3D ultrasound computer tomography for medical imaging

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Abstract

Our 3D ultrasound computer tomograph (USCT) is an device method aimed at early breast cancer diagnosis. It is capable of producing images with sub-millimeter resolution and high signal-to-noise ratio. This method is universally useable for the 3D analysis of sufficient small bodies similar to the size of a breast, which may be immersed in a liquid coupling medium. In this paper, an overview of the developed method and the first results for static test examples (phantoms) and the perspective of our 3D-USCT at Forschungszentrum Karlsruhe are presented.

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1. Introduction

Early breast cancer diagnosis is still a major 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 images of anisotropic resolution, which are hard to read due to the high noise level. MRI is expensive and has low specificity [2]. In conclusion, a low-cost 3D ultrasound imaging with approximately the resolution of X-ray mammography would be the instrument of first choice, because it avoids radiation doses and breast compression.

At the Forschungszentrum Karlsruhe, in the Institute for Data Processing and Electronics, we are developing a new low-cost ultrasound computer tomograph (USCT) in order to support reproducible 3D imaging with sub-wavelength resolution. Additionally, from the same data the local distribution of speed of sound and attenuation can be derived, which are known to be different in cancer tissues. In this paper, a short introduction into the principles of USCT is given, the configurations and the results of a 2D and a 3D demonstration system are presented. Based on these results, the necessary steps for an in vivo measurement system are given.

2. Ultrasound computer tomography and its problems

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 1980s [3]. Building such a device for clinical practice never quite succeeded; mostly due to the huge data rate and time-consuming image reconstruction. Several scientists have been working on ultrasound tomography [4–14]. Norton and Linzer [4,5] have introduced a survey of the principles of ultrasonic reflectivity...
Computed tomography, based on circular and spherical arrays. They have discussed analytical expressions of the point-spread function based on different transducer activations. Greenleaf et al. [6,7] have discussed transmission and diffraction computed tomography. Schomberg [8] 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 [9] has developed an ultrasound tomograph based on four transducer arrays for testicle and mamma screening. Ashfaq and Ermert [10,12] 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 [14]. 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 3D image took approximately 20 min. For high-quality images, movement of the imaged tissue must be avoided. Also, the observation of fast dynamical effects, e.g. the tracing of contrast agents, is not possible.

The manifold challenges to build a system for clinical use can be summarized as follows:

1. A large number of ultrasound sensors are needed to obtain the necessary resolution. These sensors have to be low cost to restrict the overall price of the system. Additionally, they have to be very similar in their characteristics (resonance frequency and bandwidth) to ensure homogenous insonification and recording of the object. To submit a spherical wave, the emitters have to be small, ideally smaller than the submitted wavelength. Thus, the amplitude of the emitted pressure is low, resulting in a low signal-to-noise ratio.

2. For sub-millimeter resolution in 3D, the data acquisition time has to be as short as possible to correct for patient movement. Though post-beam-forming enables fast recording.

3. The image reconstruction, i.e. the post-beam forming, has a high computational complexity, i.e. the number of A-scans (pressure over time at one receiver) to be recorded is given by the number of receivers times the number of senders and each of these A-scans has to be processed for each 3D image point (voxel).

4. The dynamic range of the received signals is large. The transmission signals may be attenuated by the breast or only go through water; the attenuation of the pressure signal in the breast may be as large as 58 dB for 20 cm at 3 MHz. On the other hand, the changes in impedance between normal and cancerous tissue are only small leading to a reflected wave of the order of approximately 0.03%.

5. To reach sub-millimeter resolution and a high signal-to-noise and signal-to-artifact ratio, coherent imaging has to be applied. For adding the pulses in phase to archive additive interference at the imaged point, the accuracy of calculating the position of a reflection has to be better than $\lambda/2$ (for 3 MHz and 1500 m/s in water $<0.25\text{mm}$). Therefore, a highly accurate correction for speed of sound variation and refraction effects is necessary, additionally to a precise spatial calibration of the sensor positions.

3. Methods

Conventional 2D sonography scanners consist of a rectangular or curved phased array of up to several hundred ultrasound transducers (see Fig. 1a) and the back reflected signal is recorded. Many small emitters are combined to create one scan line focused to one point; an acoustic lens usually fixes the second focus dimension, the elevation plane. During reception, beam forming is carried out. As result of the single focus during emission, the image lines show only the highest resolution in the vicinity of this focal point. The resulting 2D images (B-scans) are composed of the columns of the envelope functions of these signals. Sub-wavelength scatterers cause interferences in the ultrasound field and lead to so-called speckle noise, which is visible as bright granular structure causing a low signal-to-noise ratio in the images.

Conventional 3D sonography usually combines conventional B-scans acquired from different angles. A spatial locator records the actual position and enables the relative summation of the B-scans. The imaged tissue adopts different states of deformation due to the contact pressure of the scanner during each scan. Thus, elastic registration is necessary in overlapping B-scan sections to correct the differences in the images [26]. Alternative methods are dedicated 3D scanners or extensive beam steering, but these methods scan only small volumes from a limited angle domain. The advantages of conventional 3D ultrasound are production of 3D volumes, speckle reduction and decreased shadowing. The disadvantages are long scanning times, blurring of the resulting images due to registration inaccuracies and, analogous to conventional 2D sonography, only sparse focal points. USCT can overcome all this disadvantages due to its fixed setup, no deformation and post-beam forming.

Our approach to USCT (see Fig. 1b) combines two basic concepts: synthetic aperture for post-beam forming and ultrasound emission and reception from many different directions.

Therefore, shadowing and speckle noise are strongly reduced leading to a superior image quality. Our cylindrical prototype is built to analyze the freely suspended breast, without compression, at prone position of the patient simplifying the matching to other methods for diagnosis like MRI and mammography.

Synthetic aperture bases on the idea to combine many low-resolution images to one high-resolution image.
Ultrasound is sequentially emitted from small elements as unfocused spherical wave. The received signals are combined by post-beam forming so that the image information is focused at any image point and the overall image resolution is increased. Mathematically, beam forming with phased arrays and post-beam forming are equivalent \[23\].

Ultrasound emission from many directions increases the signal-to-noise ratio, decreases the speckle noise, and levels the usually anisotropic point-spread function of ultrasound. Additionally, the shadowing effects of attenuating objects are reduced, so that the outlines of the objects become visible in 3D. When used with a manually operated ultrasound transducer or with a positioning device, the resolution is limited by the precision of the transducer position and the deformation of the breast, which is omitted by the fixed setup used in USCT.

Surrounding the breast with a fixed setup of ultrasound transducers has further advantages: with the same image acquisition step data for reflection and transmission tomography is recorded, enabling the reconstruction and perfect registration of three different tissue properties: reflection, speed of sound and attenuation \[15\]. 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. Additionally, imaging of an undeformed breast in 3D allows the registration of the images with other modalities, e.g. MRI, and older images of the same modality for follow up comparison.

The applied principle of imaging with USCT is given in Fig. 2. 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.

The here applied reconstruction algorithm for reflection images is the so-called sum-and-delay algorithm \[16\]. For each point in the image to be reconstructed, the amplitude or some preprocessed value of the acquired A-scans at the position corresponding to the distance between sender, point and receiver is accumulated:

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

where \(f\) denotes the reflection image, \(\vec{x}\) the coordinates of the reconstructed point, \(T\) the preprocessed A-scan \(A_{(j,k)}\) 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. This approximation is valid for constant speed of sound, small attenuation, weak point scatterers and spherical emittance. Preprocessing steps may be, e.g. band pass filtering, envelope calculation, or deconvolution.

To apply coherent imaging, the accuracy of the calculated position of a reflection has to be better than \(\lambda/2\). At the moment this time–space relationship cannot be always obtained with this precision due to the errors introduced by the assumption of constant speed of sound over distances up to 20 cm, inaccuracies in the position of sender and receiver, and the long measuring time yielding calibration errors due to temperature fluctuations. Additional possible sources of error are refraction and changes...
in the pulse shape due to frequency depending attenuation. Therefore, the images calculated at the moment are based on incoherent imaging using the envelope of the A-scans. This results in a broader point-spread function and a lower signal-to-noise ratio due to the non-zero mean of the noise envelope.

4. 2D-USCT

The experimental 2D setup for USCT (Fig. 3) consists of two transducer arrays in a water tank, a pulse generator, an amplifier and a digital oscilloscope connected to a computer. The used ultrasound pulse had a centre frequency of 2.5 MHz in order to show that at frequencies appropriate to penetrate breast tissue of 10 cm in diameter our method could reconstruct structures of 0.1 mm diameter (a fifth of the wavelength).

In Fig. 4, an exemplary result obtained 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 of the A-scans, it was possible to reconstruct the nylon threads [17]. In comparison to conventional images with ultrasound the quality of the reflection images with USCT is dramatically enhanced, the images show sub-wavelength resolution and the speckle noise is reduced to the level of background noise in the image.

4.1. 2D reflection images

In the simple phantom used above all objects are surrounded by water, thus the assumptions of constant speed of sound and low attenuation is approximately fulfilled. But for realistic phantoms or the breast itself the speed of sound may vary by 5–10%, limiting the precision of reconstruction. Therefore, we analyzed a simple ultrasound phantom made from gelatin (\(c \approx 1530\) m/s) immersed in water (\(c \approx 1490\) m/s) at room temperature (see Fig. 5). It consists of a plastic box with a lid, which is divided into four departments by plastic film. Each department is filled with gelatin, but with different concentration of contrast agent. Fig. 6 shows the reconstructed image of a cross-section. The outlines are clearly recognizable, but the separating lines between the departments appear twice and have low contrast. The doubled lines are caused by the local variations of speed of sound. Additionally, the contrast is low due to absorption differences of the ultrasound pulse on its different path lengths in the medium.

That means local speed of sound and absorption maps are necessary to improve reconstruction. These maps can...
be calculated using standard reconstruction algorithms for transmission tomography, which are modified in respect to our environment [18]. Fig. 7 displays the speed of sound map of the examined phantom. The resolution of the maps is low, because we measured only 100 emitting positions, but sufficient for corrections. During reconstruction the correct speeds of sound are determined by summing and weighting the pixel values of the speed of sound map along the signals travel line. The computation of the absorption value is similar, but the whole A-scan is multiplied with a correction factor dependent on the determined absorption, thus increasing the signal strength. The resulting image is displayed in Fig. 8. The sharpness of the inner structures is increased by a factor 5.9. We define sharpness as ratio of the peaks height to width at half-maximum. The results
show that the use of speed of sound and absorption maps can significantly increase the quality of images reconstructed with a full aperture sum-and-delay algorithm. At this stage of development we were confident to start the work on a 3D-setup.

5. 3D-USCT

Our experimental 3D setup consists of a cylinder (see Fig. 9) (18 cm diameter and 15 cm height), filled with water and containing the object for diagnosis. At the surface of the wall, 384 sending and 1536 receiving transducers are embedded and arranged in three rings, each 5 cm high. The sensors are grouped in 48 transducer array systems (TAS) each consisting of eight senders and 32 receivers. The cylinder can be rotated by a motor in six steps, emulating a complete cover of the cylinder with transducers adding up to 2304 virtual senders and 9216 virtual receivers. Sequentially, each emitter sends an unfocused wave front with a centre frequency of 2.7 MHz. 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 quadratic aperture of $(1.4 \text{ mm})^2$ is a compromise between emitted signal amplitude and desired spherical emission characteristics, resulting in an opening angle of approximately 30°. The sensors are connected to a data acquisitions system for high data rates developed for the fluorescence detector of Auger [24]. All needed hardware devices, including the 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 [19].

Due to the large number of transducers, the data rate of the system is high. The 3D demonstration system records 192 channels in parallel; resulting in a data rate of 4 Gbytes/s. For each transducer, a high-speed data acquisition and data processing unit has to be used. With the progress in digital electronics and programmable gate arrays during the last years, such processing units can now be built. The USCT crate contains a controller board and eight digitization boards. The controller board supervises the TAS, digitization, and the readout. Each digitization board samples 24 channels with 10 MHz and 12 bits. The boards are connected via a VME-Bus backplane.

Due to sequential emission the overall data acquisition time for one volume is 1.4 s, assuming 2304 emitting positions, a maximum diagonal distance of 23 cm, speed of sound in water of 1500 m/s, and allowing 0.3 ms for the ultrasound to fade away after each shot. With the actual setup (192 parallel readout channels, 16-fold averaging and 6 rotations) a lower limit of 3 min for data acquisition can be reached. The bottleneck is the available memory and the speed of the on-line data readout. This results in an insufficient readout time of several hours. Thus living tissue cannot be imaged with this version.

For one breast, 20 Gbytes of raw data have to be recorded and processed. Therefore, the reconstruction time for sub-millimeter resolution is immense. Reconstruction of the whole USCT volume with voxels of 0.4 mm edge length leads to 80 million voxels. Using the measured 3.5 million A-scans for this high-resolution reconstruction results in approximately 1 month processing time on a standard PC or several days using 8–16 PCs in parallel [20]. At the moment sub-volumes, volumes of lower resolution and high-resolution slice images are reconstructed to get images in acceptable time.
5.1. 3D calibration

The performance of the actual 3D setup was tested in different measurements. First the system was calibrated, where the USCT is only filled with water. This measurement was necessary to evaluate the reproducibility of the ultrasound transducers and the geometry of the sensors. The recorded A-scans were used to calculate the transducer properties. The average of the resonance frequency was measured to 2.7 MHz with a standard deviation of 0.2 MHz (6%). The bandwidth was in average 3.2 MHz. The average of the maximum pulse height at 0 amounted to 93.9% of the total maximum amplitude and had a standard deviation of 7.4%.

The transducers had a mean aperture angle of approximately 7.30°. The first and second side lobes are located at approximately 7.40° and 7.65°, respectively. Due to these results, the transducers of the developed TAS satisfy the requirements and are satisfactorily reproducible.

5.2. Imaging results: nylon threads of 0.15 mm diameter

Four nylon threads of 0.15 mm diameter and 15 cm length were suspended from the top of the 3D-USCT. The maximum distance of two nylon threads was approximately 6 cm. They were weighted down by metallic flat washers at the lower end of the 3D-USCT, so that the nylon threads spanned the whole sensor height and the flat washers were just below the sensor area. The experiment was imaged with all sensors, but using only one motor position. The volume reconstruction (Fig. 10) with 0.4 mm voxel edge length clearly reproduces the nylon threads over the whole height. The artifacts at the bottom are due to reflections from the flat washers and at the image edges due to the USCT walls. In a reconstruction of a slice image at the center level with pixel edge length of 0.03 mm the mean diameter (FWHM) of the nylon threads is 0.16 mm (standard deviation 0.06 mm), a very good match to the real size of the nylon threads.

5.3. Imaging results: CIRS triple modality biopsy phantom

For a more realistic test, we imaged a clinical breast phantom (CIRS triple biopsy breast phantom [25]) with our USCT. The phantom is breast shaped and contains cancer and cyst mimicking masses of 2–10 mm in diameter. Its physical characteristics resemble at mean a 50% glandular breast. For comparison, a MRI volume of the phantom (1.5 T Siemens Magnetom Vision, double breast coil, T1-weighted, (1.37 mm)³ voxel size), a X-ray mammogram, and a sonogram were acquired.

For first tests only the central layer of sensors was used, resulting in a focal region of approximately 2 cm height at the centre level of the cylinder. In this region, the surface of the phantom (“skin”) and the masses (“cysts” and “cancers”) are clearly visible and show good agreement with the MRI data (see Fig. 11). As expected the speckle noise of the system is reduced to the level of background noise due to the spatial compounding effect of the system.

Fig. 10. Imaging results: Four nylon threads of 0.15 mm diameter each, equalizing approximately One-fifth of the wavelength. The 15 cm long nylon threads are suspended into the USCT. Their largest mutual distance is approximately 6 cm. They are clearly reproduced over their whole length in spite of using only one motor position.

Fig. 11. Comparison of MRI slice (a) and USCT slice (b) of the CIRS breast phantom in the focal region of the USCT.
The results of imaging of the breast phantom with USCT are encouraging. In the focal region the interfaces of the masses appear to be even clearer than imaged in the MRI. Still, artifacts due to strong reflectors inside and outside the focal region are present in the image. The bright blob in the center of the image obscures partly two of the cysts. It is caused by a cluster of strong reflectors (“cancer”) near the “nipple” of the phantom outside the focal region. Due to the ellipsoidal back-projection, the ellipsoids originating from reflection/scatter outside the focal region are also drawn into the focal region. In case of reflections much stronger than the ones in the focal region, even single ellipsoids can be brighter than the sum of all ellipsoids imaging a point of weak reflection. Similar artifacts can also be observed for stronger reflectors inside the focal plane (e.g. “cancer” structure in the top left of the image). This effect is analogous to grating and side lobes as described for linear arrays. Thus, increasing the number of sensors even further could be a solution to this problem, or in the 3D case, a combination of increasing the number of sensors and changing the 3D aperture in azimuthal direction towards a hemisphere could lead to a more optimal global point-spread function of the system.

6. Conclusion and future perspective

So far structures down to 0.15 mm size could be recognized with our 2D and 3D ultrasound computer tomography.

For clinical applicability, the reconstruction time for a high-resolution volume has to be reduced to less than an hour. Fortunately, the reconstruction algorithm can be parallelized easily [20]. We are working on a reconstruction scheme for Grid computing as well as on parallel reconstruction using the FPGA hardware with some 100 GOPS/s computing power available on the digitization boards under development.

The applied reconstruction algorithm is based on a number of oversimplifying assumptions, which cause artifacts in the images, e.g., the assumption of constant speed of sound is causing structures to be doubled. Adapting the reconstruction to more realistic preconditions leads to higher image quality and resolution as shown in the 2D-USCT case. Our 2D approach is actually expanded to the 3D data. Using a 3D simulation tool for ultrasound signals [21], we could show that it is possible to reconstruct point scatterers at 0.05 mm in diameter, which corresponds to one-tenth of the wavelength [22].

A new generation of data acquisition hardware is under development to reduce the duration of data acquisition to 1.5 min. Sub-images may be recorded in 100 ms and allow to eliminate artifacts due to patient movements and temperature variations. Furthermore, an enhanced precision of calibration and the introduction of coherent analysis will improve the resolution and facilitate a stronger suppression of artifacts. Also the sensor configuration has to be optimized and additional sensors will cover the bottom of the cylinder. Fast acquisition of a volume will open additional applications for USCT, e.g. real-time imaging of contrast agent distribution in the breast. Additionally, Doppler and elastography imaging with contrast agents are possible.

Our USCT images can be easily registered to MRI volumes, due to the 3D imaging of the undeformed breast in prone position. This combination could support early breast cancer diagnosis by providing complementary information about the tissue properties. Furthermore, it may provide a valuable tool for image-guided biopsy. In future USCT could be extended to additional applications like diagnosis of testicular cancer and hip dysplasia in newborns and search for inclusions and defects in bulk materials. Due to the large number of emitters present in USCT and the possibility to focus them at any point in 3D, therapy by ultrasound hyperthermia may be also a future application.

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References

[25] CIRS Incorporated, Norfolk, VA, USA.