The study of the high frame rate imaging method and its application to the strain and strain rate imaging

Hong Chen

Follow this and additional works at: http://utdr.utoledo.edu/theses-dissertations

Recommended Citation
Chen, Hong, "The study of the high frame rate imaging method and its application to the strain and strain rate imaging" (2012). Theses and Dissertations. 283.
http://utdr.utoledo.edu/theses-dissertations/283
A Dissertation

entitled

The Study of the High Frame Rate Imaging Method and Its Application to the Strain and Strain Rate Imaging

by

Hong Chen

Submitted to the Graduate Faculty as partial fulfillment of the requirements for the Doctor of Philosophy Degree in Engineering

_____________________________________
Dr. Jian-yu Lu, Committee Chair

_____________________________________
Dr. Ronald Fournier, Committee Member

_____________________________________
Dr. Samer Khouri, Committee Member

_____________________________________
Dr. Scott Molitor, Committee Member

_____________________________________
Dr. Stephen Callaway, Committee Member

_____________________________________
Dr. Patricia R. Komuniecki, Dean
College of Graduate Studies

The University of Toledo

May 2012
An Abstract of

The Study of High Frame Rate Imaging Method and its Application to Strain and Strain Rate Imaging

by

Hong Chen

Submitted to the Graduate Faculty as partial fulfillment of the requirements for the Doctor of Philosophy Degree in Engineering

The University of Toledo
May 2012

The high frame rate imaging method is expected to be a leading technology of ultrasound imaging in the future. In this dissertation, the high frame rate imaging method is further studied, and the applications of the high frame rate imaging method are explored. To achieve the real-time 3D ultrasound imaging with a full 2D ultrasound transducer in a high frame rate imaging system, a new model of an ultrasound probe is proposed using the data compression technique. Raw echo data is first compressed by the JPEG algorithm, and then transferred to the image processing system with a reasonable number of wires in the cable connecting an ultrasound probe and the image processing system. The results show that the quality of the image after the JPEG data compression technique is still close to the image before the data compression. In addition, the effects of phase aberration and noise on the high frame rate imaging method is quantitatively studied. The parameters of energy ratio and sidelobe ratio are proposed and combined with the traditional -6-dB lateral resolution to analyze the image quality over the entire imaging area. The results show that the high frame rate imaging method has higher lateral...
resolution than the D&S imaging method, and the high frame imaging method is influenced less by the phase aberration and noise than the D&S imaging method. The results also indicate that the proposed parameters of energy ratio and sidelobe ratio could potentially be used for the correction of phase aberration. Finally, the high frame rate imaging method is applied to the strain and strain rate imaging of soft tissues. The two dimensional speckle tracking technique is used to measure the values of strain and strain rate. In the speckle tracking technique, the algorithm of the summation of absolute difference is used to search matched blocks between two consecutive images. Both simulation and experimentation studies were conducted to verify the method of applying the high frame rate imaging method to estimate the strain and strain rate. It is found that the high frame rate imaging method can be used to accurately measure the strain and strain rate. Considering the long computation time of speckle tracking, a new kernel using two crossing diagonal lines is proposed. The results of the preliminary study show that it is possible for the new kernel to reduce the computation time while retaining the accuracy of the estimation of strain and strain rate various imaging methods.
To my wife and daughter!
Acknowledgements

This dissertation is the result of not only my own effort, but also a lot of effort from my advisor, Dr. Jian-yu Lu, who provides me with helpful guidance, great support and encouragement. I want to express my sincere thanks to the committee members of my dissertation defense: Dr. Ronald Fournier, Dr. Samer Khouri, Dr. Scott Mollitor and Dr. Stephen Callaway for their patient guidance and helpful suggestions.

My thanks are extended to my friends, Rui Zheng, Huan Zhou, Bin Li, and Xinren Yu, for their help during my study in Toledo.

Last but not least, I want to thank my wife from the bottom of my heart for her unselfish love to me and tremendous contribution to our family, allowing me to focus on my research and work. Meanwhile, I also acknowledge my parents and my parents-in-law for their unconditional love and support.
# Table of Contents

Abstract .......................................................................................................................... iii

Acknowledgements ........................................................................................................... vi

Table of Contents .............................................................................................................. vii

List of Tables ..................................................................................................................... xiii

List of Figures .................................................................................................................... xiv

List of Abbreviations ......................................................................................................... xix

List of Symbols .................................................................................................................. xx

1 Introduction....................................................................................................................... 1

1.1 Background ................................................................................................................ 1

1.1.1 Ultrasound Imaging ............................................................................................... 1

1.1.2 Wave Propagation Theory .................................................................................... 3

1.1.3 Limited Diffraction Beam ...................................................................................... 6

1.1.4 High Frame Rate (HFR) Imaging Method .............................................................. 12

1.1.5 Reconstruction of Full View Images in Fourier Domain .................................... 17
1.1.6 Phase Aberration and Noise ............................................................... 21
1.1.7 Elasticity Imaging ............................................................................. 31
1.2 Motivation and Significance ................................................................. 35
1.3 Research Objectives ........................................................................... 37
1.4 Dissertation Organization ................................................................. 38
2 Study of Data Compression for Ultrasound Imaging .............................. 40
  2.1 Introduction ......................................................................................... 40
  2.2 Preliminary Analysis .......................................................................... 41
  2.3 Experiments and Results .................................................................... 45
  2.4 Discussion .......................................................................................... 51
  2.5 Conclusion ......................................................................................... 52
3 Quantitative Study of Effects of Phase Aberration and Noise on the High Frame Rate Imaging ................................................................. 53
  3.1 Introduction ......................................................................................... 53
  3.2 Parameters and Conditions ................................................................ 54
    3.2.1 Parameters for Assessing Quality of Images ................................. 54
    3.2.2 Addition of Phase Aberration and Noise ...................................... 58
3.3 Simulation and Results ................................................................. 61
  3.3.1 Simulation Conditions .............................................................. 61
  3.3.2 Simulation Results ................................................................. 63
3.4 Experiment and Results ............................................................... 74
  3.4.1 Experiment Conditions ............................................................ 74
  3.4.2 Experiment Results ................................................................. 77
3.5 Discussion ...................................................................................... 86
3.6 Conclusion ....................................................................................... 87
4 Strain and Strain Imaging Using High Frame Rate Imaging Method .............. 89
  4.1 Introduction ................................................................................... 89
  4.2 Theoretical preliminary ................................................................. 90
    4.2.1 Speckle tracking technique ......................................................... 90
    4.2.2 Strain rate calculation ............................................................... 93
  4.3 Simulation study and results ......................................................... 94
    4.3.1 Simulation study conditions ....................................................... 94
    4.3.2 Simulation results ................................................................. 96
  4.4 Experimental study ................................................................. 99
C.6 Market Size and Trends.................................................................................. 140

C.7 Estimated Market Share and Sales..................................................................... 142

C.8 Competition........................................................................................................ 144

C.9 Regulatory Issues ................................................................................................. 145

C.10 Management Team and Organizational Structure ............................................ 146

C.11 Marketing Plan .................................................................................................... 146

C.11.1 Market Entry and Growth Strategy ................................................................. 146

C.11.2 Advertising and Promotion.............................................................................. 147

C.11.3 Price.................................................................................................................. 147

C.11.4 Sales Tactics.................................................................................................... 147

C.11.5 Distribution...................................................................................................... 147

C.11.6 Service and Warranty Policies ......................................................................... 148

C.12 Patents and Proprietary Issues .......................................................................... 148

C.13 Product Design and Development Plans ............................................................ 148

C.14 Manufacturing and Operations Plans ................................................................. 149

C.14.1 Manufacturing Plan ......................................................................................... 149

C.14.2 Operation Plans .............................................................................................. 150
C.15  Financial Plan ........................................................................................................ 152

C.15.1 Break Even Analysis ......................................................................................... 152

C.15.2 Yearly Pro forma Income Statements ............................................................. 153

C.16  Overall Venture Schedule .................................................................................. 154

C.17  Critical Risks and Assumptions ....................................................................... 154

C.17.1 Critical Assumptions: ....................................................................................... 154

C.17.2 Critical Risks: ................................................................................................... 155

C.18  Harvest Strategy ................................................................................................ 156

C.19  Proposed Company Offering ............................................................................ 156
List of Tables

1.1 Sound speed of pertinent materials and biological tissues........................................... 26

2.1 Compression ratio of 11 slices of echo data. ............................................................... 51

3.1 Differences of -6dB lateral beam widths between the HFR and the D&S imaging methods for the simulated images before adding phase aberration and noise.............. 67

3.2 Relative changes and their averages of the -6dB lateral beam width, ER, and SR for the HFR imaging methods based on the simulated data. ................................. 70

3.3 This table is the same as Table 3.2 except that it is for the three point scatterers located at lateral distances of 0, 20, and 40 mm at a depth of 50 mm. ....................... 74

3.4 This table is the same as Table 3.1 except that it is obtained with experiment data... 82

3.5 This table is the same as Table 3.2 except that it is obtained with experiment data... 83

3.6 This table is the same as Table 3.3 except that it is obtained with experiment data... 86

C.1 The market size of major ultrasound medical equipment markets.........................141

C.2 The yearly income statements of the period of 5 years............................................. 153

C.3 Proposed desired financing and offering................................................................. 157
List of Figures

1-1 B-mode image of a fetus................................................................. 1

1-2 M-mode image of the heart wall..................................................... 2

1-3 Doppler color flow image of a patient with mitral regurgitation in the left atrium...... 3

1-4 The coordinate system used for calculating simulated echo data......................... 6

1-5 X-waves at different depths from simulation and experiments............................ 8

1-6 The one-way and two-way Blackman window functions................................. 10

1-7 3D spatial Fourier-domain coverage of a pulse-echo imaging system................ 15

1-8 Spatial Fourier domain coverage for imaging of radiation sources..................... 16

1-9 Limited-diffraction beam imaging with multiple transmissions and steered plane
    wave imaging with multiple transmissions ........................................... 17

1-10 Mapping between the Fourier transform of echo signals and the Fourier transform of
    an object function.................................................................................. 19

1-11 The first order phase aberration corresponding to 5% overestimation in sound speed
    at the focal range................................................................................... 22
1-12 The impact of the first order phase aberration on the system point spread function (PSF)……………………………………………………………………………………………………..23

1-13 One realization of a simulated 50 ns RMS second order aberrator with a spatial autocovariance FWHM of 3mm. ………………………………………………………………………24

1-14 The impact of the second order phase aberration on the system PSF …………….. 25

1-15 Influence of noise on the construction of images of a line target with the Fourier and the conventional (delay-and-sum) methods using a linear array transducer.……….28

2-1 The schematic of the proposed ultrasound probe. ………………………………………42

2-2 The example of four slices of echo data acquired at four different steering angles. 43

2-3 The ATS 539 tissue mimicking phantom and the imaging area for the images used in the data compression analysis. ……………………………………………………………….46

2-4 The 11 slices of echo data without the addition of noise. ……………………………..47

2-5 B-mode image of Phantom ATS539 reconstructed with the HFR imaging method. 48

2-6 The 11 slices of echo data after data compression with the JPEG algorithm.………..49

2-7 The B-mode image reconstructed from the 11 slices of echo data after being processed through the data compression technique. …………………………………………..50

3-1 Procedures for obtaining a maximum envelope plot of the PSF of a B-mode image of a point scatterer. ………………………………………………………………………….56

3-2 A phase screen that is used to introduce phase aberration. ………………………....59
3-3 Flow charts for introducing the phase aberration and the noise into the high-frame-rate (HFR) and the delay-and-sum (D&S) imaging methods ........................................ 60

3-4 Imaging area that includes 8 point scatterers in the simulation study .......................... 62

3-5 Images reconstructed by different imaging methods and different transmission waves from simulated echo data ........................................................................................................ 64

3-6 Results of the -6 dB beam width, ER, and SR of the simulated images along the axial direction for the D&S and the HFR imaging methods .................................................. 66

3-7 Relative changes of the absolute changes of three parameters after adding the phase aberration and the noise at different depths for the simulated data ............................. 69

3-8 This figure is the same as Figure 3-6 except that it is for the three point scatterers located at lateral distances of 0, 20, and 40 mm at a depth of 50 mm. ............................. 72

3-9 This figure is the same as Figure 3-7 except that it is for the three point scatterers located at lateral distances of 0, 20, and 40 mm at a depth of 50 mm. ............................. 73

3-10 A modified AIUM 100-mm test object ........................................................................ 76

3-11 This figure is the same as Figure 3-5 except obtained with experiment data .......... 78

3-12 This figure is the same as Figure 3-6 except obtained with experiment data .......... 79

3-13 This figure is the same as Figure 3-7 except obtained with experiment data .......... 80

3-14 This figure is the same as Figure 3-8 except obtained with experiment data .......... 81

3-15 This figure is the same as Figure 3-9 except obtained with experiment data .......... 85
4-1 Block matching procedure for speckle tracking in two consecutive images........... 92
4-2 Length change of a bar for strain and strain rate calculation................................. 93
4-3 Imaging area in simulation study and the positions of two point scatterers............. 95
4-4 Consecutive images captured each time 1 ms elapsed, which are reconstructed from
simulated echo data with one transmission.. ......................................................... 96
4-5 Measured velocities and velocity errors of point scatterer (speckle) 1 and 2 in
simulation study.. ........................................................................................................ 97
4-6 Estimated 2D strain based on D&S and HFR methods in simulation study. ............ 98
4-7 Estimated 2D strain rate based on D&S and HFR methods in simulation study ...... 98
4-8 Overall setting for experimental study. ..................................................................... 99
4-9 Imaging area in experimental study and the positions of two point scatterers with
moving direction shown.. ........................................................................................... 100
4-10 Motor controlling system......................................................................................... 101
4-11 High frame imaging system.................................................................................... 102
4-12 11 consecutive images based on experimental echo data acquired at 11 moments
when motor moves 1 mm horizontally.. ....................................................................... 103
4-13: Estimated moving step of two point scatterers (speckles) in experimental study.. 103
4-14 Estimated 2D strain based on D&S and HFR methods in experimental study. ..... 104
4-15 The two crossing diagonal lines (shown as red) used in the SAD calculation ...... 105
4-16 The 11 slices of B-mode images after adding noise in simulation study .......... 106
4-17 The velocity measurement of speckle 1 and 2 in Figure 4-3 ......................... 107
4-18 The measurement of accumulated strain with two different kernel types ....... 108
4-19 The estimated strain rate with two different types of kernel ....................... 108
4-20 The 11 slices of B-mode images after adding noise in experimentation study .. 109
4-21 The measurement of the relative speed of speckle 2 .................. 110
4-22 The estimated strain with two different kernels ..................................... 110
A-1 The interface of “bscan” software used for reconstructing 2D B-mode images .. 131
C-1 Schematic showing desired mobility vs. computing power ...................... 136
C-2 Prototype of medical HFR imaging system ............................................. 138
C-3 The latent demand across regions across the US .................................... 143
C-4 The manufacturing flow chart of the HFR ultrasound imaging system ........... 150
C-5 Cash conversion procedures and length ................................................ 151
C-6 The break even analysis for the HFR ultrasound imaging equipment ............ 152
C-7 The overall venture schedule for the product ........................................ 154
## List of Abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>B</td>
<td>brightness mode</td>
</tr>
<tr>
<td>CW</td>
<td>continuous wave</td>
</tr>
<tr>
<td>DCT</td>
<td>discrete cosine transform</td>
</tr>
<tr>
<td>D&amp;S</td>
<td>delay-and-sum</td>
</tr>
<tr>
<td>ER</td>
<td>energy ratio</td>
</tr>
<tr>
<td>FFT</td>
<td>Fast Fourier Transformation</td>
</tr>
<tr>
<td>HFR</td>
<td>high frame rate</td>
</tr>
<tr>
<td>JPEG</td>
<td>Joint Photographic Experts Group</td>
</tr>
<tr>
<td>LDB</td>
<td>limited-diffraction-beam</td>
</tr>
<tr>
<td>M</td>
<td>motion mode</td>
</tr>
<tr>
<td>PSF</td>
<td>point spread function</td>
</tr>
<tr>
<td>RF</td>
<td>radio frequency</td>
</tr>
<tr>
<td>SAD</td>
<td>summation of absolute difference</td>
</tr>
<tr>
<td>SNR</td>
<td>signal-to-noise ratio</td>
</tr>
<tr>
<td>SR</td>
<td>sidelobe ratio</td>
</tr>
<tr>
<td>SRI</td>
<td>strain rate imaging</td>
</tr>
<tr>
<td>SPW</td>
<td>steered plane wave</td>
</tr>
<tr>
<td>TDI</td>
<td>Tissue Doppler Imaging</td>
</tr>
<tr>
<td>1D</td>
<td>one dimensional</td>
</tr>
<tr>
<td>2D</td>
<td>two-dimensional</td>
</tr>
<tr>
<td>3D</td>
<td>three-dimensional</td>
</tr>
</tbody>
</table>
List of Symbols

\( \varepsilon \) ....................... strain
\( \rho \) .......................... the distance away from the aperture’s center axis:
\[ \rho = \sqrt{x^2 + y^2} \]
\( \varsigma \) .......................... axicon angle
\( \dot{\varepsilon} \) ....................... strain rate

\( \nabla^2 \) .................. Laplace operator
\( k \) ......................... wave number
\( S \) ....................... transducer aperture area
\( r_{01} \) ......................... the space distance between points \((x_0,y_0,z_0)\) and \((x_1,y_1,0)\)
\( \vec{n} \) ....................... the outward normal to the aperture surface \( S \)
\( J_0 \) .......................... the zero-order Bessel function of the first kind
\( J_n \) .......................... the \( n \)-th order Bessel function of the first kind
\( u \) .......................... the strength of an ultrasound field
\( U \) .......................... the complex form of the strength of an ultrasound field
\( B \) .......................... the transfer function of a transducer array
\( D \) .......................... transducer aperture diameter
\( \bar{R} \) .......................... the temporal Fourier transform of echo signals
\( A \) .......................... the amplitude of a plane wave
\( T \) .......................... the transfer function of ultrasound transducer
\( F_{BL} \) .......................... the band limited version of object function
\( L_0 \) .......................... the original material length
\( L \) .......................... the material length after deformation
\( \Delta L \) .......................... the length change
\( V \) .......................... estimated moving velocity
\( V_0 \) .......................... motor moving velocity
Chapter 1

Introduction

This chapter introduces the background of the dissertation work. This includes ultrasound wave propagation, the limited diffraction beam, high frame rate (HFR) imaging method, phase aberration effects, noise effects, and elasticity imaging. Following that, the motivation for this dissertation, its significance and research objectives will be presented. The final part of this chapter presents the organization of this dissertation.

1.1 Background

1.1.1 Ultrasound Imaging

Audible sound waves have a frequency between 20 Hz and 20 KHz. When a sound frequency goes higher than 20 KHz which cannot be heard by the human ear, it is called ultrasound. Ultrasound waves can penetrate human soft tissues and can be reflected when encountering the boundary of two different tissues having different density. This reflected ultrasound wave, or so-called echo, can be collected by an ultrasound transducer to reconstruct images which can vividly show the structure of human organs [1]. These images can be shown in two different modes such as brightness mode (B-mode) or motion mode (M-mode). An example of a B-mode image is shown in
Figure 1-1. An M-mode image is obtained by monitoring a specific range over a length of time. An example M-mode image is shown in Figure 1-2. The Doppler Effect also can be used to detect blood flow in human hearts and arteries and color can be added over a B-mode image to show velocity magnitude and direction of blood flow Figure 1-3.

Figure 1-1: B-mode image of a fetus. The dark region is the uterus, which is filled with fluid.
Figure 1-2: M-mode image of the heart wall, used for assessment of cardiac wall motion during contraction. The black region is blood, the bright reflection is the pericardium (i.e., a membrane around the heart), and the gray region in between is the heart muscle itself. (Courtesy of Professor J. D’hooge, Department of Cardiology. Reprinted with permission of Leuven University Press.) [2]
Figure 1-3: Doppler color flow image of a patient with mitral regurgitation in the left atrium. The bright green color corresponds to high velocities in mixed directions because of a very turbulent flow leaking through a small hole in the mitral valve. (Courtesy of the Department of Cardiology.) [2]

1.1.2 Wave Propagation Theory

The wave equation in (1.1) gives a general description of a wave, which is applicable to ultrasound waves and assumes the propagation medium is homogeneous.

\[ \nabla^2 u - \frac{1}{c^2} \frac{\partial^2 u}{\partial^2 t} = 0 \]  

where \( u \) is the wave field, and \( \nabla^2 \) is the Laplace operator given in (1.2).
\[ \nabla^2 = \frac{\partial^2}{\partial x^2} + \frac{\partial^2}{\partial y^2} + \frac{\partial^2}{\partial z^2} \quad (1.2) \]

By substituting the complex expression of wave field, the Helmholtz equation, seen in (1.3), which is time independent, can be derived [3].

\[ (\nabla^2 + k^2)U = 0 \quad (1.3) \]

In this formula, \( U \) is the complex expression of \( u \); \( k = 2\pi / \lambda \) is the wave number and \( \lambda \) is the wavelength.

A wave generated by an aperture with finite size will inevitably diffract as it is propagating. Wave diffraction theory has been developed for a long time, initially proposed by Huygens in a hypothetical form. In Huygens' hypothesis of diffraction, every wave front is considered a secondary wave source, which induces Huygens wavelets to construct a new wavefront [3]. Later, Fresnel based his mathematical proof of the principle of the Huygens wave construction on the assumptions of amplitude and phase of secondary wavelets [3]. Those assumptions have been proven to be natural consequences of the nature of light waves by Kirchhoff. The diffraction formula is given in (1.4), which is referred to as the Fresnel-Kirchhoff diffraction formula. During the derivation of this formula, Green’s function is selected as the spherical expanding wave function, which is then used in Green’s theorem [3].

\[
U(x_0, y_0, z_0) = \frac{1}{4\pi} \iiint_S \frac{\exp(jkr_{01})}{r_{01}} \left[ \frac{\partial U(x_1, y_1, 0)}{\partial n} - jkU(x_1, y_1, z_1)\cos(\vec{n}, \vec{r}_{01}) \right] ds \quad (1.4)
\]
In this formula, $S$ is the aperture area; $r_{01}$ is the space distance between points $(x_0, y_0, z_0)$ and $(x_1, y_1, 0)$; $U(x_1, y_1, 0)$ is the wave source on the aperture; $\vec{n}$ is the outward normal to the aperture surface $S$. To successfully apply the Fresnel-Kirchhoff diffraction formula, it requires two boundary conditions, called Kirchhoff boundary conditions which are simultaneously applied to $U$ and $\partial U/\partial n$. However, these conditions cause inconsistency between theory and physical phenomena. To mitigate this problem, Sommerfeld proposed a new Green’s function that does not need the requirements of the Kirchhoff boundary conditions on both $U$ and $\partial U/\partial n$, rather, the boundary condition is only applied to either $U$ or $\partial U/\partial n$ to obtain a simpler diffraction formula (1.5) [3]. This formula is called the Rayleigh-Sommerfeld diffraction formula,

$$U(x_0, y_0, z_0) = \frac{1}{j\lambda} \int_S U(x_1, y_1, 0) \exp\left(\frac{jkr_{01}}{r_{01}}\right) \cos(\vec{n}, r_{01}) ds$$

(1.5)

where $\lambda$ is the wavelength; $S$ is the aperture area; $r_{01}$ is the space distance between points $(x_0, y_0, z_0)$ and $(x_1, y_1, 0)$; $\vec{n}$ is the outward normal to the aperture surface ($S$).

In the image quality assessment and strain/strain rate imaging in later chapters, the Rayleigh-Sommerfeld diffraction formula above is used to simulate two-dimensional (2D) B-mode images. Computer programs that process both the transmission and reception beams using steered plane wave (SPW) and other transmission waves are developed based on the Rayleigh-Sommerfeld diffraction formula (see Figure 1-4). The term $\cos(\vec{n}, r_{01})$ is substituted by $z/r_{01}$ in (1.5) using trigonometry [4]. Riemann sum is used to substitute double integral in (1.5) to simplify the computation complexity, which is shown with dashed lines in Figure 1-4. $U(x_1, y_1, 0)$ in (1.5) can be a plane wave ($A_0$), a
Gaussian wave \( e^{-\frac{(x^2+y^2)}{\sigma^2}} \), a focused wave \( A_0 e^{-j\frac{k}{\sqrt{F^2+x_0^2-F}}} \) where F is the focal distance or a limited diffraction beam which will be introduced in the next section. The detailed procedures of the simulation of the echo generation are given in Appendix B.

Figure 1-4: The coordinate system used for calculating simulated echo data. The transducer surface of the 1D array is on the \( x'-y' \) plane. The 1D array is virtually subdivided in elevation direction (dash lines) for simulation in order to substitute the integral with summation during field strength calculation. \((x_1, y_1, 0)\) is the coordinate values of a virtually subdivided element. \( z \) is the wave propagating direction. \((x_0, y_0, z_0)\) is a point located within imaging area. \( r_{01} \) is the space distance between the point scatterer and the virtually subdivided element.

1.1.3 Limited Diffraction Beam

As introduced in Section 1.1.2, waves (i.e. light waves or sound waves) are diffractive as they propagate. However, under certain assumed conditions such as infinite
aperture size, waves can go to infinity without diffraction, but in practice, the waves
diffact in a limited rate along axial and lateral directions with finite aperture size. In
1941, Stratton first theoretically discovered a type of limited diffraction beam, called the
Bessel beam [5]. Such a beam is a solution derived from the isotropic-homogeneous wave equation (1.1). Durnin in 1987 further studied the Bessel beam and verified it in optical experiments [6]. The limited diffraction beam formula derived by Durnin is shown in (1.6), where \( U(\vec{r},t) \) is the scalar distribution of a wave field; \( \rho = \sqrt{x^2 + y^2} \) is the distance away from aperture’s center axis; \( J_0 \) is the zero-order Bessel function of the first kind and \( k^2 = \alpha^2 + \beta^2 \).

\[
U(\vec{r},t) = J_0(\alpha \rho) \cdot \exp(j(\beta z - \omega t))
\]  

(1.6)

Since the wave equation (1.1) can be applied to both light and ultrasound waves, the solution of the Bessel beam to the wave equation exists for ultrasound waves as well. In 1990, Lu and Greenleaf designed an ultrasonic nondiffracting transducer which can be used for ultrasound medical imaging based on Durnin’s Bessel beam [7]. The nondiffracting transducer consists of 10s annular rings with the inner and outer radius equal to those zeros of the zero-order Bessel function of the first order. The in vivo and in vitro experiments were conducted to research the annular transducer’s capability [8]. Even though the Bessel beam possesses the benefits of a larger imaging depth of field and comparable axial resolution as compared to conventional beams such as the Gaussian beam, imaging contrast with the Bessel beam is lower than the conventional beams due to a higher sidelobe.
By solving the homogeneous wave equation (1.1) in a different way, Lu proposed a new family of limited diffraction beams in 1992, which was called the X-wave due to its X-shape waveform in the plane through the axial axis [9, 10]. Figure 1-5 gives an example of the X-wave with a specific series of parameters shown above the top row of the figure.

Figure 1-5: X-waves at different depths from simulation (top row) and experiments (bottom row) [10].

A key set of formulas [9] used to derive the X-wave are listed as follows:
\[ \Phi_\zeta(s) = \int_0^\infty T(k) \left[ \frac{1}{2\pi} \int_{-\pi}^{\pi} A(\theta) f(s) d\theta \right] dk \] (1.7)

\[ \Phi_K(s) = \int_{-\pi}^{\pi} D(\zeta) \left[ \frac{1}{2\pi} \int_{-\pi}^{\pi} A(\theta) f(s) d\theta \right] d\zeta \] (1.8)

\[ \Phi_L(r, \phi, z - ct) = \Phi_1(r, \phi) \Phi_2(z - ct) \] (1.9)

where

\[ s = \alpha_0(k, \zeta) r \cos(\phi - \theta) + b(k, \zeta)[z \pm c_1(k, \zeta)t] \] (1.10)

and

\[ c_1(k, \zeta) = c \sqrt{1 + [\alpha_0(k, \zeta)/b(k, \zeta)]^2} \] (1.11)

Here, \( k \) and \( \zeta \) are free parameters, \( \alpha_0(k, \zeta) \) and \( b(k, \zeta) \) are arbitrary complex functions and \( A(\theta) \), \( f(s) \) and \( D(\zeta) \) are well-behaved functions. However, only (1.7) is used to derive a new family of limited diffraction beams. Bessel beam is also a special case of the general solution (1.7). X-wave beam is derived by setting \( T(k) = B(k) e^{-a_0k} \), \( A(\theta) = j^n e^{jnt} \), \( f(s) = e^s \), \( \alpha_0(k, \zeta) = -jk \sin \zeta \), \( b(k, \zeta) = jk \cos \zeta \), giving the final X-wave formula (1.12) where \( n \) is the order of X-wave and is an integer equal to or larger than 0. \( J_n \) is the nth-order Bessel function of the first kind. \( B(k) \) is the transfer function of a transducer array, which is usually the Blackman window function (1.13) for a bandwidth limited transducer.

9
\[ \Phi_{X_n} = e^{-jn\phi} \int_0^\infty B(k)J_n(k r \sin \zeta) e^{-k[a_0 - j(z \cos \zeta - \sigma)]} \, dk \] 

(1.12)

\[ B(k) = \begin{cases} 
  a_0 \left[ 0.42 - 0.5 \cos \frac{\pi k}{k_0} + 0.08 \cos \frac{2\pi k}{k_0} \right], & 0 \leq k \leq 2k_0 \\
  0, & \text{otherwise} 
\end{cases} \]

(1.13)

The plots of one-way and two-way Blackman window functions are shown in Figure 1-6.

![Blackman Window Function](image)

Figure 1-6: The one-way and two-way Blackman window functions [11].
Since practical transducers are all band limited and have finite apertures, the formula (1.14) of the nth-order X-wave for limited bandwidth and finite aperture is obtained by substuting (1.12) into the Rayleigh-Sommerfeld diffraction formula (1.5).

\[
\Phi_{Rn}(\vec{r},t) = F^{-1}[B(k)*F^{-1}\left[ \frac{1}{j\lambda} \int_0^{2\pi} d\phi' \int_0^{D/2} r'dr' \left( \frac{2\pi}{c} J_\nu(kr'\sin\zeta) H(k)e^{-\frac{\mathrm{e}^{jkr_0\sin\zeta}}{r'_{01}^2}} \right) \right] ]
\]

(1.14)

The X-wave, similar to the Bessel beam, has a large imaging depth of field: for example, the imaging depth is 178.8 mm given the aperture diameter \((D)\) equal to 25 mm and an Axicon angle \((\zeta)\) of 4 degrees [9]. One interesting thing is that the maximum imaging depth of X-wave depends on the aperture size and axicon angle only. This is different from Bessel beam which depends on the frequency of transmission waves as well. However, X-wave also has a higher sidelobe than a conventional focused beam at a focal plane but with larger imaging depth of field. Even though at present the application of X-wave is limited to an annular transducer which requires mechanical movement to obtain a B-mode image, steered X-waves with 2D array have already been studied by Lu and could be promising to be applied to B-mode imaging in the future once the technology of a fully sampled 2D transducer array matures [12, 13].

Research on the reduction of the sidelobe of limited diffraction beams has also been conducted by Lu. In 1993, the summation-subtraction method was applied to the Bessel beam by summing two A-lines obtained by two rotated 2nd order Bessel beams and then the result is subtracted from a 0th order Bessel beam with the compromise of reducing the imaging frame rate [14]. In 1995, the spatial derivative of limited diffraction solution
with respect to one transverse direction was taken to obtain a new family of limited diffraction beams which possess much lower sidelobe in this direction (the so-called bowtie limited diffraction beams due to their bowtie shape field pressure) [15-17]. The limited diffraction array beam was also developed in 1997 by Lu by taking the derivative of Bessel beam [18]. Using the linear combination of four specific limited diffraction solutions to the scalar wave equation (1.1), the grid array beam was developed [18]. In addition, using the linear combination of only two specific limited diffraction solutions (1.15), the layered array beam (1.16) was developed [18, 19]. All these new limited diffraction beams are promising in the applications of medical imaging, nondestructive material inspection, and the Doppler velocity measurement [20].

\[
\Phi_G(x, y, z - c_t t) = \cos(k_x x) \cos(k_y y)e^{ik_z(z-c_t t)}
\]

(1.15)

\[
\Phi_L(x, y, z - c_t t) = \cos(k_x x)e^{ik_z(z-c_t t)}
\]

(1.16)

where \( k_z = \sqrt{k^2 - k_x^2 - k_y^2} \); \( c_t = \omega / k_z \) and \( k_y = 0 \) for layered array beam.

1.1.4 High Frame Rate (HFR) Imaging Method

Based on his discovery of the broadband limited diffraction array beam (1.17) in 1997, Lu developed 2D and 3D (three-dimensional) high frame rate (HFR) imaging method [21]. HFR imaging method utilizes only one or two transmission, which is either a plane wave or a limited diffraction beam, and reconstructs images in Fourier domain. In (1.17), \( T(k) \) is the transducer transfer function that is proportional to the Blackman window function and \( H(k) \) is the Heaviside step function.
\[ \Phi_{\text{array}}(\vec{r}, t) = \frac{1}{2\pi} \int_{-\infty}^{\infty} T(k)H(k)e^{ik_x x + ik_y y + ik_z z} e^{-i\omega t} dk \]  

(1.17)

This method is operated in the Fourier domain and therefore, Fast Fourier Transformation (FFT) can be implemented to boost the computation speed for image reconstruction. Generally, there are three steps in HFR imaging with limited diffraction beams, as given in the following [21]:

- Firstly, use the FFT to transform echo signals in the temporal domain and the spatial domain;

- Secondly, map the frequency component obtained in the first step into a new spatial Fourier domain using an interpolation method;

- Thirdly, use inverse FFT to transform the spatial frequency components into the spatial domain;

When a plane wave with amplitude of \( A(k) \) is applied as the transmission and the receiver is weighted by the limited diffraction array beam (the same transducer is used as both transmitter and receiver), the object function is given by (1.18) and (1.19) where \( k'_z = k + k_z \) and \( \tilde{R}_{k_x, k_y, k'_z}(\omega) \) is the temporal Fourier transform of echo signals. To make the two equations more practical, an approximation is made by combining \( T(k) \) and \( F(k_x, k_y, k'_z) \) to form a band limited version of object function \( F_{BL}(k_x, k_y, k'_z) \).

\[
 f(\vec{r}) = \frac{1}{(2\pi)^3} \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} F(k_x, k_y, k'_z)e^{-i k_x x - i k_y y - i k'_z z} dk_x dk_y dk'_z 
\]  

(1.18)
\[ \widetilde{R}_{k_x, k_y, k_z'}(\omega) = \frac{A(k)T(k)H(k)}{c^2} F(k_x, k_y, k_z') \]  (1.19)

For 2D B-mode imaging, \(k_y\) is zero and consequently equation (1.20) gives the image reconstruction formula. The frequency mapping relationship is given in (1.21).

\[ f(x, z) = \frac{c^2}{(2\pi)^2} \int_{0}^{\infty} \int_{k > |k_z|} dk_x dk_y' F_{Bl}(k_x, k_y') e^{-ik_x x - ik_y' z} \]  (1.20)

\[ k = \frac{k_z'^2 + k_x^2 + k_y^2}{2k_z'}, \quad k_z' \geq \sqrt{k_x^2 + k_y^2} \]  (1.21)

For the case of applying the same limited diffraction beam as transmission and reception (called two-way dynamic focusing with the limited diffraction array beam [22]), the object function is given in (1.22) and (1.23) where \(k_x' = 2k_x, k_y' = 2k_y, k_z' = 2k_z\). The band limited version of the object reconstruction formula can be obtained by combining \(T(k)^2\) and \(F(k_x', k_y', k_z')\) which is similar to plane wave transmission.

\[ f(\vec{r}) = \frac{1}{(2\pi)^3} \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} F(k_x', k_y', k_z') e^{-ik_x' x - ik_y' y - ik_z' z} dk_x' dk_y' dk_z' \]  (1.22)

\[ \widetilde{R}_{k_x, k_y, k_z'}(\omega) = \frac{T(k)^2 H(k)}{c^2} F(k_x, k_y, k_z') \]  (1.23)

With the transmission of a limited diffraction beam, the HFR imaging method can gain larger coverage of spatial Fourier domain than plane waves as transmission (comparing Figure 1-7 and Figure 1-8 from [21]), resulting in higher spatial resolution, and there is no need of steering beams to cover large field of view because of the specific selection of
frequencies of transmission waves. However, the frame rate is reduced by two times because multiple transmissions are required to reconstruct one image [22]. The object functions exist in the spherical coordinate system when using both plane waves and the limited diffraction array beam as transmission. However, FFT cannot be applied in this coordinate system and integral computation is required which results in large computation time.

![Diagram](image_url)

**Figure 1-7:** 3D spatial Fourier-domain coverage of a pulse-echo imaging system where a plane wave pulse (broadband) is used in transmission and limited diffraction beams of different parameters are used in reception. (a) is 3D view and (b) is 2D view at the $k_x$-$k_z$ plane. $k_a$, $k_b$, and $k_c$ are different values of $k$ (wave number) [21].
Figure 1-8: Spatial Fourier domain coverage for imaging of radiation sources (k′=k, 
k_x′=k_x, k_y′=k_y, k_z′=k_z, k_1′=k_1) or for pulse-echo imaging using limited 
diffraction array beams in both transmission and reception (k′=2k, 
k_x′=2k_x, k_y′=2k_y, k_z′=2k_z, k_1′=2k_1) [21].

The HFR imaging method with limited diffraction beam has been verified as a superior imaging method to the conventional delay-and-sum (D&S) imaging method and as a robust imaging method which is affected less by phase aberration and noise [23-25]. Since no beamformer is required for the HFR imaging method and the FFT can be implemented, the hardware requirement for an imaging system adopting HFR imaging method is much simpler, especially for a 3D ultrasound imaging system. The HFR imaging system architecture has also been proposed in 1997 by Lu [21] and has been
successfully made in his lab in 2004 [26]. A patent of HFR ultrasound imaging system with limited diffraction beam was issued to Lu in 1998 [27].

1.1.5 Reconstruction of Full View Images in Fourier Domain

As mentioned in section 1.1.3, the transmitting beam, i.e. plane wave, used in the HFR imaging method can be steered electronically to gain a large field of view [21]. In 2006, the work to extend the HFR imaging method with limited diffraction beams was accomplished by Cheng and Lu [28, 29]. In the extended HFR imaging method, steered plane waves at different angles and multiple transmissions of limited diffraction array beams with different parameters are used. The beam steering schematics of the extended HFR imaging for plane waves and limited diffraction array beams are shown in Figure 1-9 [11].

![Steered Plane Wave Imaging with N Transmissions](image)

![Limited-Diffraction Array Beam Imaging with N Transmissions](image)

Figure 1-9: Steered plane wave imaging with multiple transmissions (a) and Limited-diffraction beam imaging with multiple transmissions (b) [11].
These were utilized in their home made imaging system and demonstrated a new way to produce ultrasonic images. If multiple transmissions of limited diffraction array beams are used, as they did, the time-delay unit, which is the core part of the conventional D&S imaging method is not needed. Such implementation increases the lateral resolution and image contrast while maintaining the large imaging depth. However, the compromise of the extended HFR imaging method is the reduction of frame rates that are inversely proportional to the number of transmissions.

The formulas for extended HFR imaging method are given in (1.24) where superscript “T” means transmitting. The variables in Equations (1.25) and (1.26) are defined as follows: \( k'_x = k_{xT} + k_x \), \( k'_y = k_{yT} + k_y \) and \( k'_z = k_{zT} + k_z = \sqrt{k^2 - k_{xT}^2 - k_{yT}^2} + \sqrt{k^2 - k_{xT}^2 - k_{yT}^2} \) [28]:

\[
\Phi^T_{array}(\vec{r}, t) = \frac{1}{2\pi} \int_{-\infty}^{\infty} A(k)e^{ik_{xT}\hat{x}+ik_{yT}\hat{y}+ik_{zT}\hat{z}}e^{-i\omega t} \, dk \tag{1.24}
\]

\[
f(\vec{r}) = \frac{1}{(2\pi)^3} \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} F(k'_x, k'_y, k'_z)e^{-ik'_x\hat{x}-ik'_y\hat{y}-ik'_z\hat{z}} \, dk'_x \, dk'_y \, dk'_z \tag{1.25}
\]

\[
\tilde{R}_{k_x,k_y,k'_z}(\omega) = \frac{A(k)T(k)H(k)}{c^2} F(k'_x, k'_y, k'_z) \tag{1.26}
\]

In this extended HFR method, the band limited object function in spatial Fourier domain is used to approximate the real object function, which is the same as the HFR imaging method with single transmission. In the extended HFR imaging theory, it is found that limited diffraction array beams weighted at the reception transducer is equivalent to the spatial Fourier transformation along x and y axis which produces 3D ultrasound images.
Therefore, the band limited object function can be obtained by doing temporal Fourier transformation and spatial Fourier transformation along \( x \) and \( y \) axes to collected echo signals. The frequency mapping functions from \( (k_x, k_y, k_z) \) domain to \( (k'_x, k'_y, k'_z) \) is given in (1.27) and also shown in Fig. 2 in [28]. In the extended HFR imaging method, bilinear interpolation is used.

\[
\begin{align*}
  k'_x &= k_x + k_{xT} \\
  k'_y &= k_y + k_{yT} \\
  k'_z &= k_z + k_{zT} = \sqrt{k^2 - k_x^2 - k_y^2} + \sqrt{k^2 - k_{xT}^2 - k_{yT}^2}
\end{align*}
\]  

(1.27)

Figure 1-10: Mapping between the Fourier transform of echo signals (a) and (c) and the Fourier transform of an object function (b) and (d), for limited-diffraction array beam (a) and (b) and steered plane wave (c) and (d) transmissions. Note that \( \theta \) in this figure is the steering angle of SPW [28].
Additionally, by assigning \( k_{xT} = 0 \) and \( k_{yT} = 0 \) (see equation (1.15)), a special case can be created whereby a plane wave of a limited diffraction beam is created. Another case where a two-way dynamic focusing imaging method with limited diffraction array beams can be achieved by letting \( k_x = k_{xT} \) and \( k_y = k_{yT} \) in (1.25) and (1.26). While using a SPW as the transmission beam of the extended HFR imaging method, the spherical coordinate system is used for frequency mapping relationships. Axicon angles therefore become the steering angles and are fixed for each transmission. The steering angle (Axicon angle) is obtained by equally dividing half maximum field of view by the number of transmissions. When using limited diffraction array beams as the transmission, \( k_{xT} \) is fixed for each transmission and the maximum value of \( k_{xT} \) is determined by \( 2\pi / \text{transducer\_pitch} \). The number of transmission determines the \( k_{xT} \) value used for each transmission. It is worth noting that one \( k_{xT} \) is used for two transmissions with sine and cosine weighting respectively.

In 2006, square wave aperture weighting instead of sine and cosine weighting was applied to extended HFR imaging systems to further reduce the hardware requirements for ultrasound systems [30]. With this method, only one or two transmitters are required to construct images. Such a simplification will be proven valuable in the future when HFR imaging systems are used for 3D imaging, for example, if a 128*128 2D array transducer is used, the number of transmitters will be decreased from 16384 to only 1 or 2. Furthermore, for a steered plane wave, the HFR imaging system is simplified again by rotating the Cartesian coordinate system of wave transmission by steering angles (for 2D imaging) [31]. Under the coordinate rotation, the unwanted high frequency components
produced by off-axis wave vectors are reduced so that the computation task is decreased because the number of data points used in FFT is reduced.

1.1.6 Phase Aberration and Noise

Phase aberration degrades the image quality of B-mode images. This phenomenon is mainly caused by the inhomogeneous speed of sound of human tissues. There are two types of phase aberrations: the first order and the second order phase aberrations [32]. The first order phase aberration results from the sound speed error between the sound speed used in the ultrasound system and the actual sound speeds for different persons. Since the beamformer, such as a focused beam of ultrasound, is designed according to a presumed general sound speed, a misalignment of the receiving beamformer will result in a phase error, which degrades image quality. The example of the first order phase aberration is shown in Figure 1-11 and its affect on point spread function is shown in Figure 1-12. The second order phase aberration arises from the inhomogeneity of biological tissues in a human body, which results in distortions of wave fronts for transmission waves. Therefore, the second order phase aberration can degrade image quality greatly and such degradation persists even after the first order phase aberration is compensated for. The example of the second order phase aberration is shown in Figure 1-14.
Figure 1-11: The first order phase aberration corresponding to 5% overestimation in sound speed at the focal range [32].
Figure 1-12: The impact of the first order phase aberration (sound speed error) shown in Figure 1-11 on the system point spread function (PSF). The aberrated case (bottom) is compared to the unaberrated control (top). The “sharpness” of the PSF is markedly reduced, reflecting losses of spatial and contrast resolution. The images area not displayed with the same gray-scale, so they do not show the 75% reduction in PSF amplitude that the aberration causes [32].
Figure 1-13: One realization of a simulated 50 ns RMS second order aberrator with a spatial autocovariance FWHM of 3mm. This particular aberrator creates 41.5 ns RMS phase error across the array, and has a spatial autocovariance FWHM of 2.7 mm [32].
Figure 1-14: The impact of the second order phase aberration shown in Figure 1-13 on the system point spread function. The aberrated case (bottom) is compared to the unaberrated control (top). As in the case of first order aberration, the “sharpness” of the PSF is reduced, reflecting losses of spatial and contrast resolution. Once again, the images are not displayed with the same gray-scale, so they do not show the 91% reduction in PSF amplitude caused by the aberration [32].
The discussion on wave front distortion caused by phase aberration dates back to 1974 when Marcus found that the measurement of tissue sound absorption had a large error when inhomogeneous tissue was encountered [33]. O’ Donnell also measured phase aberration in the liver from different human bodies, which showed that phase aberration affects image quality [34]. Table 1.1 gives the sound speeds of relevant materials to ultrasound research and human tissues [35]. From Table 1.1, a maximum sound speed difference of 8% exists between fat tissue and kidney tissue, which potentially distorts the designed ultrasound beams and then degrades image quality.

Table 1.1: Sound speed of pertinent materials and biological tissues at room temperature (20-25 °C) [35].

<table>
<thead>
<tr>
<th>Medium</th>
<th>Speed (m/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>1484</td>
</tr>
<tr>
<td>Aluminum</td>
<td>6420</td>
</tr>
<tr>
<td>Air</td>
<td>343</td>
</tr>
<tr>
<td>Plexiglas</td>
<td>2670</td>
</tr>
<tr>
<td>Blood</td>
<td>1550</td>
</tr>
<tr>
<td>Myocardium</td>
<td>1550</td>
</tr>
<tr>
<td>Fat</td>
<td>1450</td>
</tr>
<tr>
<td>Liver</td>
<td>1570</td>
</tr>
<tr>
<td>Kidney</td>
<td>1560</td>
</tr>
<tr>
<td>Skull</td>
<td>3360</td>
</tr>
</tbody>
</table>

Phase aberration can be corrected to obtain higher ultrasound image quality. Before performing corrections, the phase error across the transducer aperture through the
imaging depth needs to be measured. Methods utilizing signals from a targeted point scatterer, or signals from random collections of scatterers, can be used to estimate phase error to provide a guide for the correction of phase aberration [36, 37]. The mean of echo magnitude squared or higher power, studied by Zhao and Trahey, was shown to be effective quality factors which can guide the correction of phase aberration [38]. The impact of phase aberration was detected on ultrasound breast images from a group of females and then a phase aberration profile was obtained with the help of a mammogram to correct phase aberration, improving image quality [39]. Adaptive focusing, utilizing a synthetic receiving aperture, was used to compensate for phase aberration by Nock and Trahey [40, 41] and by Karaman [42]. Zhao and Trahey proposed a method utilizing the difference between consecutive echo signals from moving targets, such as blood flow, as a quality factor to guide the correction of phase aberration [43, 44]. In 2003, Fernandaze and Trahey implemented a 1.75 D array into an adaptive imaging system for the phase aberration correction by applying the least-mean-squares algorithm to determine two dimensional phase aberration [45]. Urban et al., in 2006, conducted phase aberration correction by optimizing radiation force to obtain the optimal focused ultrasound beam [46, 47]. In this method, the optimization of radiation force was achieved by monitoring the velocity of the targeted object.

For ultrasound imaging methods, noise is another factor that degrades image quality. Ultrasound is absorbed, scattered and reflected [33]. The reflected or scattered ultrasound signals are captured by an ultrasound transducer and used to reconstruct ultrasound images. However, the absorption of ultrasound energy attenuates the ultrasound signal when sound waves propagate. As a result, the signal captured by a
transducer becomes weak and the signal-to-noise ratio (SNR) is limited, especially, for large imaging depths \[11\]. The reduced SNR enhances speckle noise in final B-mode images and lowers the image resolution and contrast \[25, 48\]. An example of noise effect on B-mode image is given in Figure 1-15, which clearly shows the increase of speckle noises in images after the addition of noise with amplitude equal to 50% of maximum amplitude in images without noise. The sources of noise could come from thermal noise, on-chip electronic noise, and or amplifier noise \[11\].

![Influence of Noise in Image Construction](image)

**Figure 1-15:** Influence of noise on the construction of images of a line target with the Fourier and the conventional (delay-and-sum) methods using a linear array transducer. Noises that the upper and lower grayscale bars are in linear and log scales, respectively. The speed of sound is assumed to be 1450 m/s \[25\].
Extensive research has been conducted to improve the performance of ultrasound imaging systems by reducing noise. Chinrungrueng et al. reduced speckle noise by applying the two dimensional Savitzky-Golay filter based on the least squares polynomial surface fitting to image intensities [49]. Rabbani et al. derived the minimum mean square error of the echo signal and the averaged maximum posterior estimator based on the assumption that wavelet coefficients of the logarithm of images have local mixture distribution, to suppress ultrasound speckle [50]. Yan et al. proposed a method to nonhomomorphically filter ultrasound images by using the combination of the generalized likelihood ratio and the local Wiener filter technique [51]. Kwon et al. used multi-resolution coherence measurement to enhance edge, which is used for speckle noise reduction based on adaptive wavelets [52]. Roomi et al. developed a method to reduce speckle noise in ultrasound images by obtaining new pixel values from three different values, which are median values of diagonal elements, maxima values of horizontal elements, and vertical elements. These are weighted differently based on Particle Swarm Optimization [53]. Noise reduction technique was also explored for Doppler imaging. Newhouse and Amir proposed repetitious pulse Doppler processing to improve the SNR while monitoring a flow measurement. Afterwards, they switched to the continuous wave (CW) Doppler operation and found that it further enhanced the output SNR [54]. Sasaki et al. utilized a grey-level concentrative filter based process to reduce noise in CW Doppler images [55]. With this method, noise is significantly reduced from cardiac Doppler histograms. Olhede et al. applied multiple complex-value Morse wavelets to reduce ultrasound signal noise and demonstrated the efficacy of their method by implementing it into quadrature Doppler ultrasound blood flow imaging [56]. Wang et al.
stated an approach for noise reduction in Doppler ultrasound signals by means of the combination of adapted local cosine transformation and the non-negative garrote thresholding [57]. Noise imposed harm to elasticity imaging as well. Chaturvedi et al. applied two dimensional compounding technique to improve the correlation between the signals before and after compression, which ultimately improved the accuracy of strain estimation [58]. Hall proposed the combination of three different techniques to reduce strain imaging noise: first, select appropriate frame pairs with the criterion of reasonable frame-average strains; second, use statistical arguments and a moving linear regression to predict large displacement errors; third, use strain energy as the smoothness constraint function to model higher order motion [59]. Konofagoul et al. applied variable applied strain for dynamic range expansion of elastography and used an iterative temporal stretching algorithm for the windowed post-compression radio frequency (RF) echo signal to improve SNR [60]. SNR of ultrasound imaging systems was improved by O’Donnell through the implementation of pseudochirp code excitation and equalization filtering [61]. Benkhelifa et al. improved the output SNR of ultrasound imaging systems by transmitting a continuous coded signal and then receiving the echo signal with a high and short pulse which came from the compression of coded excitation [62].

In short, the phase aberration due to inhomogeneity of biological tissues and noise existing in ultrasound imaging systems affects image quality of ultrasound imaging systems, so it is important to understand the effect of phase aberration and noise on different ultrasound imaging methods. Moreover, the quantitative study on those effects is more significant than their qualitative study because it can be a guide for the correction of phase aberration and the improvement of the SNR of ultrasound imaging systems [63].
1.1.7 Elasticity Imaging

B-mode ultrasound images can provide quality information on organ shapes and boundaries inside the human body. However, it is difficult to visually detect some pathological tissues, from B-mode ultrasound images, but the mechanical properties or elasticity changes due to diseased tissues can be observed using elasticity imaging. For example, scirrhous carcinoma of the breast increases the stiffness of breast tissue due to hard nodules [64]; atherosclerosis changes the compliance of arteries, progressing into ischemia or thrombus [65]. Palpation has been a basic diagnostic tool to detect tissue pathology for centuries [66], but it encounters great limitation when it endeavors to diagnose diseases of tissues or organs located deep in the body, or the pathological lesion of small size. Dickson and Hill, in 1982, first monitored the soft tissue motion using correlation coefficient between pairs of A-lines [67]. In the same year, Wilson and Robison also used the technique of correlation of phase between consecutive M-mode signals to record the small displacement and small deformation of cardiac structures and chest wall movement in the fetus [68]. It is worthy of mentioning that the Doppler technique cannot be applied to detect slow tissue motion, such as that which is present in the artery wall caused by cardiac pulsation. Based on the idea of measuring tissue displacement, using correlation technique, Ophir proposed a method to measure the tissue elasticity in axial direction through the assessment of Young’s modulus of targeted tissue in 1991 [64]. In Ophir’s method, external stresses were applied to imaging objects with different elasticity, and then strain profiles were obtained by comparing the pairs of pre-compression and post-compression A-lines with correlation technique. In 1993, Ophir’s method of measuring tissue elasticity had been verified with \textit{in vivo} experiments on
muscle and breast tissue by Céspedes and Ophir, et al., and scirrhous carcinoma nodules in breast tissue were clearly shown in the resulting elastogram [69]. It was proven, by Belaid et al., that the contrast level of an elastogram in his study was higher than that of an echogram. In this simulation study, a disc-shape tissue with different stiffness and reflectivity than the surrounded medium was used [70]. Moreover, the elasticity imaging technique, utilizing cross-correlation, was optimized by utilizing the temporal stretching method to align the echo signals peaks and utilizing the multiple small compression averaging method in order to reduce the strain estimation error caused by decorrelation between pre-compression and post-compression A-lines [71]. In 1998, Konofagou et al. added the lateral strain estimation to the existing axial estimation by interpolating neighboring A-lines to produce high lateral tracking precision. In doing so, they further obtained the Poisson’s ratio of tissues to differentiate their mechanical property [72]. Another method to assess the axial strain value based on the assessment of centroid spectral shift of A-lines at the mean center frequency was proposed by Konofagou et al. [73, 74]. The spectrum based strain-estimation method was more resistant to decorrelation errors due to the insensitivity to phase decorrelation noise, but it was less precise, especially for small strains. The entire spectrum, rather than only the centroid shift of center frequency, was used by Varghese et al. to estimate the strain profile based on the RF A-lines of pre-compression and post-compression with the cross-correlation technique [75]. Konofagou et al. also applied the cross-correlation based elastography on M-mode signals to assess the myocardium function [76].

Color flow Doppler imaging of the blood flow in organs was developed by Kasai et al. in 1985 using a combination of the conventional pulsed Doppler technique and the
autocorrelation technique [77]. Based on the color flow Doppler imaging method, the Tissue Doppler Imaging (TDI) to detect the motion of myocardium was developed by McDicken et al. in 1992 [78]. Tissue motion velocity using TDI was obtained by revising the filter of the conventional color flow Doppler imaging system to retain the information of lower Doppler shift resulting from the myocardium movement. TDI was further investigated by Sutherland et al. in 1994 [79]. Zamorano et al., as well, utilized TDI to reveal the relationships between different myocardium velocity patterns and the myocardial phases of systole and diastole [80]. In fact, the study of myocardium motion to diagnose the heart health condition has already been conducted by Chu and Raeside in 1978 [81]. They sought the Fourier pattern of echocardiogram (or M-mode signal) to detect heart diseases. In 1990, Yamakoshi et al. also measured the elasticity information of soft tissue using the Doppler effect but not with the autocorrelation technique [82]. In 1998, Heimdal and Stoylen et al. found that it was possible for TDI to be applied to detect the mechanical parameter of strain rate of myocardium and the left ventricle was used as the subject for study to obtain the strain rate imaging (SRI) diagram [83]. Since the internal stress of heart wall is usually unknown, the strain or strain rate is difficult to be further used for the estimation of Young’s modulus. In echocardiography, strain or strain rate imaging is the basic tool to assess the mechanical properties of myocardium. The SRI was then applied to in vivo experiments to effectively localize dysfunction of left ventricle and localize lesion in coronary artery [84-86].

However, the strain or strain rate imaging has the problem of angle-dependence which allows the estimations of strain and strain rate to lie only in the direction of ultrasound beam propagation. To overcome that problem, strain or strain rate imaging
based on speckle tracking applied into B-mode images is proposed. Initially, speckle tracking based on the assessment of cross-correlation between two kernels in consecutive B-mode images was proposed to measure the velocity of blood flow by Trahey in 1988 [87]. Since speckles in a series of B-mode images were tracked in both axial and lateral directions, the obtained velocity was a vector having exact decomposed values along axial and lateral directions. Even though the cross-correlation based speckle tracking technique possessed higher estimation precision of speckle displacement, the computation time was too high, and so the summation of absolute difference (SAD) was used to substitute cross-correlation as the core algorithm of block match [88, 89]. In 1993, the SAD based speckle tracking technique was utilized by Walker et al. to measure the soft tissue displacements induced by the external mechanical waves [90]. O’Donnell et al., in 1994, utilized speckle tracking technique to acquire strain imaging of stiff inclusion inside soft medium [91]. In 1998, Chaturvedi et al. applied 2D compounding into strain imaging based SAD speckle tracking technique to reduce the decorrelation error between the pre-compression and post-compression echo data so that the accuracy of strain estimation was improved [58]. Kaluzynski et al. used 2D speckle tracking to estimation strain rate imaging of the mimicking heart phantom, providing the information of stiff structure inside a soft tissue [92]. The speckle tracking based strain assessment was verified and its applications were further explored by many in vivo experiments, for example, Amundsen et al. used dog’ heart as subject to noninvasively measure myocardial strain [93]; Fernanda et al. using 2D speckle tracking echocardiography to determine the mechanics of impaired left ventricle dysfunction [94]; Left ventricle
torsion was also measured by Notomi using the 2D speckle tracking technique and results showed consistence with those obtained with tagged magnetic resonance imaging [95].

Tissue mechanical parameter of viscoelasticity can also be used to measure tissue stiffness. Radiation force was applied as internal force to trigger shear waves and then strain caused by shear wave was recorded for diagnosis [96, 97].

1.2 Motivation and Significance

Although 2D ultrasound imaging is at present widely used in clinics and in research laboratories, high quality real-time 3D ultrasound imaging is presently a technical challenge. To reduce the complexity of system for 3D high frame rate imaging, data compression on the raw echo data collected from the receiving transducer is needed.

As mentioned in the previous section, human tissues have inhomogenous speed of sound, which causes phase aberration to ultrasound beams and thus distorts images. Electrical noise of ultrasound imaging systems cannot be completely removed, which degrades image quality. Although the effects of phase aberration and noise on the extended HFR imaging method have been studied qualitatively on image contrast, affects on the image resolution and change of energy distribution between mainlobe and sidelobes of a single point scatterer have not been studied quantitatively. Since biological soft tissues can be modeled as a group of multiple single point scatterers, it is necessary to quantitatively study the effects of phase aberration and noise on images of individual point scatterers for the HFR imaging method. This process can be analyzed with traditional parameters such as image resolution and a set of new parameters proposed. A combination of the traditional parameters with the set of new parameters can also be used
as criteria for a more comprehensive assessment of image quality for various imaging methods such as the HFR, the D&S, and any other imaging methods. Moreover, a quantitative analysis of the effects of phase aberration on image resolution, mainlobe, and sidelobe of a point scatterer can provide useful information for correction of phase aberration.

As introduced in Section 1.1, strain and strain rate can provide useful information to assist doctors with their diagnosis of lesion tissues or abnormal contraction of myocardium. To overcome angle dependent limitation caused by the inherent shortcoming of the Doppler method, 2D speckle tracking technique is used for displacement measurement. With conventional D&S imaging method to estimate strain and strain rate, the problem of skewed moving objects in a B-mode image exists and reduces the estimation accuracy. However, HFR imaging method possesses the feature of obtaining a B-mode image with only one transmission which ensures the snapshot of moving objects. The method combining 2D speckle tracking based on SAD and HFR imaging method is helpful to improve the accuracy while measuring tissue strain and strain rate.

When the HFR imaging method is used for assessing tissue deformation, usually the tissue displacement in human body is very small. In order to increase the estimation accuracy, the spatial resolution of B-mode images should be small as well. However, the higher image resolution will increase the image size and then increase the computation time for speckle tracking. To reduce the computation time, a new kernel for the block matching procedure for speckle tracking is also proposed. The new kernel is set to be square and utilizes only two crossing diagonal lines to calculate SAD. The reduction of
computation time will make possible real time application of strain and strain rate measurement into HFR imaging system.

1.3 Research Objectives

The preliminary study of data compression applied for reduction of data size of echo signals has been conducted. Since a full 2D array transducer with the elements of 128*128 is not available to us, echo data acquired from a 1D array transducer are used for study. Lossy data compression scheme is applied to raw echo data. The objective of this study is to find out if the image quality is still acceptable after the lossy data compress technique is applied to echo data before it is transferred from ultrasound probe to workstation.

The research objective of quantitative study of the influence of phase aberration and noise on HFR imaging system is to test the efficacy of newly developed parameters for assessing image quality. The new parameters are energy ratio (ER) which is the ratio between the total energy of sidelobe and the total energy of mainlobe, and sidelobe ratio (SR) which is the ratio between the peak value of the sidelobe and the peak value of the mainlobe of a point spread function for an imaging system. The newly proposed parameters are combined with traditional -6-dB lateral resolution to quantitatively measure the effects of phase aberration and noise on HFR imaging method. The conclusion that the HFR imaging method is superior to the D&S imaging method is obtained.

The objective of the research on strain and strain rate imaging is to prove that 2D speckle tracking technique based on SAD is able to be applied to measure strain and
strain rate by using high frame rate imaging method with one transmission. Both simulation and in-vitro experiment were conducted to verify this method. The benefit of modified block match scheme by substituting full 2D kernel calculation with two crossing diagonal lines was also be explored with both simulation and experimental data sets. The goal is to show that new block matching scheme can provide accurate estimation but reduce computation time by several times.

1.4 Dissertation Organization

Introduction to the background of ultrasound imaging, wave propagation and HFR imaging method, and elasticity imaging is given in Chapter 1. Meanwhile, Chapter 1 also provides the research motivation, significance, and objectives in section 1.2 and 1.3.

Chapter 2 introduces the lossy data compression technique and its application to the 2D echo data. Possibility is explored for the application of the lossy data compression into an ultrasound probe to realize real-time 3D ultrasound imaging with a full 2D array.

In Chapter 3, the influences of phase aberration and noise on the high frame rate imaging method is quantitatively studied. Computer simulation to generate ultrasound B-mode images including eight point scatterers is conducted for the quantitative study. In-vitro experiments, using modified standard phantom, are also performed to examine the efficacy of the newly proposed parameters. The results from HFR imaging method and conventional D&S imaging method are compared.

In Chapter 4, strain and strain rate are measured for a deformed virtual tissue consisting of two glass beads using the method of high frame rate imaging and 2D speckle tracking. Simulation of two point scatterers and experimentation in a water tank
are conducted to examine the method. The estimation errors of speckle moving velocity and moving stepwise are compared between the HFR imaging method and the D&S imaging method. The D&S method is applied using the dynamic focusing in reception. In this chapter, modified speckle tracking, using two crossing diagonal lines for block matching, is also investigated. Preliminary study is performed with simulation and experimental data sets which is contaminated with noise. Results are compared between the HFR imaging method and the dynamic focusing D&S.

Finally, a summary of the whole research work and the future studies are provided, and the interface of software, which is able to be used for the localization of the point of interest in a B-mode image, is presented in the Appendix.
Chapter 2

Study of Data Compression for Ultrasound Imaging

2.1 Introduction

Due to the technology bottleneck on the limited number of transducer elements attached to an ultrasound imaging system, real-time quality 3D ultrasound imaging is still a challenge. Currently, most ultrasound imaging systems and ultrasound imaging researches are focusing on 2D imaging. In the ultrasound imaging system based on the HFR imaging method, the square-wave aperture weighting technique can be adopted to reduce the ultrasound transmitters’ number, which is equal to the total number of transducer element in D&S imaging method, to only one or two [98]. However, the number of the electronic wires connecting transducer elements and the circuits collecting echo signals are still equal to that of transducer elements, resulting in an unacceptable size of the cable connecting ultrasound probe and image processing station. To provide a possible way for the HFR method utilized in future 3D ultrasound imaging systems, data compression techniques can be applied onto raw ultrasound echo data which is collected by a receiving transducer. The Joint Photographic Experts Group (JPEG) algorithm is
used to compress echo data. The compressed echo data will be transferred through an Ethernet cable in real-time.

The possibility analysis on the application of data compression technique to simplifying ultrasound imaging system is presented in Section 2.2. In Section 2.2, details of JPEG algorithm is also given. Section 2.3 shows the in vitro experiment which is conducted to preliminarily study the feasibility of the proposed idea. Discussion and conclusion are given in Section 2.4 and 2.5 respectively.

2.2 Preliminary Analysis

In clinics, ultrasound imaging systems are equipped with 1D array transducer, containing 128 elements. This type of transducer will be taken as the example for the feasibility analysis and the following study. To make the quality of 3D ultrasound images comparable to that of 2D imaging, full 2D array transducers with the size of 128 by 128 are required. Apparently, the transducer element number of a 2D array transducer is increased by 128 times than that of 1D array transducer. This increase results in not only the 128 times’ increase of the cable which connects an ultrasound probe and an image processing system, but also the 128 times’ increase of the echo data volume for a 3D ultrasound imaging system. These increases for the diameter of connecting cable and echo data volumes are disasters for real-time 3D ultrasound imaging. In order to solve these problems, the quantity of wires inside the cable connecting an ultrasound probe and image processing system is desired to be firstly reduced to an acceptable number. A novel ultrasound probe model was proposed (see Figure 2-1), in which data buffers and data compression units are built for 128 groups of 128 transducer elements inside an ultrasound probe. Echo data acquired by a receiving transducer is compressed in data
compression units. In this way, echo data volume is reduced greatly, and an acceptable size of cables connecting ultrasound probe and image processing system is achievable.

Figure 2-1: The schematic of the proposed ultrasound probe. “Amp” represents time-gated amplifiers. “A/D” represents the converters from analog to digital signals. “Data Buffer”, “Data Compression”, and “Data Transferring” are the core parts for the new model for an ultrasound probe, and they are used for buffering echo data, compressing echo data, and transferring compressed echo data, respectively.

One 2D B-mode ultrasound image is reconstructed from multiple echo data slices acquired at multiple steering angles. One echo data slice consists of 128 A-lines. Each slice is directly viewed as an image (see Figure 2-5), and then a lossy compression scheme, JPEG, is used to compress the slice of echo image.
Figure 2-2: The example of four slices of echo data acquired at four different steering angles. Each slice of echo data contains 128 A-lines.

The procedures of JPEG algorithm are given as follows:

- Firstly, the image is divided to be many blocks of data with the size of N*N. Usually, the N is taken as 8 for the usage of FFT.
Secondly, In N×N data blocks, the discrete cosine transform (DCT), which is the core component in JPEG algorithm, transforms the image signals from the spatial domain into the frequency domain. The formula of DCT is given as the following:

\[
F(i, j) = \frac{1}{\sqrt{2N}} C(i)C(j) \sum_{x=0}^{N-1} \sum_{y=0}^{N-1} \text{pixel}(x, y) \cos \left( \frac{(2x + 1)i\pi}{2N} \right) \cos \left( \frac{(2y + 1)i\pi}{2N} \right) \tag{2.1}
\]

In the equation (2.1), \( C(x) = \frac{1}{\sqrt{2}} \) if \( x = 0 \), and \( C(x) = 1 \) if \( x \) is larger than 0.

Thirdly, Quantum data matrix with the size of N×N is designed, in which all the values are between zero and the maximum pixel value of an image. When the data in the quantum data array is further away from the up-left corner value at (0, 0), it becomes less important to the image, so the quantum value becomes larger. The quantum data matrix is used to divide the values in N×N data blocks after the DCT (see equation (2.2)), and all resulted values are rounded to be integers. In effect, this processing will result in many high frequency components to be zeros. The same array of quantum values is used to recover the values in frequency domain during the data compression recovering stage. This step is called the quantization, which is the only step where data is lossy.

\[
\text{Quantized}\text{ }\text{value}(i, j) = \frac{F(i, j)}{\text{Quantum}(i, j)} \tag{2.2}
\]

Fourthly, in the quantized 8×8 data block, data is compressed by the Run-Length Encoding (RLE) algorithm in the zig-zag sequence from the up-left corner. The encoded data is then further encoded by Huffman coding algorithm. This method
of combining the RLE and Huffman coding algorithms is called entropy encoding. At this time, the data is fully compressed and ready for transferring.

- Fifthly, after the compressed data is transferred to a new location, it is time to recover the image. The encoded data is decoded. The quantum matrix used in compression processing is also used to recover the quantized data. It is worth mentioning that the recovered data is not the same as the original data due to the error introduced during the quantization. Inverse DCT ((2.3)) is then applied to the data which is recovered from the quantization.

\[
pixel(x, y) = \frac{1}{\sqrt{2N}} C(i)C(j) \sum_{i=0}^{N-1} \sum_{j=0}^{N-1} F(i, j) \cos \left( \frac{(2x+1)i\pi}{2N} \right) \cos \left( \frac{(2y+1)i\pi}{2N} \right) \tag{2.3}\]

In Equation (2.3), \(C(x) = \frac{1}{\sqrt{2}}\) if \(x\) is 0, and \(C(x)=1\) if \(x\) is larger than 0.

### 2.3 Experiments and Results

By observing the ultrasound B-mode images, the background may have big similarity (see Figure 2-2). In the fourth step of JPEG data compression algorithm, tremendous high frequency components containing small information become zero and are discarded in the RLE, so the compression of echo data is achieved. In this study, only the echo data of 2D image from the phantom in Figure 2-3 is used as the analyzed subject. Only 11 slices from 91 slices of echo data with the steering angle stepwise of 9 degrees are used for the image reconstruction, which are shown in Figure 2-4. The reconstructed B-mode image is shown in Figure 2-5, which covers the view field of 90 degrees and has
the imaging depth of 120 mm. This image is reconstructed with the HFR imaging method and it is reconstructed from echo data without the data compression.

Figure 2-3: The ATS 539 tissue mimicking phantom and the imaging area for the images used in the data compression analysis. The imaging area is indicated by red dashed frame, and its size is 153.6 mm by 120 mm.
Figure 2-4: The 11 slices of echo data without the addition of noise.
Figure 2-5: B-mode image of Phantom ATS539 reconstructed with the HFR imaging method. Even though 91 transmissions are sent during data acquisition, only echo data from 11 transmissions were used for the image reconstruction.

Data compression algorithm is applied to those 11 slices of echo data images shown in Figure 2-4. For the data compression with JPEG method used in this study, the block size (N) is set to be 16. The quantum matrix is select by following the formula (2.4).

\[ Quantum(i, j) = 1 + (1 + i + j) \times quality\_factor \]  \hspace{1cm} (2.4)

The quality factor in (2.4) was set to be 50. The larger quality factor will result in higher data compression ratio, however, the image quality become poorer. These values of the
block size and the quality factor are the optimum empirical ones for the images reconstructed by the HFR imaging method. The compressed echo data is shown in Figure 2-6 and it is not much different from the original echo data without being processed by the data compression. The B-mode image reconstructed from the compressed echo data is shown in Figure 2-7.

Figure 2-6: The 11 slices of echo data after data compression with the JPEG algorithm.
Figure 2-7: The B-mode image reconstructed from the 11 slices of echo data after being processed through the data compression technique.

By comparing Figure 2-5 and Figure 2-7 it can be found that the degradation of image quality after data compression is slight from the judgment by human eyes. The resulted compression ratios of those 11 slices of echo data are presented in Table 2.1. The average compression ratio is calculated as 94.50%, which means an echo data volume of 100M bits can be reduced to be only 5.5M bits.
Table 2.1: Compression ratio of 11 slices of echo data.

<table>
<thead>
<tr>
<th>Slice NO.</th>
<th>0</th>
<th>9</th>
<th>18</th>
<th>27</th>
<th>36</th>
<th>45</th>
</tr>
</thead>
<tbody>
<tr>
<td>Compression Ratio (%)</td>
<td>94.47</td>
<td>94.38</td>
<td>94.29</td>
<td>94.41</td>
<td>94.43</td>
<td>94.46</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Slice NO.</th>
<th>54</th>
<th>63</th>
<th>72</th>
<th>81</th>
<th>90</th>
<th>Average.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Compression Ratio (%)</td>
<td>94.52</td>
<td>94.49</td>
<td>94.48</td>
<td>94.66</td>
<td>94.87</td>
<td>94.50</td>
</tr>
</tbody>
</table>

2.4 Discussion

From the results shown in the Section 2.3, it is seen that the image quality is retained even though high compression ratio is achieve (94.50%). For 3D ultrasound imaging using the HFR imaging system with a full 2D array transducer [30], the A/D converters are 12 bits, and the sampling rate is 40 MHz. The element number of a full 2D array transducer is 16384. If the imaging range is 120 mm, the echo data volume for 3D imaging is 157 MB. With the lossy data compression introduced in this study, the data volume is reduced to be about 8.6 MB. Thus, the required transferring rate of total cables is reduced from 3932 Gb/s to 216 Gb/s. If the cable connecting an ultrasound probe and image processing system with 1 Gb/s is utilized, the number of required cables will be only 216, which could be feasible with the existing technology. Apparently, the results give very strong support for the hypothesis of the novel probe model.
2.5 Conclusion

A new model of probe (Figure 2-1) is proposed in this work. This probe is equipped with data buffers which are used for echo data collection over a certain period, and data compression units which are used for echo data compression. This type of probe is potential to be used for the future ultrasound imaging system for 3D imaging, especially, the HFR imaging system because real-time 3D imaging systems are more easily realizable after that the square-wave weighting function is applied.
Chapter 3

Quantitative Study of Effects of Phase Aberration and Noise on the High Frame Rate Imaging

3.1 Introduction

As introduced in Chapter 1, human tissues have inhomogenous speed of sound, which causes phase aberration to ultrasound beams and thus distorts images. Electrical noise of ultrasound imaging systems cannot be completely removed, which degrades image quality. To compare the effects of phase aberration and noise on the HFR imaging method and the D&S imaging method, limited investigations that only studied experimentally the effects of phase aberration on grayscale objects and studied qualitatively the effects of noise on a point target with a computer simulation were conducted in 2000 for a single plane-wave transmission [24, 25]. In 2007, the effects of phase aberration and noise on the extended HFR imaging method were qualitatively studied with an experiment and compared with the D&S imaging method [48]. This study, however, is mainly focused on the contrast of a cystic target of an ATS539 tissue-mimicking phantom.
Since biological soft tissues can be modeled as a superposition of multiple single point scatterers, it is necessary to quantitatively study the effects of phase aberration and noise on images of individual point scatterers for the HFR imaging method with traditional parameters such as image resolution and a set of new parameters that are to be introduced below.

In this chapter, detailed explanation on the newly proposed set of parameters that are used to assess the quality of ultrasound B-mode images will be given in Section 3.2. A process to add phase aberration and noise into simulated and experimentally obtained images will also be given in this section. Simulation of an object containing a total of 8 point scatterers in axial and lateral directions and its results are given in Section 3.3. Conditions for In-vitro experiments based on a modified AIUM 100-mm standard test object and the experiment results will be given in Section 3.4. In Sections 3.5 and 3.6, a discussion and a conclusion will be presented respectively.

3.2 Parameters and Conditions

3.2.1 Parameters for Assessing Quality of Images

Instead of assessing image quality qualitatively by the human eyes, a set of parameters is proposed in this study to quantitatively measure the image quality for the effects of phase aberration and noise on the HFR and the D&S imaging methods. The set of parameters include the -6-dB lateral resolution, the ratio of sidelobe energy to mainlobe energy (energy ratio or short for ER), and the ratio of maximum sidelobe peak value to mainlobe peak value (sidelobe ratio or short for SR). It is important to know that the proposed set of parameters is focused on the point spread function (PSF) of imaging
methods since a quantitative analysis of image contrast based on images of a cyst target of an ATS 539 tissue-mimicking phantom has already been studied [24]. Moreover, only the maximum envelope of the PSF along the lateral direction is studied because the -6 dB resolution, ER, and SR along axial direction depends mainly on the bandwidth of the transducer used. A detailed explanation of the -6-dB lateral resolution, ER, and SR is given below.

Resolution of an image is defined as the minimum distance between which two point scatterers can be distinguished in the image. Therefore, resolution is an important parameter to assess image quality. In ultrasound imaging, it is desired to have a high resolution for more accurate diagnoses of diseases. In this study, the -6 dB lateral beam width of the maximum envelope of the PSF of the imaging methods will be used as one of the parameters to quantitatively measure the quality of ultrasound images. A -6 dB lateral beam width is defined as the lateral distance between two points whose values are half of that of the peak of the mainlobe in a plot of the maximum envelope of the PSF over the lateral direction (see Figure 3-1(c)). A smaller -6-dB lateral beam width corresponds to a higher lateral image resolution or a high image quality, i.e., the lateral image resolution is inversely proportional to the lateral beam width.
Figure 3-1: Procedures for obtaining a maximum envelope plot of the point spread function (PSF) of a B-mode image of a point scatterer. An imaging area that contains 9 point scatterers is shown in (a). A magnified area that is cropped from the B-mode image is shown in (b). A plot of the maximum envelope of the PSF over the lateral distance is in (c). The vertical value of the plot is the maximum value of a column of the cropped image. The -6-dB lateral resolution, mainlobe, sidelobes, and the areas under the mainlobe and sidelobes are illustrated in the plot. The boundaries ("1" and "2") between the mainlobe and the sidelobes are determined by the intersections of the lines that are extended from the peak of the mainlobe through the -10-dB points of the mainlobe with the lateral axis.
Sidelobe is produced by edge waves of transducers and appears on both sides of
the mainlobe. It produces artifacts in ultrasound images and lowers image contrasts.
Therefore, it is necessary to assess the influence of sidelobe on image quality. The second
parameter used for image quality analysis is the energy ratio or ER, which describes the
energy distribution of the maximum envelope of the PSF between the areas under the
sidelobe and the mainlobe. The formula for calculating the ER is given in (3.1), where
\( \text{value}_{-}\text{PSF}_i \) is the value of the maximum envelope of the PSF at the \( i \)th lateral position.
A lower ER value means a higher image quality since sound energy is more concentrated
in the mainlobe of the image of a point scatterer.

\[
ER = \frac{\sum_{i} \text{value}_{-}\text{PSF}_i^2}{\sum_{j} \text{value}_{-}\text{PSF}_j^2}
\tag{3.1}
\]

As shown in Figure 3-1(c), the lateral positions of the boundaries of the mainlobe area is
determined by the intersections of the lines that are extended from the peak of the
mainlobe through the -10 dB points of the mainlobe peak on both sides of the mainlobe.
The sidelobe areas are defined as all areas that are not in the mainlobe. Apparently, ER
will increase in ultrasound imaging when there is phase aberration [99] that may cause a
split of the mainlobe, resulting in a shrunk mainlobe area and an expanded sidelobe area.

The third parameter that is used for the assessment of image quality is the
sidelobe ratio or SR. This parameter is calculated with equation (3.2) where
\( \text{sidelobe}_{-}\text{peak} \) and \( \text{mainlobe}_{-}\text{peak} \) are the peak values of the sidelobe and mainlobe,
respectively, in the plot of the maximum envelope of the PSF (see Figure 3-1 (c)). Image
quality is degraded when SR becomes larger after phase aberration or noise is introduced during an imaging process.

\[ SR = \frac{\text{sidelobe}_{\text{peak}}}{\text{mainlobe}_{\text{peak}}} \]  \hspace{1cm} (3.2)

Although the plot of the maximum envelope of the PSF over the lateral distance (see Figure 3-1 (c)) has been used to view both the mainlobe and sidelobes as in the study of effects of motion on a simulated PSF of the HFR imaging method in [100], it is not quantitative. SR provides a convenient single-parameter assessment on the size of sidelobe relative to the mainlobe.

### 3.2.2 Addition of Phase Aberration and Noise

Phase aberration is caused by a local variation of speed of sound in different human tissues[99]. The most common source of phase aberration in ultrasound imaging is the fat layer that is close to an ultrasound transducer, as described in [48]. Phase aberration affects both ultrasound transmission and reception in the imaging process. In this study, phase aberration is added into the imaging process in both transmission and reception using a phase screen model. The phase screen will cause a peak-to-peak change of the phase of \(3\pi/2\) (or 0.75 wavelengths) as shown in Figure 3-2. This phase screen is selected from one of the two phase screens used in [24, 48] for an easier comparison of the results of this study with previous studies. A flow chart for the addition of phase aberration in both transmission and reception in imaging is shown in Figure 3-3(a).
Figure 3-2: A phase screen that is used to introduce phase aberration. The range of the phase shift of the phase screen is from $-3\pi/4$ to $3\pi/4$ ($-3/8$ to $3/8$ wavelengths), giving a peak-to-peak phase shift of $3\pi/2$ ($3/4$ wavelengths). The center ultrasound wavelength used for the phase screen in both the simulation and the experiment is 0.6 mm.
Figure 3-3: Flow charts for introducing (a) the phase aberration and (b) the noise into the high-frame-rate (HFR) and the delay-and-sum (D&S) imaging methods. The phase aberration is added to both transmission and reception and the noise is added only to the received radio-frequency (RF) echo signals.

Random noise is also a factor in reducing ultrasound image quality. The noise is mainly from the electrical noise of an imaging system. In this study, pseudo random noise
pattern is used so that the same pattern can be used in both simulation and experimental studies. The noise bandwidth is equal to the two-way bandwidth of the 1D array transducer used in the experiments, i.e., 58% of transducer center frequency. To achieve such a noise bandwidth, a two-way Blackman window function is used to filter the pseudo-random noise in the frequency domain. The maximum amplitude of the noise was set to be 50% of that of the entire echo data set (global maximum), which gives a signal-to-noise ratio of 6 dB. The addition of the noise is shown in the flow chart in Figure 3-3(b).

3.3 Simulation and Results

3.3.1 Simulation Conditions

In the simulation study, a total of 8 point scatterers are assumed in the imaging area (see Figure 3-4). 6 point scatterers are located at depths of 10, 30, 50, 70, 90, 110 mm, respectively, along the axial axis of the transducer, and 2 point scatterers are located at a depth of 50 mm and at lateral positions of 20 and 40 mm, respectively. Such an arrangement of point scatterers allows us to study the effects of phase aberration and noise on the quality of images at depths ranging from 10 to 110 mm and lateral positions ranging from 0 to 40 mm, covering most of the imaging area.
Figure 3-4: Imaging area that includes 8 point scatterers in the simulation study. 6 point scatterers are located at depths of 10, 30, 50, 70, 90, 110 mm, respectively, and 2 point scatterers are located on lateral positions of 20 and 40 mm, respectively, at a depth of 50 mm. The area of the final B-mode image, indicated by the dashed frame, has a width of 153.6 mm and a depth of 120 mm.

In addition, the speed of sound is assumed to be 1,450 m/s that is the same as that of the ATS539 tissue-mimicking phantom used in previous studies [24]. A 1D phased array transducer with a center frequency of 2.5 MHz, 128 elements (19.2 mm x 14 mm
aperture size), and about a 58% -6-dB two-way fractional bandwidth that is obtained by squaring the Blackman window function is also assumed (notice that the electromechanical transfer function of a real ultrasound transducer can be approximated with a Blackman window function [101]). The parameters of the transducer are similar to those of a commercial Acuson V2 probe (Acuson, Mountain View, California, USA) that was used with the experiment.

The imaging area is of a sector shape (see Figure 3-4) and has a field of view of 90 degrees consisting of 91 transmissions that are steered beams focused at a fixed depth of 70 mm (for the D&S imaging), steered plane waves (for the HFR imaging), or limited-diffraction beams (for the HFR imaging). This focal depth is chosen so that the transmit beam is focused around the middle section of the images for an optimum imaging quality when using a single focus in the D&S imaging method. Images reconstructed have a size of 153.6 mm and 120 mm in the lateral and axial directions respectively as they are shown in the rectangle in Figure 3-4 to include all point scatterers.

The simulation conditions above are chosen to be as close as possible to those of the experiment for comparison.

3.3.2 Simulation Results

Images before adding the phase aberration and the noise for the D&S, steered plane wave (SPW), and limited-diffraction-beam (LDB) imaging methods are shown in Figure 3-5(a), (b) and (c), respectively. Images after adding the phase aberration are given in Figure 3-5(d), (e) and (f), and images with the noise added are in Figure 3-5(g), (h) and (i). After adding the phase aberration, image quality is degraded dramatically as
the sidelobes around the point scatterers are increased. The noise added fills out the otherwise clear background.

Figure 3-5: Images reconstructed by the D&S (first row), steered plane wave (SPW) HFR (middle row), and limited-diffraction beam (LDB) HFR (bottom row) imaging methods from simulated echo data before adding a two-way phase aberration and the noise (left column), after adding the two-way phase aberration (middle column), and after adding the pseudo-random noise (right column). All images are log compressed at 50 dB to show details.
The -6-dB lateral resolution, ER, and SR are used to quantitatively assess image quality for all images in Figure 3-5. Because the lateral distance between two neighboring point scatterers is 20 mm (see Figure 3-4), the area for getting the maximum envelope of the PSF (see Figure 3-1(a)) is set to be 20 mm by 20 mm to avoid significant influence from neighboring point scatterers. Parameters for assessing the image quality are calculated for scatterers arranged in both the lateral and axial directions (see Figure 3-4).

### 3.3.2.1 Simulation Results for Point Scatterers along Axial Direction

The simulation results for point scatterers along the axial direction are given in Figure 6. The -6-dB lateral beam width (a smaller beam width means a higher resolution) of the images increases with the depth (see Figure 3-6(a), (d), and (g)). To show the beam width relative to that of the D&S imaging method, the differences are calculated with (3.3) and the results are given in Table 3.1(a). A negative value in the table means that the corresponding lateral resolution of the HFR imaging methods is higher than that of the D&S imaging method.

\[
beamwidth_{\text{diff}} = beamwidth_{\text{HFR}} - beamwidth_{\text{D & S}}, \quad (3.3)
\]

where \( beamwidth_{\text{HFR}} \) and \( beamwidth_{\text{D & S}} \) are the -6-dB lateral beam widths of the HFR and the D&S imaging methods respectively.
Figure 3-6: Results of the -6 dB beam width (1st row), ER (2nd row), and SR (3rd row) of the simulated images of point scatterers in Figure 3-5 along the axial direction for the D&S and the HFR imaging methods. The depths are at 10, 30, 50, 70, 90, and 110 mm. Panels (a), (b), and (c) in the left column show the results for the -6dB lateral beam width, ER, and SR, respectively, of images before adding the phase aberration and the noise. Panels (d), (e), and (f) in the middle column show the results after adding the phase aberration. Results in Panels (g), (h), and (i) are those after adding the noise. All panels in the figure contain three curves representing the results from the D&S, SPW, and the LDB imaging methods, respectively.
Table 3.1: Differences of -6dB lateral beam widths between the HFR and the D&S imaging methods for the simulated images before adding the phase aberration and the noise. Results based on Figure 3-6(a) for point scatterers along the axial direction at depths of 10, 30, 50, 70, 90, and 110 mm are shown in Panel (a), and results based on Figure 3-8(a) for point scatterers along the lateral direction at lateral distances of 0, 20, and 40 mm at a depth of 50 mm are shown in Panel (b). SPW and LDB denote the HFR imaging methods with steered plane wave and limited-diffraction-beam transmissions, respectively.

<table>
<thead>
<tr>
<th></th>
<th>10</th>
<th>30</th>
<th>50</th>
<th>70</th>
<th>90</th>
<th>110</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPW</td>
<td>-0.1550</td>
<td>-0.3080</td>
<td>-0.2375</td>
<td>0.1315</td>
<td>0.0477</td>
<td>-0.3209</td>
</tr>
<tr>
<td>LDB</td>
<td>-0.1327</td>
<td>-0.3180</td>
<td>-0.2760</td>
<td>0.0742</td>
<td>-0.0299</td>
<td>-0.4090</td>
</tr>
</tbody>
</table>

(a)

<table>
<thead>
<tr>
<th></th>
<th>0</th>
<th>20</th>
<th>40</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPW</td>
<td>-0.2375</td>
<td>-0.0994</td>
<td>-0.0244</td>
</tr>
<tr>
<td>LDB</td>
<td>-0.2760</td>
<td>-0.1352</td>
<td>0.1701</td>
</tr>
</tbody>
</table>

(b)

It is seen from Table 3.1(a) that when there are no phase aberration and noise, the -6-dB lateral resolution of the HFR imaging methods is higher than that of the D&S imaging method at all depths except at 70 mm (0.1315 mm for SPW and 0.0742 mm for LDB) and at 90 mm (0.0477 mm for SPW). This is because 70 mm is the focal depth of the D&S imaging method and thus the highest resolution is achieved at this and nearby depths. The ER and SR in Figure 6(b) and (c) are obtained without phase aberration and noise, which only have small variations over the depth.
Comparing the results of the 2nd columns with the 1st column in Figure 3-6, it is clear that the -6-dB lateral resolution, ER, and SR become worse due to the phase aberration. After adding the noise, the results shown in the 3rd column of Figure 3-6 also become worse.

Due to the phase aberration and the noise, there are changes of the three parameters as they are compared to those without the phase aberration and the noise. To assess whether the HFR imaging methods are more resistant to the phase aberration and the noise than the D&S imaging method, the following formula is used to calculate the changes (relative_change) relative to the changes (change_for_D&S) of the D&S method for the changes (change) of an imaging method for the three parameters:

\[
\text{relative\_change} = |\text{change}| - |\text{change\_for\_D\&S}|, \quad (3.4)
\]

where change and change_for_D&S are the differences of parameters after and before the addition of the phase aberration or the noise for an imaging method and the D&S imaging method, respectively. The relative changes of the -6-dB lateral beam width, ER, and SR after adding the phase aberration and the noise for the point scatterers along the axial direction are shown in Figure 3-7. Apparently, if \( \text{change} = \text{change\_for\_D\&S} \), relative_change \( \equiv 0 \). This produces horizontal lines at value of 0 in Figure 3-7.
Figure 3-7: Relative changes of the absolute changes of the -6-dB lateral beam width (Panels (a) and (d)), ER (Panels (b) and (e)), and SR (Panels (c) and (f)) after adding the phase aberration (1st row) and the noise (2nd row) at depths ranging from 10 to 110 mm for the D&S imaging and the SPW and LDB HFR imaging methods for the simulated data in Figure 3-6. The absolute changes of the D&S imaging method are used as references and thus their relative changes are all zeros.
Table 3.2: Relative changes and their averages (last columns) of the -6dB lateral beam width, ER, and SR at 6 depths of 10, 30, 50, 70, 90, and 110 mm for the HFR imaging methods based on the simulated data in Figure 3-6. The relative changes of the absolute changes (see (3.4)) of the three parameters after adding the phase aberration and the noise are shown in Panels (a) and (b), respectively. The absolute changes of the D&S imaging method are used as references in calculating the relative changes.

<table>
<thead>
<tr>
<th></th>
<th>10</th>
<th>30</th>
<th>50</th>
<th>70</th>
<th>90</th>
<th>110</th>
<th>Average</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPW_Width</td>
<td>-0.2816</td>
<td>-0.0039</td>
<td>-0.0919</td>
<td>-0.0216</td>
<td>-0.0673</td>
<td>-0.1522</td>
<td>-0.1031</td>
</tr>
<tr>
<td>SPW_ER</td>
<td>0.1541</td>
<td>-1.1024</td>
<td>-0.0457</td>
<td>0.0127</td>
<td>-0.0172</td>
<td>-0.0472</td>
<td>-0.1743</td>
</tr>
<tr>
<td>SPW_SR</td>
<td>-0.2029</td>
<td>-0.2233</td>
<td>-0.0367</td>
<td>0.0066</td>
<td>-0.0345</td>
<td>-0.0854</td>
<td>-0.0960</td>
</tr>
<tr>
<td>LDB_Width</td>
<td>-0.2629</td>
<td>-0.0021</td>
<td>-0.1181</td>
<td>-0.0063</td>
<td>-0.0495</td>
<td>-0.1402</td>
<td>-0.0965</td>
</tr>
<tr>
<td>LDB_ER</td>
<td>0.3659</td>
<td>-1.0466</td>
<td>0.0049</td>
<td>0.0500</td>
<td>0.0239</td>
<td>-0.0113</td>
<td>-0.1022</td>
</tr>
<tr>
<td>LDB_SR</td>
<td>-0.1429</td>
<td>-0.1858</td>
<td>0.0247</td>
<td>0.0893</td>
<td>0.0526</td>
<td>0.0219</td>
<td>-0.0267</td>
</tr>
</tbody>
</table>

(a) Add phase aberration

<table>
<thead>
<tr>
<th></th>
<th>10</th>
<th>30</th>
<th>50</th>
<th>70</th>
<th>90</th>
<th>110</th>
<th>Average</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPW_Width</td>
<td>-0.0060</td>
<td>-0.0054</td>
<td>-0.0523</td>
<td>-0.0034</td>
<td>-0.0221</td>
<td>0.0023</td>
<td>-0.0145</td>
</tr>
<tr>
<td>SPW_ER</td>
<td>-0.0469</td>
<td>-0.0230</td>
<td>-0.0159</td>
<td>-0.0029</td>
<td>-0.0018</td>
<td>-0.0013</td>
<td>-0.0153</td>
</tr>
<tr>
<td>SPW_SR</td>
<td>-0.0059</td>
<td>-0.0016</td>
<td>0.0027</td>
<td>-0.0025</td>
<td>-0.0142</td>
<td>-3.4900e-04</td>
<td>-0.0036</td>
</tr>
<tr>
<td>LDB_Width</td>
<td>-0.0067</td>
<td>-0.0021</td>
<td>-0.0462</td>
<td>-0.0028</td>
<td>0.0059</td>
<td>0.0203</td>
<td>-0.0053</td>
</tr>
<tr>
<td>LDB_ER</td>
<td>-0.0462</td>
<td>-0.0223</td>
<td>-0.0129</td>
<td>-1.6000e-04</td>
<td>7.6000e-05</td>
<td>5.9300e-04</td>
<td>-0.0135</td>
</tr>
<tr>
<td>LDB_SR</td>
<td>-0.0051</td>
<td>0.0071</td>
<td>-0.0040</td>
<td>-0.0078</td>
<td>-0.0134</td>
<td>0.0050</td>
<td>-0.0030</td>
</tr>
</tbody>
</table>

(b) Add noise

The relative changes and their averages for the three parameters over all depths are given in Table 3.2. From both Table 3.2 and Figure 3-7, it is clear that most of the relative changes are negative and the averages of the relative changes are all negative. This demonstrates that the HFR imaging methods are less susceptible to the phase
aberration and the noise than the D&S imaging method. In addition, the noise has less influence on the quality of images than the phase aberration (see Table 3.2(b)).

### 3.3.2.2 Simulation Results for Point Scatterers along Lateral Direction

The results of the -6-dB lateral resolution, ER, and SR for point scatterers along the lateral direction at a depth of 50 mm and lateral positions of 0, 20, and 40 mm are given in Figure 3-8. Comparing the 2\textsuperscript{nd} and 3\textsuperscript{rd} columns with the 1\textsuperscript{st} on corresponding rows, it is clear that the quality of image becomes worse as either the phase aberration or the noise is introduced. Table 3.1(b) shows the differences of the -6-dB lateral beam widths relative to that of the D&S imaging method (using equation (3.3)) for the plots in Figure 3-8(a). Values (0.1701 mm for LDB and -0.0244 mm for SPW) in the Table 3.1(b) are only positive or close to 0 for the point scatterer at 40 mm lateral distance that is near the edge. This is because fewer images are superposed near the edge of the image for the HFR imaging methods[28].
Figure 3-8: This figure is the same as Figure 3-6 except that it is for the three point scatterers located at lateral distances of 0, 20, and 40 mm at a depth of 50 mm.

The relative changes of the -6-dB lateral beam width, ER, and SR after adding the phase aberration and after adding the noise for the point scatterers along the lateral
direction are shown in Figure 3-9. The average relative changes for the three parameters over the three lateral positions at 0, 20, and 40 mm are given in Table 3.3. From Table 3.3, it is seen that the average relative changes over the three lateral positions has a maximum absolute value of 0.0454. This means that the overall effects of the phase aberration and the noise in the lateral direction are not as much as those for the point scatterers in the axial direction.

Figure 3-9: This figure is the same as Figure 3-7 except that it is for the three point scatterers located at lateral distances of 0, 20, and 40 mm at a depth of 50 mm.
Table 3.3: This table is the same as Table 3.2 except that it is for the three point scatterers located at lateral distances of 0, 20, and 40 mm at a depth of 50 mm.

<table>
<thead>
<tr>
<th></th>
<th>0</th>
<th>20</th>
<th>40</th>
<th>Average</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPW_Width</td>
<td>-0.0919</td>
<td>0.0659</td>
<td>0.1247</td>
<td>0.0329</td>
</tr>
<tr>
<td>SPW_ER</td>
<td>-0.0457</td>
<td>-0.0174</td>
<td>-0.0136</td>
<td>-0.0256</td>
</tr>
<tr>
<td>SPW_SR</td>
<td>-0.0367</td>
<td>-0.0380</td>
<td>-0.0025</td>
<td>-0.0257</td>
</tr>
<tr>
<td>LDB_Width</td>
<td>-0.1181</td>
<td>-0.0059</td>
<td>0.0607</td>
<td>-0.0211</td>
</tr>
<tr>
<td>LDB_ER</td>
<td>0.0049</td>
<td>0.0454</td>
<td>0.0190</td>
<td>0.0231</td>
</tr>
<tr>
<td>LDB_SR</td>
<td>0.0247</td>
<td>0.0413</td>
<td>0.0702</td>
<td>0.0454</td>
</tr>
</tbody>
</table>

(a) Add phase aberration

<table>
<thead>
<tr>
<th></th>
<th>0</th>
<th>20</th>
<th>40</th>
<th>Average</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPW_Width</td>
<td>-0.0523</td>
<td>-0.0234</td>
<td>0.0082</td>
<td>-0.0225</td>
</tr>
<tr>
<td>SPW_ER</td>
<td>-0.0159</td>
<td>-0.0167</td>
<td>-0.0118</td>
<td>-0.0148</td>
</tr>
<tr>
<td>SPW_SR</td>
<td>0.0027</td>
<td>-0.0065</td>
<td>5.2000e-05</td>
<td>-0.0012</td>
</tr>
<tr>
<td>LDB_Width</td>
<td>-0.0462</td>
<td>0.0209</td>
<td>0.0590</td>
<td>0.0112</td>
</tr>
<tr>
<td>LDB_ER</td>
<td>-0.0129</td>
<td>-0.0075</td>
<td>0.0169</td>
<td>-0.0012</td>
</tr>
<tr>
<td>LDB_SR</td>
<td>-0.0040</td>
<td>-0.0065</td>
<td>0.0117</td>
<td>3.9200e-04</td>
</tr>
</tbody>
</table>

(b) Add noise

3.4 Experiment and Results

3.4.1 Experiment Conditions

To evaluate the performance of the imaging methods for data acquired from an actual imaging system, an experiment was conducted. In the experiment, a modified AIUM 100-mm standard test object was used. A homemade HFR imaging system was
used to acquire RF echo data. Details on the development and the capability of the imaging system are given in [26, 28, 98, 102]. To be consistent with the simulation, 4 nylon wires have been added to the standard AIUM 100-mm standard test object as shown in Figure 3-10. In the imaging area, there is a group of 6 point scatterers clustered together near the lower right corner of the sector area. However, these pointer scatterers are excluded from analyses because they are not assumed in the simulation. In the experiment, an Acuson V2 probe that is a 1D phased array transducer having 128 elements and 2.5-MHz center frequency was used. As mentioned before, these and other parameters in the experiment are the same as those assumed in the simulation.
Figure 3-10: A modified AIUM 100-mm test object. “1”, “2”, “3” and “4” are 4 nylon wires added to the standard AIUM 100-mm test object, but only “1” (at the depth of 10 mm) and “2” (at the depth of 30 mm) are within the imaging area for the experiment in this study. 6 point scatterers in the axial direction are located at depths of 10, 30, 50, 70, 90, and 110 mm, respectively, and 2 point scatterers in the lateral direction are located at lateral positions of 20 and 40 mm, respectively. Point scatterers that are clustered near the bottom-right corner of the fan-shaped area are not used in the study since they are not assumed in the simulation. The dashed rectangle gives an area of final B-mode images, which has a size of 153.6 mm in width and 120 mm in depth.
3.4.2 Experiment Results

Images obtained from the experiment without adding the phase aberration and the noise, after adding the phase aberration, and after adding the pseudo-random noise are shown in Figure 3-11. It is clear that the image quality is decreased due to the addition of the phase aberration and the noise. The three parameters, the -6-dB lateral beam width, ER, and SR are calculated for all images in Figure 3-11 and the results for point scatterers along the axial and lateral directions are shown in Figure 3-12, Figure 3-13 and Figure 3-14, respectively.
Figure 3-11: This figure is the same as Figure 3-5 except that it is obtained with experiment data.
Figure 3-12: This figure is the same as Figure 3-6 except that it is obtained with experiment data.
Figure 3-13: This figure is the same as Figure 3-7 except that it is obtained with experiment data.
Figure 3-14: This figure is the same as Figure 3-8 except that it is obtained with experiment data.

3.4.2.1 Experiment Results for Point Scatterers along Axial Direction

Comparing Figure 3-12(a) with Figure 3-6(a), it is seen that the -6-dB lateral beam width at the depth of 10 mm is around 2.5 mm, which is much larger than 0.4 mm
at the same depth in the simulation. This is due to the leaking of transmit pulses into the receiver amplifier at the beginning of data acquisition. As is in the simulation, the differences of the -6-dB lateral beam widths between the HFR and the D&S imaging methods can be calculated with (3.3) (see Table 4 (a)). The all-negative values in this table indicate that the resolution of the HFR imaging methods is higher than that of the D&S imaging method. This is different from the simulation and may be caused by multiple factors that may not be considered in the simulation. Without the phase aberration and the noise, ER and SR have only relatively small variations over the depth (see Figure 12(b) and (c)), which is similar to those in the simulation. Again, after adding the phase aberration and the noise, ER and SR increase significantly.

**Table 3.4: This table is the same as Table 3.1 except that it is obtained with experiment data.**

<table>
<thead>
<tr>
<th></th>
<th>10</th>
<th>30</th>
<th>50</th>
<th>70</th>
<th>90</th>
<th>110</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPW</td>
<td>-0.3342</td>
<td>-0.3701</td>
<td>-0.1953</td>
<td>-0.2069</td>
<td>-0.3356</td>
<td>-0.5376</td>
</tr>
<tr>
<td>LDB</td>
<td>-0.3460</td>
<td>-0.3736</td>
<td>-0.2344</td>
<td>-0.3103</td>
<td>-0.4584</td>
<td>-0.6869</td>
</tr>
</tbody>
</table>

(a)

<table>
<thead>
<tr>
<th></th>
<th>0</th>
<th>20</th>
<th>40</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPW</td>
<td>-0.1953</td>
<td>-0.1568</td>
<td>-0.1915</td>
</tr>
<tr>
<td>LDB</td>
<td>-0.2344</td>
<td>-0.1800</td>
<td>-0.2022</td>
</tr>
</tbody>
</table>

(b)

The relative changes (see equation (3.4) above) of the -6-dB lateral beam width, ER, and SR of the images of point scatterers along the axial direction are shown in Figure 12.
3-13. The relative changes of ER after adding the phase aberration or the noise for point scatterers close to the surface of the transducer are much smaller for the HFR imaging methods than for the D&S imaging method (see Figure 3-13(b) and (e)). This is because there are more sub-images superposed coherently near the surface of the transducer to reduce the effects of the phase aberration or the noise (it is superposed close to 91 times due to 91 transmissions).

The relative changes and their averages of the three parameters for the point scatterers over the 6 depths are listed in Table 3.5. The averages of the relative changes are almost all negative in all cases. This demonstrates that the changes caused by the phase aberration or the noise for the HFR imaging methods are smaller than those for the D&S imaging method. In other words, the HFR imaging methods are more resistant to the phase aberration and the noise than the D&S method.

**Table 3.5:** This table is the same as Table 3.2 except that it is obtained with experiment data.

<table>
<thead>
<tr>
<th></th>
<th>10</th>
<th>30</th>
<th>50</th>
<th>70</th>
<th>90</th>
<th>110</th>
<th>Average</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPW_Width</td>
<td>-0.4655</td>
<td>-0.1940</td>
<td>0.0350</td>
<td>0.0660</td>
<td>-0.0624</td>
<td>-0.0690</td>
<td>-0.1181</td>
</tr>
<tr>
<td>SPW_ER</td>
<td>-0.0418</td>
<td>-0.0590</td>
<td>-0.2132</td>
<td>-0.0747</td>
<td>-0.1047</td>
<td>-0.0795</td>
<td>-0.2288</td>
</tr>
<tr>
<td>SPW_SR</td>
<td>0.1911</td>
<td>-0.2195</td>
<td>-0.1091</td>
<td>0.0620</td>
<td>-0.0393</td>
<td>-0.0706</td>
<td>-0.0309</td>
</tr>
<tr>
<td>LDB_Width</td>
<td>-0.5264</td>
<td>-0.1807</td>
<td>0.0305</td>
<td>0.0808</td>
<td>-0.0917</td>
<td>-0.0397</td>
<td>-0.1212</td>
</tr>
<tr>
<td>LDB_ER</td>
<td>0.0420</td>
<td>-0.8243</td>
<td>-0.1814</td>
<td>-0.0120</td>
<td>-0.0440</td>
<td>-0.0615</td>
<td>-0.1802</td>
</tr>
<tr>
<td>LDB_SR</td>
<td>0.0899</td>
<td>-0.1662</td>
<td>-0.0786</td>
<td>0.1517</td>
<td>0.0617</td>
<td>-0.0380</td>
<td>0.0034</td>
</tr>
</tbody>
</table>

(a) Add phase aberration

<table>
<thead>
<tr>
<th></th>
<th>10</th>
<th>30</th>
<th>50</th>
<th>70</th>
<th>90</th>
<th>110</th>
<th>Average</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPW_Width</td>
<td>-0.2534</td>
<td>-0.0300</td>
<td>-0.0065</td>
<td>0.0037</td>
<td>-0.0186</td>
<td>-0.0081</td>
<td>-0.0521</td>
</tr>
<tr>
<td>SPW_ER</td>
<td>-0.6628</td>
<td>-0.3203</td>
<td>-0.0596</td>
<td>-0.0338</td>
<td>-0.0359</td>
<td>-0.0056</td>
<td>-0.1863</td>
</tr>
<tr>
<td>SPW_SR</td>
<td>-0.2079</td>
<td>0.0117</td>
<td>-0.0106</td>
<td>0.0050</td>
<td>0.0022</td>
<td>-0.0122</td>
<td>-0.0353</td>
</tr>
<tr>
<td>LDB_Width</td>
<td>-0.1701</td>
<td>0.0081</td>
<td>0.0099</td>
<td>0.0025</td>
<td>-0.0066</td>
<td>-0.0317</td>
<td>-0.0340</td>
</tr>
<tr>
<td>LDB_ER</td>
<td>-0.6356</td>
<td>-0.3144</td>
<td>-0.0540</td>
<td>-0.0340</td>
<td>-0.0331</td>
<td>-0.0066</td>
<td>-0.1796</td>
</tr>
<tr>
<td>LDB_SR</td>
<td>-0.2241</td>
<td>0.0036</td>
<td>-0.0041</td>
<td>-0.0017</td>
<td>0.0062</td>
<td>-0.0062</td>
<td>-0.0377</td>
</tr>
</tbody>
</table>

(b) Add noise
3.4.2.2 Experiment Results for Point Scatterers along Lateral Direction

The -6-dB lateral beam width, ER, and SR of the images of the point scatterers at the depth of 50 mm and at lateral positions of 0, 20, and 40 mm in Figure 3-11 are given in Figure 3-14. The differences of the -6-dB beam widths between the HFR and the D&S imaging methods are calculated from Figure 3-14(a) with (3.3) and are shown in Table 3.4(b). The all-negative values means that the HFR imaging methods have a higher lateral resolution than the D&S imaging method at all lateral positions. In addition, the ER and SR values for the D&S imaging method are also generally larger than those of the HFR imaging methods (see the 2nd and the 3rd rows of Figure 3-14).

Figure 3-15 shows the relative changes of the -6-dB lateral beam width, ER, and SR for point scatterers along the lateral direction after the addition of the phase aberration and the noise. The relative changes and their averages of the three parameters for the point scatterers at 0, 20, and 40 mm are given in Table 6. The average relative changes in the table are almost all negative, indicating that the HFR imaging methods are more resistant to the phase aberration and the noise.
Figure 3-15: This figure is the same as Figure 3-9 except that it is obtained with experiment data.
Table 3.6: This table is the same as Table 3.3 except that it is obtained with experiment data.

<table>
<thead>
<tr>
<th></th>
<th>0</th>
<th>20</th>
<th>40</th>
<th>Average</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPW_Width</td>
<td>0.0350</td>
<td>-0.1166</td>
<td>-0.0106</td>
<td>-0.0307</td>
</tr>
<tr>
<td>SPW_ER</td>
<td>-0.2132</td>
<td>-0.0617</td>
<td>-0.0907</td>
<td>-0.1219</td>
</tr>
<tr>
<td>SPW_SR</td>
<td>-0.1091</td>
<td>0.0046</td>
<td>-0.0235</td>
<td>-0.0457</td>
</tr>
<tr>
<td>LDB_Width</td>
<td>0.0305</td>
<td>-0.0932</td>
<td>0.0389</td>
<td>-0.0079</td>
</tr>
<tr>
<td>LDB_ER</td>
<td>-0.1814</td>
<td>-0.0415</td>
<td>-0.0341</td>
<td>-0.0857</td>
</tr>
<tr>
<td>LDB_SR</td>
<td>0.0786</td>
<td>0.0585</td>
<td>0.0654</td>
<td>0.0151</td>
</tr>
</tbody>
</table>

(a) Add phase aberration

<table>
<thead>
<tr>
<th></th>
<th>0</th>
<th>20</th>
<th>40</th>
<th>Average</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPW_Width</td>
<td>-0.0065</td>
<td>-0.1329</td>
<td>-0.0663</td>
<td>-0.0665</td>
</tr>
<tr>
<td>SPW_ER</td>
<td>-0.0596</td>
<td>-0.0515</td>
<td>-0.0695</td>
<td>-0.0602</td>
</tr>
<tr>
<td>SPW_SR</td>
<td>-0.0106</td>
<td>-0.0415</td>
<td>-0.0033</td>
<td>-0.0185</td>
</tr>
<tr>
<td>LDB_Width</td>
<td>0.0099</td>
<td>-0.1311</td>
<td>-0.0297</td>
<td>-0.0503</td>
</tr>
<tr>
<td>LDB_ER</td>
<td>-0.0540</td>
<td>-0.0485</td>
<td>-0.0582</td>
<td>-0.0536</td>
</tr>
<tr>
<td>LDB_SR</td>
<td>-0.0041</td>
<td>-0.0272</td>
<td>-0.0152</td>
<td>-0.0155</td>
</tr>
</tbody>
</table>

(b) Add noise

3.5 Discussion

From the results of the simulation and the experiment in Sections 3.3 and 3.4, it is clear that the set of parameters (i.e., the -6-dB lateral resolution, ER, and SR) can be used to quantitatively assess the effects of phase aberration and pseudo-random noise on imaging methods. From Table 3.4, it is clear that in the experiment, the HFR imaging methods have higher -6-dB lateral resolutions at all depths and all lateral positions than the D&S imaging method.
The results in Figure 3-7 (simulation) and Figure 3-13 (experiment) show that the image quality of point scatterers near the surface of the transducer is more susceptible to phase aberration and noise than at deeper depths. In general, curves of the HFR imaging methods in Figure 3-7 and Figure 3-13 are smaller than zero. This means that the HFR imaging methods are affected less by the phase aberration and the noise than the D&S imaging method. This is also supported by Tables 2 and 5 where the average relative changes have mostly negative values.

As for the results from the lateral direction (see Figure 3-9 and Figure 3-15, and Table 3.3 and Table 3.6), the experiment produced mostly negative average relative change values. This means that the experiment demonstrates that the HFR imaging methods are less susceptible to the phase aberration and the noise than the D&S imaging method.

3.6 Conclusion

This study uses a set of parameters to assess image quality. These parameters include the traditional -6-dB lateral resolution, and the newly developed parameters ER and SR. From the study, it is found that the HFR imaging methods have higher lateral image resolution than the D&S imaging method in the experiment and this is also true in the simulation except at the depths near the focus of the transmit beams of the D&S imaging method. The results also show that the HFR imaging methods are less affected by phase aberration and noise. These results are consistent with those obtained from previous studies. However, the studies carried out in this study are more comprehensive covering image depths from 10 to 110 mm and lateral positions from 0 to 40 mm at a depth of 50 mm. In addition, the newly developed parameters are useful not only in the
studies here, but also can be used to quantitatively assess quality of other imaging methods.

It is worth noting that although the HFR imaging method is promising for clinical use, commercialization of this technology may require a fundamental change to the traditional D&S beamforming architecture that has dominated the market over the past few decades and need a large capital investment. Unless it is significantly profitable, commercial companies may be reluctant to make such a transition. However, as the technologies such as microelectronics advance and thus the power consumption and costs of electronics are lowered, the HFR imaging method will be more attractive to commercial vendors.
Chapter 4

Strain and Strain Imaging Using High Frame Rate Imaging Method

4.1 Introduction

As discussed in Chapter 1, strain and strain rate can provide important diagnosis information about the tissue deformation and deformation rate, especially for heart disease diagnosis and lesion detection. Tissue Doppler technique is one of the most prevalent methods. However, with this technique the strain rate is ignored in lateral direction, which is also informative. Even though at present speckle tracking technique is applied to B-mode images to measure strain and strain rate, the accuracy of their results are low because of the skewed images generated by D&S imaging method. This problem exists when multiple transmissions are used to obtain one B-mode image. To overcome these problems of the angle dependent limitation for the Tissue Doppler technique and the skewed objects in B-mode images obtained with D&S, 2D speckle tracking is combined with the HFR imaging method. The HFR imaging method utilizes only one transmission. With only one transmission, the frame rate can achieve as high as about 6000 frames per second if imaging depth is at 120 mm.
A new type of kernel used in the speckle tracking technique is proposed as well. The new kernel uses two crossing diagonal lines instead of a full kernel to search the moved kernel block. The preliminary study has been finished and presented in this chapter.

Both simulation and experiment are performed to verify the new method. Because D&S with multiple transmissions have obvious disadvantages of skewing problem and low detectable velocity for moving objects due to long echo acquisition time, the comparison study here is restricted only to D&S with one transmission. To obtain the image using one transmission for D&S method, dynamic focusing is applied to all points throughout imaging area. This study is organized as follows. Theoretical preliminary is given in Section 4.2. Sections 4.3 and 4.4 give the simulation and experimental designs and results, respectively. Section 4.5 gives the preliminary study of the measurement of strain and strain rate with two crossing diagonal lines. The testing results, based on the same data sets as those used in the strain and strain rate measurement with the full kernel, are also shown in Section 4.5. Discussion and conclusion are given in Section 4.6 and 4.7, respectively.

4.2 Theoretical preliminary

4.2.1 Speckle tracking technique

For speckle tracking technique, a 2D kernel region including interested speckles at the center is first defined as a rectangular region (see Figure 4-1). Height and width of the kernel region depend on axial and lateral resolutions, assuring to include the whole interested speckles. A search region is then defined by centering the 2D kernel region.
The height and width of search region depend on the velocity of moving targeted object. After kernel and search regions are manually defined, block match method is used to find out matched speckle in the next frame of image by searching the same speckle through the whole search region. SAD given in (4.1) is utilized as a similarity measure for block matching.

\[
SAD = \sum_{i=0}^{N-1} \sum_{j=0}^{M-1} (x_{i,j} - x_{(l+i)(m+j)})
\]  

(4.1)

Where \(x_{i,j}\) and is pixel value in the 2D kernel of the image with stationary speckles; \(x_{(l+i)(m+j)}\) is the corresponding pixel value in a searching block of search region, with \(l\) and \(m\) indicating the axial and lateral shifts; \(N\) and \(M\) are the height and width of 2D kernel region that are 32 and 200 respectively. The height and width of search region are 1200 and 800 respectively. The height and width of images analyzed in simulation and experiment studies are 6666 and 1066, respectively.
Figure 4-1: Block matching procedure for speckle tracking in two consecutive images.

After searching the whole search region in the consecutive image with block matching algorithm, the block in next image having smallest SAD is taken as the moved 2D kernel including interested speckle. The coordinate values of both original and moved speckles’ centers are recorded for the calculation of strain and strain rate.
4.2.2 Strain rate calculation

Strain and strain rate give the respective deformation and deformation rate of a material, which is human tissue here. The strain calculation is given by (4.2) and illustrated in Figure 4-2.

\[ \varepsilon = \frac{L - L_0}{L_0} = \frac{\Delta L}{L_0} \]  \hspace{1cm} (4.2)

Where \( \varepsilon \) denotes strain; \( L \) is the material length after deformation; \( L_0 \) is the original material length; \( \Delta L \) is the length change.

Figure 4-2: Length change of a bar for strain and strain rate calculation.

A tissue segment can be either shortened or elongated, causing the strain to be negative and positive respectively. There is no unit for strain value. From strain value, strain rate can be derived by (4.3). So once the strain curve is drawn, strain rate curve can be easily obtained by differentiating strain value with respect to time. Strain rate means the deformation rate of a material, which if applied to assessment of human tissue, gives the information of whether the tissue is still in good condition or not functioning normally.
\[ \xi = \frac{de}{dt} \] 

(4.3)

In this formula, \( \xi \) denotes strain rate; \( t \) is time.

4.3 Simulation study and results

4.3.1 Simulation study conditions

Simulation study is conducted to first verify the new method which is compared with speckle tracking based on D&S with single transmission. Plane wave is utilized in the simulation because only one transmission is applied for imaging and all the imaging area needs be “illuminated” by ultrasound wave. One thing needs to be noticed is that limited diffraction beam is the same as plane wave when using only one transmission along the axial direction, without steering the transmission beam [19, 98]. Two point scatterers are assumed at two different depths in the simulation study. They are initially at (0, 15 mm) and (-4.0, 70.0 mm). The reason for selecting two point scatterers at the depths of 15 mm and 70 mm are that we want to mimic the measurement of longitudinal strain and strain rate of the heart tissue. The two point scatterers are assumed at two ends of a segment of tissue. The imaging area and point scatterer positions are given in Figure 4-3. The final reconstructed image has a width equal to 19.2 mm which is the width of transducer, and the depth is selected as 120 mm which is the usual imaging range. 1D phased array transducer is assumed for simulation study. It has 2.5 MHz center frequency, about 58\% of two-way fractional bandwidth, and 128 elements (19.2 mm by 14 mm aperture size).
Figure 4-3: Imaging area in simulation study and the positions of two point scatterers. Point scatterer 1 moves along the axis z at the speed of 0.5 m/s. Point scatterer 2 moves at the speed of 1.0 m/s.

As shown in Figure 4-3, point scatterer 1 is assumed moving at a constant speed of 0.5 m/s along the axial direction, and point scatterer 2 is assumed moving at a constant speed of 1.0 m/s along the direction rotating counterclockwise 30 degrees from axial direction. The movements of two point scatterers are virtually captured at 11 moments with time step of 0.001 s (or 1 ms) from the initial status (frame 0). Here, the image frame rate is set to be 1000 frames per second (1/0.001 s). With 1000 frames per second, the maximum detectable velocity is 121.5 m/s. Obviously, none of human tissue can move in that fast speed, so it is reasonable to assume frame rate to be 1000 frames per second.
4.3.2 Simulation results

11 consecutive echo data sets are automatically captured by HFR system. The 11 images reconstructed from dynamic focusing D&S method and from high frame imaging method are shown in Figure 4-4 (a) and (b) respectively. Image on the left most shows the initial state of two point scatterers and the trajectory of the two point scatterers are clearly seen in both Figure 4-4 (a) and (b).

![Figure 4-4](image_url)

(a) (b)

Figure 4-4: Consecutive images captured each time 1 ms elapsed, which are reconstructed from simulated echo data with one transmission. In (a), images are reconstructed from the D&S method. In (b), images are reconstructed from the HFR method.

Velocities of point scatterer 1 and 2 for 10 image frames at 10 image-capturing moments except starting moment are calculated by dividing the displacement between the original and new position over the time span. Results based on images from D&S and HFR methods are given in Figure 4-5. Velocity estimation error is also calculated based
on the assumption that speckles 1 and 2 moving constantly at 0.5 m/s and 1.0 m/s respectively. Measured strain is given in Figure 4-6. By calculating the derivative of strain with respect to time, estimated strain rate is given in Figure 4-7.

![Estimated Velocity](image)

(a) Estimated Velocity (b) Estimated Velocity

**Figure 4-5:** Measured velocities and velocity errors of point scatterer (speckle) 1 and 2 in simulation study. The exact moving velocities for Speckle 1 and 2 are 0.5 and 1 m/s respectively. Figure (a) gives the velocity based on images from D&S method. Figure (b) gives the velocity based on images from HFR method.
Figure 4-6: Estimated 2D strain based on (a) D&S method and (b) HFR method in simulation study.

Figure 4-7: Estimated 2D strain rate based on (a) D&S method and (b) HFR method in simulation study.
4.4 Experimental study

4.4.1 Experiment conditions

To further verify the new method of measuring strain and strain rate, experiments are designed. Homemade HFR imaging system is used to acquire echo data [98]. Two glass beads are immersed into water in a water tank to mimic two point scatterers. The experiment settings are shown in Figure 4-8. The transducer is tilted at 30 degrees to get the oblique movement of point scatterer because the motor controlling glass bead movement can only move in horizontal direction.

![Figure 4-8: Overall setting for experimental study. Figure (a) shows the controlling motor, tilted transducer above water tank. Figure (b) shows two glass beads immersed in water.](image)

Since it is difficult to accurately position glass beads at exact coordinates and also not important to require exact positioning, these two point scatterers are roughly positioned at...
(0, 15 mm) and (-6, 70 mm) (see Figure 4-9). Point scatterer 1 keeps stationary during capturing 11 consecutive images. Point scatterer 2 moves at an unknown constant speed $V_0$ along horizontal direction, but at a known movement step of 1 mm. The motor controlling system is shown in Figure 4-10.

Figure 4-9: Imaging area in experimental study and the positions of two point scatterers with moving direction shown. Point scatterer 1 is static during the experiment. Point scatterer 2 moves at an unknown velocity $V_0$, but its moving step size is known as 1 mm.
Figure 4-10: Motor controlling system. (a) shows the software for controlling motor movement and sending triggers for HFR system. (b) shows the counter for counting trigger number.

The motor controlling system is connected with HFR system [98]. By setting the motor movement step as 1 mm, the motor controlling system sends trigger to HFR system (see Figure 4-11), to capture moving point scatterers each time the motor moves 1 mm in horizontal direction.
4.4.2 Experiment results

Based on acquired experimental echo data via homemade high frame imaging system, 11 consecutive images reconstructed from D&S and HFR methods are shown in Figure 4-12. From the result, we can see point scatterer 1 is stationary and point scatterer 2 moves at the designed trajectory. Estimated moving velocity (V) of point scatter 2 is divided by motor moving velocity (Vo) to obtain relative moving speed which is used for assessing velocity estimation accuracy. The estimated relative moving velocity of point scatterer 2 is shown in Figure 4-13. Because point scatterer 1 is assumed stationary, its velocities at frames from 1 to 10 are all zeros. The velocity calculations are conducted only for 10 images frames except frame 0 captured the starting moment. Figure 4-14 shows the estimated strain based on D&S and HFR methods.
Figure 4-12: 11 consecutive images based on experimental echo data acquired at 11 moments when motor moves 1 mm horizontally. Only one transmission is used to generate echo. Images in (a) are reconstructed with D&S method. Images in (b) are reconstructed with HFR method.

Figure 4-13: Estimated moving step of two point scatterers (speckles) in experimental study. The exact value of relative moving velocity for Speckle 2 is 1 m/s. (a) is based on D&S method. (b) is based on HFR method.
Figure 4-14: Estimated 2D strain based on (a) D&S method and (b) HFR method in experimental study.

4.5 Preliminary Study of a New Type of Kernel

When the HFR imaging method is used for the measurement of strain and strain rate, high spatial resolution of B-mode images is required. As shown in the Section 4.2.1, the image size with the of the B-mode images generated with one transmission is 1066 by 6666. To accurately search similar kernel patterns in two consecutive images, kernels of a large size are required. The kernel with a large size will result in the long computation time, so simpler kernel is desirable to reduce the computation time. The kernel using the two crossing diagonal lines (see Figure 4-15) can help achieve this goal. In this new type of kernel, the shape of kernel is required to be square and only two diagonal lines crossing the square kernel are used in the SAD calculation. In this way, the data points used in arithmetic operations during the SAD calculation are reduced from N*N to 2*N, so the reduction of computation time is N/2 times. Apparently, this reduction of
computation time will be significant when the N becomes larger, for example, if the N is 300, the computation time will be decreased by around 150 times.

![Diagram showing the decrease in arithmetic operations](image)

**Figure 4-15:** The two crossing diagonal lines (shown as red) used in the SAD calculation. The arithmetic operations are decreased from $N^2$ times to $2N$ times.

The data sets obtained from the simulation and experimentation studies in Section 4.3 and 4.4 are also used to verify the efficacy of this new type of kernel. Additionally, the noise with the amplitude of 50% of the maximum value in a B-mode image is added to ensure the studies closer to the practical applications.

### 4.5.1 Simulation Study and Results

For the simulation study, original B-mode images without adding any noise are shown in Figure 4-4, and the B-mode images after adding noise is shown in Figure 4-16. The size of the 2D full kernel used in the SAD calculation is 100 by 100. The velocity measurements based on the new type of kernel are presented in Figure 4-17. The measurements of strain and strain rate with both the new kernel and the 2D full kernel are
presented in Figure 4-18 and Figure 4-19, respectively. Overall, the results obtained from
the HFR imaging method have higher accuracy than those obtained with the D&S
imaging method. The measurements with the 2D full kernel

![Images in figure (a) are reconstructed with the D&S imaging method. Images in figure (b) are reconstructed with the HFR imaging method.](image)

Figure 4-16: The 11 slices of B-mode images after adding noise in simulation study.

Images in figure (a) are reconstructed with the D&S imaging method.

Images in figure (b) are reconstructed with the HFR imaging method.
Figure 4-17: The velocity measurement of speckle 1 and 2 in Figure 4-3. Both the results from the images reconstructed with the HFR and D&S imaging methods are presented. Figure (a) shows the results obtained with the kernel consisting of two crossing diagonal lines, and figure (b) shows the results obtained with the full 2D kernel.
Figure 4-18: The measurement of accumulated strain with two different kernel types. Figure (a) shows the results obtained with the two crossing diagonal lines. Figure (b) presents the results calculated based on the 2D full kernel.

Figure 4-19: The estimated strain rate with two different types of kernel. They have the same format as Figure 4-18.
4.5.2 Experimentation Study and Results

The B-mode images after adding noise into the experimental B-mode images are shown in Figure 4-20, comparing to the original B-mode images in Figure 4-12.

Figure 4-20: The 11 slices of B-mode images after adding noise in experimentation study. The figure format is the same as Figure 4-16.

The same as the experimental study in Section 4.4, the measurement of the relative speed of speckle 2 (see Figure 4-9) are presented in Figure 4-21. From Figure 4-21(a), the estimation results obtained with the HFR imaging method is much higher than that with the D&S imaging method. The measurement results of accumulated strain are also presented in Figure 4-22. Since the time interval between two consecutive B-mode images is unknown, the strain rate is not calculated.
Figure 4-21: The measurement of the relative speed of speckle 2. The exact value of the moving step is 1 mm. The format of this figure is the same as Figure 4-17.

Figure 4-22: The estimated strain with two different kernels. The format of this figure is the same as Figure 4-18.
4.6 Discussion

In both simulation and experimental study, the quality of image reconstructed with the HFR method is almost the same as D&S method if only judged by human eyes. Moreover, HFR method reconstructs images in Fourier domain with fast Fourier transform, so the computation efficiency of HFR method will be superior to D&S method, especially when 3D image is reconstructed [21, 23]. In the simulation study, velocity estimation based on high frame imaging method is accurate, having 2.80% as the highest error for speckle 1 and 3.99% for speckle 2. The results are comparable with that of the conventional D&S method. In the experimental study, estimated relative moving velocity of speckle 2 has the highest error of 9.4 % for both D&S and high frame imaging methods, which means the new method combining speckle tracking with the high frame imaging method is effective to capture and assess motion.

About strain estimation in simulation study, the values are all positive because the virtual segment defined by two point scatterers are elongating based on the movement assumption. Since we assume that two speckles move at constant speeds and time interval between two consecutive images are the same, the length change of virtual tissue segment accumulate linearly. Strain estimation in experimental study has all negative values because the virtual tissue segment is shortening, which is apparently shown in Figure 9. By observing equations (4.2) and (4.3), it is easy to get that the accuracy of strain and strain rate estimation completely depends on the accuracy of displacement estimation which is directly given in experimental study and indirectly given by calculating the accuracy of velocity for simulation study. Therefore, small displacement error or velocity error ensures an accurate estimation of strain and strain rate with the proposed method.
When the two crossing diagonal lines are applied into the calculation of the SAD during the speckle tracking technique, the computation time is reduced by \((N/2-1)\). It is noticed that the accuracy of the velocity estimation with the new type of kernel both in simulation and experimentation studies is lower than that with the 2D full kernel. However, the errors are still acceptable because the estimation of strain and or strain rate in Figure 4-18, Figure 4-19 and Figure 4-22 shows the same trend between two different types of kernels. The results of velocity estimation in both simulation and experimentation studies show the consistency that the HFR imaging method results in better velocity and displacement measurements than the D&S imaging method.

4.7 Conclusion

This study proposes a new method, combining speckle tracking technique and HFR method, to measure strain and strain rate of human tissue. Through simulation and experimental studies by comparing the new method with the method based on the conventional D&S method, it is proven that the new method is effective and efficient to measure 2D strain and strain rate of fast moving objects. The snapshot of moving object in imaging area also assures object’s shape is not skewed and thus assures the accuracy of strain and strain rate imaging. If the proposed method is used for heart tissue assessment, it will provide full and accurate information of heart tissue elasticity, which is helpful for doctors. In the future, the method also has a potential for measuring blood flow velocity due to its capability of measuring quickly moving object. It is also proven that the simplification of the kernel used for the SAD calculation during the speckle tracking technique is able to save the computation time while using the HFR imaging method for
estimating strain and strain rate with speckle tracking technique. That can help ease the real-time application of strain and strain rate estimation in the HFR imaging system.
Chapter 5

Summary

This dissertation presents three different projects, which are the feasibility study of the application of data compression with the new design of an ultrasound probe for future 3D ultrasound imaging systems, the quantitative study of the effects of phase aberration and noise on the HFR imaging method, and the study of the measurement of strain and strain rate with the HFR rate imaging method.

Chapter 1 provides the basic knowledge of biomedical ultrasound imaging technology, the background about the development of the HFR imaging method, and the background of the measurement of strain and strain rate in soft tissues. Discussion of research motivation, significance and objectives are also presented in Chapter 1.

The study of data compression is presented in Chapter 2. The algorithm following the JPEG standard is used as the data compression technique. The size of blocks in images is set to be 16 instead of the standard value of 8 due to the consideration of high background similarity in ultrasound echo data. Meanwhile, the larger size of blocks will result in more efficient entropy encoding followed by higher data compression ratios. The results obtained from the study show that the new model of an ultrasound probe with the
embedding of the data compression technique is promising for future 3D ultrasound imaging systems. In the future, more ultrasound images of different human soft tissues and organs can be used to further study the feasibility of this new type of ultrasound probe design. Furthermore, the hardware based on the design schematic (Figure 2-1) can be built to test the new method.

In Chapter 3, a combination of the traditional -6-dB lateral resolution and the proposed parameters of ER and SR is used to quantitatively study the influence of phase aberration and noise on the HFR imaging method. Both the simulation and experimentation studies show that the HFR imaging method provides better lateral resolution along both the lateral and the axial directions in B-mode images. It is also found that the HFR imaging method is more resistant to (less influenced by) the phase aberration and noise than the D&S imaging method. In the future, the proposed parameters can be used to examine the ultrasound B-mode images quantitatively. For example, they can be used to measure the degradation of B-mode images after the echo data has been processed through data compression. Additionally, the parameters of ER and SR can also be used to quantitatively assess the imaging quality of any ultrasound imaging methods, and they can be used as guides for the phase aberration correction as well. A revised version of the materials presented in this chapter has been submitted to the Journal of Ultrasonics for review.

The study of the application of the HFR imaging method into the estimation of strain and strain rate of soft tissues is presented in Chapter 4. The 2D speckle tracking technique is utilized to measure the displacement and velocity of moving targets in B-mode images. The results based on images from both simulation and experimentation
studies indicate that the HFR imaging method is applicable to estimating the strain and strain rate of soft tissues. A conference paper based on this part of study has been accepted and published in the Proceedings of Meetings on Acoustics. In addition to the study of combining the HFR imaging method with the speckle tracking technique using the full 2D kernel, a new kernel using two crossing diagonal lines is proposed and studied preliminarily in Chapter 4. The results of estimating velocity, strain, and strain rate using the new type of kernel are compared with those using the full 2D kernel. It is found that the kernel using two crossing diagonal lines can be used for accurate speckle tracking and to reduce the computation time greatly. In the future, measuring strain and strain rate using the HFR imaging method can be extensively studied in in vivo experiments. The algorithm of real-time estimation of strain and strain rate can be incorporated into the existing HFR imaging system. The software of “bscan” coded with VC++ 6.0 can be used to localize the speckles of interest. The interface of “bscan” is shown in the Appendix A.

The business plan based on the researches in this dissertation is presented in Appendix C.

The publications and presentations resulted from the research work in this dissertation are listed as follows:

- Hong Chen and Jian-yu Lu, “Quantitative Assessment of Effects of Phase Aberration and Noise on High Frame Rate Imaging”, (to be published).

• Jian-yu Lu and Hong Chen, “High frame rate imaging with diverging beam transmission and Fourier reconstruction”, 2011 IEEE International Ultrasonics Symposium Proceedings (to be published).

• Hong Chen and Jian-yu Lu, “Modified Speckle Tracking Technique for Strain and Strain Rate Estimation based on High Frame Rate Imaging”, 3rd Annual Midwest Graduate Research Symposium, 2011, Toledo, OH, Oral presentation. (Abstract)

• Hong Chen, “Estimation of 2D Strain and Strain Rate with High Frame Rate Imaging Method Combined with Speckle Tracking”, 2nd Annual Midwest Graduate Research Symposium, 2011, Toledo, OH (Abstract).
References


[57] X. Wang, Y. Shen, and Z. Liu, "Noise reduction for doppler ultrasound signal based on the adapted local cosine transform and the garrote thresholding method,"


Appendix A

The “bscan” Software Interface

The interface of “bscan” is shown here. The small red dashed frame indicates the parameter setting that can be used to exactly localize the interested speckles. As shown in the red rectangular frame in the B-mode image (indicated by the big red dashed frame), the speckle of interested is flexibly positioned by clicking the left button on the computer mouse while pressing the “ctrl” button on the computer keyboard. Another application of this software is that it can be used to calculate the ER and SR of interested points.
Figure A-1: The interface of “bscan” software used for reconstructing 2D B-mode images. The contrast of a specified area and the resolution of a specified point can be calculated in this software.
Appendix B

The Simulation of Echo Collection

The procedures of generating echoes and collecting them are as follows:

- First, to simulate ultrasound imaging, one and a half sine waves, which are weighted by exponential decay function (see equation (B.1)), are used as the transmitting waves. After virtually transmitting the one and a half sine waves, the transducer is switched to reception mode to receive echo covering the range of 148.48 mm. Therefore, zeros are added to extend the signal duration to be 204.8 us, resulting in 8192 data points in this 1D signal array.

\[
f(t) = e^{-t} \cdot \sin(\omega_0 t)
\]  

(B.1)

- Second, the 8192 data elements in the 1D data array are converted to be frequency components by FFT. The transmitting sine wave is then filtered by the Blackman window which approximates the transfer function of an ultrasound transducer. The filtered signal is further delayed if the steered plane wave or a focused beam at 70 mm is applied.
Third, for a point scatterer in the space, its ultrasound field responses for all the frequency components of the filtered transmission sine wave are calculated by the Fourier transformed Rayleigh-Sommerfeld function ((1.5)). All the ultrasound field responses, at one point scatterer, from the 128 transducer elements are summed to obtain the response at that point scatterer.

Fourth, the reflection coefficients of scatterers in the imaging area are assumed to be one. The point scatterer in the imaging area is treated as a secondary ultrasound source. Using the reciprocal principle, the ultrasound field response on the surface of each transducer element from each scatterer in the imaging area can be obtained. The response of a transducer element is obtained by summing all the responses on the same transducer element from all the scatterers in the imaging area. This response is, in fact, an echo signal in the temporal Fourier domain.

Fifth, the echo signal in the temporal Fourier domain is converted to be echo signal in temporal domain using the inverse FFT.

Finally, the above five steps are repeated for all the 128 transducer elements and all the steering angles to obtain all the echo signals.

The simulated echo data can be either reconstructed by the D&S imaging method or the HFR imaging method.
Appendix C

Business Plan

C.1 Executive Summary

Conventional ultrasound imaging systems suffer from the problems of the low frame rate and complex hardware. The high frame rate (HFR) imaging system developed in ultrasound lab at the University of Toledo adopts the HFR imaging method, improving the ultrasound image quality and boosting the imaging frame rate. It is proven that the HFR imaging system provides better image quality than the conventional ultrasound imaging system. Especially, the HFR ultrasound imaging system assures the image quality in the cases of imaging fast-moving human organs, such as hearts. Additionally, the HFR ultrasound imaging system can provide the strain rate imaging, which can be used to diagnose abnormal human tissue in early phase. In short, the HFR ultrasound imaging system can outperform the current ultrasound imaging system with a lower cost.

The gross margin of one 2D HFR ultrasound imaging system is projected to be around 33%. The initial target market will be hospitals and clinics in the Lucas County of Ohio state. For the first year, 5 units are expected to be sold. The growth rate is projected to be around 60% when the business grows stable and covers all the hospitals in Ohio.
The operation plan and financing strategies have been proposed. The proposed return to investors and founders are also presented.

C.2 Opportunity Rationale

Ultrasound imaging is a non-ionizing imaging tool, and has been proven harmless to human body and even to fetus. Existing ultrasound imaging instruments are based on the conventional technology. The conventional technology requires the beamformers, which are core units in current ultrasound imaging systems. The beamformer is one of the major costs to manufacture one current ultrasound imaging system. In addition to that, large number of high-power transmitters in current imaging systems also drives the high cost of an ultrasound imaging instrument. Our ultrasound imaging systems based on the HFR imaging method can solve these problems and eventually, bringing in low-cost and high-quality ultrasound imaging systems.

In Figure C.1, the existing conventional ultrasound imaging system is compared with the HFR ultrasound imaging system in the aspects of performance and cost. The lower right quadrant in Figure C.1 is the advantage and final goal of the product of the HFR ultrasound imaging system. In terms of the performance such as imaging resolution and resistance to noise, the ultrasound imaging system with the HFR imaging method is proven higher than current technology, D&S imaging method. Additionally, with the simplicity of hardware design in the HFR ultrasound imaging system, the manufacturing cost is lower than the existing ultrasound imaging instruments.

Even though other imaging methods, such as parallel beamforming, synthetic apertures, are proposed to achieve high frame rate and high quality ultrasound image, the
imaging systems’ complexity are increased as well. That results in higher product price than generic equipments.

![Diagram showing desired mobility vs. computing power.](image)

**Figure C-1: Schematic showing desired mobility vs. computing power.**

The ultrasound imaging system based on the HFR imaging method can create unique benefits: higher image quality, system simplicity, and system extendibility. The HFR ultrasound imaging method is not only used for the direct 2D or 3D organ or tissue imaging, but also used for exploring the applications of the ultrasound imaging method. Strain rate imaging, which can be used for detecting the mechanical property of soft tissues, is researched. The investigated strain rate imaging utilizes 2D ultrasound images generated by the HFR imaging method. It is found that when the strain rate imaging,
based on the HFR imaging method, is applied, the accuracy of strain estimation of soft
tissues is higher than that with the D&S imaging method.

The HFR imaging system can find its niche market in the business field of ultrasound imaging equipments. Not only does it provide better ultrasound image quality, but also provide easier extended applications into the HFR imaging platform.

C.3 The Company

The prototype of the HFR ultrasound imaging system has been built in the Ultrasound Lab at the University of Toledo. Moreover, the patent of this system has been issued to Dr. Jian-yu Lu. Once the business plan is verified to be feasible, a limited liability company (LLC) with the core members of Dr. Jian-yu Lu and Hong Chen will be registered. Since the prototype system has been already finished, the product level of the HFR ultrasound imaging system will be developed once funding is ready.

C.4 The Product

The product is the ultrasound imaging system based on the high frame rate imaging method. With this ultrasound imaging system, traditional 2D ultrasound images of human organs or fetuses, and color Doppler Imaging can be produced. 3D ultrasound imaging with a full 2D array transducer is also proven feasible when data compression is applied to echo data in a scanning probe. The initial product will be the 2D HFR ultrasound imaging system, and 3D HFR ultrasound imaging system will be kept researching and developing. Additionally, our product of the ultrasound imaging system can provide strain rate imaging showing tissue elasticity information. This product possesses the big advantage of high frame rate, which meet the requirement of its
application into the echocardiology. The prototype of the HFR ultrasound imaging system is shown in Figure C.2.

![Prototype of medical HFR imaging system.](image)

**Figure C-2: Prototype of medical HFR imaging system.**

Along with the increasing need of home health care, portable ultrasound imaging devices will be a branch of trend in the future. The secondary application of this HFR imaging technique is portable ultrasound imaging device. The portable version of the HFR ultrasound imaging system can be designed to meet the rising home health care services. Due to the feature of lower hardware requirement for the HFR ultrasound imaging system than the D&S ultrasound imaging system, the portable version of the system is easier to be realized thanks to its absence of beamformer units.
The initial product of the HFR imaging system will not incorporate the functionality of 3D imaging, but the 3D imaging will be incorporated in the next generation of product. The product will be priced at $80,000 per unit. This price is much lower than that of the existing products at the same quality level (around $100,000). If an ultrasound probe for a 3D HFR imaging system is successfully developed, the cost of 3D HFR imaging system will be projected to be $90,000. The selling price will be marked as $120,000.

C.5 Industry overview

The ultrasound imaging equipment is one segment of the medical imaging industry which also includes the X-ray imaging instruments and magnetic resonance imaging (MRI) instruments, et al. The medical imaging industry is expected to keep growing at a moderate pace through 2015 in the global range. The global industry has experienced growth from $9.65 billion in 2002 to around $20 billion dollars in 2011. Even though industry growth has been hammered by the economic downturn since 2008, it is still projected to rise to $24 billion by 2012. The largest market of the medical imaging devices is US, following that, Europe is the second largest market. These two markets are matured and are near to saturation due to the long lifetime of ultrasound imaging equipments. It is worth noting that the increasing aging population in US, resulting in higher patient volume, presents a good opportunity to the industry. Meanwhile, technological advancements can also boost the growth of the medical imaging industry. However, it is worth noting that the increasing aging population in US, resulting in higher patient volume, can be a good opportunity to the industry.
The regional market of Asian is believed to be untapped and it will be a major engine of industry growth due to their government policies and health care reforms. China is apparently the fastest growing market for medical ultrasound imaging equipments. According to the 2010 research report about the ultrasound equipments from the Icon Group International, in the first global 20 international cities ranked with the market size, 6 cities are located in China.

C.6 Market Size and Trends

In the medical imaging industry, ultrasound imaging instruments have become the second largest sector globally and it is the fifth largest sector in US. Even though the global market of ultrasound imaging equipments has suffered from decline from 2007 to 2009, especially, the drop of 6% between 2008 and 2009 due to the economic recession, the market is still estimated to grow around 20%, reaching $6 billion, by 2015. The market sizes of US, Europe, China and Japan are given in Table 1. In 2009, the US market is the world’s largest ultrasound market. The ultrasound market value of Europe is estimated $1.75 billion in 2011. However, the ultrasound market in US is highly saturated as well as the Europe. According to the analysts from the InMedica, China will become the major market for ultrasound imaging equipments in the coming years. China will be the largest market in Asian pacific region.
Table C.1: The market size of major ultrasound medical equipment markets

<table>
<thead>
<tr>
<th>Country</th>
<th>Market Size</th>
<th>Year</th>
</tr>
</thead>
<tbody>
<tr>
<td>US</td>
<td>$1.26 billion</td>
<td>2009</td>
</tr>
<tr>
<td>Europe</td>
<td>$1.75 billion</td>
<td>2011</td>
</tr>
<tr>
<td>China</td>
<td>$1.00 billion</td>
<td>2014</td>
</tr>
<tr>
<td>Japan</td>
<td>$445.97 million</td>
<td>2009</td>
</tr>
<tr>
<td>World</td>
<td>$6.00 billion</td>
<td>2015</td>
</tr>
</tbody>
</table>

The color Doppler ultrasound imaging system weights more and more on the ultrasound market. The market share of Color Doppler ultrasound systems accounts for 93% of the whole ultrasound market. Our HFR ultrasound imaging system will also incorporate the functionality of the color Doppler technology to increase the product competency. Additionally, the strain rate imaging will also provide the diagnosis information in the color format. The real-time 3D ultrasound imaging is also a trend for the ultrasound imaging market. It not only provides vivid 3D images of fetus or organs, but also the 3D images can be used to extract useful and more accurate diagnosis information. However, the current products for 3D imaging from three big companies: General Electric (GE), Philips, and Siemens have complex system design and high product prices.

Handheld or portable ultrasound imaging system is believed to be the most important trend in the future. The growth of the segment of handheld/ portable ultrasound is projected to be $1.4 billion globally by 2016. To catch the business development wave
of portable ultrasound, our company’s next research goal is to develop the portable version of the HFR ultrasound imaging system. The system complexity is expected to be reduced thanks to the technology utilized in the HFR imaging method, so the cost of manufacturing a portable ultrasound imaging equipment will be lower than the products based on D&S imaging method.

C.7 Estimated Market Share and Sales

According to the estimated market size, which is around $5 billion in 2012, the number of sold units of ultrasound imaging systems is more than 20,000. In this proposal, 5 units of the HFR ultrasound imaging system is projected to be sold in the first selling year, moreover, these 5 units are only sold in local hospitals in Lucas County, OH. Selling our products to local or nearby hospitals can be easier than selling in states far away from Toledo, because it is convenient to demonstrate the advantages of our HFR ultrasound imaging system, and convenient to provide training to the physicians or sonographer technicians. Moreover, the purpose to firstly explore the local or near-neighbor hospitals is to monitor closely the performance of our product and to provide excellent technical support if system problems appear. Additionally, business exploration in the local hospitals in the initial stage can save the overhead for our field engineers travel for solving system problems. The first year sale value will be $400,000 if the sale goal of 5 units is successfully achieved. That will account for around 0.015% of the global market share. Since our HFR imaging system possess all the functionalities current systems have and can even outperform current systems, it will be not difficult to accomplish the selling goal. In the following five years, the business will be expanded gradually to cover the whole Ohio area. The growth rate is projected to be 60%, which
means 8 units will be sold in the second year, 13 units in the third year and so forth. At the fifth year, the sale volume is projected to be 33 units, which can produce the sale value of around $2.64 million. The market of Ohio locates in the Great Lake market region, which is a middle size market accounting for around 14% of the whole US ultrasound market as shown in Figure C.3. In this region, the sale volume of our market plan will not be small enough to keep low profile in initial business phase, meanwhile; it will not be too small to maintain the business.

![Latent Demand across Regions across the US for Medical Ultrasound Equipment (USS Million): 2011](image)

**Figure C-3: The latent demand across regions across the US for medical ultrasound equipment (US $ Million) in 2011.**

The handheld or portable ultrasound imaging system will also be developed based on the HFR imaging system. Our company will assure the time length of research and development to be short enough to compete with existing portable ultrasound imaging products. The sale project will be changed if the portable system is developed successfully, and undoubtedly, it can boost sales greatly.
C.8 Competition

The competition of the ultrasound market is considered to be intensive even though the manufacturer number is not large. Among the ultrasound manufacturers, four companies occupy almost two third market share of the global ultrasound market. They are respectively GE Healthcare, Philips Healthcare, Siemens Healthcare and Toshiba Medical Systems. It is worth mentioning that GE and Philips accounts for 43% of the ultrasound market in the global. Moreover, the biggest global electronic company, Samsung, also tried to enter the medical imaging industry by acquiring the Aloka. All of these aforementioned companies impose the direct competition to our company and they will be the biggest threat to our business’ viability. There are also mid-tier competitors who also manufacture ultrasound imaging systems. They are Hitachi medical systems, Sonosite Inc., Esaote North America, Tomtec Imaging system, BK Medical, and Zonare Medical Systems.

Most of the competitive products from these companies, however, are based on the conventional medical imaging method, so it is proven by Hong Chen that the imaging quality is lower than our product of the HFR ultrasound imaging system. Real time 3D ultrasound imaging will be a trend for the ultrasound imaging system. In fact, ultrasound imaging systems from GE, Philips, and Siemens already have integrate the 3D functionality. For example, the iE33 xMatrix from Phillips Healthcare Division can function well in 3D imaging; the Vivid E9 from GE Healthcare Division can perform 4D (real time) imaging; the ACUSON SC2000 from Siemens Healthcare Division can execute 4D imaging as well. However, as mentioned in the “product” part, our ultrasound imaging system has the advantage of system simplicity to achieve 3D ultrasound imaging.
than existing products because only one or two transmitters and no beamformer units are adopted for the HFR imaging system. The lower hardware requirement of our ultrasound imaging system means a lower manufacturing cost. Besides the technology advantages of our HFR ultrasound imaging system, we also target at providing premium customer service which is always a weakness for big companies. Those big companies could sell good quality ultrasound imaging systems to hospitals and research institutes; however, their technical support will usually not be timely if equipment suffers from technical problems. As a start-up and small size company, our premium and timely customer service is attractive to customers.

Indirect competition also exists for the ultrasound imaging industry. The ionizing imaging systems utilizing X-ray, the MRI, and PET impose big challenges to ultrasound imaging technology. These imaging systems can provide clearer images than ultrasound to some extent. However, the high prices of these systems are the main problem for many hospitals and research institutes to buy these systems. Furthermore, these imaging technologies have the risk of radiation, which does not exist for ultrasound imaging technology. So it is believed that the business of ultrasound imaging system will still keep growing because the ultrasound technology never stop advancing to provide higher quality images. Meanwhile, the safety factor of an imaging system is always a big concern to patients.

C.9 Regulatory Issues

The HFR ultrasound imaging system needs to be proven safe on human bodies by the Food and Drug Administration (FDA). Since the transmitting power applied on the
human bodies is the same as the conventional ultrasound imaging system and has already been proven safe in the Ultrasound Lab at the University of Toledo, it will not be an issue to stop the release of our product.

C.10 Management Team and Organizational Structure

A LLC will be registered by Hong Chen. Hong Chen, as the founder of the company, will be temporarily the chief executive officer (CEO). Jian-yu Lu will be the chief technical officer (CTO) thanks to his expertise in the HFR ultrasound imaging system. The company will be headquartered at Toledo, OH. Once the venture capital is obtained, an experienced CEO in medical imaging industry will be hired to well manage the company. An accountant will be hired to handle the financial section of the company when the company grows. Two more engineers possessing excellent knowledge of ultrasound imaging system will be hired from universities to develop the 3D imaging part in the HFR ultrasound imaging system. The first year salary will be only $25,000. However, the 5% company stake will be provided to either of them.

C.11 Marketing Plan

C.11.1 Market Entry and Growth Strategy

At the first step, the HFR ultrasound imaging system will be demonstrated in hospitals and research institutes in Lucas County, Ohio. The business will grow by selling our products to hospitals and research institutes in Ohio. In three years, the markets of adjacent states are targeted. In five years, the business will be spread nationwide.
C.11.2 Advertising and Promotion

The main advertising method is the demonstration in hospitals and research institutes. The demonstration and training will be provided once one month at the same location. Meanwhile, the trainings of how to effectively and efficiently manipulate our ultrasound imaging equipment will be provided for free. During the training course, manuals featuring our product will be provided for free. Radio will also be taken as the channels of introducing the new ultrasound imaging system to the local public. The total budget for the radio is $4,000 for a full year, so it is estimated one spot for one day.

C.11.3 Price

The price of the HFR imaging system will be $80,000, which is much lower than existing high-level ultrasound imaging systems. The cost of this product is estimated as $60,000, so the gross margin is calculated as 33%. Along with the system upgrading with the incorporation of more functionality, the price will be increased as well.

C.11.4 Sales Tactics

Direct sale to the hospitals, clinics, and research institutes will be major tactic. Founders will be the salesman in the initial phase to search for customers. Then, interested customers will be invited to visit our office to try our product.

C.11.5 Distribution

The circuit part, such as manufacturing print circuit board, soldering and testing circuit boards will be outsourced to electronic factories in US. The equipment housing to accommodate the electronic systems can be procured from China. The HFR imaging
system will be assembled in the company and distributed directly to hospitals or research institutes.

C.11.6 Service and Warranty Policies

Purchased equipments will be installed and tested by technicians from our company to assure the equipments run normally. Lifetime warranty of the products will be issued to the customers. 24/7 free technical support will be available.

C.12 Patents and Proprietary Issues

The patent of HFR imaging system has been issued in June, 2012. Novel techniques found during the research and development (R&D) of 3D imaging and portable ultrasound devices will be filed for patents in time.

C.13 Product Design and Development Plans

The prototype of 2D HFR ultrasound imaging system has been developed. More peripheral devices need to be further developed. The development steps are as follows:

1. The small size image reconstruction system, which can fit into the instrument housing, will be designed.
2. The 3D imaging will be developed and integrated into the HFR imaging system.
3. The Color Doppler imaging will be developed and integrated into the HFR imaging system.
4. Integrate strain rate imaging into the HFR imaging system.
5. Develop the portable HFR imaging system by reducing the system functionality and shrinking the system size.
The above development plan will be executed step by step. Time span for the development plane from step 1 will be around four months. Step 2 is estimated to take a year. Step 3 and 4 are estimated to take four months. Step 5 is expected to be finished in two years.

C.14 Manufacturing and Operations Plans

C.14.1 Manufacturing Plan

The product of the HFR ultrasound imaging system requires moderate R&D. The R&D will be conducted in the rent office. Hong Chen and Jian-yu Lu will mainly be in charge of the R&D. A station for soldering and testing circuits is a necessity. Circuit testing equipments, such as the oscilloscope, power supply etc. will be purchased. The circuit soldering will be outsourced to a specific company in US and the assembly will be conducted in our office. Since the company is in US, it is easier to communicate with them to control the quality of soldered circuit boards. If there is any problem about circuit soldering, the problem circuit boards will be sent back to the contracted electronic factory. The instrument housing will be purchased oversea from China since the materials are cheap there. Since the final product is assembled and packaged in our company, we will have enough control over the quality of final products. The manufactured HFR ultrasound imaging systems will be tested with standard phantoms and excised human tissues to ensure the system working perfectly. The manufacturing flow chart is shown in Figure C.4.
C.14.2 Operation Plans

The distributing channel of our HFR ultrasound imaging equipments is selling them directly to hospitals and research institutes. The operating cycle for the HFR ultrasound imaging equipment will not be long because most of the manufacturing steps are finished in US. Only the order of instrument housing takes around 15 days from order taken since it is shipped to US from China to reduce the manufacturing cost. Considering the high cost of medical imaging equipment, the equipment other than the demonstration one will not be built until the order is placed from hospitals and research institutes. In this
way, there will be no need for spacious warehouse to stock inventory. The equipment will be delivered by our technical staffs, who will also install the delivered imaging systems, using the rental truck. Renting facilities is always a good way to save monetary for startup companies.

The cash conversion scheme for our operation plan is shown in Figure C.5. With this cash conversion scheme, it will take about one month to convert cash after the procurement payment. The total operation length is 53 days for our ultrasound imaging system. Since our initial targeted customers are local or near hospitals and research institutes, it will take only one day to deliver our products. If the demand of our ultrasound imaging system surges, some inventory of electronic components and instrument housings will be held to shorten the waiting period for our customers to get their products. In this case, only 15 days are needed to manufacture one imaging system, compared with the original 23 days. Even though a small amount of inventory is needed, our office will still be large enough to hold them because electronic components and instrument housings are of small size.

![Figure C-5: Cash conversion procedures and length.](image)

151
C.15 Financial Plan

The break even analysis of the first year is shown. The projected yearly cash flow and income statements are presented for 5 years as well.

C.15.1 Break Even Analysis

The break even analysis is shown in Figure C.6. The fixed cost for the first year is projected to be $80,000, which includes the cost of rent (around $12,000), equipments (around $10,000), wages (around $50,000), and other expenses ($8,000) for two hired technical persons. The cost of one ultrasound imaging instrument is estimated to be $60,000, and the price markup is assumed to be 33.33%, ending up with the gross revenue of $20,000. The sale goal of the first year is 5 units, and the goal of the second year is 8 units with the growth of 60%. As we can see from Figure C.6, the breakeven point is reached when only four units are sold. If the economic scale of volume is achieved, the cost of the equipment housing, the electronic components will be further reduced, resulting in the final decrease of the price of our product.

Figure C-6: The break even analysis for the HFR ultrasound imaging equipment.
C.15.2 Yearly Proforma Income Statements

The proforma income statements over the period of 5 years are provided with the assumption of the linear growth of 60% each year. The itemized costs and earnings for yearly income statements are shown in table, respectively, in which the unit is US dollar. The starting up fund is projected to be $200,000, which includes fixed cost and the cost of the first year manufacturing of two imaging equipments. From the proforma income statement table, we can see the completely positive sign to our business. The net income will achieve $1 million in the fourth year (see Table 2). Considering the net income growth and the whole market size of ultrasound medical equipment, it is believe that our business will be viable and will have big room to expand.

Table C.2: The yearly income statements of the period of 5 years.

<table>
<thead>
<tr>
<th></th>
<th>Start-up</th>
<th>Year 1</th>
<th>Year 2</th>
<th>Year 3</th>
<th>Year 4</th>
<th>Year 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Net sales</td>
<td>0</td>
<td>400,000</td>
<td>640,000</td>
<td>1,040,000</td>
<td>1,680,000</td>
<td>2,640,000</td>
</tr>
<tr>
<td>Cost of sales</td>
<td>0</td>
<td>300,000</td>
<td>480,000</td>
<td>780,000</td>
<td>1,260,000</td>
<td>1,980,000</td>
</tr>
<tr>
<td>Gross profit</td>
<td>0</td>
<td>100,000</td>
<td>160,000</td>
<td>260,000</td>
<td>420,000</td>
<td>660,000</td>
</tr>
<tr>
<td>Advertisement</td>
<td>0</td>
<td>4,000</td>
<td>4,000</td>
<td>4,000</td>
<td>4,000</td>
<td>4,000</td>
</tr>
<tr>
<td>Office Rent</td>
<td>3,000</td>
<td>12,000</td>
<td>12,000</td>
<td>12,000</td>
<td>12,000</td>
<td>12,000</td>
</tr>
<tr>
<td>Wages</td>
<td>0</td>
<td>50,000</td>
<td>100,000</td>
<td>120,000</td>
<td>120,000</td>
<td>120,000</td>
</tr>
<tr>
<td>Equipments</td>
<td>10,000</td>
<td>0</td>
<td>0</td>
<td>10,000</td>
<td>0</td>
<td>10,000</td>
</tr>
<tr>
<td>Utilities</td>
<td>0</td>
<td>3,000</td>
<td>4,500</td>
<td>7,000</td>
<td>11,000</td>
<td>17,000</td>
</tr>
<tr>
<td>Others</td>
<td>3,000</td>
<td>1,000</td>
<td>1,500</td>
<td>2,438</td>
<td>3,938</td>
<td>6,188</td>
</tr>
<tr>
<td>Income total</td>
<td>-16,000</td>
<td>30,000</td>
<td>38,000</td>
<td>104,563</td>
<td>269,063</td>
<td>490,813</td>
</tr>
<tr>
<td>Income tax</td>
<td>0</td>
<td>10000</td>
<td>16000</td>
<td>26000</td>
<td>42000</td>
<td>66000</td>
</tr>
<tr>
<td>Net income</td>
<td>-16,000</td>
<td>20,000</td>
<td>22,000</td>
<td>78,563</td>
<td>227,063</td>
<td>424,813</td>
</tr>
<tr>
<td>Net income (% of Sales)</td>
<td>5%</td>
<td>3%</td>
<td>8%</td>
<td>14%</td>
<td>16%</td>
<td></td>
</tr>
</tbody>
</table>
C.16 Overall Venture Schedule

The objectives included in the overall venture schedule are the R&D of 3D imaging, the selection of PCB manufacturer, the selection of components and equipment housing manufacturers, and also the timing to start the advertisement, and so on. The details schedule is shown in Figure C.7. The time span of this schedule is 12 months. It is believed that bank relation could be established once our first product is sold because of the large profit. Advertisement will start one month right before the beginning of the product of HFR ultrasound imaging systems.

![Table](image)

**Figure C-7**: The overall venture schedule for the product of the HFR ultrasound imaging system.

C.17 Critical Risks and Assumptions

C.17.1 Critical Assumptions:

1. The development of 3D ultrasound imaging component is successful.
The 3D imaging of our product will be an outstanding feature for us to compete with other products due to the novel imaging method. That will make our business expandable because the price for the HFR 3D ultrasound imaging equipment incorporating 3D imaging will be low. This is not the case for the existing ultrasound imaging equipments.

C.17.2 Critical Risks:

1. Intense competition comes from existing rivals

   Evaluation: Intense competition mainly comes from those big companies manufacturing ultrasound imaging system, i.e. GE, Philips, and Siemens and so on. However, we are targeting a small and local market and trying to avoid intense competition.

   Contingency: It is believed that such a small sales volume will not catch the eyes of big players in the ultrasound imaging business.

2. Sales lower than expected volume

   Evaluation: Our sales goal for the first year is 5 units. There are more than 10 hospitals and research institutes in Toledo. Salesman will try their best to attract customers.

   Contingency: If this happens, we will double the investment in advertisement and hire one more salesman.

3. PCB manufacturer quits
Evaluation: There are plenty of PCB manufacturers in US. The competitions in those two industries are relatively high currently. It is unlikely the situation could happen.

Contingency: If PCB manufacturer quit doing business with us, it is easy to find a substitute in China or in other low-wage countries.

C.18 Harvest Strategy

By following up the development trajectory of our company, there will be two harvest strategies to harvest. First, if the company’s sales always face very intense competition from big players and we might not be strong enough to directly compete with them, the company will be sold to one of those big manufacturers of ultrasound imaging equipment. This situation is expected to happen in 5 years. An alternative strategy is that if our company grows big and strong enough to compete with those big players and holds the leading role in the field of 3D ultrasound imaging products, an IPO for our company will be proposed for further growth.

C.19 Proposed Company Offering

Funding is needed for the R&D phase. Marketing also plays an important role for the potential success of this product, so more marketing personnel will be hired as the sales increase and more advertisements will be designed to increase the brand recognition by the public. Common shares will be issued to investors and founders and contributors’ efforts for return. In the first year, love money and funds from angel investors will be sought. After two years of the first selling, the funds from venture capitalists will be sought to accelerate the company expansion. The details of proposed offering are given in Table C.3.
Table C.3: Proposed desired financing and offering.

<table>
<thead>
<tr>
<th>Sources</th>
<th>Funding Sought ($)</th>
<th>Percent of Ownership (%)</th>
<th>Interest Return (%)</th>
<th>Usage of Funding</th>
</tr>
</thead>
<tbody>
<tr>
<td>Love Money</td>
<td>200,000</td>
<td>0</td>
<td>10</td>
<td>Initial R&amp;D, Rent Working Space</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Buying facilities</td>
</tr>
<tr>
<td>Angel Investors</td>
<td>600,000</td>
<td>10</td>
<td>10</td>
<td>R&amp;D, Hiring technicians, Marketing, File patents, Setup online sale website</td>
</tr>
<tr>
<td>Venture Capitalists</td>
<td>1,500,000</td>
<td>40</td>
<td>0</td>
<td>R&amp;D, Hiring more technicians, Expand manufacturing volume</td>
</tr>
</tbody>
</table>