STATE OF THE ART AND CHALLENGES IN ULTRASOUND COMPUTER TOMOGRAPHY

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Abstract: Ultrasound computer tomography is a new imaging system for early breast cancer diagnosis capable of producing images with high spatial resolution and image quality. The large number of transducers in a fixed setup produces high data rates of up to 4 GByte/s. Also the reconstruction of images based on the complex interaction of ultrasound and breast tissue acquires high computational complexity. In this paper the state of the art of USCT at Forschungszentrum Karlsruhe and the future optimization of the described method are presented.

Introduction

Early breast cancer diagnosis is still a mayor challenge. The standard screening methods often detect cancer in a state when metastases have already developed [1]. The presence of metastases decreases the probability of survival significantly. A more sensitive tool for breast cancer diagnosis could lead to diagnoses in an earlier state, i.e. before metastases are generated.

The standard method for breast cancer diagnosis is x-ray mammography assisted by sonography and sometimes Magnetic Resonance Imaging (MRI) for further investigation. X-ray mammography produces images with high spatial resolution, but has the drawback of being a 2D projection. Thus the lesions appear embedded in a highly heterogeneous background, which makes the detection of small lesions difficult. Additionally, x-ray mammography is not feasible for younger women with predominant glandular tissue. Conventional ultrasound imaging produces low resolution images, which are hard to read due to the high noise level. MRI is expensive and has low sensitivity [2].

We are developing a new imaging method for breast cancer diagnosis, ultrasound computer tomography (USCT), which allows recording of reproducible 3D images with high spatial resolution and tissue contrast. Additionally, quantitative measurements of physical parameters, i.e. sound speed and attenuation, which are known to be different in cancer tissues, are possible. In this paper a short introduction into the principles of USCT is given, the setup and the results of a 2D and a 3D demonstration system are presented. Based on these results the necessary steps for a real time 3D-USCT system are discussed.

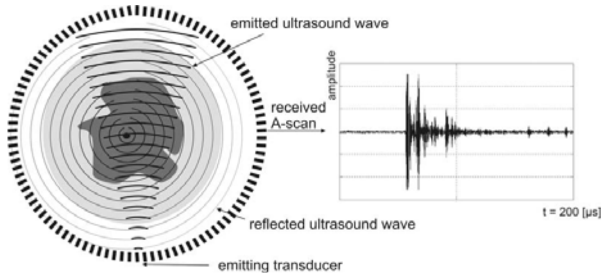

Figure 1: The principle of imaging with USCT. Left: One transducer emits a spherical pulse; all the other transducers receive the transmitted, scattered, and reflected ultrasonic waves. Right: Received ultrasound signal (A-scan) at one exemplary receiver.

Methodology

Conventional sonography scanners consist usually of a rectangular array of several hundred ultrasound transducers. The transmitters are coupled to emit a focused beam and record only the backreflected signals. The resulting layer images are composed of the columns of the envelope functions of these signals. Sub resolution scatterers cause interferences in the ultrasound field and lead to so-called speckle noise, which is visible as bright granular structures causing a low signal-to-noise-ratio in the images. Ultrasound computer tomography, i.e. arranging ultrasound transducers around the imaged object in a fixed setup, is not a new idea. The first publications in this field date back to the early 80s [3]. Building such a device for clinical practice never quite succeeded; mostly due to the huge data rate and time consuming image reconstruction, but different groups achieved impressing results, in transmission tomography as well as in reflection tomography, e.g. [4]. The here applied principle of imaging with USCT is given in Figure 1. The object is completely surrounded by ultrasound transducers. One by one each transducer emits an unfocused spherical pulse, until the object is illuminated with ultrasound from all directions. During each shot, all other transducers receive the transmitted, scattered and reflected signals. Thus the recorded signals yield a high level of redundancy. In comparison with images of conventional ultrasound the quality of the reflection images with USCT is dramatically enhanced, the reflection images show sub wavelength resolution and the speckle noise is reduced to the level of the background noise of the image.

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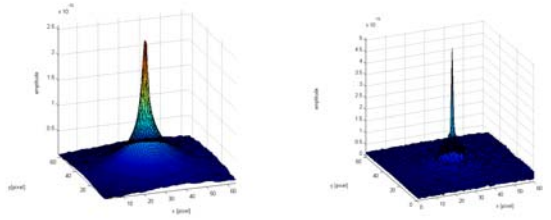

(a) (b)

Figure 2: Simulated psf in the central layer of the 3D USCT demonstrator. (a) Point spread function (psf) as result of reconstruction with envelop of A-scan. (b) Psf as result of reconstruction with the original A-scan.

The recorded signals (A-scans) enable to reconstruct images of structural information using the scatter and reflection information, as well as reconstruction of physical parameters, i.e. sound speed and attenuation [5]. These physical parameters for breast cancer tissue are known to differ from normal tissue [3], so that these images will be valuable additional diagnostic information.

The here applied reconstruction algorithm for reflection images is a so-called sum-and-delay algorithm [6]. For each point in the image to be reconstructed the amplitude or some preprocessed version of all acquired A-scans at the position corresponding to the distance between sender, point and receiver is accumulated:

$$f(\vec{x}) = \sum_{(j,k)} T(A_{(j,k)}(\frac{a_j + b_k}{c})),$$

where f denotes the reflection image, $\vec{x}$ the coordinates of the reconstructed point, T are preprocessing steps, $A_{(j,k)}$ the A-scan acquired at sending position $\vec{x}_j$ and receiving position $\vec{x}_k$. c is the speed of sound in water, and a_j and b_k are the distances from the reconstructed point to the sender and receiver, respectively. To use this reconstruction method certain simplifying assumptions have to be made: All interactions between ultrasound and tissue are scattering, which can be modeled as Huygens point sources, the attenuation can be neglected and the sound speed is constant. Additionally, the influence of the system's electronics is ignored and an ideal spherical pulse is assumed.

Preprocessing steps T can be e.g. band pass filtering, envelop calculation or deconvolution. These operations greatly influence the point spread function (psf) of the imaging system as shown in Figure 2.

Experimental 2D Setup for USCT

The experimental 2D setup for USCT (Figure 3) consists of two transducer arrays in a water tank, a pulse generator, an amplifier and a digital oscilloscope connected to a computer.

Figure 3: Experimental 2D setup for USCT. The white tank is filled with water and contains the object to be imaged. Two unfocused ultrasound transducer arrays are manually moved on a ring with 12 cm diameter to simulate 100 sending and 1600 receiving positions in one circular layer around the object.

The used ultrasound pulse has a center frequency of 2.5 MHz in order to penetrate breast tissue of 10 cm in diameter.

In Figure 4 an exemplary result of data acquisition with this setup is given. A circular plastic container (approximately 8 cm in diameter) filled with water holds several plastic objects. Additionally, an L shaped row of nylon threads, each 0.1 mm in diameter and 0.5 mm apart, was inserted. Using the sum-and-delay reconstruction with spiking deconvolution for preprocessing, it was possible to reconstruct the nylon threads, giving a sub wavelength resolution [7].

The drawbacks of this system are the long data acquisition time of approximately 12 hours due to the sequential data acquisition and the limitation of the system to 2D slice images.

Experimental 3D Setup for USCT

The experimental 3D setup for USCT (Figure 5) consists of a cylindrical container with 18 cm in diameter and 14 cm in height, filled with water as coupling medium. The sensors are connected to a special dedicated data acquisitions system for high data rates. All needed hardware devices, including the ultrasound transducer arrays (TAS), are in-house products. In order to achieve a system with low costs and highly reproducible transducer characteristics, the manufacturing process of the TAS is almost automatic [8].

In the full equipped setup 48 transducer array systems are mounted in 3 stacked layers at the container walls. Each TAS consists of 8 senders and 32 receivers. The cylinder can be mechanically rotated in six steps, emulating the complete coverage of the container with 2304 virtual senders and 9216 virtual receivers.

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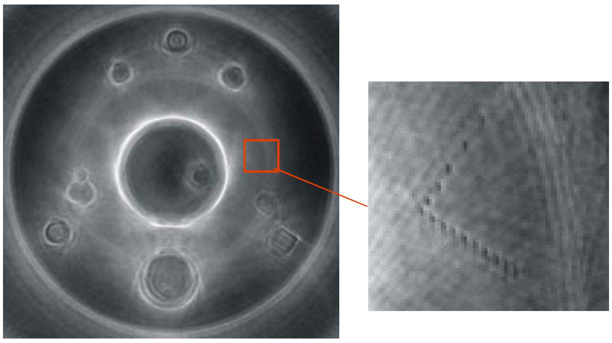

(a) (b)

Figure 4: Results of the experimental 2D setup. (a) A circular plastic container with different structures was filled with water. (b) Nylon threads with a diameter of 0.1 mm and a distance of 0.5 mm are clearly visible.

Included in the TAS are pre-amplifiers, a multiplexer and a microcontroller for addressing. Coded excitation can be used to shape the emitted pulse with arbitrary waveforms. The transducers are designed to send an unfocussed beam with a center frequency of 2.7 MHz and a bandwidth of approximately ±1 MHz. The quadratic aperture of $(1.4\ \text{mm})^2$ is a compromise between emitted signal amplitude and desired spherical emission characteristics.

Due to the large number of transducers, the data rates of the system are high. The 3D demonstration system is recording 192 channels in parallel, resulting in a data rate of 4 GByte/s. When reducing the data acquisition time and thus the artefacts due to patient movements, data rates of up to 20 GByte/s will have to be recorded. The USCT crate contains a controller board and 8 digitization boards. The controller board controls the measuring process of the digitization boards and the TAS. Each digitization board supports 24 digitization channels sampling at 12 bit and 10 MHz. The boards are connected via a VME-Bus backplane.

Until now one layer of TAS is installed in the 3D setup, recordings with 128 senders and 512 receivers are available. With 6 different motor positions 768 sending and 3072 receiving positions can be achieved.

To test the performance of the actual 3D setup different measurements where made. First for calibration an empty measurement was performed, where the USCT is only filled with water. This measurement was applied to evaluate the reproducibility of the ultrasound transducers. The recorded A-scans were used to calculate the transducer properties, i.e. the center frequency, the bandwidth and the maximum amplitude are compared for individual senders. The average of the center frequency was found at 2.7 MHz with standard deviation of 0.2 MHz (6%). The bandwidth was in average 1.6 MHz. The average of the maximum pulse amplitudes at 0° amounted to 93.9% of the total maximum amplitude and had a standard deviation of 7.4%.

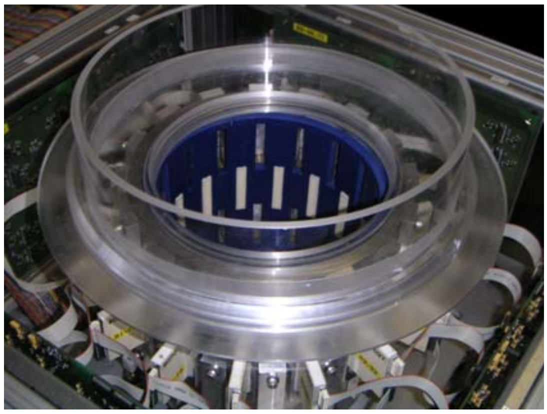

Figure 5: Experimental 3D setup for USCT. The metallic tank is filled with water and contains the object to be imaged. One TAS layer is already introduced. Until now 128 senders and 512 receivers are available. With 6 different motor positions 768 sending and 3072 receiving positions can be achieved. In the full equipped 2304 virtual senders and 9216 virtual receivers are available.

The transducers had a mean aperture angle at approximately ±30°. The first and second side lobes are located at approximately ±40° and ±65°, respectively. Due to these results, the transducers of the built TAS satisfy the requirements and are satisfactorily reproducible.

. The wire is clearly visible and had a diameter of 1.8 mm at the full-width-half-maximum. Until now no dedicated calibration and different motor positions are applied.

Further Work

So far impressive results in image quality and resolution could be achieved with our 2D approach to ultrasound computer tomography. The experimental 3D setup is running in a minimum configuration. It could be shown that it is working and promises similar results as the 2D setup. The challenges in building dedicated transducer array systems, being cheap and reproducible, were mastered.

The actual data aqcquisition system was not build for the accumulation of 4 GByte/s of data without data reduction. The bottleneck is the available memory and the speed of the on-line data readout. This results in an insufficient readout time for living and moving objects. We tackle this problem by the next generation of data acquisition system under construction, which will contain additional 4 GByte of RAM per digitization board, to buffer the accumulated data without deadtime. For the expected data rates of a full equipped system without multiplexing we are developing a real time compression system for online data reduction.

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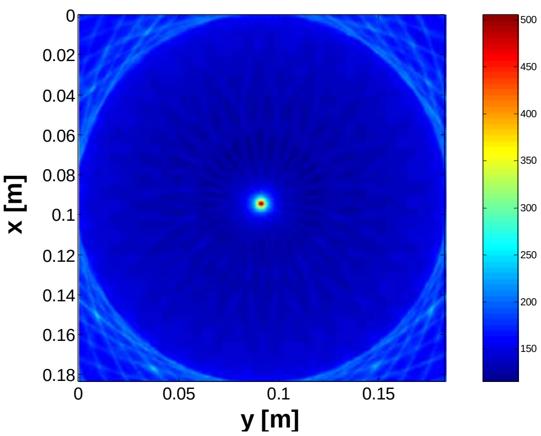

Figure 6: First results with the experimental 3D setup for USCT. A thin copper wire of 1.6 mm in diameter was placed in the middle of the USCT. Here a slice image of the 3D volume is displayed. The data was recorded at only one motor position and no dedicated calibration was carried out.

The reconstruction time for a high resolution volume has to be reduced considerably. Fortunately, the reconstruction algorithm can be parallelized easily. We are working on a reconstruction scheme for Grid computing as well as on parallel reconstruction in hardware using the processing FPGA hardware available on the digitization board.

The applied reconstruction algorithm is based on a number of oversimplifying assumptions which cause artifacts in the images, e.g. the assumption of constant sound speed is causing structures to be doubled [9]. Adapting the reconstruction to more realistic preconditions will lead to even better image quality and resolution as already achieved. Using a 3D simulation tool for ultrasound signals [10], it could be shown that given ideal circumstances it is possible to reconstruct point scatterers at 0.05 mm in diameter which corresponds to one tenth of the wavelength (

Figure 7).

For further optimization the sensor configuration has to be optimized. To enable faster image acquisition and thus the minimization of artefacts due to patient movements, the number of channels has to be increased and the USCT has to be covered completely with transducers. Therefore the transducer array systems have to be miniaturized further to allow mounting them directly next to each other. Fast acquisition of a volume will open additional applications for USCT, e.g. real time imaging of contrast agent distribution in the breast.

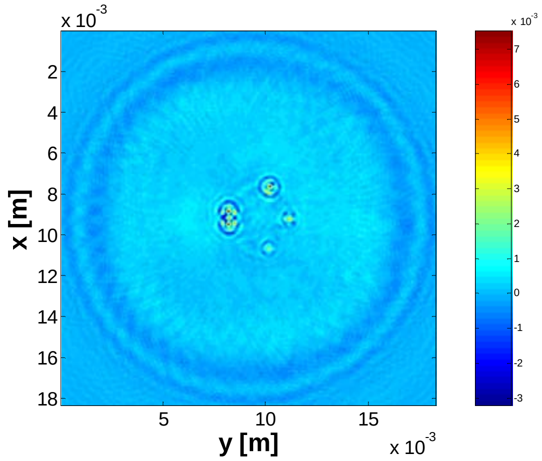

Figure 7: Reconstruction of simulated A-scans in the configuration of the experimental 3D setup. The scattering of ultrasound on nylon threads of 0.2, 0.1 and 0.05 mm diameter in water were simulated. For an ideal calibration and no noise it was possible to reconstruct point scatterers of 0.05 mm, corresponding to one tenth of the wavelength of a 3 MHz pulse in water.

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