Development of transducer arrays for ultrasound–computertomography

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ABSTRACT

Ultrasound computer–tomography (USCT) is a novel ultrasound imaging method capable of producing volume images with both high spatial and temporal resolution. Several thousand ultrasound transducers are arranged in a cylindrical array around a tank containing the object to be examined coupled by water. Every single transducer is small enough to emit an almost spherical sound-wave. While one transducer is transmitting, all others receive simultaneously.

Our experimental setup, using only a few transducers simulating a ring-shaped geometry, showed even nylon threads (0.1 mm) with an image quality superior to clinical in-use ultrasound scanners.

In order to build a complete circular array several thousand transducers, with cylindrical sound field characteristics, are needed. Since such transducer arrays are hardly available and expensive, we developed inexpensive transducer arrays consisting of 8 elements. Each array is based on a plate of lead titanate zirconate ceramics (PZT) sawn into 8 elements of 0.3 mm width, 3.8 mm height and 0.5 mm pitch. Each element has a mean frequency of 3.8 MHz and can be triggered separately.

The main challenge was the development of production steps with reproducible results. Our transducer arrays show only small variances in the sound field characteristics which are strongly required for ultrasound tomography.

Keywords: ultrasound, transducer array, ultrasound computer–tomography, diffraction, circular array, breast imaging

1. ULTRASOUND–COMPUTERTOMOGRAPHY (USCT)

1.1. Diagnosis of breast cancer

Breast cancer is one of the most widespread cancer types among females in the western world. Approximately every tenth woman is threatened by breast cancer in Germany[1, 2]; in the USA this rate is a little higher. X–ray mammography, magnetic resonance imaging and ultrasound are established methods of breast cancer diagnosis. Each method is sensitive to specific tissue types and changes. In medical check–ups mainly X–ray mammography is used, therefore several tissue changes are hardly detectable. In many cases ultrasound examinations lead to additional diagnostic information about e.g. cysts and fibro adenomas. Additionally ultrasound does not harm biological tissue and may be applied frequently. Further advantages of ultrasound examinations are cost and speed.

The disadvantages of conventional ultrasound imaging methods[3] are both poor spatial and temporal resolution. The contrast and the resolution depend highly on the frequency used, as well as on the distance between the transducer array and the region of interest within the breast. The medical doctor operates the transducer array manually and deforms the tissue trying to get as close as possible to desired region. Therefore the image contents and image quality are highly operator dependent, and almost impossible to reproduce. Exact measurements of tissue structures, e.g. tumor size, are hardly possible.

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1.2. USCT at Forschungszentrum Karlsruhe

Ultrasound computer-tomography (USCT)[4–19] is a new imaging method which allows the recording of reproducible images with higher resolution and tissue contrast. In conventional ultrasound imaging a linear transducer array is operated manually and only the tissue reflections are recorded. In USCT the transducers are arranged in a fixed geometry around the object to be examined. For breast examinations the breast is placed in a tank filled with water as a coupling medium. The transducers are mounted in a cylindrical array at the tank walls (see fig. 1).

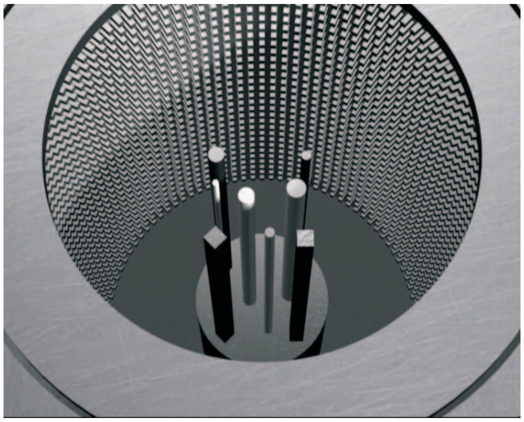

Figure 1. Fixed arrangement of transducers for 3D ultrasound–computertomography. The transducers are arranged in a cylindrical array around a tank containing the object and water as the coupling medium.

One transducer is acting as a sender and emits a short pulse which is scattered by the structures inside the object. Every transducer is small, emitting a nearly undirected beam (spherical wave front). All other transducers measure the transmitted, reflected and scattered signals (A–scan) simultaneously. The received signals are amplified, digitized and stored. Then the next transducer will transmit an ultrasound pulse while all others receive the signals and so on. Figure 2 shows the ring architecture of a two–dimensional ultrasound computer–tomography system and an A–scan of one receiving element.

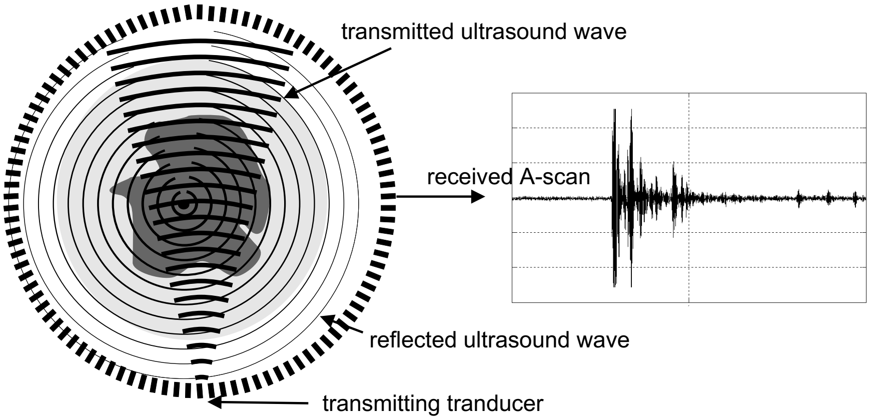

Figure 2. Architecture of the USCT system built in FZK, shown in 2D. A ring (cylinder) of ultrasound transducers encloses the object (left). One transducer emits a short ultrasound pulse, all other transducers receive simultaneously. The A–scan at the right side shows the directly transmitted and scattered signals.

At Forschungszentrum Karlsruhe, we are developing an ultrasound computer–tomography system for breast imaging. We have built an experimental setup[18] (fig. 3) to evaluate ultrasound computer tomography for medical imaging.

Figure 3. Experimental prototype at Forschungszentrum Karlsruhe. The white tank which is filled with water, contains the object to be examined. Two ultrasound transducer arrays are mounted on rotating rings to simulate all emitter and receiver positions in two dimensions.

It consists of two commercial transducer arrays (2.5 MHz) in a water tank, a pulse generator, an amplifier and a digital oscilloscope connected to an external computer. The signal processing and image reconstruction is done by the computer. Both transducer arrays can be positioned independently on the ring to emulate a full circular array (1600 elements, diameter 12 cm). An array consists of 16 elements each 0.2 mm wide, 10 mm high and a pitch of 0.25 mm. One array is used as the emitter, the other as the receiver. Every receiving element is treated separately. For every emitter position, all possible receiver positions on the circle are evaluated and the correspondent signals are recorded successively.

Based on the data recorded by our experimental setup we are able to reconstruct two–dimensional tomographic images. The reconstruction is based on a full aperture sum–and–delay algorithm described in[18] on the assumption of constant sound speed in the water and the object. Furthermore, no corrections of the angle–dependent sensitivity of the transducers have been applied.

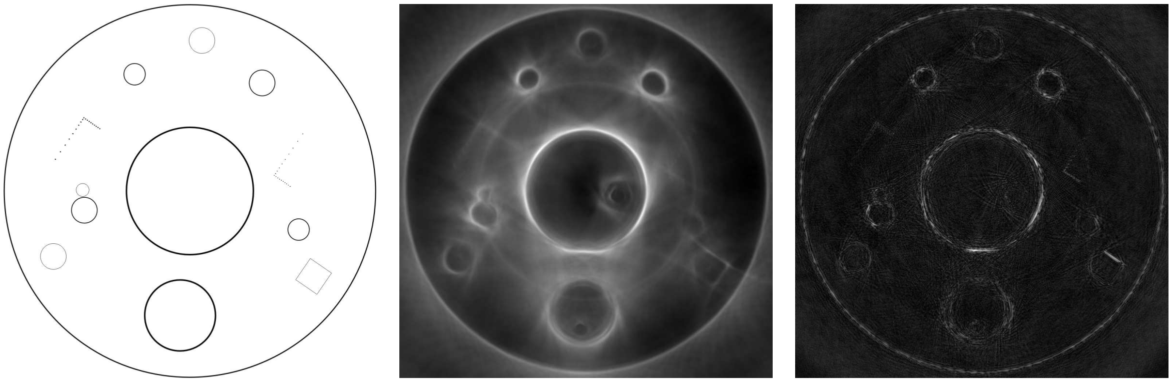

Figure 4. Ultrasound phantom and reconstructions. Left: rough plan of the phantom, the diameter is 8.5 cm. The phantom contains plastics tubes, straws and nylon threads as smallest structures. Middle: reconstruction using only amplitude information. Right: with additional phase information.

Figure 4 and 5 show the results of an image reconstruction of a phantom. Two different types of images can be reconstructed: one using only the amplitude information and a second one including the phase information. Amplitude images show a high contrast and low noise but their image resolution is limited to the length of the

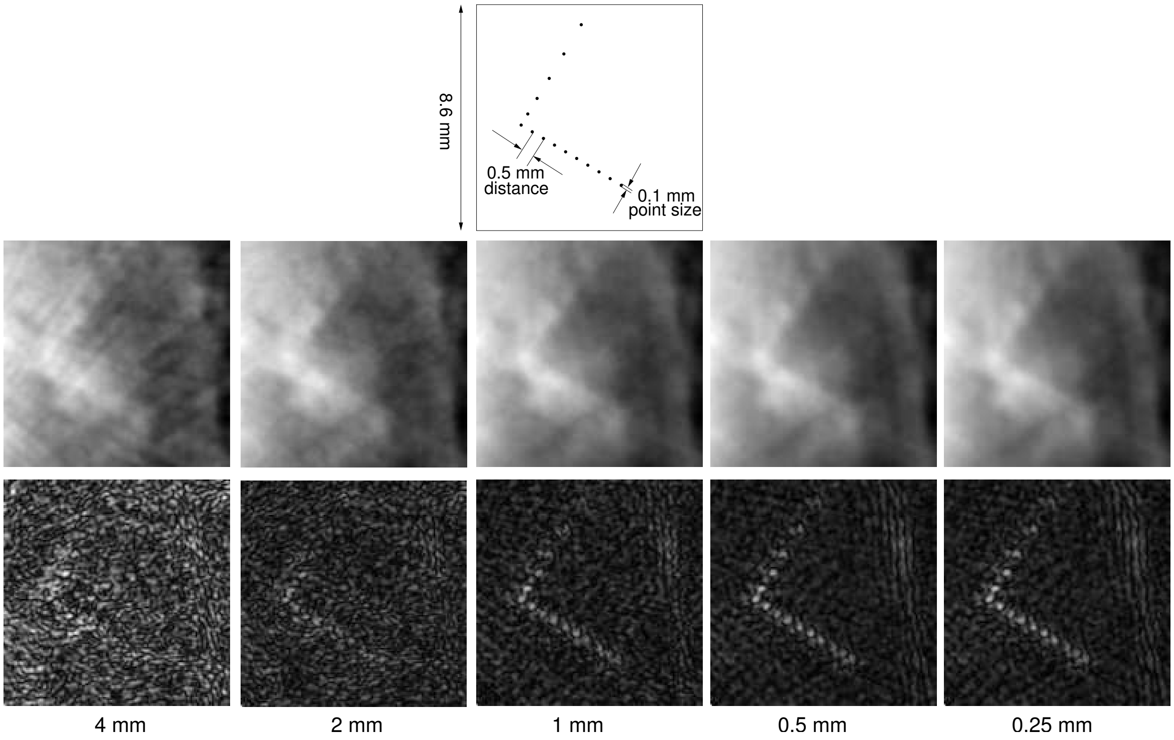

Figure 5. Resolution of ultrasound–computertomography. Top: region enlargement of the plan of the phantom shown in fig. 4. The points are nylon threads with a diameter of 0.1 mm. Middle: reconstructions using only the amplitude information. Bottom: reconstructions with phase information. From left to right: the sensor arrangement on the ring is modified from a sensor distance of 4 mm to 0.25 mm. A sensor distance of 1 mm is sufficient to identify single nylon threads.

ultrasound pulse (approx. 1.5 mm). Including phase information allows the imaging of small structures of the size of 0.1 mm, but are more noisy.

The resolution depends on the sensor geometry, e.g. the distance of receivers on the ring. Figure 5 shows the influence of the sensor distance to the reconstruction quality. The lower nylon threads have a width of 0.1 mm and a gap of 0.5 mm. Even with a sensor distance of 1 mm the single threads are still distinguishable in the phase image.

2. TRANSDUCERS FOR USCT

2.1. Requirements

Commercial sensor arrays cost several hundred dollars each. For our next generation of ultrasound computer–tomography systems we will need approximately 1000 elements to equip a ring with a diameter of 20 cm. The commercial transducer arrays we used, described in section 1.2, are expensive and show a poor reliability. So we have developed our own inexpensive transducer arrays.

We have chosen an ultrasound frequency of 2.5 – 3.5 MHz as a compromise between larger absorption at higher frequencies, and lower resolution due to larger wavelength at lower frequencies. To achieve very short pulses, the transducers need to be damped to produce broadband pulses.

To get a nearly cylindrical wave profile the width of the sensor surface should be smaller than 1/2 of the wavelength, e.g. 0.3 mm at 2.5 MHz. On the other hand the sensitivity of the sensor is reduced with the size of the surface, too. For this reason we have chosen a length of 3.8 mm.

2.2. Production Steps

The design of ultrasound transducer arrays has been described in the literature.[20,21] We used the standard composition and introduce some modifications simplifying the fabrication. The general structure of our transducer array is shown in figure 6. The ultrasound excitation source is a piezo–ceramic plate consisting of modified lead zirconate titanates (PIC 155, PI Ceramic GmbH). The plate has a thickness of 0.39 mm resulting in a resonance frequency of 5 MHz. This frequency is higher than the desired frequency due to the fact that nearly all processing steps will move the resonance to lower frequencies.

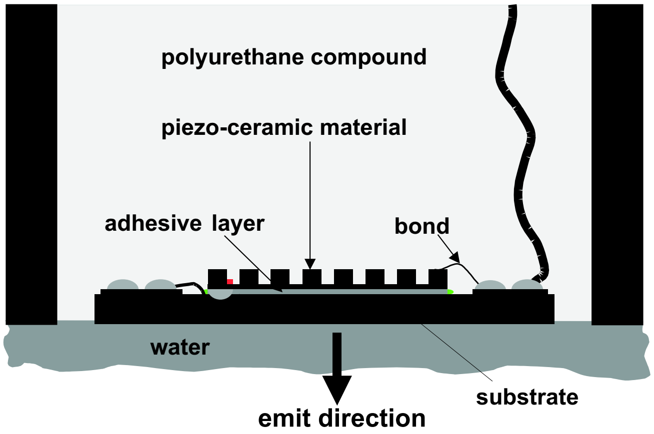

Figure 6. Principle design of an ultrasonic transducer array. The heart consists of a piezo–ceramic plate structured in single elements. Each element is individually connected to conducting paths on a ceramic substrate which is simultaniously used as the circuit board and the ultrasound adaption layer. The housing is filled with a polyurethane compound to damp the oszillations.

An optimal adaption layer[20] between water with an impedance of $Z_W = 1.5$ MRayl and the piezo–ceramic plate $Z_K = 30.4$ MRayl follows the equation

$$Z_A = \sqrt{Z_W * Z_K} \tag{1}$$

where Z_A denotes the impedance of the adaption layer. Temperature stable microwave laminate is one of the most suitable materials with an impedance of $Z_A = 6.4$ MRayl. To be able to customize the substrate thickness to a quarter of the desired wavelength (e.g. 0.2 mm), various material thicknesses are available.

The temperature stable microwave laminate is used as the circuit board, as the carrier for the piezo-ceramic plate and as the adaption layer between the piezo–ceramic plate and water. The different production steps are:

1. The lower electrode of the piezo–ceramic plate is connected to the circuit board.
2. The piezo–ceramic plate is positioned with high precision flip–chip–placer and is glued on the circuit board.
3. After gluing the piezo-ceramic plate is structured using a dicing saw into 8 elements. Each element has a size of 0.3×3.8 mm^2 with a gap of 0.2 mm. In figure 7 (left) the structure of the piezo-ceramic plate and the laminate carrier with the circuit board are shown.
4. Each element is wired to the circuit board with an automatic ultrasonic wire bonder.
5. The outgoing cables are connected to the circuit board and all parts are integrated in a housing filled with ployurethane compound. The filling material shields the tiny strutcures of the transducer array against impacts, seals the array and serves as backing layer to damp the ultrasonic oscillations. The quality of the backing layer can be influenced by different loadings of additives to increase the absorption of acoustic energy, e.g. aluminium oxyde.

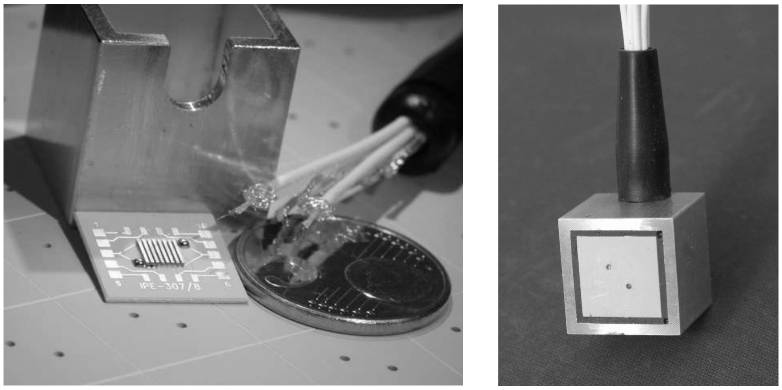

Figure 7. Left: Prototype of an ultrasonic transducer array consisting of 8 elements. Each element has a mean frequency of 3.8 MHz, 3.8 mm length and a pitch of 0.5 mm. Right: Assembled ultrasonic transducer array.

3. RESULTS

The main challenge was the development of production steps to give reproducible results. We used our machinery in operation for packaging of integrated circuits for precise manufacturing. We produced five ultrasonic transducer arrays (figure 7 right) as prototypes with the following characteristics:

number of elements per array	8
element length	3.8 mm
element width	0.3 mm
element pitch	0.5 mm
mean frequency	3.8 MHz
bandwidth	1.35 MHz
reproducibility	± 2 %

The low price (5 – 10 USD per array) and the high reproducibility encourage us to produce several hundred transducer arrays. A complete ring of transducers will enhance our next prototype of ultrasound computertomography. In addition with massive parallel data aquisition hardware we will be able to record the ultrasonic signals of all receiving positions in real–time.

4. FUTURE DEVELOPMENT

In future we will develop a new ultrasonic transducer array for three–dimensiomal ultrasound tomography. The transducer arrays will cover the tank walls to form a cylindrical array. We will use small square tranducer elements with a mean frequency of 1.5 – 2.5 MHz. The shape of the transducer elements will be designed to receive ultrasound signals in three dimensions. Due to the required size of 0.5 – 1.0 mm^2 of the elements, the sensitivity of the elements will be angle dependent. This can be compensated for by a greater distance between the elements and the object (e.g. greater diameter of the cylinder) and by software.

The circuit board of the linear transducer array described in section 3 is merely used to connect the transducer elements to the cables. We want to equip the transducer housing with integrated amplifier electronics for a better electrical impedance matching to reduce the noise and to enhance the sensitivity. Also included are emitting units and the electronics for pulse generation. Multiplexing will reduce the amount of cables. The circuit board will be designed consisting of three parts (figure 8). The middle part will carry the structured piezo–ceramic plate with the transducer elements and their connecting bonds to the circuit board. Both other parts will carry the amplifier, multiplexer and pulse generation electronics. The three–part circuit will be folded: The middle part will form the transducer surface, both other sides will form the housing filled with damping material.

Most of the parts of these ultrasound transducer array systems can be automatically manufactured in series, using our machinery for packaging of integrated circuits, with high precision and low costs.

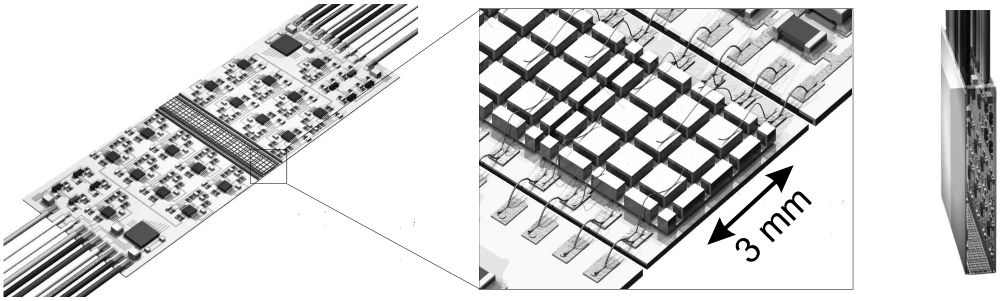

Figure 8. Next generation of ultrasonic transducer array systems. A system consists of a piezo–ceramic transducer array with square elements and a three–part circuit board carrying the transducer array, the amplifier electronics and multiplexer to control the receiving elements and a pulse generator for emitting. The middle picture shows the elements and their connecting bonds in detail. The tree–part circuit board (right) is folded and the inner volume is filled with damping material. The bottom part carries the tranducer elements and is used as the adaption layer.

REFERENCES

1. P. Bannasch, *Cancer Diagnosis. Early Detection.*, Springer–Verlag, 1992.
2. D. von Fournier, H. J. H.-W. Anton, and G. Bastert, "Breast cancer screening," in *Cancer Diagnosis Early Detection*, pp. 78–87, Arbeitsgemeinschaften der Grossforschungseinrichtungen (AGF), Springer, 1992.
3. Z.-H. Cho, J. P. Jones, and M. Singh, *Foundations of Medical Imaging*, John Wiley & Sons, Inc., 1993.
4. H. Schomberg, "An improved approach to reconstructive ultrasound tomography," *J. Phys. D.: Appl. Phys.* **11**, 1978.
5. S. J. Norton and M. Linzer, "Ultrasonic reflectivity tomography: Reconstruction with circular transducer arrays," *Ultrasonic Imaging* **1**, pp. 154–184, 1979.
6. S. J. Norton and M. Linzer, "Ultrasonic reflectivity imaging in three dimensions: Reconstruction with spherical transducer arrays," *Ultrasonic Imaging* **1**, pp. 210–231, 1979.
7. K. A. Dines and A. Kak, "Ultrasonic attenuation tomography of soft tissues," *Ultrasonic Imaging* **1**(1), pp. 16–33, 1979.
8. J. Greenleaf, J. J. Gisvold, and R. Bahn, "Computed transmission ultrasound tomography," *Medical Progress through Technology* **9**, pp. 165–170, 1982.
9. J. R. Jago and T. A. Wittingham, "Experimental studies in transmission ultrasound computed tomography," *Med. Phys. Biol.* **36**, pp. 1515–1527, 1991.
10. M. Nguyen, H. Bressmer, P. Kugel, and U. Faust, "Improvements in ultrasound transmission computed tomography," in *Technology and Health Care, Proceedings of the second European Conference of Engineering and Medicine in Stuttgart*, pp. 189–190, 1993.
11. J. Greenleaf, "3D and tomographic ultrasound," in *Medical CT & ultrasound: current technology and applications*, L. W. Goldman, ed., pp. 267–284, Madison, Wisc.: Advanced Medical Publ., 1995. ISBN 1-883526-03-5.
12. M. Krueger, A. Pesavento, and H. Ermert, "A modified time–of–flight tomography concept for ultrasonic breast imaging," *1996 IEEE Ultrasonics Symposium* , pp. 1381–1385, 1996.
13. H. Schlaberg, M. Yang, B. Hoyle, M. Beck, and C. Lenn, "Wide-angle transducers for real–time ultrasonic process tomography imaging applications," *Ultrasonics* **35**, pp. 213–221, 1997.
14. M. Krueger, A. Pesavento, and H. Ermert, "Ultrasonic breast imaging assisted by acousic velocity reconstruction," *Acoustical Imaging* **23**, pp. 101–106, 1997.
15. M. Yang, H. I. Schlaberg, B. S. Hoyle, M. S. Beck, and C. Lenn, "Real–time ultrasound process tomography for two-phase flow imaging using a reduced number of transducers," *IEEE Transactions on Ultrasonics, Ferroelectrics and Frequency Control* **46**, pp. 492–501, May 1999.

16. R. Stotzka, J. Würfel, K. Scholte–Holubek, H. Gemmeke, and W. Kaiser, “USCT: Ultraschall–Computertomographie,” in *35. Jahrestagung der Deutschen Gesellschaft für Biomedizinische Technik*, 2001.
17. M. Ashfaq and H. Ermert, “Ultrasound spiral CT for the female breast — first phantom imaging results —,” in *35. Jahrestagung der Deutschen Gesellschaft für Biomedizinische Technik*, 2001.
18. R. Stotzka, J. Würfel, and T. Müller, “Medical imaging by ultrasound-computertomography,” in *SPIE's Internl. Symposium Medical Imaging 2002*, pp. 110 – 119, 2002.
19. R. Leach, S. Azevedo, J. Berryman, H. Bertete-Aguirre, D. Chambers, and J. Mast, “Comparison of ultrasound tomography methods in circular geometry,” in *SPIE's Internl. Symposium Medical Imaging 2002*, pp. 362 – 377, 2002.
20. R. McKeighen, “Design guidelines for medical ultrasonic arrays,” in *SPIE's Internl. Symposium Medical Imaging 1998*, pp. 2 – 18, 1998.
21. J. Krautkrämer and H. Krautkrämer, *Ultrasonic Testing of Materials*, Springer, 4 ed., 1990.