Improved Image Quality in Phased Array Ultrasound by Deconvolution

Patrik BROBERG, Mikael SJÖDAHL, Anna RUNNEMALM

1 Department of Engineering Science, University West; University West, SE-46186, Trollhättan, Sweden; Phone: +46 520 22 33 69, Fax: +46 520 22 30 99; patrik.broberg@hv.se , mikael.sjodahl@hv.se , anna.runnemalm@hv.se

Abstract
High contrast and resolution in phased array ultrasonic images are of importance for accurate evaluation. The spread of the ultrasonic beam is one cause of the images being unsharp. One technique for reducing the influence of the beam spread, and thereby improving the image quality, is by deconvolving the data with the point spread function of the ultrasonic beam. By assuming that the material is homogeneous, the point spread function of the beam can be simulated using diffraction theory. Results from a deconvolution performed on data acquired from a side drilled hole in a steel calibration block are presented. It is shown that a significant improvement in sharpness and contrast can be achieved.

Keywords: ultrasound, phased array, image quality, deconvolution, point spread function

1. Introduction

In non-destructive testing (NDT) using ultrasound, the result is negatively affected by the width of the beam. This is made evident during inspection using a phased array ultrasound system where the spread of the beam can be seen as a blur in the image and the response from detected defects will be spread perpendicular to the beam. This effect can be reduced by focusing the beam; however focusing is usually only used at one distance and it can result in poor results at other distances. The consequence of the spread is that the shape of defects and details will be obscured. It can also result in the echo from two adjacent defects being misinterpreted as if it originated from one larger defect or defects close to larger reflections, such as corners being overlooked.

Images affected by the beam spread can be improved by using deconvolution if the point spread function (PSF) of the beam in the material is known [1]. One technique used in medical imaging with ultrasound, for acquiring the PSF, is blind deconvolution [2-4], where the PSF is estimated from the measured data. In this paper we instead acquire the PSF by simulating the beam with the assumption that the material is homogeneous. Data from a phased array ultrasound sector scan of a steel block with a side drilled hole is deconvolved with the simulated PSF. The results show that the width of echoes, caused by the spread of the ultrasound beam, is reduced and echoes will become more distinct and easier to separate.

2. Theory

A 2D image acquired using a phased array ultrasound, \( g(x,y) \), by varying the angle of the beam, can be seen as a convolution between the point spread function (PSF), \( h(x,y) \), of the sound beam and the sound reflecting surfaces in the material, \( f(x,y) \) [5]. If the noise is represented by \( e(x,y) \), the image is described by the equation

\[
g(x, y) = f(x, y) \ast h(x, y) + e(x, y).
\]  

Only the tangential spread caused by the ultrasound pulse will be treated here and Equation (1) is therefore reduced to
\[ g(x) = f(x) \ast h(x) + e(x). \] (2)

It is convenient to deal with the signal in the frequency domain and Equation (2) then transforms into

\[ g(k) = f(k) \cdot h(k) + e(k). \] (3)

If the PSF of the ultrasound beam and the noise is known, it is possible to calculate the shape of the object that caused the reflection. Since the PSF of an ultrasound pulse has a shape that is close to a Gaussian distribution, it will greatly increase the high frequency noise if Equation (3) is used to calculate \( f(k) \). A Wiener filter is commonly used to suppress noise when working with deconvolution. The Wiener filter is given by

\[ H(k) = \frac{h^*(k)}{|h(k)|^2 + \Gamma}, \] (4)

where \( \Gamma \) is the ratio between the noise spectrum squared and the squared signal spectrum \([4]\) and \( * \) is the complex conjugate. Replacing \( 1/h(k) \) in Equation (3) with \( H(k) \) from Equation (4) gives

\[ f(k) = \frac{g(k) \cdot h^*(k)}{|h(k)|^2 + \Gamma} - \frac{e(k) \cdot h^*(k)}{|h(k)|^2 + \Gamma} \] (5)

which, with the noise term omitted, is used for the deconvolution of the ultrasound images.

The PSF used in the convolution was calculated numerically for the ultrasound beam in the material using an exact numerical solution to the diffraction equation. The material was assumed to be homogeneous and the reflected wave was assumed to travel back to the transducer with only a loss in amplitude and no change in direction or scattering in the material. The PSF was calculated for the same coordinates as the data in the images, with the same resolution and one PSF was acquired for each angle in the scan.

The deconvolution was performed in Cartesian coordinates by plotting the radial coordinate on the y-axis and the angular coordinate on the x-axis. The data had to be sheared to compensate for the different length that the beams had to travel in the wedge, for different angles, before entering the material, as shown in Figure 1. Since the deconvolution is performed in the Fourier domain, it is sensitive to discontinuities, such as those found at the borders of the data. Due to this, all data was extrapolated by fitting a Gaussian curve to the edges, thereby removing the discontinuity.

The noise reduction term \( \Gamma \) in Equation 5 was calculated using averages from data containing only noise and from an area containing a large echo.
Figure 1. All data from the sector scans were transformed into a Cartesian coordinate system, sheared and extrapolated to prepare it for the deconvolution.

3. Experimental Setup

Data was collected from a test block containing a side drilled hole in order to test the method presented above. The experimental setup is shown in Figure 2, where A is the phased array ultrasound probe and B is the test block. The thickness $h$ of the test block was 25.4mm and the side drilled whole was positioned in the middle of the block. The phased array system used was a GE Phasor XS with a 4MHz, unfocused, 16 element probe, that was scanned with a scanning angle $\alpha$, from 45° to 70° with a 0.5° step.

Figure 2. Experimental setup showing the probe, B, and the test block, A, with a side drilled hole.

The data collected was 512 samples in length with 51 angles in width and was rectified to remove the 4MHz oscillations.

4. Results

The ultrasound beam was simulated for different angles using a diffraction equation and the result is shown in Figure 3a. Since the reflected signal was assumed to travel back to the transducer with only a loss in amplitude, the intensity of the beam, at the same coordinates as in the data, was used as the PSF. In order to perform the deconvolution the PSF was
transformed to a Cartesian coordinate system in the same way as with the data, compare Figure 1, and the result can be seen in Figure 3b.

The deconvolution was calculated in the angular direction through the whole data set and the result was transformed back to a polar coordinate system. The original and deconvolved images are shown in Figure 4.

A comparison of the width of the echo from the side drilled hole shows that the width at 50% amplitude has decreased by about 35%, making the echo more distinct. The two echoes, close to each other, on the bottom right side have become more distinct and easier to separate in the deconvolved image. It can also be seen that the deconvolved image contains more speckle noise.

5. Conclusions and Discussion

A phased array ultrasound sector scan has been deconvolved using a PSF obtained by simulating the ultrasound beam. The results show the width of echoes, caused by the spread of the ultrasound beam, is reduced and echoes become more distinct and easier to separate. Due to the imperfections in, among other things, the PSF and noise reduction term, there is an amplification of the noise in the image.

Figure 5b shows some improvements in comparison to Figure 5a in terms of the width of the echo from the side drilled hole, but also more noise. There are several things that could be
changed in order to improve on the results. The PSF is an approximation of the real PSF and will be improved if more material properties are added, such as scattering. The noise reduction term in the deconvolution has not been optimised but tests have shown that this term greatly affects the results, if it is too large there will be no improvement on the size of echoes, and if it is too small the deconvolution will greatly enhance noise and add oscillations around large echoes. The extrapolation of the data is needed to remove the discontinuity at the borders. This extrapolation was performed by fitting a Gaussian curve to the borders but did not always produce a completely smooth transition and did thereby give rise to additional noise. The deconvolution in itself uses a Wiener filter; several other methods have been developed that gives better results and is less sensitive to noise, such as in [5]. Deconvolution in medical imaging is most often performed using blind deconvolution; similar algorithms could possibly be used to fine-tune the PSF in order to take inhomogeneities in the material into account.

Acknowledgements

The work presented above was performed in cooperation with DEKRA Industrial AB, GE Sensing & Inspection Technologies, Innovatum Teknikpark, Volvo Construction Equipment, Volvo Aero Corporation and Fuji Autotech AB and was financed by the Knowledge Foundation, Sweden.

References