Prototype of a new 3D ultrasound computer tomography system: transducer design and data recording

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ABSTRACT

Ultrasound computer tomography is an imaging method capable of producing volume images with both high spatial and temporal resolution. The promising results of a 2D experimental setup of an ultrasound computer tomography system with at least 0.25 mm resolution encouraged us to build a new 3D demonstration system. It consists of three parts: a tank containing the sensor system, a data acquisition hardware and a computer workstation for image reconstruction and visualization.

For the sensor system we developed and manufactured our own low–cost transducer array emitting or receiving ultrasound signals in three dimensions. To optimize the transducer geometry in respect to aperture angle and pressure amplitude the pressure field was simulated using the ultrasound simulation program Field II. Each transducer array system carries 8 emitting and 32 receiving elements with integrated amplifier and address electronics.

192 A–scans can be recorded in parallel by the data acquisition hardware. 48 multiplexing steps are needed to store all A–scans of the 1536 receiving transducers. After recording the data is transmitted to the computer workstation.

Keywords: ultrasound computer tomography, ultrasound, transducer array, simulation, data acquisition, 3D array, breast imaging

1. ULTRASOUND COMPUTER TOMOGRAPHY (USCT)

1.1. Breast Cancer

Breast cancer is the most frequent cancer of women. For a successful therapy it is crucial to detect the tumor before external metastases are produced.[1] In breast cancer diagnosis, ultrasound examination provides useful additional diagnostic information. Moreover ultrasound does not harm biological tissue and can be applied frequently. But conventional ultrasound imaging methods lack both high spatial and temporal resolution. Additionally, the scanner is operated manually and the tissue is deformed while getting as close as possible to the regions of interest. Therefore, image contents and image quality depend strongly on the operator.[2] Interpretation of ultrasound images, e.g. discrimination of benign and malignant structures, is very difficult and requires many years of experience. Exact measurement of tissue structures, like tumor size, is not possible.

1.2. Principle of USCT

Ultrasound computer tomography (USCT) is a new imaging method which allows the recording of reproducible images with high resolution and tissue contrast.[3,4] In conventional ultrasound imaging a linear transducer array is operated manually and only the reflections are recorded. In USCT the object is placed in a tank filled with water as coupling medium. The tank walls are covered with transducers in a fixed geometry (figure 1), building a cylindrical array surrounding the imaged object.

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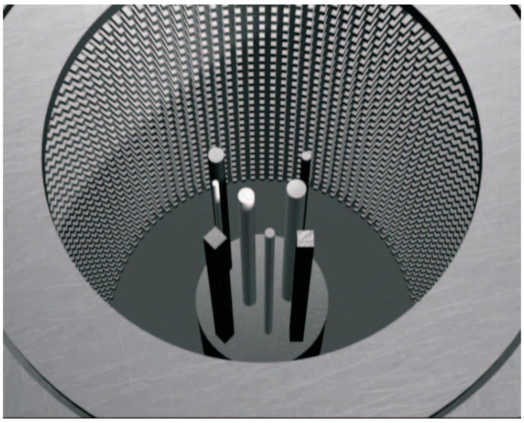


Figure 1. Fixed arrangement of transducers for 3D ultrasound computer tomography. The transducers are arranged in a cylindrical array around a tank containing the object and water as the coupling medium.

One transducer acts as a sender and emits a short pulse which is scattered by the structures inside the object. Every transducer emits a nearly undirected beam (spherical wave front). All other transducers measure the transmitted, reflected and scattered signals (A–scans) simultaneously. The received signals are amplified, digitized and stored. Then a different 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 a receiving element.

At Forschungszentrum Karlsruhe we have built a 2D experimental setup to evaluate a multi–sensor system which regards transmitted, scattered and reflected signals at the same time. It consists of two transducer arrays (2.5 MHz) in a water tank, a pulse generator, an amplifier and a digital data acquisition system 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 and 10 mm high with a pitch of 0.25 mm. One array is used for emitting, the other for receiving. 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.

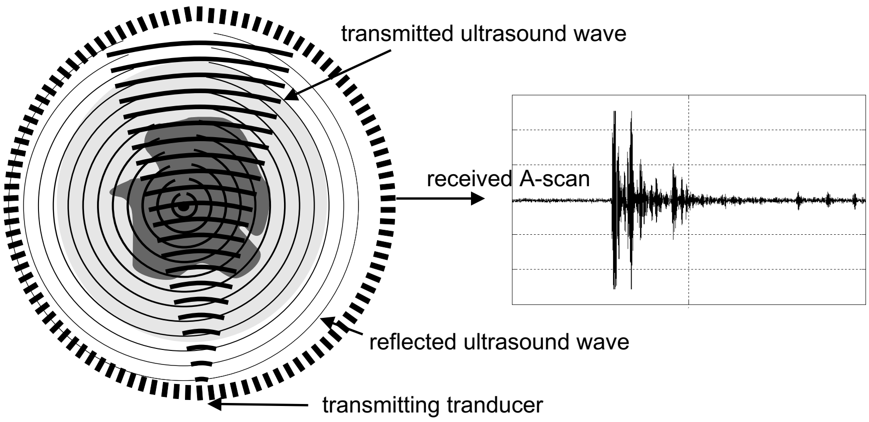


Figure 2. Architecture of the USCT system built in our institute, shown in 2D. A ring (cylinder) of ultrasound transducers encloses the object (left). One transducer emits a short ultrasound pulse, all other transducers receive simultaneously. The A–scan at the right side shows the directly transmitted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

1.3. Preliminary Results

Since the measuring time for the experimental setup takes up to 10 hours, we needed static phantoms mimicking biological tissue. We created several phantoms from a gelatine and olive–oil emulsion with straws and threads as inner structures. The whole imaging system is two–dimensional in a cylindrical slice with a height of about 1 cm and a diameter of 12 cm. For the phantom, we built a model as sketched in figure 3 which we submerged inside a plastic cup with very thin walls. The smallest structures within the phantom are nylon threads with a diameter of 0.1 mm each, corresponding to approximately 0.2 wavelengths of the emitted ultrasound signal. We have recorded the ultrasound signals of the phantom for 100 emitter positions and 1000 receiver positions each, gathering approximately 5 GBytes ultrasound data. Then signals were preprocessed using a low–pass filter and a blind deconvolution method similar to Jensen[5] to sharpen the scattered signals.

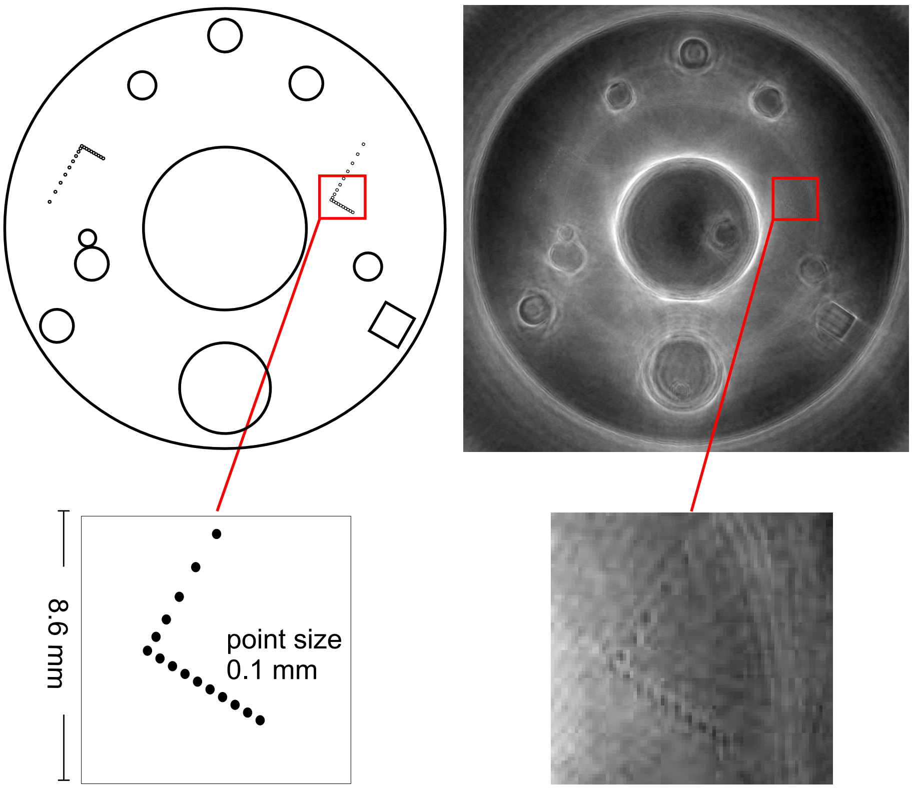


Figure 3. Phantom and reconstructed images. Left: Rough plan of the phantom. The diameter is 8.5 cm. The smallest structures consist of nylon threads with a diameter of 0.1 mm and a spacing of 0.5 mm shown in the region enlargement below. Right: Reconstruction using a full aperture sum–and–delay algorithm. The nylon threads are clearly visible.

The reconstruction itself is based on a full aperture sum–and–delay algorithm 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 3 shows a reconstructed cross–section of the phantom. The nylon threads with a diameter of 0.1 mm and a pitch of 0.5 mm are clearly visible and discernible, resulting in a spatial resolution superior to 0.25 mm.

2. DEVELOPMENT OF A 3D SYSTEM

Using the two–dimensional experimental setup described in the previous chapter it is not possible to image living biological tissue. Yet the promising results encourage us to develop a new ultrasound computer tomography demonstration system which is capable to record the ultrasound data in real–time and in three dimensions.

2.1. Transducer Design

Several hundred transducers are necessary to cover the tank walls with a dense population of transducers according to figure 1, since the signal–to–noise ratio of the reconstructed images correlates directly with the amount of received A–scans. Since commercial transducer systems with sufficient low costs and very reproducible sound characteristics are not available we decided to develop low–priced transducer arrays.

Our first transducer array[6] was designed to equip the 2D demonstration system described above. The manufacture steps have been optimized to assemble the transducer arrays almost automatically to archieve reproducible sound characteristics (± 2 %). The geometry of the sensor arrays are optimized to three–dimensional emission and receiving operation. Therefore a new design of an ultrasonic transducer array system has been developed.

2.1.1. Requirements

A transducer array system for 3D ultrasound computer tomography requires the following ultrasound characteristics:

Frequency We have chosen an ultrasound frequency of 2.5 – 3.5 MHz as a compromise between larger absorption at higher frequency, and lower resolution due to the greater wavelength at lower frequencies.

Bandwidth To generate short ultrasonic pulses, the transducers need to be damped to reach a bandwidth of 50 %.

Aperture angle Spherical sound field characteristics are ideal for the computer tomography, to irradiate the complete object and to detect all scattered and reflected signals from the object. For this case the size of the transducer surface should be smaller than 1/2 of the wavelength, e.g. 0.3 mm at 2.5 MHz. Increased surfaces result in smaller aperture angles.

Sound pressure Small transducers offer small sound pressure amplitudes for emitting transducers and low sensitivity for receiving ones. Consequently the chosen geometry of the transducer surface has to be a compromise between the sound amplitude and the aperture angle.

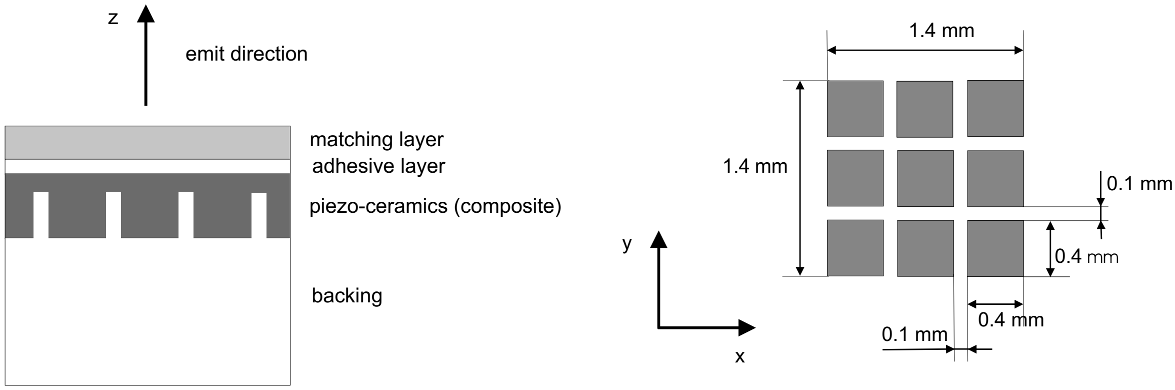
To develop the optimal geometric design of the transducer elements we followed a two–step approach consisting of building a prototype transducer element, measuring its sound fields and simulating the sound fields (see section 2.1.3). The prototype transducer element enables us to determine the modelling parameters for the simulation. In the simulation we optimize geometric parameters, e.g. the size of the transducer surface, to find a compromise between the sound pressure and the aperture angle.

2.1.2. Prototype transducer element

The architecture of the transducer element is shown in figure 4. The ultrasound excitation source is a piezo–ceramics plate consisting of modified lead zirconate titanate (PIC 155, PI Ceramic GmbH). Its thickness of 0.42 mm leads to an initial resonance frequency of approx. 4 MHz. The piezo–ceramics is structured as a composite by sub–dicing.[7] The unstructured side of the piezo–ceramics is glued on a temperature stable laminate, which is used as the circuit board for the electrical contacts and as the matching layer.

The outgoing cables are connected to the circuit board and all parts are integrated in a housing filled with a polyurethane compound. The filling material shields the tiny structures of the transducer array against impacts, seals the array and serves as backing layer to damp the ultrasonic oscillations.

Every manufacturing step leads to a frequency downshift until a final resonance frequency of 3.25 MHz was reached. By impedance measurements the oscillation behavior was determined. We used a PVDF needle hydrophone to measure the sound pressure field in water. The results are shown in figure 7(b).



(a) Cross–section of the prototype transducer element

(b) Surface of the sub–diced piezo–ceramics

Figure 4. Design of the prototype transducer element. Left: cross–section of the sub–diced piezo–ceramics with adhesive layer, matching layer and backing. Right: The surface of the piezo–ceramics is sub–diced in 9 sub–elements.

2.1.3. Simulation of the sound pressure field

To find the appropriate geometry of the transducer surface according to the requirements we applied a computer simulation to generate the sound pressure field. In a first step we modelled and simulated the prototype transducer element, as shown in figure 4. Secondly we verified the simulation by comparing its results to the measured sound field.

For simulating the ultrasound pressure field, we used the Matlab library Field II.[8] It is based on the Tupholme–Stepanishen method,[9, 10] which gives an exact solution for a transducer element, modelled as a planar piston vibrating in an infinite rigid baffle. The sound medium is assumed as homogeneous with linear propagation. To calculate the pressure field Huygens‘ principle[11] is used: Every point of the oscillating transducer surface acts as the origin of an outgoing spherical wave. The sound pressure at every field point is calculated by superposition of all these spherical waves.

The parameters of the Field II model depend on the sound medium, the transducer and its excitation voltage:

The **sound medium** is specificated by its sound speed, density and attenuation. In our simulation we used water as sound medium and its parameters specified in table 1.

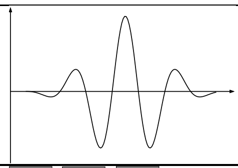
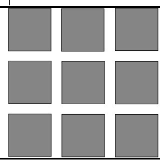
The **transducer** itself is modelled by defining the aperture geometry of the oscillating area (which corresponds to its surface) and the electromechanical impulse response. The electromechanical impulse response is defined as the acceleration of the transducer surface by driving the transducer with a Dirac pulse. In the simulation the real acceleration of the transducer surface is calculated from the convolution of the exciting voltage and the electromechanical impulse response. We modelled the electromechanical impulse response as a gaussian pulse. The amplitude A of the waveform is given by

$$A = \frac{1}{\sigma\sqrt{2\pi}} e^{-\frac{(f-f_0)^2}{2\sigma^2}} \tag{1}$$

where f denotes the frequency, f_0 the mean frequency and σ^2 the variance. The bandwidth B is determined by $B = 2.355\,\sigma$. We fitted the mean frequency and bandwidth to the impedance measurements of the prototype transducer element. Its amplitude was scaled by adapting the simulated pressure to the measured pressure by the needle hydrophone. The aperture of the oscillating area is determined by the sub–diced surface (figure 4(b)).

The model and its parametrization used in the Field II simulation package does not consider lateral oscillations within the piezo–ceramics and effects caused by the adhesive layer or the matching layer. Furthermore very small

Table 1. Simulation parameters of the prototype transducer element in Field II.

sound speed	$1498\,m/s$
density	$1000\,kg/m^3$
attenuation	$0\,dB/m$
electromechanical impulse response	 $f_0 = 3.25\,MHz$ $B = 1.7\,MHz$
aperture geometry of the the transducer surface	 width $= 1.4\,mm$ area $= 1.44\,mm^2$
sampling frequency	100 MHz

variations in the manufacturing process (e.g. thickness of the adhesive) cause changes in the emitted sound field distribution. Thus we did not expect an exact match of the measured sound field of the prototype transducer element and the simulations.

2.1.4. Simulation results

With the parameterized prototype transducer element we simulated the sound pressure at every coordinate in the field. The coordinates are shown in figure 5. To assure simulations in the far field only we chose a minimum distance of 1 cm.

Figure 6 shows the simulated sound pressure field in the xz–plane. Due to the symmetry of the transducer element it is similar to the yx–plane. The highest pressure is reached in the direction of the transducers z–axis. The aperture angle with an amplitude decrease of 50 % is about 30°, which is appropriate to the requirements.

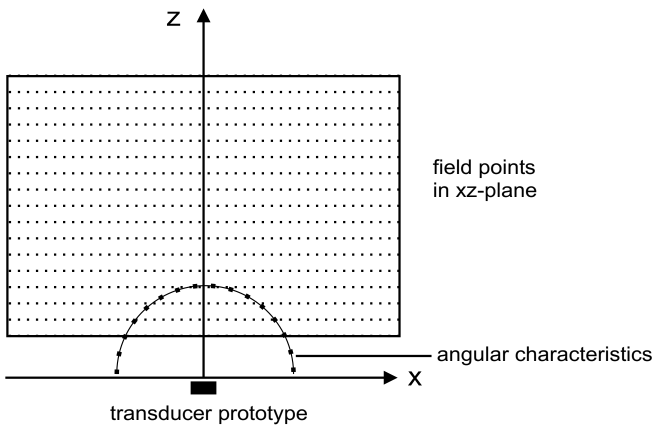


Figure 5. Coordinates of the simulated field in the xz–plane. The transducer surface is positioned in the xy–plane with its edges parallel to the axes. Its center point is positioned in the origin of the coordinate system. The coordinates to measure the angular characteristics are located on a semicircle in the xz–plane, radius 20 mm.

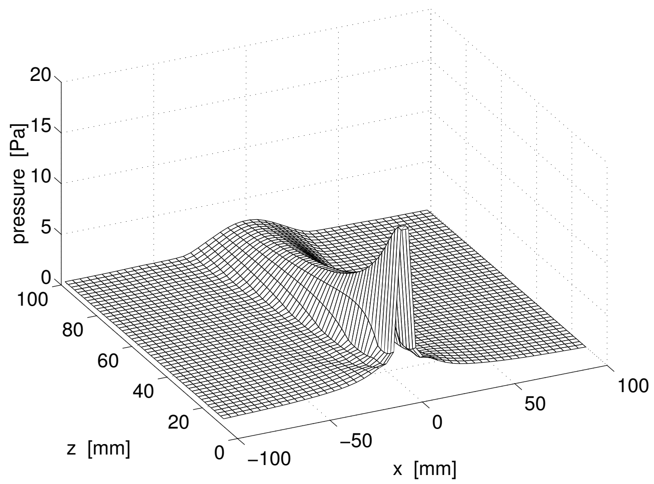
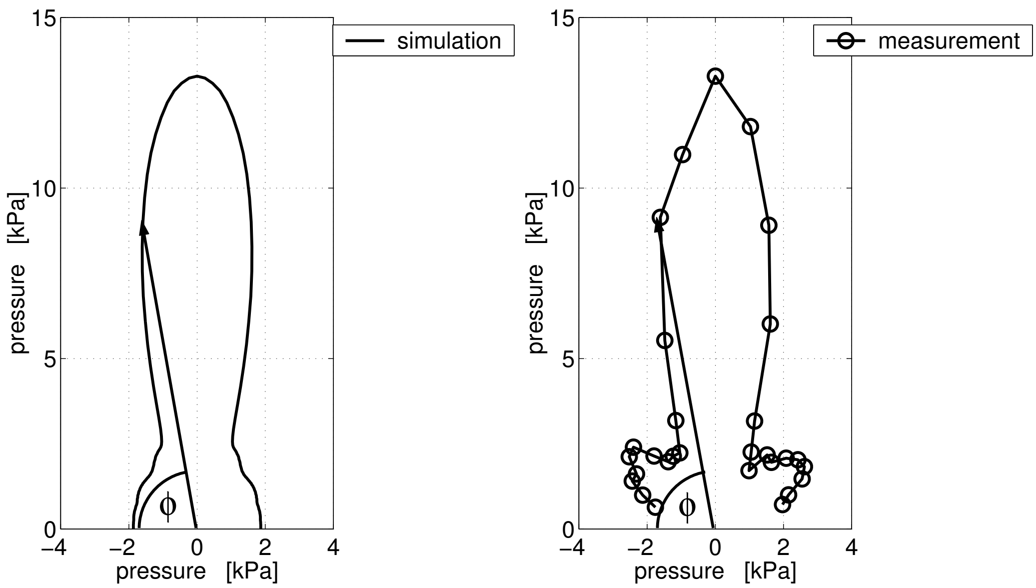


Figure 6. Simulated sound pressure in the xz–plane. The transducer is located in the origin of the coordinate system.

To verify the simulations we compared the angular characteristics of the simulated field to the measured angular characteristics of the prototype transducer element (figure 7). The maximum error of the measurement caused by noise of the hydrophone and the electrical amplifiers has been determined as $\pm 1.5\,kPa$, which corresponds to approximately 10% of the maximum pressure. If we compare the simulated angular characteristics



(a) Simulated angular characteristics. (b) Measured angular characteristics.

Figure 7. Simulated (a) and measured (b) angular characteristics of the prototype transducer element. The distance from the origin to a point of the curve corresponds to the maximum pressure in the direction described by the angle Φ as shown in the graphs. Under consideration of the measurement error the simulated and the measured characteristics are almost identical.

(figure 7(b)) and the measured ones and consider the maximum error, both curves match in an aperture range of more than 60°. The results are sufficient to model new transducer geometries and to predict their sound pressure fields.

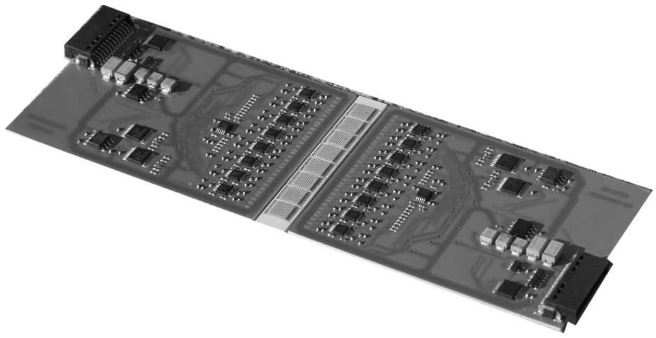
In our case the geometry of the prototype transducer element satisfies the requirements for 3D ultrasound computer tomography.

2.1.5. Architecture of the transducer array system

We used the geometry of the prototype transducer element described above to design the transducer arrays.

One transducer array system (figure 8) consists of 40 transducer elements with integrated amplifier electronics to achieve very short feed lines to the transducers. Each element is individually addressable, 32 of them are used as receivers and 8 as emitters. Multiplexing reduces the amount of cables and assemblies.

(a) Folded and sealed transducer array system



(b) Interior view of the transducer array system

Figure 8. Ultrasonic transducer array system with 8 emitters, 32 receivers at the front and integrated controller and amplifier electronics. Figure (b) shows the interior of a transducer array system. The structured piezo–ceramics plate is placed in the center. It is connected to the controller and amplifier electronics at both sides.

The circuit board of the system (figure 8(b)) forms the transducer surface and the housing. The middle part carries the sub–diced piezo–ceramics plate with the transducer elements, which are bonded to the electronics. Both sides with the electronic are folded and form the housing. The inside is filled up with polyurethane compound for damping and for watertight sealing.

2.2. Data Recording

The new 3D ultrasound computer tomography system consists of three parts: a tank containing the sensor system, a data acquisition hardware and a computer workstation.

The observation tank has a diameter of 18 cm and a height of 15 cm. 48 transducer array systems will be mounted into the tank walls carrying 32 receiving and 8 emitting elements each. The transducer elements can be accessed individually. The resulting cylindrical array will be rotated in 6 steps to fill the gaps between the transducers. Thus a fully covered cylindrical array will be emulated. This results in a total of 9216 receiving and 2304 emitting positions. The received signals and the control signals are gathered in four blocks containing preamplifiers, the address generation and control logic.

The data acquisition electronics is a modified design of the system for the Auger fluorescence detector.[12] It is based on 9 9U Eurocard boards connected by a modified VME–bus. One board controls the data acquisition process, the data storage and the transfer to the computer workstation. The other 8 boards are carrying 24 data recording channels each. A channel consists of the analog signal processing, an A/D converter (10 MHz, 12 Bit)

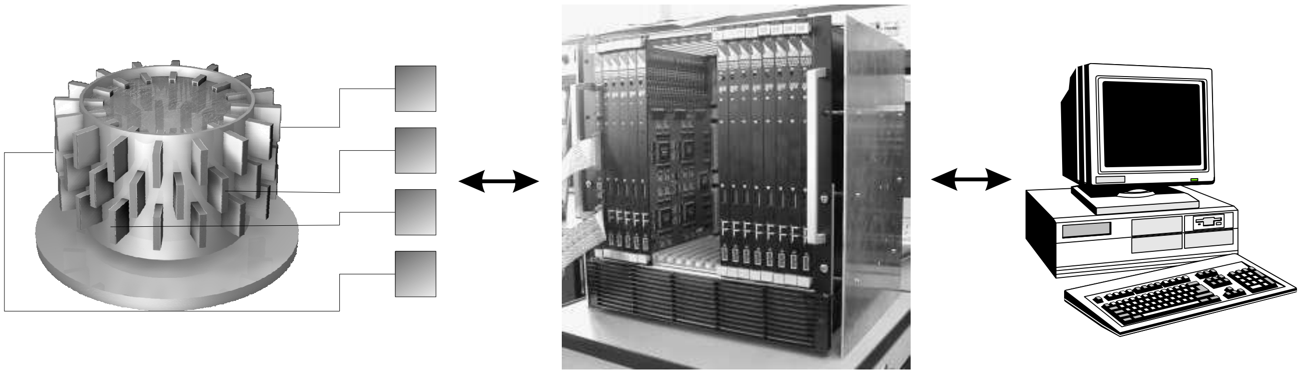


Figure 9. Overview of the data recording system. Left: Tank with 48 mounted transducer array systems. Middle: Data acquisition unit with 192 channels. Right: Computer system for image reconstruction.

and the digital signal processing. The digital processing is based on an array of four FPGAs executing the online data storage, noise reduction and simple data reduction.

While one emitting transducer is activated with coded excitation, all receiving transducers receive simultaneously the scattered signals. 192 channels are sampled and recorded in parallel. 48 multiplexing steps are needed to record all A–scans of the receiving transducers. The data is transfered to the computer workstation and the next emitting transducer is selected.

The data acquisition system and the workstation are connected via a PCI interface. In future a Firewire II interface will be used. The archival data storage, the image reconstruction and the visualization of the results are accomplished on the computer workstation.

3. DISCUSSION

We have shown that the simulation accuracy of composite transducer elements is sufficient to parameterize the geometry of transducer elements without manufacturing and testing new prototypes. Effects caused by lateral oscillations within the transducer and by the layers attached to the piezo–ceramics are only visible beyond an aperture angle of approximately 60°. These can be safely neglected for our application. Using the simulations we are able to find a proper transducer element geometry in respect to our requirements. Furthermore the reflections of point scatterer can be simulated for arbitrary transducer geometries and transducer positions. This allows to determine the point spread functions for different transducer arrangements in ultrasound computer tomography.

Our new 3D ultrasound computer tomography demonstration system is currently under construction. The ultrasound transducer array systems are ready designed and currently manufactured. The nearly entire automatic manufacture process guaranties high quality and high reproducibility of the transducer characteristics (± 2 %) and low costs (100 − 150 USD per system including frontend electronics). The development of the data acquisition boards and the corresponding control software will be completed in summer 2004. The reconstruction algorithms for 3D ultrasound computer tomography are already implemented in Matlab.

In summer 2004 we plan to start our first measurements. The data recording for a quick cross section consisting of 96 different emitting positions and 60 receiving positions each corresponding to $3.5 * 10^4$ A–scans. Each will take less than 15 seconds. This quick mode will allow to image even living biological tissue. The decreased number of recorded A–scans (compared to the 2D experimental setup) would lead to a reduced quality of the images, but will be partly compensated by the higher quality of the recorded signals. The signal–to–noise ratio of the A–scans will be increased significantly by the embedded amplifiers of the new transducer array systems.

After the initial startup of the 3D ultrasound computer tomography demonstration system we will focus on the acceleration of the ultrasound computer tomography system and on additional image reconstruction algorithms.

REFERENCES

1. P. Bannasch, *Cancer Diagnosis. Early Detection.*, Springer–Verlag, 1992.
2. Z.-H. Cho, J. P. Jones, and M. Singh, *Foundations of Medical Imaging*, John Wiley & Sons, Inc., 1993.
3. 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.
4. R. Stotzka, J. Würfel, and T. Müller, " Medical Imaging by Ultrasound–Computertomography," in *SPIE's Internl. Symposium Medical Imaging 2002*, pp. 110 – 119, 2002.
5. J. Jensen, "Deconvolution of ultrasound images," *Ultrasonic imaging* **14**, pp. 1–15, 1992.
6. R. Stotzka, G. Göbel, and K. Schlote-Holubek, " Development of transducer arrays for ultrasound–computertomography," in *SPIE's Internl. Symposium Medical Imaging 2003*, pp. 513 – 520, 2003.
7. W. Smith, "Piezocomposite materials for acoustical imaging transducers," *Acoustical Imaging* **21**, pp. 121 — 138, 1995.
8. A. J. Jensen, "Field: A Program for Simulating Ultrasound Systems," in *Medical & Biological Engineering & Computing*, **34 (1-1)**, pp. 351–353, 1996.
9. P. Stepanishen, "Transient radiation from pistons in an infinite planar baffle.," *J Acoustical. Society of America* **49**, 1971.
10. G. Tupholme, "Generation of acoustic pulses by baffled plane pistons.," *Mathematica* **16**, 1969.
11. M. Fink, W. A. Kupermann, J.-P. Motagner, and A. Tourin, *Imaging of Complex Media with Acoustic and Seismic Waves*, ch. Ultrasound imaging and its modeling. Springer Verlag, 2000.
12. H. Gemmeke, M. Kleifges, A. Kopmann, N. Kunka, A. Menshikov, and D. Tcherniakhovski, "First measurements with the auger fluorescence detector data acquisition system," in *Proceedings of the ICRC 2001*, Kopernikus Gesellschaft, (Hamburg), 2001.