Ultrasound Tomography Imaging: Results of Breast Phantom Study and Resolution Estimation

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Abstract

In order to improve breast cancer detection rates, new and better imaging methods are required. Currently, the ultrasound tomography (UT) as non-invasive and safe hybrid method may contribute to achieving a new standard for breast cancer diagnostics. The aim of the paper was to analyse the imaging ability of tissue-like media structure found in female breast using the developed novel ultrasound computer-assisted tomographic scanner. Measurements was performed on commercial breast biopsy phantoms due to their well-defined structure with inclusions mimicking glandular tissue with lesions, as well as on the simple agar phantom. Obtained magnetic resonance images (MRI), conventional ultrasound images (US) or X-ray computed tomography (CT) images of the measured media sections were used for comparison. The analysis of the obtained results and carried out theoretical considerations have allowed to estimate the resolution of soft tissue UT imaging.

Keywords: ultrasound tomography, breast phantoms, estimation of imaging resolution

1. Introduction

It is estimated that the detection of the breast cancer with a size less than one inch gives a 98% probability of survival. Referring physician has at his/her disposal, apart from palpation, imaging tests: mammography (MMG), conventional ultrasound B-mode scanning (US) and magnetic resonance imaging (MRI). Each of these methods has its advantages but also disadvantages and limitations [1].

In order to improve breast cancer detection rates, new and better imaging methods are required. Currently, the ultrasound tomography (UT) as non-invasive and safe hybrid method may contribute to achieving a new standard for breast cancer diagnostics [2-5]. The hybrid method means that 3 complementary ultrasound images, representing a distribution of transmission, reflection and scattering acoustic parameters are simultaneously reconstructed. The fusion of these images allows a qualitative and quantitative characterization of breast tissue in the entire breast volume [5]. There is no need any contrast agent in the UT method. The aim of the paper was to analyse the imaging ability of tissue-like media structure found in female breast using the novel ultrasound computer-assisted tomographic scanner developed by DRAMIŃSKI S.A. company in cooperation with Wroclaw University of Science and Technology team [5, 6]. Measurements was performed on commercial breast biopsy phantoms due to their well-defined structure with inclusions mimicking glandular tissue with lesions, as well as on the simple self-made agar phantom. Optical, magnetic resonance, conventional US or X-ray computed tomography (CT) images of the measured media sections were used for comparison. The analysis of obtained results and carried out theoretical considerations have allowed to estimate the resolution of soft tissue UT imaging.

2. Estimation of spatial resolution

The estimation of the smallest malignant tumour which can be detected in the breast of woman using an ultrasound tomography scanner is very important due to effective breast cancer screening. First, we have to consider resolution limitations resulting from a given scan resolution and properties of ultrasound wave.

Ultrasound tomography transmission quantitative images (which are the basis for the automatic recognition of tumours and estimating their malignancy) are reconstructed from projection measurements of the ultrasound pulse runtime and amplitude transmitted through the breast in water [5]. The horizontal plane resolution in the ultrasound transmission tomography (UTT) can be define as the pixel size. The minimal pixel size is determined by the number of ultrasonic transducers in the tomography ring array and by the distance between its centres (so called pitch), as well as by the ultrasonic wavelength in the breast tissue. The constructed tomography array [6] consists of 1024 piezoelectric transducers evenly distributed on the inside of the 260 mm diameter ring with the pitch about 0.8 mm and 2 MHz resonant frequency. The ultrasound wavelength in the breast for 2 MHz is $\lambda \approx 0.75$ mm. The half of the wavelength can be estimated as 0.4 mm. This is the basis for determining the horizontal plane resolution 0.4 x 0.4 mm in our ultrasound tomography transmission quantitative images. However, it does not mean that such small breast tissue lesions are always detected. This horizontal plane resolution (in the XY plane) should be understood only as a limitation of the device. It means that the minimum size of the heterogeneity visualised in the ultrasound transmission tomography image cannot be smaller than the pixel size 0.4 x 0.4 mm. The same applies to the vertical resolution (slice thickness) resulting from the height of piezoelectric transducer and the vertical scan step (array movement) of the measurement of coronal breast sections in the *Z*-axis direction. The vertical scan step is 1 or 2 mm (depending on the breast size) but the ultrasound beam in the vertical plane is wider, due to the large height of the elementary transducer (b = 18 mm). The effective angle of the vertical beam divergence resulting from the 3 dB decrease of the ultrasound wave pressure relative to the pressure on the beam axis can be calculated as about 2 using the formula $2\varphi_{z3dB} = 2\arcsin(0.442\cdot\lambda/b)$. The vertical ultrasound far field of ultrasonic rectangular transducer with height of 18 mm calculated from the equation $l_{0z} = 0.35b^2/\lambda$ is about 150 mm. Thus, the real vertical resolution can be geometrically estimated as about 5 mm. Finally, we obtain the voxels of breast tissue limited to size 0.4 x 0.4 x 5 mm.

3. Estimation of contrast resolution

The foregoing considerations, while important and necessary, but do not give the answer to the question: What the smallest malignant tumours can be detected in breasts of women using the ultrasound tomography scanner? Answering this question, we have to estimate the contrast resolution of ultrasound tomography images and verify it by medical testing on a large group of patients healthy and with lesions of different sizes and malignancy degrees. The problem is that the contrast resolution is not constant and depends not only on the scanning and image reconstruction methods but also on the structure of the imaged tissue. This applies of course not only to the UT method but also to all of the previously mentioned.

In the paper, we have estimated the contrast resolution of ultrasound tomography images by means of calculations of projection values of the ultrasound speed and attenuation after passing the ultrasound pulse through the path from the emitter to receiver along the diameter of the array ring of transducers (Fig. 1):

$$c_{projection} = \frac{L_{path}}{t_{water} + t_{breast} + t_{lesion}} = \frac{L_{water} + L_{breast} + L_{lesion}}{L_{water}/c_{water} + L_{brest}/c_{breast} + L_{lesion}/c_{lesion}} \tag{1}$$

$$\alpha_{projection} = \frac{\alpha_{water}\cdot L_{water} + \alpha_{breast}\cdot L_{breast} + \alpha_{lesion}\cdot L_{lesion}}{L_{water} + L_{breast} + L_{lesion}} \tag{2}$$

The model of the simulation of projection measurements consists of the circular section of a homogenous breast glandular tissue located in the plane of the ring array filled with water. Then, the small circular section of a homogeneous lesion can be inserted in the breast centre (Fig. 1). The acoustic parameter values of the *in vivo* breast tissue were adopted for calculations from the literature [7]. Unfavorable measuring conditions for a large coronal section of the breast at its base were assumed. In this way, the share of the lesion in the projection speed and attenuation values is small. Acoustic parameters of water correspond to the temperature of 25°C. Figure 2 shows the calculation results of the accuracy of ultrasound speed and attenuation measurements of projection values. The *Y*-axis represents the values of the difference $\Delta p = (p_{BreastWithLesion} - p_{BreastWithoutLesion})$, where p is the ultrasound speed or attenuation projection value. This difference defines the accuracy of the projection value measurement that is necessary to detect the lesion. The *X*-axis represents the lesion size from 0.1 – 10 mm. The calculations were carried out for 6 different values of the ultrasound speed and attenuation representing breast tissue of different densities, while the ultrasound speed and attenuation of lesion is constant (Fig. 1). These values are shown in Fig. 2 in the form of (c_{lesion} - c_{breast}) or (α_{lesion} - α_{breast}) on the right of graphs. The interpretation of graphs from Fig. 2 is now very clear. For example, if we want to detect the lesion with the size ≥ 5 mm in ultrasound tomography images and the local value of the ultrasound speed in the lesion differs from the value in the surrounded glandular breast tissue by about 5 m/s, we need to ensure the ultrasound speed measurement accuracy at least $\Delta c \approx 0.1$ m/s. We can refer possible to detect lesion sizes to these accuracy values using graphs in Fig. 2.

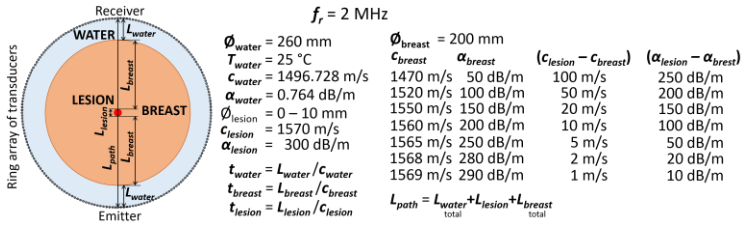

Figure 1. The model of the simulation of projection measurements used to estimate the contrast resolution of UT images

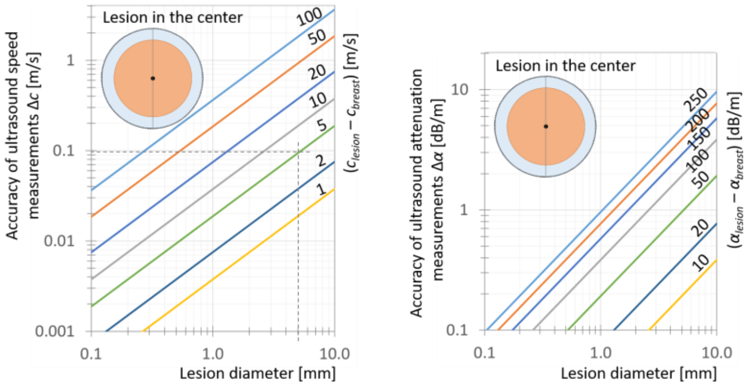

Figure 2. Calculation results of the accuracy of ultrasound speed and attenuation measurements of projection values performed during the UT breast scanning

4. Measurement results

The designed phantom of the agarose gel with 3 holes filled with water (Ø 7 mm) and one filled with water and pieces of the agar gel (Ø 4 mm) has been measured (Fig. 3) in order to assess the accuracy of the projection measurements of the ultrasound speed and attenuation using the ultrasound tomography scanner developed by DRAMIŃSKI S.A. The agar gel characterizes by the ultrasound speed similar to the soft tissue and by very low attenuation so that the impact of noise and signal distortions as well as adverse phenomena accompanying the ultrasonic wave propagation on measured values are small. The measured projection values are presented after the transformation from the divergent scanning geometry to the parallel one. Projection values of ultrasound speed were calculated from measurements of the ultrasonic pulse runtime, and attenuation projection values – from measurements of the pulse amplitude after transition.

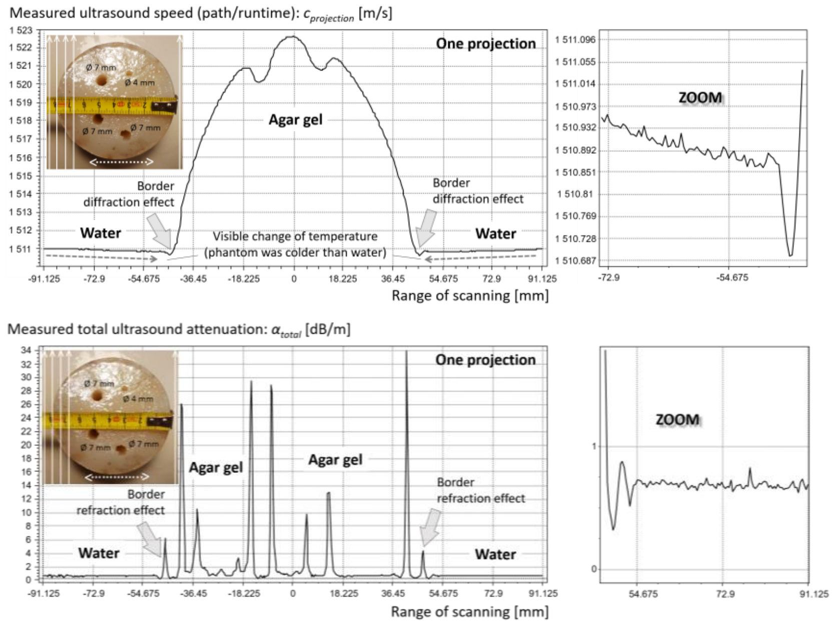

Figure 3. Projection measurements of the ultrasound speed and attenuation using the ultrasound tomography scanner developed by DRAMIŃSKI S.A.

The changes of the pulse amplitude reflects not only the attenuation of ultrasound but also its weakness after passing through the boundaries of different acoustic impedances ($\alpha_{total} = \alpha_{projection} + \alpha_{weakness}$) and are impossible to separate. Therefore, we can expect distortions of reconstructed local values of the ultrasound attenuation distribution in the tomography images. Based on Fig. 3 (see ZOOMs) we can estimate the accuracy of measurements of the ultrasound speed projection as $\Delta c \approx 0.02$ m/s and the accuracy of measurements of the ultrasound attenuation projection as $\Delta\alpha \approx 0.1$ dB/m. Figure 4 shows ultrasound tomography images of the tested agar gel phantom section reconstructed from projection measurements as presented in Fig. 3 together with the distributions of local values of the ultrasound speed and attenuation along the dotted *y*-line marked in the images. Based on Fig. 4 (see ZOOMs) we can estimate the accuracy of the reconstructed local ultrasound speed as $\Delta c \approx 0.3$ m/s and local ultrasound attenuation as $\Delta\alpha \approx 8$ dB/m in this case.

Three commercial breast biopsy phantoms with various acoustic parameters were selected to verify possibilities of the ultrasound tomography imaging in the context of the spatial and contrast resolution (Fig. 5 – Fig. 8).

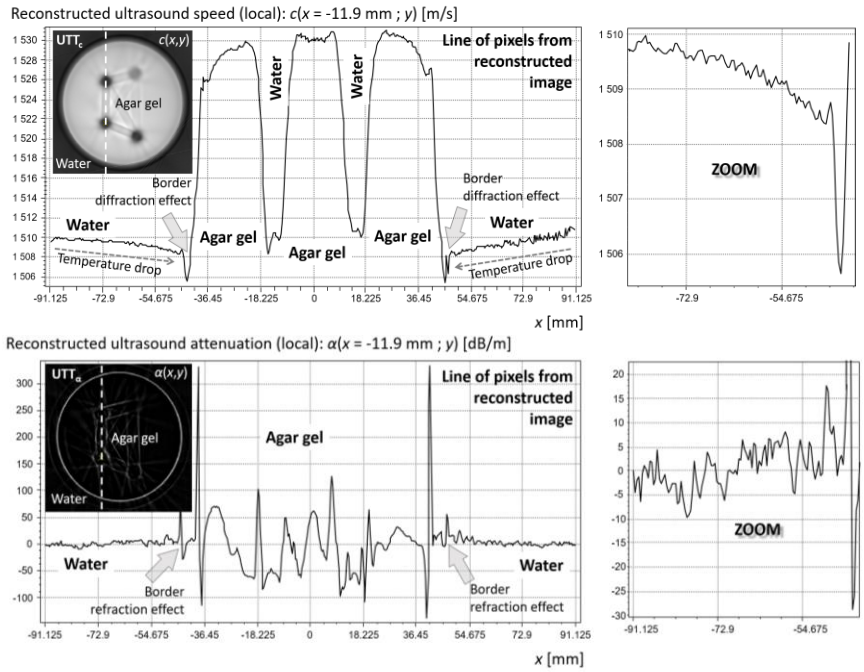

Figure 4. Ultrasound tomography images of the tested agar gel phantom section reconstructed from projection measurements together with distributions of local values of the ultrasound speed and attenuation along the dotted *y*-line marked in the images

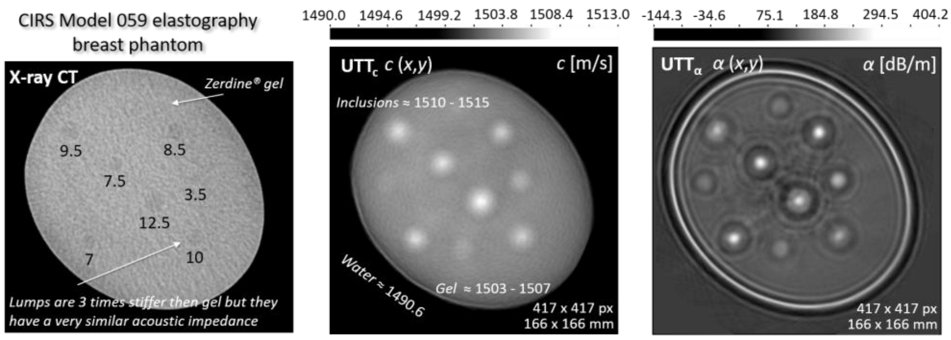

Figure 5. Comparison of X-ray CT image with UTT images of the same section of CIRS Model 059 elastography breast phantom with inclusions of the acoustic impedance very similar as in the phantom gel (low attenuation of ultrasound)

The reconstructed ultrasound images of the ultrasound speed and attenuation distribution in the selected section of these objects were compared with reference X-ray CT image, magnetic resonance images and the conventional US image.

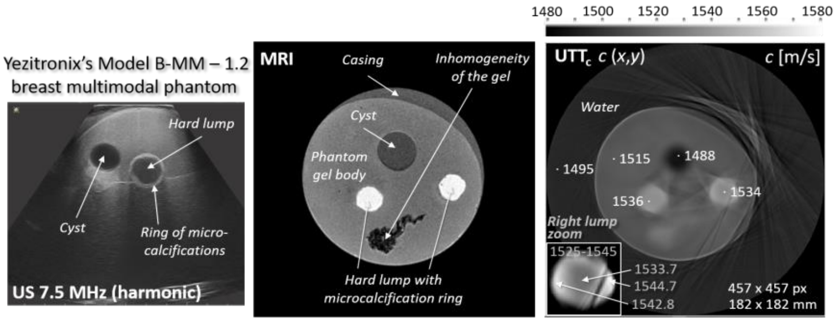

Figure 6. Comparison of harmonic conventional US and magnetic resonance image with sound speed UTTc image of the same section of Yezitronix's Model B-MM – 1.2 breast multimodal phantom with inclusions mimicking cyst and hard lumps surrounded by micro-calcifications (mid attenuation of ultrasound)

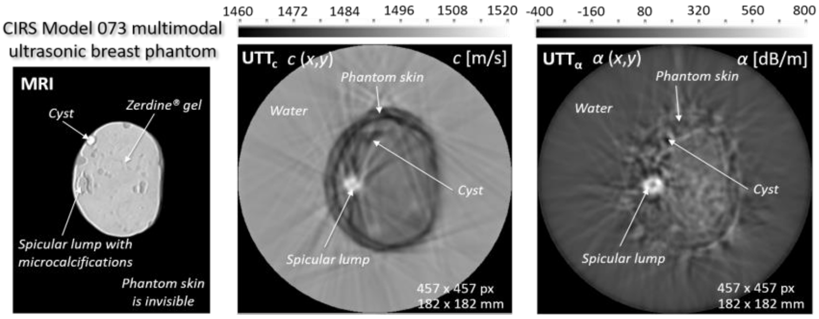

Figure 7. Comparison of magnetic resonance image with UTT images of the same section of CIRS Model 073 breast multimodal phantom with a skin and inclusions mimicking cyst and spicular cancer lump (high attenuation of ultrasound)

5. Discussion and Conclusions

Calculations show that the detection of heterogeneities in coronal breast sections of size about $d_{lesion} \geq 5$ mm differing by the ultrasound speed barely 1 m/s from the surrounding tissue is possible using the ultrasound tomography scanner developed by DRAMIŃSKI S.A. in cases of not noisy projection measurements of the ultrasound speed (Fig. 2). It was confirmed by measurements of the elastography breast phantom (Fig. 5). The size of a detected heterogeneity decreases linearly with an increase in the difference of speeds and in general is $d_{lesion} \geq 0.5$ mm for $(c_{lesion} - c_{breast}) \geq 10$ m/s due to the limitation of the horizontal plane resolution. Edges of small heterogeneities with a similar ultrasound speed in relation to their environment are blurry in UTTc images (Fig. 5, Fig. 6) because ultrasound beam rays pass then by short chords of quasi-circular shapes. This reduces the contrast resolution in such places, in accordance with calculations (Fig. 2). Despite this, it is possible to detect micro-calcifications due to their large value of the ultrasound speed (Fig. 6). The relatively low vertical resolution (5 mm) is not a disadvantage, because it allows us to increase the vertical step of the ultrasound ring array movement to 2 mm and thus reducing the time needed to scan the whole breast. A few millimeter lesions above and below the scanned coronal breast section will be still detected (see Fig. 5). In the case of extremely dense breasts, highly absorbing ultrasound, we should expect a deterioration in contrast resolution due to the decrease of the signal-to-noise ratio and the impact of phenomena accompanying the ultrasonic wave propagation (refraction, scattering, reflection, diffraction) to precise measurements of the pulse runtime and amplitude as well as affecting adversely the image reconstruction process (see distortions of UTT images in Fig. 7).

According to calculations, the detection of heterogeneities in coronal breast sections of size about $d_{lesion} \geq 2.5$ mm differing by the ultrasound attenuation about 10 dB/m from the surrounding tissue is possible in cases of not noisy projection measurements of the ultrasound attenuation (Fig. 2). The size of this way detected heterogeneity decreases linearly with an increase in the difference of attenuations. However, we should expect a falsification of reconstructed local values of the ultrasound attenuation because the projection measurement of the total weakness of the transmitted signal. This is shown in Fig. 4, where large attenuation values are exposed in projection measurements despite a very low attenuation of ultrasound in the agar gel. It is the result of weakening the signal passing through the boundaries of the water and gel. For this reason, very low ultrasound attenuation values (e.g. in water) can be reconstructed as negative (Fig. 5, Fig. 7).

Quantitative UTTc and UTTα images are currently used to develop the algorithm of the automatic recognition of breast tumors and estimation their degree of malignancy in screening test, taking into account the age of the woman and her breast density.

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