Full angle ultrasound spatial compound imaging

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Summary
Multi-angle spatial compound images (MACI) are usually generated by averaging the recorded single-angle images obtained by means of a conventional B-mode ultrasonic scanning. In this technique, images are recorded from a number of different angles (typically from 3 to 9), and these single-angle images are then combined to form the compound image. Compound images can be recorded with a linear array of piezoceramic transducers connected to an ultrasound system with a high number of channels allowing for a precise beam-steering. MACI offers noticeable advantages in image improvements, due to the reduced angle-dependence and the reduced speckle in the compound image compared with a conventional B-mode imaging. This work approaches the ultrasound compound imaging with a full angle spatial concept. A circular array of many elementary piezoceramic transducers of a small size to cover the full angle around a stationary object submerged in water allows obtaining the ultrasound spatial compound image of its cross-section. Such an arrangement is well suited for multi-modal examination (including an ultrasound transmission and reflection tomography) and early detection of malignant lesions in women's breasts. Preliminary work was carried out with a mechanical rotation of ultrasonic conventional linear arrays working with a B-mode ultrasonic scanner around standard breast phantoms imitates average values of acoustic parameters of tissues occurring in woman's breast and designed to train a thin-needle biopsy assisted by real-time ultrasonography. High resolution and contrast compound images were obtained with an artifact-free capability, as the result.

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1. Introduction

A significant improvement in the quality of ultrasound B-mode image can be obtained using the MACI (Multi-Angle Compound Imaging) method, also known as CI (Compound Imaging). In this method, ultrasound B-mode images (typically a total of 3 to 9) are recorded at different angles, then appropriately compounded [1,2]. The use of this technique in ultrasonography for in vivo diagnostics in real time is known [2], but practical implementations are being continuously improved as the result of the dynamic development of technology and the increase of computing power of digital ultrasonic systems. The essence of CI is to acquire multiple image frames at different angles, which are then digitally superimposed in accordance with the angle of scanning, e.g., by the proper averaging of pixels [1]. The implementation of the CI method in conventional B-mode ultrasonic scanners requires two basic modifications in software of the beam-steering system electronics and the image processor. There has to be the capability to deviate the ultrasound beam by a fixed arc step or steps to the left or right in relation to the axis of the linear ultrasonic array. The number of recorded image frames corresponding to angular steps depends on the parameters of the ultrasonic array and clinical applications. Keep in mind that the more frames there are in the CI sequence the better the image quality at the expense of slowing down the process of acquisition. The image processor must be capable of compounding the image in the CI sequence of frames obtained at different scanning angles, which requires matching the geometry of the rotated pixel grids and averaging their values. CI image is updated after each acquisition of a new image frame, so the number of playback frames does not change in time, but their processing introduces a motion blur effect in the case of a fast moving tissue or ultrasonic array. The more frames are compounded in the CI sequence, the greater the motion blur. CI development is oriented at estimation of motion in CI image sequences by creating vector maps of offsets of each pixel using correlation techniques [3,4,5], which are used for the correction of motion blur. Coefficient of pixel area decorrelation is directly proportional to their offset.

Significant advantages of CI imaging include [2]: reduction of spots and noise, which improves tissue differentiation and increases the clarity of pathological changes of low contrast; reduction of multiple reflections affecting the clear visualization of the content of cysts, better detection of microcalcifications, increasing the depth at which useful data is visualized; continuity of structures reflecting the ultrasonic wave affecting the highlighting of outlines of cysts, canals and connective tissue, better visualization of systems of linear striations in fat and muscles, improved presentation of the internal architecture of hard lesions; keeping the central acoustic shadow and acoustic amplification behind shading structures enabling to maintain the recognized diagnostic criteria for cysts and nodules; reduction in refractive shadows enabling to quickly identify suspicious shadows and improve the visibility of structures located, e.g., behind ligaments. With all these advantages, CI imaging can be very useful for detecting and distinguishing between malignant and benign lesions in the breast tissue in women [2].

CI imaging can be significantly extended using the sub-aperture of circular 1-D ultrasonic array developed with the participation of the author [6]. The array is a ring of piezoelectric transducers spaced evenly on its inner side and is designed especially for diagnostics of breast tissue in women in vivo in the early detection of cancer lesions [7,8]. Considering such an application, this study has involved preliminary imaging of a female breast biopsy phantom using a B-mode ultrasonic scanner with ultrasonic probe mechanically rotated around these phantoms by fixed angular steps. High resolution and contrast full angle ultrasound spatial compound images were obtained with an artifact-free capability, as the result.

2. Materials and Methods

Figure 1 shows the block diagram of the developed research set-up which allows echography imaging of breast phantoms in all directions around.

Measurements were made using DUS 101 B-mode ultrasonic scanner with a 5 MHz sector ultrasonic probe made by the Polish company DRAMIŃSKI S.A.

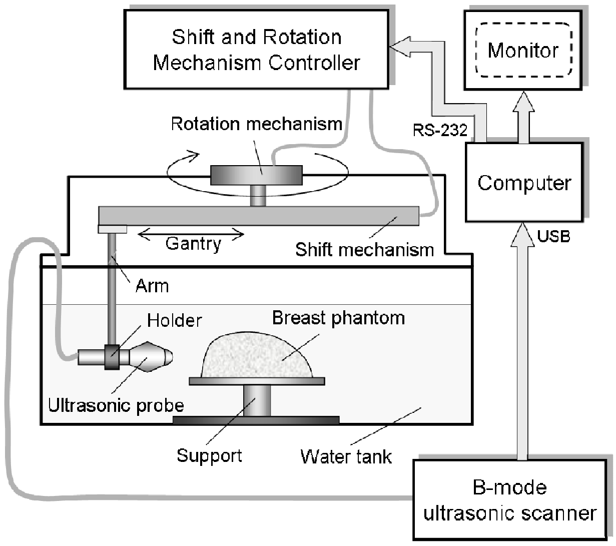

Figure 1. Block diagram of the CI research set-up.

This paper discusses the examination of a breast biopsy phantom CIRS Model 052A. It imitates average values of acoustic parameters of tissues occurring in female breast and is designed to train a thin-needle biopsy assisted by real-time ultrasound B-mode imaging. The size (150:120:70 mm, volume 600 cm^3) and the shape of a phantom simulate a woman's breast in the supine body position. The phantom is made of Zerdine^{TM} gel which imitates tissue and contains liquid and solid inclusions imitating lesions as cysts and nodules (compact masses), respectively. CIRS Model 052A phantom contains 6 amorphous (non-spherical shapes) inclusions of sizes 8 – 15 mm and green tint that imitate cysts and 6 amorphous inclusions of sizes 6 – 12 mm and black tint imitating nodules. The location of inclusions in the phantom is random. The advantage of a phantom, given its use in ultrasound testing in the water, is a smooth surface which minimizes weakening of the ultrasonic wave at oblique incidence.

Ultrasound B-mode images were recorded in a coronal section of the phantom (at a height of 26.5 mm from the base) for 32 angles of rotation in increments of 11.25°. The axis of rotation of the sector ultrasonic probe was set around the geometric centre of the phantom base, and the distance from the axis to the probe surface was 125 mm (Fig. 2). This distance is roughly equal to the radius of the developed circular ultrasonic arrays [6,8,9]. The water temperature during the measurement was 20.9 °C.

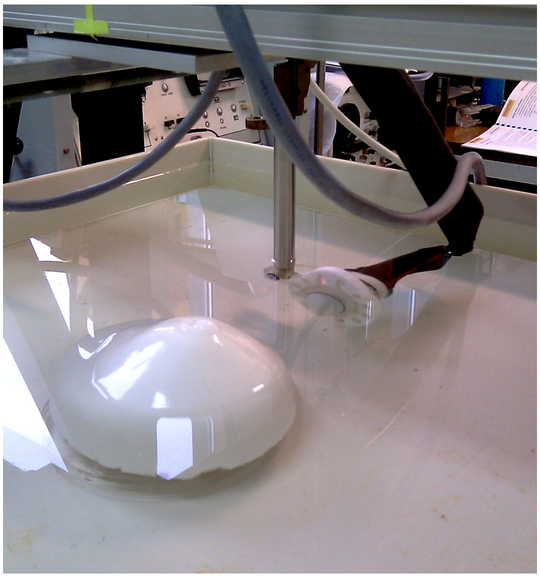

Figure 2. Setting the sector ultrasonic probe relative to the examined phantom.

3. Results

Figure 3 shows selected ultrasound B-mode images of the examined coronal section of the CIRS Model 052A breast phantom obtained for angles of rotation of the sector ultrasonic probe of 0°, 90°, 180°, 270°, respectively.

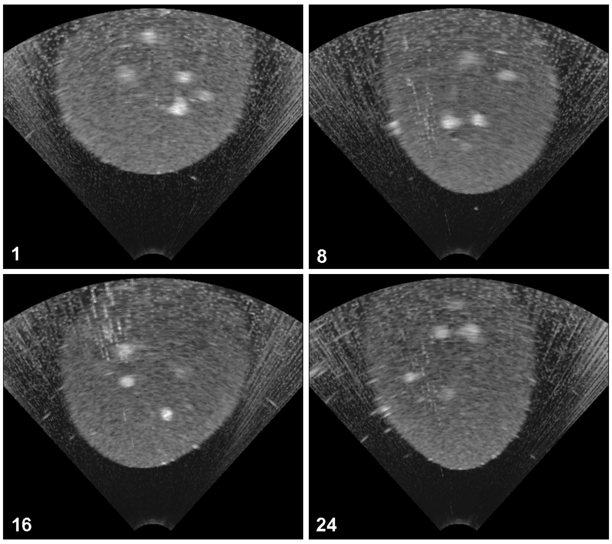

Figure 3. Ultrasound B-mode images of the examined breast phantom obtained for the following angles of rotation: *1* - 0°, *8* - 90°, *16* - 180°, *24* -270°.

Ultrasound B-mode images show the presence of inclusions (heterogeneities) of the characteristics of a compacted mass (nodule) with a density higher than ambient (bright spots). Cyst type heterogeneities of a density lower than ambient (dark spots) are difficult to see. The boundaries of spots are fuzzy, and therefore it is very difficult to determine the shape of their section and estimate their size (Fig. 5). Distortions visible on ultrasound images are caused by electromagnetic interference introduced by the pulses supplied the shift and rotation mechanisms in the measurement system. Due to the use of a sector ultrasonic probe, angular resolution of obtained ultrasound images deteriorates with the distance from the surface of the probe. Short range of the examination, chosen for its optimal image resolution does not allow visualization of the sectional area of the phantom which is the farthest from the probe. In turn, offsetting the phantom from the array surface is advantageous in terms of achieving the greatest possible area of overlapping of angular images in the examined section. Figure 4 shows full angle spatial compound images (FASCI) of the examined section of the breast phantom consisting of 32 ultrasound images (Fig. 3) obtained at angles from 0° to 348.75°.

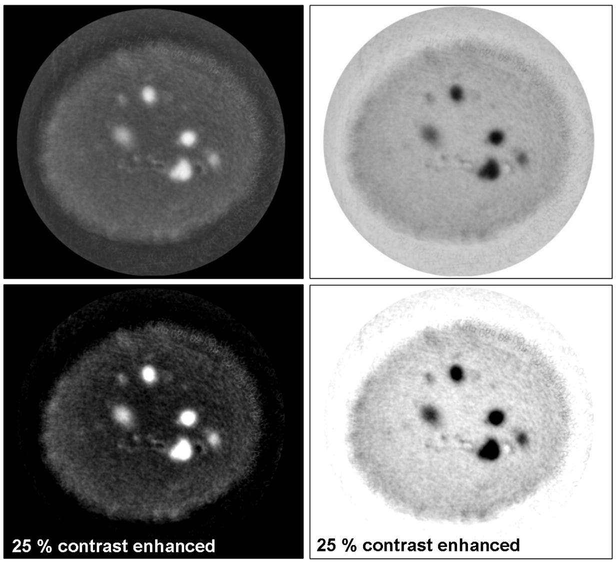

Figure 4. FASCI images of the examined section of the breast phantom consisting of 32 ultrasound B-mode images: top row – normal contrast, bottom row – contrast increased by 25 %, left column – grayscale from black to white, right column – grayscale from white to black.

FASCI images (Fig. 4) were obtained as a result of superimposing of ultrasound B-mode images with transparency of 95 %. Figure 5 shows FASCI images of the examined section of the phantom in the ROI of distribution of heterogeneities as compared to the same region in ultrasound B-mode images obtained for an angle of 0°. Images are shown in normal contrast and the contrast increased by 25 %.

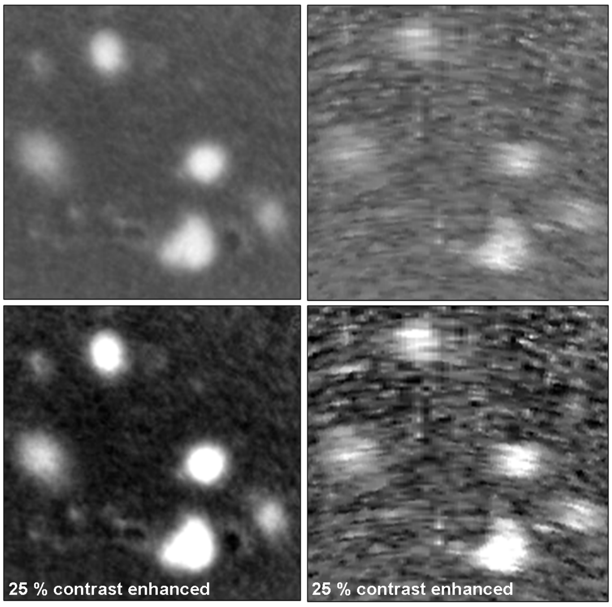

Figure 5. FASCI images of the examined phantom section consisting of 32 ultrasound B-mode images in the ROI of distribution of heterogeneities as compared to the same region in ultrasound B-mode images obtained for an angle of 0°.

The resulting FASCI images are of high resolution independent of the distance from the sector ultrasonic probe and significantly lower granularity as compared to individual ultrasound B-mode images. Noise and distortions are significantly minimized by averaging pixels of 32 superimposed ultrasound B-mode images. Shapes of inclusions (heterogeneities) in FASCI images are clear, cyst type inclusions (dark spots) appear that are invisible on individual ultrasound B-mode images.

It should be noted that the CI method is used with normalization of pixel values in the composition of ultrasound B-mode images, and there are various possible ways of averaging (e.g., arithmetic average, geometric average, median, RMS) [1]. In addition, in order to sharpen CI images, advanced algorithms can be used to create vector maps representing offsets of overlapping pixels in images for different angles of acquisition. Such an operation allows equalizing the positions of pixels. Minor changes in this position are due to the phenomenon of refraction of rays of the ultrasonic beam and the adoption of a constant velocity of ultrasound in soft tissue (c = 1540 m/s) when converting time to distance in ultrasound B-mode imaging [9], which blurs the edges of heterogeneities. The same algorithms can also be used in a FASCI method.

4. Conclusions

Full angle spatial compound imaging (FASCI) offers noticeable advantages in image improvements, due to the reduced angle-dependence and the reduced speckle in the compound image compared with a conventional B-mode imaging. Recording of 32 individual ultrasound B-mode images around the examined object is sufficient to obtain good quality of a CI image. In turn, this enables to split 1024 piezoelectric transducers of the circular ultrasonic array into 32 independent ultrasound curvilinear probes. The FASCI method is well suited for the implementation in ultrasonic tomography with the circular array of piezoelectric transducers as an additional modality which is very important from the point of view of the imaging of breast tissue in vivo.

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