# ULTRASONIC ATTENUATION IN COMPUTED TOMOGRAPHY

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## ABSTRACT

Ultrasonic Computed Tomography (USCT) is a relatively new imaging modality primarily aimed at breast cancer diagnosis. The examined object is placed in a tank, covered with several thousands of ultrasonic transducers. Each of these transducers is used for emitting and receiving radiofrequency signals, which are then used for tomographic reconstruction of the object. This paper focuses on ultrasonic attenuation imaging. Two approaches to estimating the ultrasonic attenuation coefficients are described and the reconstructed attenuation images are presented.

## 1 INTRODUCTION

Ultrasonic attenuation parameters of human tissue are closely related to their type and pathological state [1]. Estimation of these parameters can be therefore used for tissue characterization (i.e. discrimination between benign and malignant structures).

Techniques for estimation of attenuation coefficients using the standard B-mode ultrasound have been developed. However, the estimates can only be made in fairly large regions [2], and thus the resulting attenuation images are of poor spatial resolution.

Attenuation coefficients can also be estimated from the transmission signals obtained from a USCT system. In B-mode ultrasound, only the reflected ultrasonic waves are recorded, whereas in USCT, also the signals corresponding to the directly transmitted waves are available. Thus in B-mode ultrasound, the waves are attenuated twice, on the way from the transmitter to the reflector and then back from the reflector to the transmitter. The signal to noise ratio is therefore much smaller than in the USCT. Even though the results of attempts to reconstruct attenuation images in the USCT never came out in such a high quality as the reflection images (images of the reflectivity parameter) they have the potential to be at least used for a qualitative assessment of the examined object.

In this article, two approaches to estimating the attenuation coefficients are described and compared. Both of these approaches are then used to process signals obtained from a USCT system containing an ultrasonic phantom. Attenuation images of the phantom are reconstructed using the backprojection algorithm.

## 2 ESTIMATING ATTENUATION COEFICINETS

Techniques for estimating the ultrasonic attenuation in the Ultrasonic Computed Tomography have been published in [3]. I have implemented two of them: the energy ratio method, and the method of log spectra differences. It is convenient to say in advance, that there are always two sets of signals available from a USCT measurement. The first set is recorded when the USCT tank is empty (only the coupling medium – water – is present). The second set is recorded when the measured object is in the tank. Further details are described in the USCT system chapter.

In the energy ratio method, the estimated attenuation parameter $\beta$ is simply defined as

$$\beta = \tfrac{1}{2}\ln\left(E_0/E\right) \quad (1)$$

where $E_0$ is the energy of a signal from the empty measurement set, $E$ is the energy of a signal measured with the object in the USCT tank. It is necessary to say, that this method doesn't take the frequency dependency of the attenuation in to account. In case of a broad band ultrasonic pulse emitted in to a medium, the higher frequency components are attenuated more than the lower frequency components. Thus the attenuation will be underestimated by using this method [3].

The second method, on the other hand, utilizes the frequency shift of the attenuated ultrasonic signal towards the lower frequencies. The power spectrum $P(f)$ of an attenuated signal is [3]: $P(f) \cong \left(\prod_m T_m\right)^2 P_0(f)e^{-2\beta(f)}$, where $T_m$ are the transmission coefficients at the interfaces of different mediums (i.e. tissue-water interface), $P_0$ is the power spectrum of the empty measurement signal, and the exponential part contains the frequency dependant attenuation parameter $\beta$. Taking the logarithm of this expression yields:

$$2\beta(f) + b = \ln P_0(f) - \ln P(f) \quad (2)$$

where $b = \ln(\Pi_m T_m)^2$. The attenuation $\beta(f)$ can be modeled as a function, linearly depending on frequency: $\beta(f) = \alpha_0|f|$. Equation (2) actually states that by subtracting the logarithms of the power spectra, a linear function of frequency is obtained. The attenuation $\alpha_0$ (the slope of this function) can be computed by a simple linear fitting.

## 3 USCT SYSTEM

A USCT system is being developed in Forschungszentrum Karlsruhe, Institute for Data Processing and Electronics, Germany. Only a demonstration system is now available, allowing reconstruction of 2D images of the tomographic plane. In future, the system should be capable of scanning a 3D volume.

The imaged object is enclosed by several thousands of ultrasonic transducers placed on a ring. One at a time, each of these transducers emits a pulse wave in to the imaged volume and all of the other transducers record the received transmitted, reflected or refracted radiofrequency signals (fig.ure 1). From these signals, it is possible to reconstruct a parametric image of the object. Three different parameters can be imaged (depending on the computation procedure): the reflectivity, the speed of the propagating ultrasound, and the attenuation.

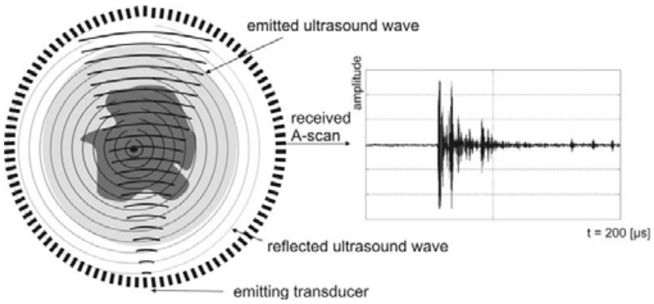


**Fig. 1:** *USCT system*

The waves emitted in to the imaged volume from the emitting transducer are undirected and thus all of the other transducers can receive the radiofrequency signals at the same time. Using this approach significantly speeds up the scanning time, but it also has its drawbacks.

Receiving transducers are recording ultrasonic beams coming from all the directions. The signals are added together and it is then impossible to distinguish them. At this moment of the project, only the directly transmitted signals (the first peaks) are taken into account, and the reflected and refracted waves that are recorded later on in the signal are neglected. Unfortunately it is sometimes hard to decide at which moment the actual directly transmitted wave hits the receiving transducer. The local speeds of the scanned object may vary. This phenomenon introduces errors in estimating the attenuation coefficient.

## 4 RECONSTRUCTION

There are several possibilities how the resulting image can be reconstructed. The most straight forward method is the filtered backprojection algorithm. It is widely used in the X-ray CT systems and there is a lot of literature on this subject available [4]. The estimated attenuation values corresponding to attenuations along ultrasonic beams are grouped in to parallel projections. Each projection is filtered with a ramp filter (a high pass filter with the transfer function $H(f) \cong |f|$ ). After this step, the projections can be backprojected to form the image. For each pixel of the image, all the contributing projection values crossing this pixel are added together. Because of the filtration, the lower frequency values are added with a smaller weight, and the reconstruction produces correct results.

Other approaches to reconstruction (so called algebraic methods) are all based on solving a set of equations. An equation for each projected beam is constructed. The pixel values (crossing the beam path) are the unknowns. Usually the set of equations has to be highly overdetermined in order to give stable results [5]. Unfortunately, highly overdetermined sets (sets of tens to hundreds of thousands of equations) are too large to be simply solved by matrix inversion or LU decomposition, etc. Most of the practical approaches introduce iterative methods.

For the reconstruction of the ultrasonic attenuation image, the backprojection algorithm was used. Because, the projections obtained from the USCT have a shape of a fan, they can't be used for the filtering right away. There is a so called rebinning process, used for the reconstruction of the 3$^{rd}$ generation X-ray CTs, where also only fan shaped projections are available. It is possible to use this rebinning process (with a slight modification due to geometry difference) for the USCT projections. It reorders the fan projections into parallel projections, making them ready for further processing.

## 5 RESULTS

A scan of an ultrasonic phantom was performed at Forschungszentrum Karlsruhe, Germany. The phantom is composed of a plastic container with a lid. Inside there are four compartments filled with gelatin each of different reflection value. Unfortunately, this phantom serves mainly as a reflection phantom, and its attenuation values aren't known. Nevertheless, we can very well assume that the attenuation values of the plastic case will differ significantly from the ones of the gelatin or water.

Figures 2-5 display the results of the reconstruction. Figures 2 and 3 were reconstructed using the attenuation values estimated by the energy ratio method, whereas figures 4 and 5 were reconstructed utilizing the attenuation estimates by the method of log spectral differences. Figures 3 and 5 were obtained by filtering the projections with a cosine filter.

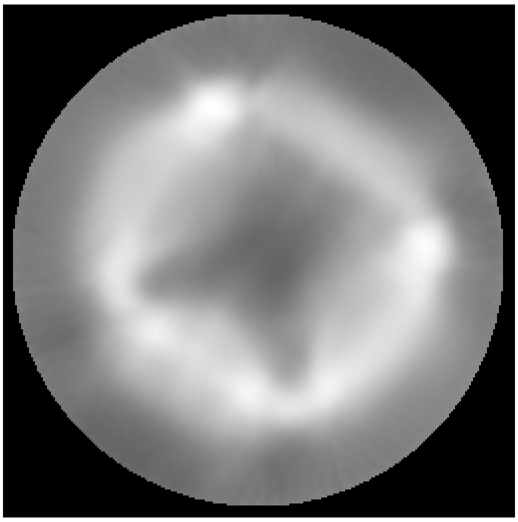

**Fig. 2:** *Simple backprojection using the energy ratio method attenuation values*

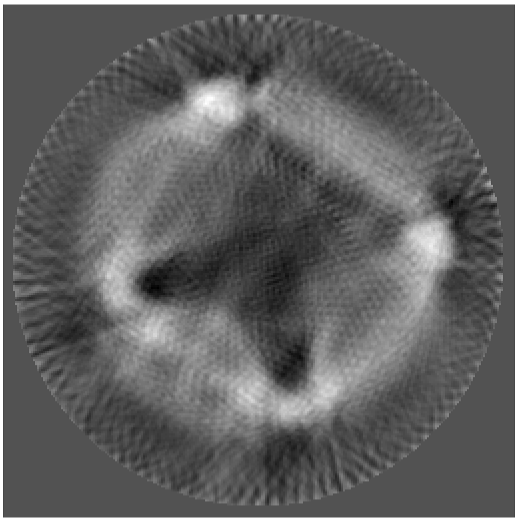

**Fig. 3:** *Filtered backprojection using the energy ratio method attenuation values*

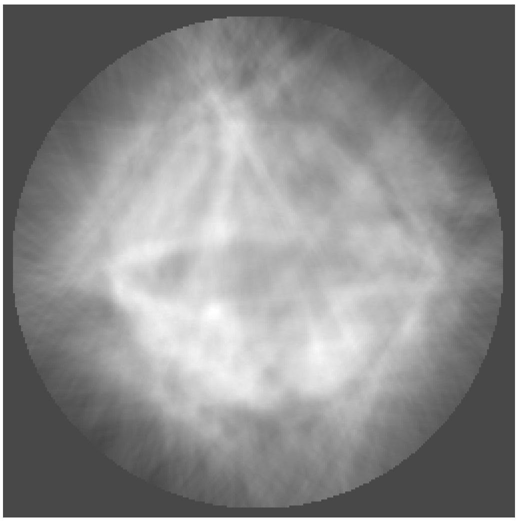

**Fig. 4:** *Simple backprojection using the log spectra differences attenuation values*

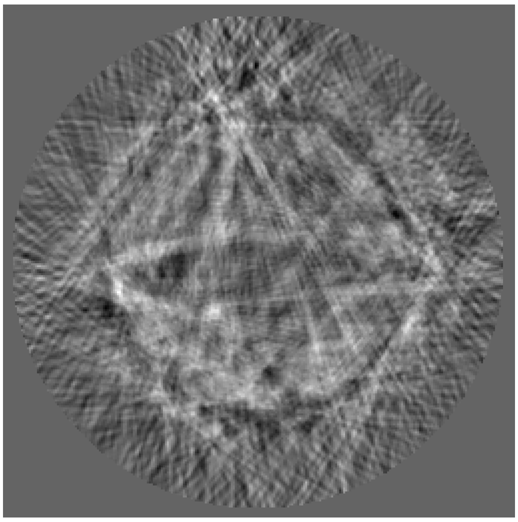

**Fig. 5:** *Filtered backprojection using the log spectra differences attenuation values*

## 6 CONCLUSION

Two methods of estimating the attenuation coefficients were presented in this paper: the energy ratio method, and the method of log spectra differences. Attenuation coefficients of an ultrasonic phantom were estimated and attenuation images were reconstructed using the backprojection algorithm. The method of log spectra differences yields much noisier results than the energy ratio method. This is probably due to the frequency dependency of the radiation patterns of the used ultrasonic transducers. At this moment, the exact radiation patterns of the transducers aren't available, and thus the errors in estimations can't be compensated for. Also both of the methods suffer from high refraction of the ultrasound beams on the edges of the phantom. This yields geometric deformations. The quality could probably be improved by a phase sensitive acquisition preprocessing. Also a different phantom with defined values of attenuation would be helpful for an assessment of the quality of the reconstructed images.

## ACKNOWLEDGEMENTS

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