q from (2) is then squared and integrated over a certain time period, forming the source intensity or energy (see eq 3).

$$I(\mathbf{x}) = \frac{4\pi}{N^2 \rho c} \int_T^{T+\Delta T} q(\mathbf{x}, t)^2 dt \quad (3)$$

This procedure needs to be repeated for every voxel within the region of interest in order to form complete cavitation maps.

2.2 Simulation

For the simulation of cavitation signals recorded by an ultrasound transducer, a numerical model was implemented. The so-called multi-bubble-model was originally created by Vokurka and has already been adapted for PAM by Coviello et al. [7,8]. To mimic the transducer bandwidth, a bandpass filter was applied to the signal. In this study, a cavitation source consisting of two bubbles at $x = 0$ mm, $y = 0$ mm and $z = 60$ mm was simulated. The excitation frequency was set to

375 kHz and the diagnostic transducer was assumed to acquire for 100 μ s with collapsing bubbles being present in each cycle over the whole time period.

2.3 Experimental part

To generate cavitation in water, a spherical focusing transducer with a diameter of about 77 mm and a centre frequency of 375 kHz was used. With a power amplifier with integrated source generator (AG series amplifier, T&C Power Conversion Inc., Rochester, USA), a CW-signal was generated. The power level was set to 3 W, which creates pressure levels that are sufficiently high to produce cavitation bubbles in water. All experiments were performed in a small tank filled with water. The receivers were aligned orthogonally to the sound field axis of the therapy transducer in an axial distance of approximately 60 mm and positioned in such a way that the aperture captured the focus area of the therapy transducer.

The pre-beamformed channel data were recorded for 102 μ s by an ultrasound research platform with 1024 channels at a sampling rate of 40 MSa/s (Digital Phased Array System, Fraunhofer IBMT). Afterwards, the data was transferred to a separate computer for further processing with MATLAB (R2020b, The Mathworks Inc., Natick, USA).

2.4 Receiving transducers

For both simulations and experiments, two 32x32 matrix arrays with different aperture sizes and centre frequencies were used. One is a 3 MHz transducer with an element pitch of 0.3 mm in both directions (Vernon, Tours, France) and the other one is a costum-made 4 MHz transducer with an element pitch of 0.94 mm in both directions. In a second step, only 121 elements of the 4 MHz transducer were used for the reconstruction. The elements were equally distributed on the aperture, forming an 11x11 matrix. In the numerical part, a costum-made 11x11 matrix array with an element pitch of 2.8125 mm and a centre frequency of 2 MHz was used additionally.

2.5 Performance evaluation

The voxel size was set to 0.3 mm in lateral, axial and elevational direction. Within the volumetric imaging area, several x-z-planes were reconstructed as cavitation maps. In this study, the x-z-plane containing the maximum intensity is displayed. For the reduced number of elements, the x-z-plane

Figure 1: Reconstructed x-z-planes of simulated cavitation signals for different transducers. (a) 3 MHz 32x32 array. (b) 2 MHz 11x11 array. (c) 4 MHz 32x32 array. (d) 4 MHz 11x11 out of 32x32 array.

which originally contained the maximum intensity in the reconstruction with the full matrix is shown.

The performance of the reconstruction with different transducer setups is evaluated by the axial Point-Spread-Function (PSF) as well as the background contrast. The PSF sizes are determined by the full width at half maximum (FWHM) values. The background contrast or signal-to-noise ratio (SNR) is calculated from the ratio between the mean of intensities that are above half of the maximum intensity and the mean of intensities that are equal to or lower than half of the maximum intensity (see eq 4).

$$SNR = 20 \log_{10}(I_{higher}/I_{lower}) \quad (4)$$

3 Results

3.1 Simulation

The reconstructed cavitation maps for the simulated cavitation source are shown in Figure 1. The corresponding FWHM values and the background contrasts (SNR) can be found in Table 1.

It can be seen that the 3 MHz 32x32 matrix array creates a PSF of considerable axial extent, bearing in mind that only one cavitation source is present. Accordingly, the FWHM in axial direction is as high as 57.9 mm. With the 4 MHz 32x32 array on the other hand, it is possible to reconstruct cavitation maps that allow the localization of cavitation bubbles with a higher precision in axial direction. The axial FWHM is with 6.3 mm more than nine times smaller compared to the 3 MHz array. The 2 MHz 11x11 array with a similar aperture size as the 4 MHz array leads to an axial FWHM of 10.8 mm, which is merely less than two times higher than for the 4 MHz array while using more than eight times less elements.

As the simulation with the 2 MHz array suggests that 121 elements on a large aperture are sufficient to produce similar image qualities as with a 1024 element transducer, the next step was to use only 121 of 1024 elements from the 4 MHz array for a direct comparison. With 11x11 of the originally 32x32 elements of the 4 MHz array, the reconstruction quality is only slightly decreased in terms of the background contrast, while the axial FWHM is with 6.0 mm even a little bit smaller.

3.2 Experiment

Analogous to the simulation part, the reconstructed cavitation maps for the experimental data are displayed in Figure 2 and the calculated FWHM values as well as the background contrasts (SNR) are shown in Table 1.

Again, the 3 MHz array shows a poor axial resolution. A localization of the cavitation source with this transducer is barely possible, making it unsuitable for monitoring purposes. Similar to the simulation, the 4 MHz array makes it possible to determine the origin of the cavitation within a range of several millimetres with contrast levels of about 40 dB, using both the full 32x32 matrix and the reduced 11x11 matrix.

Table 1: Axial full width at half maximum (FWHM) and signal-to-noise ratio (SNR) values for reconstructed images of numerical and experimental obtained data with different transducer setups.

Array type		Axial FWHM [mm]		SNR [dB]	
		Sim.	Exp.	Sim.	Exp.
3 MHz	32x32	57.9	52.8	18.94	19.15
	11x11	10.8	-	42.71	-
4 MHz	32x32	6.3	9.0	55.78	40.49
	11x11	6.0	7.2	55.17	38.35

Figure 2: Reconstructed x-z-planes of experimental data from different transducers. (a) 3 MHz 32x32 array. (b) 4 MHz 32x32 array. (c) 4 MHz 11x11 out of 32x32 array.

4 Discussion

As both the 3 MHz and the 4 MHz array consist of 1024 elements, it seems to be the aperture size that makes the biggest difference in the achievable image quality. The positive influence of a higher centre frequency needs to be taken into account as well, but since the transducers are only 1 MHz apart from each other and the 2 MHz array shows good image qualities in the simulation, the influence can be considered to be rather low.

The higher FWHM values for the 4 MHz transducer in the experimental part might be the result of a cavitating bubble cloud, which leads to a bigger cavitation area compared to the simulated point source. Furthermore, the simulation does not account for the diffraction pattern of the elements, assuming omnidirectional receivers. The fact that the reduced number of elements of the 4 MHz array leads to smaller FWHM values in the numerical and in the experimental part might be due to an elevational shift of the maximum intensity. Nevertheless, the experimental results confirm the trend derived from the numerical studies. While a high aperture area seems to be crucial in order to get satisfying images, the number of elements distributed on this area does not have to be as high as possible.

5 Conclusion

The study suggests that, compared to a complete matrix transducer, using a highly diminished number of elements on a large aperture is suitable for the reconstruction with PAM algorithms. While the time needed to perform the reconstruction decreases with a lower number of elements, the quality in terms of background contrast and PSF-sizes remains within a satisfying range. Further studies are needed in order

to assess the optimum compromise between the number of elements and the desired reconstruction quality. In addition to that, a possible shift of the reconstructed source location when using a reduced number of elements needs to be addressed. Furthermore, the performance of a reconstruction with a lower number of elements should be evaluated both numerically and experimentally when multiple cavitation sources are present.

Author Statement

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