Medical Imaging by Ultrasound–Computertomography

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ABSTRACT

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. Usually, 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. Exact measurement of tissue structures, like tumor size, is not possible.

Instead of a manually controlled linear transducer array, we use ultrasound computer tomography (USCT) to image a volume directly. A few 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. Afterwards a different transducer emits the next pulse. For volume reconstruction every transmitted, scattered and reflected signal is used. This new method allows reproducible image sequences with enhanced spatial and temporal resolution. For the benefit of more reconstructed 3D images per second, spatial resolution may be reduced offline. First tests with our prototype in a ring–shaped geometry have even showed nylon threads (0.4 mm) and an image quality superior to clinical ultrasound scanners.

Keywords: ultrasound, computed tomography, diffraction, circular array, breast imaging, transducer array, 3D

1. INTRODUCTION

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.[1,2] Malign breast tumors double their size between 80 — 300 days, reaching the size of 1 cm in 5 to 30 years. The probability of having spread metastases in the lymph nodes is about 30 % growing to over 60 % for tumors of 2 cm size.

X–ray mammography, magnetic resonance imaging and ultrasound are established methods of breast cancer diagnosis. Each method is sensible to specific tissue changes. In medical check–ups mainly X–ray mammography is used, but some tissue changes are hardly detectable. In many cases, ultrasound examinations are used to get additional diagnostic information about e.g. cysts and fibro adenoma. Furthermore, ultrasound does not harm biological tissue and may be applied frequently. Further advantages of ultrasound examinations are low costs and high speed.

The usual disadvantages of conventional ultrasound imaging methods are both poor spatial and temporal resolution. The contrast and the resolution depend highly on the used frequency as well as on the distance between the transducer array and the region of interest within the breast. The transducer array is manually operated and deforms the tissue to get as close as possible to the areas of interest. Therefore, the image contents and image quality are highly operator–dependent and almost impossible to reproduce. Exact measurements of tissue structures, e.g. tumor sizes, are hardly possible.

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Conventional ultrasound imaging systems[3] provide only two–dimensional images. In common B–scan–technique a sound pulse travels along a line into the medium. The pulse is partly reflected at several tissue borders and the reflections travel back to be detected by the same transducers from which the original pulse emerged (A–scan). For the registation of a two–dimensional image the transducer is moved, recording parallel A–scans. The time needed to register a complete image is the travel time of a pulse through the medium multiplied by the number of A–scans or number of parallel transducers used. Scanning a 10^3 cm^3 cube with a voxel size of 1 mm will take approximately 1.3 seconds. This technique is too slow for the observation of the dynamic spread of an ultrasound contrast agent. Furthermore, this method is limited in the resolution function by the wavelength of the used ultrasound. Therefore we need a different approach for fast high resolution 3D–imaging with ultrasound: ultrasound–computertomography.

2. ULTRASOUND–COMPUTERTOMOGRAPHY (USCT)

In USCT systems ultrasound reflection and/or transmission data are recorded from different angles. The imaging is based on backprojection methods to reconstruct images based on tissue properties like sound velocity, absorption and diffraction.

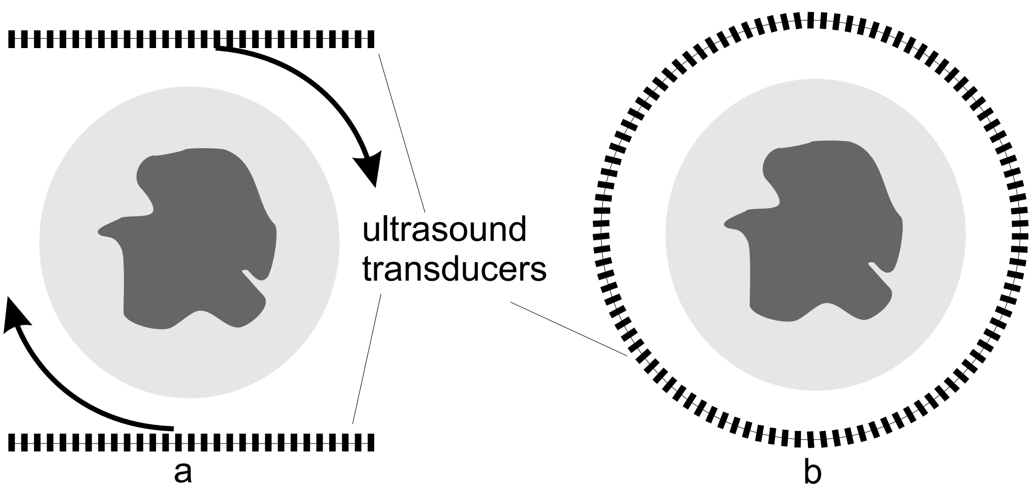

Figure 1. Architectures of ultrasound tomographic systems. a) Two linear opposite transducer arrays rotating mechanically around the object. b) Ring shaped array enclosing the object.

Principally, there are two different architectures to arrange the ultrasound transducers: Two opposite linear arrays rotating mechanically around the object (figure 1a) or a circular transducer array enclosing the object (figure 1b). Both methods record ultrasound signals of a slice of the object and can be easily extended to three dimensions by repeating data recording for different slices or using two–dimensional transducer arrays. In most cases, the architecture consisting of two opposite arrays (figure 1a) uses focused transducers, sending ultrasound pulses along a straight line directly to the opposite transducer and measures the directly transmitted and the backscattered signals. In the ring–shaped architecture, (figure 1b) all transmitted and scattered signals are recordable from all angles.

Ultrasound waves have the advantage of traveling slow enough, so that their actual travel time can be measured compared with X–rays. Therefore, there are more methods for calculating tomography images than for using X–rays, namely time–of–flight transmission tomography, reflection respectively diffraction tomography and conventional absorption transmission tomography.

In time–of–flight transmission tomography the travel time of a pulse directly from the transmitter to the receiver is measured. An image of the sound speed distribution is calculated. The algorithm is similar to the one used for absorption data, filtered backprojection, i.e. in X–ray tomography.[4]

In reflection–tomography, the scattered and reflected signals are exploited. In standard B–scan imaging only the backscattered information is used. In figure 1b the complete scattering information is measured by several transducers situated on a ring around the object. Every transducer records a signal which contains information about the whole imaging volume yielding a very high level of redundancy. This gives us the possibility to measure high–quality images in (theoretically) very short time. It allows a quantitative reconstruction of tissue properties, including density, absorption and sound speed at the same time.

Due to the limited penetration depth of ultrasound in biological tissue, most medical applications of USCT concentrate on breast and testicle imaging. In the last 25 years, several scientists have been working on ultrasound tomography.[5–19]

Norton and Linzer[6,7] have introduced a survey of the principles of ultrasonic reflectivity tomography, based on circular and spherical arrays. They have discussed analytical expressions of the point–spread function based on different transducer activations. Greenleaf[12,20] has discussed transmission and diffraction computed tomography. Schomberg[5] has built an experimental setup for ultrasound tomography of the female breast. Difficulties appeared in the processing time, data acquisition and the ultrasound properties (scattering and high absorption) of the female breast. Nguyen[11] has developed an ultrasound tomograph based on four transducer arrays for testicle and mamma screening. Ashfaq and Ermert[19] have introduced a spiral tomography system for the female breast, based on a standard medical transducer array and an ultrasound mirror mounted on the opposite side of the breast. Using this device they are capable of recording conventional ultrasound B–scan and time–of–flight data according to Richter.[21] The main advantages are the simple design and use of standard transducer arrays with no need for modification of the internal data reduction. Due to the mechanical restrictions, the recording of a three–dimensional image took approx. 20 minutes. Movement of the imaged tissue must be avoided, thus the observation of fast dynamical effects, e.g. the propagation of contrast agents, is not possible.

Ultrasound computed tomography methods have not been included into commercially medical ultrasound systems yet. The reasons are:

- The structure of USCT systems is very complex. As more transducers than in conventional ultrasound imaging are needed, costs are high.
- At the moment, the large arrays of small transducers with similar properties, which are needed, are not available.
- Conventional ultrasound arrays have a raw data rate of 1 – 3 GBytes per second which can be reduced by standard data reduction algorithms. In USCT the data rate can extend 100 GBytes per second. Data reduction algorithms are hardly applicable because more information is needed than in standard B–scan technology and the arrays are much larger. Thus, for every transducer a high–speed data acquisition and data processing unit has to be developed. With the progress in digital electronics and programmable gate arrays during the last years, such processing units can now be build.
- The medical applications are restricted to small (< 20 cm in diameter) parts of the body which may be enclosed by a ring (cylinder) of transducers or a mechanical device. In contrast to conventional ultrasound, USCT systems are not portable.

3. USCT AT FORSCHUNGSZENTUM KARLSRUHE

3.1. Architecture

At Forschungszentrum Karlsruhe, we are developing an ultrasound computer–tomography system for breast imaging providing high quality three–dimensional images with a high repetition rate. A ring (two–dimensional case) or a cylinder (three–dimensional) of ultrasound transducers encloses the object. One transducer emits an ultrasound pulse with a mean frequency of 2.5 MHz which is scattered by the structures inside the object. Every single transducer is small emitting a nearly undirected beam (tree–dimensional fan beam). All other transducers measure the transmitted, reflected and scattered signals simultaneously. The received signals are amplified, digitized and stored. In the next stage, another transducer will transmit an ultrasound pulse while all the others receive the scattered signals and so on.

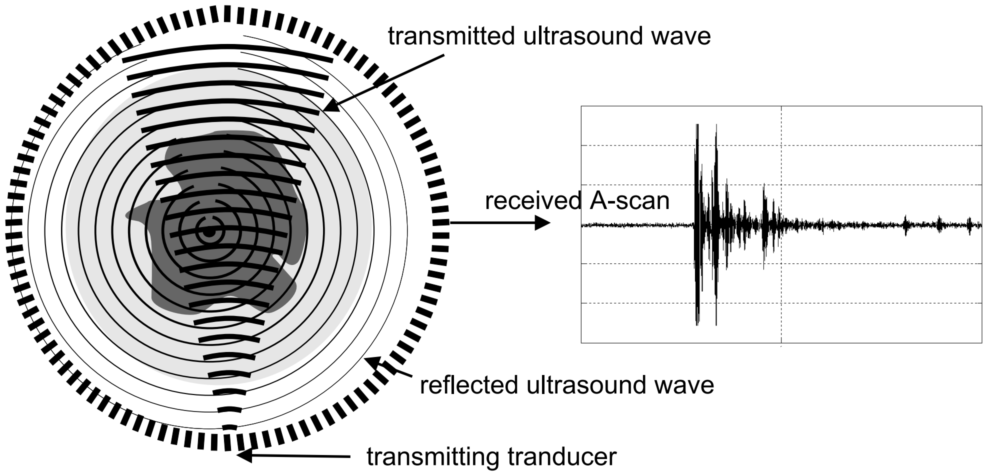

Figure 2. Architecture of the USCT system build in Forschungszentrum Karlsruhe, shown in 2D. A ring (cylinder) of ultrasound transducers encloses the object. One transducer transmits an ultrasound pulse, all other transducers receive simultaneously.

The ring (cylinder) has a diameter of 12 cm and due to absorption it is reasonable to assume that no measured signal has traveled more than 24 cm. With an average sound speed of 1500 $\frac{m}{s}$ one "shot" takes 160 μs. Theoretically, 6250 shots per second can be emitted from different transducers gathering information from the same region from many different viewpoints. The recorded ultrasound data contain very redundant image information, reducing noise and artifacts.

3.2. Data Processing

In this section, the data processing and the image reconstruction algorithms are described for the two–dimensional problem in which the ultrasound transducers are arranged on a ring.

First, several assumptions are made:

- The electronics of the whole system, as well as the medium of sound propagation, can be handled as linear systems.
- The scattering amplitudes inside the medium are weak and can be described with the Born approximation, i.e. only first order scattering is considered.
- All scattering sources can be modeled as Huygens point sources.

On the assumption that the sound speed is constant within the imaged region, the receivers integrate the echoes arising over an elliptical path whose foci coincide with the transmitter and receiver positions. Thus image reconstruction for scattering and reflection can be described as a simple filtered delay and sum operation.

The data acquisition and image reconstruction algorithms can be divided into several steps:

Data acquisition: Short ultrasound pulses with a mean frequency of 2.5 – 3.5 MHz are emitted from different transducer positions (angles) into the tank.

1. All other transducers receive the signals simultaneously. Each transducer produces a time–dependent voltage, an A–scan (figure 2).

2. The signals are amplified, optionally the effect of absorption within the object can be reduced using time–gain compensation.
3. The signals are digitized and stored.

Calibration: Since sound speed in water and biological tissue depends highly on the temperature, the temperature should be determined frequently, at least once at the beginning of an experiment. The sound speed $c(T)$ in water is a function dependent of the temperature T and follows the relationship[22]

$$c(T) = (1557 - 0.0245(74 - T)^2)\ \frac{m}{s} \tag{1}$$

On the other hand, the transducer properties do not stay constant (aging) and vary very largely from transducer element to transducer element. Thus, a calibration step should determine the shape of the transmitted beams to determine the angle dependent sensitivity and the transfer functions of all transmitter and receiver combinations. This can be done by a calibration step with a tank containing only the coupling medium (water) and recording all signals.

After this the tank can be loaded with the object to examine and the data can be recorded.

Data pre–processing: The sensitivity of real transducers depends on the angle. In the calibration step, the angle dependency of every transducer has been determined. With this information, the amplitudes of the directly transmitted signals can be corrected.

Since in our prototype system the temperature during data acquisition was not stable, we had to determine the sound speed within the water and to correct the signals.

The part of the signal which is measured before the arrival of the direct pulse does not contain any imaging information,but this part of the signal can be used to estimate the amplifier noise or to correct its voltage offset.

After these steps, we use the resulting signals (A–scans) $p(t, \phi, \theta)$ where t denotes the time, ϕ the position (angle) of the receiver on the ring and θ the position of the emitter. In the simple reconstruction method, we assume that the sound speed c of the water does not differ from the object. In this case, t can be also interpreted as a distance.

Analytical signal: The signal $A(t, \phi, \theta)$ is the analytical continuation of the original signal $p(t, \phi, \theta)$ into a complex one, having the original signal as the real part:

$$A(t, \phi, \theta) = p(t, \phi, \theta) - iH(p(t, \phi, \theta)) \tag{2}$$

where $H(p(t, \phi, \theta))$ denotes the Hilbert transform of $p(t, \phi, \theta)$. The analytical signal is a very convenient expression with $H(p(t, \phi, \theta))$ containing the phase information and $|A(t, \phi, \theta)|$ expressing the envelope function of the amplitude.[23–25]

Filtering: Some shaping filters can be applied easily in the Fourier domain. Since the Fourier transform is used to calculate the Hilbert transform, the implementation of such filters will not require much costs.

Sound speed distribution: To achieve high quality images, the travel time of the sound wave has to be known as good as possible. For this purpose, time–of–flight transmission tomography can be used to calculate an image of the sound speed distribution within the object. For this purpose, no extra measurements are necessary, since the travel time from emitter to receiver can be deduced in the A–scans from the minimum propagation time.

Grid: To reconstruct an image, we use a two–dimensional grid of 512×512 pixel with a pixel size of approximately 0.23 mm.

Calculate reflectivity values: Assuming that the sound speed c is constant in the water and the object, the reflectivity value of a pixel $f(x, y)$ can be determined by accumulating for all possible emitter–receiver combinations (see figure 6):

$$f(x,y) = |\sum A(\frac{a_\theta + b_\phi}{c}, \phi, \theta)| \tag{3}$$

where a_θ denotes the distance between the pixel at position (x, y) and the emitter at angle θ and b_ϕ the distance to the receiver at angle ϕ.

Modifying equation 3 and accumulating the absolute values of the analytical signals $|A(\frac{a_\theta+b_\phi}{c}, \phi, \theta)|$ will apply only the amplitude information, otherwise the phase information is included. The introduction of phase information into reconstruction requires a higher sampling rate of the analog signal, approximately ten times of the mean ultrasound frequency.

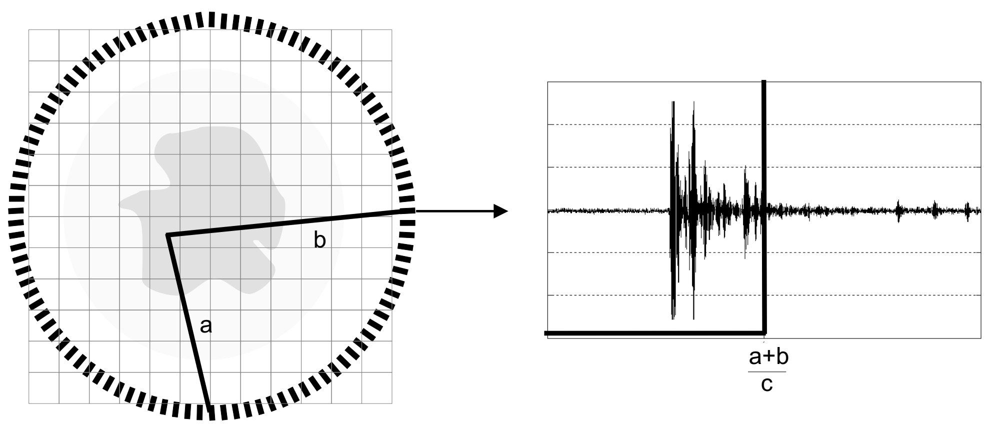

Figure 3: Reconstruction principle on the assumption that the sound speed c is constant in the water and the object.

The weakness of this algorithm lies in the (false) assumption that the sound speed is constant in the whole object. The correct sound speed depends on the material at the local position (x, y). The local sound speed can be determined by conventional time–of–flight tomography using only the information of the arrival time of the first signal in the A–scan. With the sound speed distribution within the image, the relationship of the distances a_θ and b_ϕ to the time can be determined correctly.

The angle–dependent sensitivity of the transducers can be corrected by amplifying the addends, according to their angles to transmitter and receiver.

3.3. Prototype System

We have built an experimental setup[17] to evaluate ultrasound computer tomography for medical imaging. It consists of two transducer arrays 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 a ring to emulate a full circular array successively (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 emitter, the other as 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.

The chosen ultrasound frequency of 2.5 MHz is 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 are strongly damped and produce broadband pulses. The signals of 200 μs length are digitized with a sampling rate of 50 MHz and a signal quantization of 10 bits.

Figure 4. 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. The signals are digitized, using a digital oscilloscope.

We have used only the simplest reconstruction method, diffraction tomography as described in equation 3, on the assumption of constant speed of sound in the water and the object. Furthermore, no corrections of the angle–dependent sensitivity of the transducers have been applied.

4. RESULTS

Since the measuring time for the prototype ranges from several hours to weeks, we needed fixed phantoms mimicking biological tissue. To create our phantoms we used a matrix of a galantine and olive–oil emulsion[26] with straws and threads as test structures which had to be imaged. The mixture has about the same ultrasound absorption as breast tissue.

The whole imaging system is two–dimensional in a cylindrical slice with a diameter of 12 cm and height of about 1 cm. For the phantom, we first built a model sketched in figure 5 which we submerged in the gelatine–oil–emulsion inside a plastic cup with very thin walls. The smallest structures within the phantom are four nylon threads with a diameter of 0.45 mm each, corresponding to approximately 0.7 wavelengths of the transmitted ultrasound signal. A commercial high–quality ultrasound scanner (Acuson Sequoia 512) produced the first images of the phantom (figure 5). Using 3 MHz and harmonic imaging, the shape of the phantom and the front and back reflections of the straws are clearly visible. The rectangular pillars on the right and on the left shade most of all structures from behind. The thin nylon threads are not visible at all. The whole image is disturbed with noise.

In two weeks we have recorded ultrasound signals of the phantom described in figure phantom. For 25 different transmitter positions we took all receiver positions gathering approximately 72,000 signals. In future, after having built a complete ring of transducers and the parallel data acquisition electronics, this task will take us only 0.02 seconds.

The image reconstruction results are shown in figure 6. The round shaped objects and even the nylon threads are clearly visible. Although the two pillars on the bottom have a rectangular shape and plain surfaces, they can be detected too.

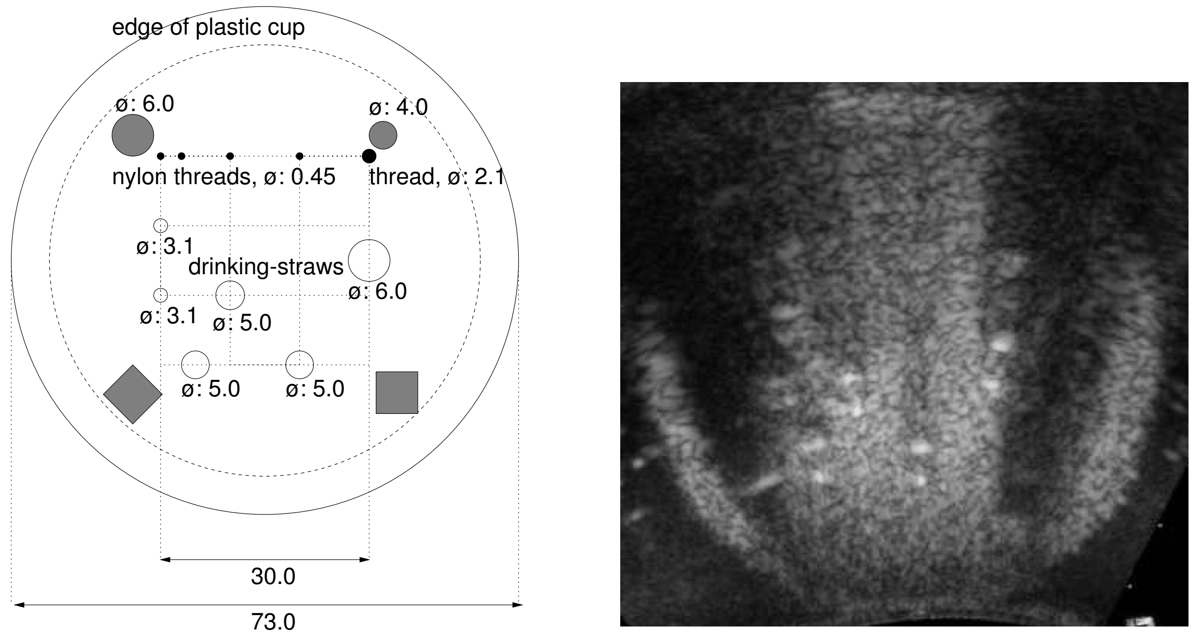

Figure 5. Ultrasound phantom. Left: plan of the phantom, units are mm. Right: conventional ultrasound image, Acuson Sequoia 512, 3 MHz.

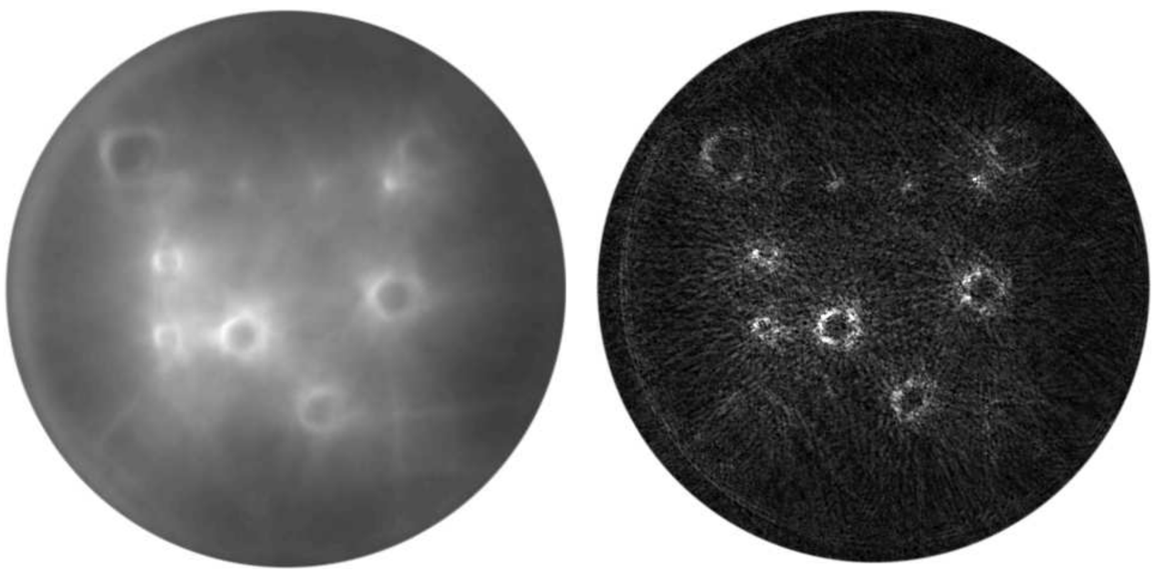

Figure 6. Ultrasound computed tomography of the phantom shown in figure 5. Left: reconstruction using only amplitude information. Right: with phase information.

Compared with the conventional ultrasound image in figure 5, the computed reconstructions show a better contrast and the lack of noise. If we take into account that we used only 25 different transmitter positions although theoretically 6250 per second are possible, this imaging method is very promising.

5. DISCUSSION AND FUTURE WORK

In less than one and a half years, we have succeeded in constructing a fully operable USCT prototype as shown in fig. 4. Due to the impressive data rates, we settled for a compromise in construction: first, instead of a completely filled circular array of ultrasound transducers, we used two linear arrays with only 16 elements. We simulated the circular array by rotating emitter and receivers step by step, reducing hardware requirements and data rate in trade–off for a longer data acquisition process. The acquired data is nevertheless identical to the one received by a parallel imaging process. Secondly we decided to image only a slice of the examined object for further data reduction. Just like in common computer tomography a three–dimensional object can be reconstructed by several two–dimensional slices. Therefore, the results can be applied to a cylindrical array too. Despite our efforts in data reduction it took 14 days to acquire 25 different emitter positions resulting in a final size of about one Gigabytes for the imaged object slice.

The reconstruction algorithms are implemented and the correctness is tested successfully. The imaging method is of superior quality in comparison to common ultrasound imaging systems. More detailed structures are visible and there are no shadows hiding the view of remote tissue features. The resolution and contrast are comparable to or even better than produced by state–of–the–art ultrasound imaging systems. Up to now, a major drawback of our system has been the high computing time for images. It takes about two hours on a standard–PC to reconstruct a full resolution image, using all received signals. The reconstruction process may be accelerated significantly in exchange for lower image quality if less received signals are used, thus allowing quick image surveying during the examination process.

The main problem of ultrasound tomography is the vast data rate. A realistic goal is to receive about 2000 pulses per second sequentially with all receivers. A circular array with 1440 sensors, signal quantization of 12 bit and a sampling rate of 50 MHz will lead to an estimated data rate of about 140 Gigabytes per second. The data has to be transferred online into a computing system for reconstruction. This is absolutely impossible for multiplexing systems and common hardware.

In order to be able to handle the huge data flow, massive parallel data processing electronics are necessary. Secondly, an online signal compression to reduce the incoming data to the essential information for image reconstruction is needed. At our institute, we are currently developing a massive parallel system in combination with the data reduction algorithms.

The image reconstruction algorithms will be improved first by determining the correct sound velocity distribution using time–of–flight transmission tomography. Using this information the accurate travel time of ultrasound within the tissue can be determined. Next, the transfer functions of the emitter and receiver pairs will be determined to design (optimal) filters for backprojection.

With the new system subtle tissue changes caused by early cancers may be visible. This will improve the detection of small cancers before having spread metastases in the lymph nodes.

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