CHAPTER 1

Introduction

1.1 Introduction

Ultrasound imaging is considered as the most prevalent medical diagnostic tool among currently available medical imaging modalities since it is non-invasive, practically harmless to human body, economic and portable. Modern diagnostic ultrasound is gradually opening up its widespread clinical usage such as B-mode imaging, colour-flow imaging, spectral Doppler etc. (York and Kim, 1999) due to following advantages: (a) Ultrasound is hazardless and allows repeated examinations of the same patient with no risk using standard equipment. (b) Since the range of medical examinations are relatively short (typically 0.1- 25 cm) (Azhari, 2010) and the speed of ultrasound wave through tissue cells is about 1540 m/sec, many waves can be transmitted within short time interval and gain sufficient information to follow the dynamic changes of the body parts in 'real time'. (c) Cost effective. Unfortunately, the image quality and resolution of ultrasound image is inherently poor due to the physical phenomenon underlying the image acquisition and imperfection of the imaging system design (Michailovich and Tannenbaum, 2006). A comparison of different diagnostic medical imaging modalities is given **Table 1.1**(Becker, 2004; Kallergi, 1998; Osuga and Han, 2004; Tavernier, 2011). Other medical imaging modalities such as X-ray, CT scanner and MRI have better resolution with respect to ultrasound image. Hence there is a scope of improving the resolution of ultrasound image. To point out the resolution problems persist in the B-mode ultrasound imaging system we first present a brief description of the imaging system.

1.2 B-mode Ultrasound image formation technique

The schematic diagram in **Figure 1.1** illustrates the processing stages of a typical diagnostic ultrasound image formation system. The ultrasonic wave pulse of appropriate frequency is transmitted by means of piezoelectric transducer. A portion of acoustic energy reflects back from the interface of materials with different acoustic impedance. The reflected pulses are received by the transducer and converted into the RF electrical signals. The amplitude of the signal gets attenuated when it propagates through the tissues. Therefore, the receiver first amplifies the received signal with the help of an amplifier whose gain is proportional to distance traversed by the pulse or the time required for pulse to return. This amplifier stage is known as **Time Gain Compensation (TGC)** stage (York and Kim, 1999).

Table 1.1: Comparison of medical imaging modalities

Imaging modalities	**Spatial Resolution**			
Latest Multi-slice CT Scanner	≈ 0.2 - 0.5 mm in X-Y-Z directions			
X-Ray Mammography	≈ 100 µm - 40 µm			
MRI	≈ 20 µm - 1 mm			
PET	≈ 0.5 - 2 mm			
Ultrasound image	Frequency (MHz)	Resolution (mm)		Penetration depth(mm)
		Axial	Lateral	
	3.5	1	2	160
	5	0.6	1.2	100
	7.5	0.4	0.8	50

The conditioned output of the TGC stage is then sampled at a high rate (typically four to five times of the transducer frequency). The demodulator removes the carrier through quadrature demodulation technique. The output of the demodulator is low-pass filtered and results in a baseband signal of a complex samples. This complex samples contains both the in-phase and quadrature components. The B-mode image is created by taking the magnitude (envelope detection) of the demodulator output. Since the difference in the echo intensity between the bright specular reflector and weakest parenchymal scatter might be as much as 60 dB and a conventional TV monitor screen capable of displaying no more 30 dB, some compression of the range of the echo amplitudes is necessary. This is achieved by amplifying the low level echoes almost linearly, but the high level echoes in a manner, which compresses them into a narrow dynamic range. The **logarithmic compression amplifier** with suitable gain can perform this operation. Hence the envelope is passed through a logarithmic amplifier whose input output characteristic is given by:

$$Y_o = D \log [X_{in}] + G \tag{1.1}$$

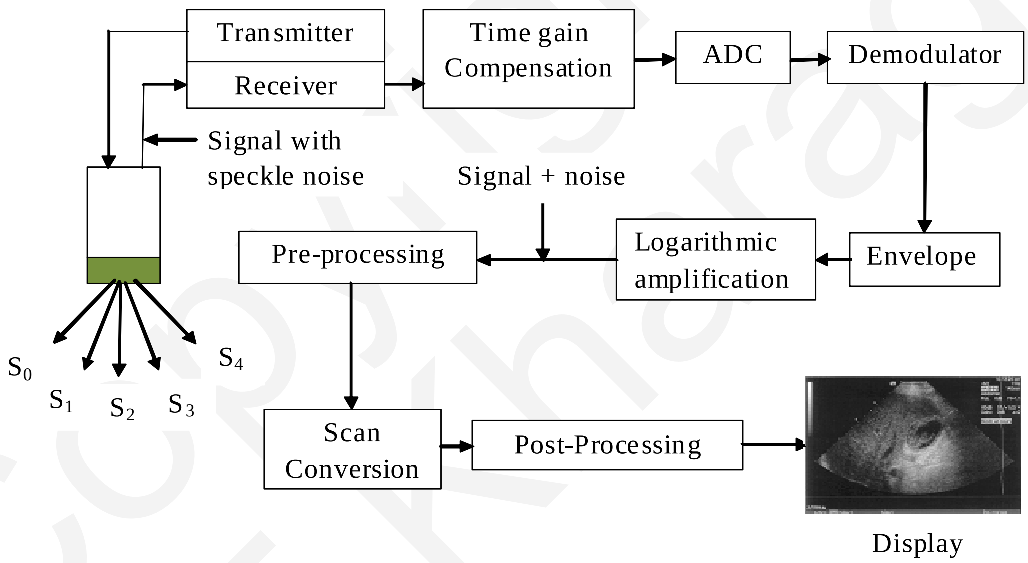

Figure 1.1: Schematic block diagram of ultrasound image formation system.

where X_{in} is the input of the amplifier Y_o is the output of the amplifier. D is the parameter of the amplifier. It determines the dynamic range of the amplifier. G represents the linear gain of the amplifier (York and Kim, 1999).

The samples of the signal obtained from received pulse constitute a vector of sample points. This is known as scan line. In recent day's phased array transducers can change the focus point of the beam as well as steer the beam by employing electronically beam steering arrangement. Beam steering is performed by changing the timings of the firing of the array elements. By steering the beams in a plane, multiple vectors or multiple scan lines can be obtained in different directions in a plane (i.e. S_0 - S_4 in **Figure 1.1**). Each sample value on the scan lines has unique position (polar co-ordinate : r, Ψ). Since conventional television monitor supports input only in Cartesian co-ordinate for displaying the image, scan conversion is essential for scan lines. Hence through scan conversion, a two dimensional image is formed in Cartesian co-ordinate. Scan conversion is done through interpolation. Ophir and Maklad (1979), Robinson and Knight (1982), Lee and Park (1985), Rechard and Arthur (1994), Berkhoff et el. (1994), Chang et el. (1997) discussed several types of scan conversion techniques. Lee et al. (1986) has given the analytical treatment of scan conversion algorithm for a real time ultrasound scanner. Ophir and Brinch, (1982) addressed Moire′ under-sampling artifacts in digital ultrasound image. A typical scan conversion technique is shown in **Figure 1.2**.

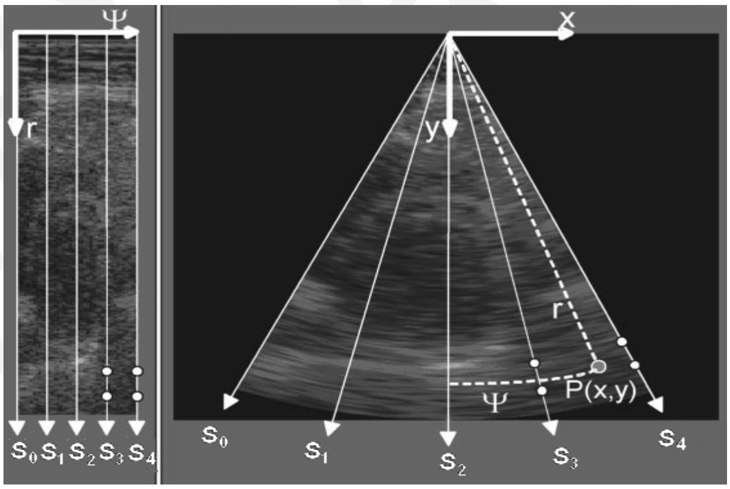

Figure 1.2: Scan conversion from scan lines

Geometrical pattern of the scan lines (for B-mode sector scan) on the rectangular raster reveals that the inter-sample distance between two successive sample points gradually becomes larger from lower depth of penetration to higher depth of penetration. It makes aerial sample density at the higher depth lower. Consequently, the image quality degrades and loses its resolution at the higher depth. Sometimes it may happen that the inter-sample distance between two successive sample points is so large that in the horizontal direction Nyquist criteria may not be satisfied causing aliasing.

1.3 Resolution

In many cases, ultrasound is used to make anatomic measurements of an organ or lesion. The precision of such measurements suffers from certain limitations, which are the fundamental consequence of the physics itself. In particular, the resolution of the image determines the precision of any measurement made from it. The resolution of an image is power of resolving two distinct objects within the image. In ultrasound imaging the resolutions are defined as follows.

- **Spatial resolution**: Spatial resolution of an image denotes the spacing between two successive pixels. It is measured in pixels per inch (ppi). In ultrasound imaging, spatial resolution divided into two categories:

  (a) Axial resolution: Axial resolution is defined as the minimum separation between two successive sample points of different objects in a direction of propagation of acoustic wave. It depends on the frequency (f_c) of the ultrasound wave. It also depends on the sampling frequency of the imaging system.

  (b) Lateral resolution: Lateral resolution is defined as the minimum separation between two samples of two separate objects aligned along the perpendicular to the propagation of ultrasound beam. Lateral resolution depends on the aperture of the transducer array (or the number of array elements) the number of scan lines within a field of view. Large number of the transducer array generates small beam-width and consequently increases the lateral resolution. Lateral resolution is significantly worse compared to axial resolution.

- **Temporal resolution**: Temporal resolution is defined as the number of frames captured per second. It is commonly called frame rate. It is related to perceptible motions between the frames.

Brightness resolution: Brightness resolution of an image refers to the number of brightness levels that is given to a pixel of the image. It is dependent on the quantization levels during digitization of image. Usually the brightness resolution is limited to 256 gray levels.

1.4 Motivation

From the **Table 1.1**, it is observed that the ultrasound images have relatively poor resolution with respect to other imaging modalities. Hence it is necessary to put efforts in research of enhancing the resolution so that it becomes comparable to the other imaging modalities. There are several hardware based techniques by which the resolution of ultrasound wave can be improved. Axial resolution can be increased by increasing the pulse repetition frequency (PRF) and the frequency of the ultrasound wave. But high frequency ultrasound waves are attenuated rapidly in the path of propagation through tissue cells. It makes difficult to image the higher depth of tissue structure. Typically, in diagnostic imaging, frequencies vary from about 2 MHz for some cardiac and deep abdominal applications through 10 MHz for imaging the superficial structures such as blood vessels, to 20 MHz or higher for intravascular imaging (Burns, 2005). For typical ultrasound imaging system, at 3.5 MHz ultrasound frequency, axial resolution is 1 mm, lateral resolution is 2 mm and the penetration depth is 160 mm. At 5 MHz ultrasound frequency, axial resolution is 0.6 mm, lateral resolution is 1.2 mm and the penetration depth is 100 mm; and at 7 MHz frequency, axial resolution is 0.4 mm, lateral resolution is 0.8 mm and the penetration depth is 50 mm (G. Schmidt, 2004). The lateral resolution increases with increasing frequency and the number of transducer array elements. It is noticed that increasing the operating frequency reduces the depth of penetration. In the case of sector scanner, the sampled scan data is available in polar co-ordinate system. To display the image in video screen having rectangular raster, scan conversion is performed by using a suitable interpolation technique. Since the samples are addressed in polar co-ordinate, the gap between the two samples of two successive scan lines at same radial distance increases with the depth of penetration. Due to this geometrical pattern of the scan lines, the quality of scan-converted image becomes poor at higher depth of penetration. Increasing the number of scan lines can solve this problem. But increasing the number of scan lines leads to increased hardware complexity and the cost associated with it.

To solve the problem of resolution, many attempts are made by the scientists and researchers. As solutions to the problem of resolution, tissue harmonic imaging (THI) and contrast harmonic imaging (CHI) has got major important to the researchers. THI can acquire higher spatial resolution image and has been used in the commercial medical ultrasound system since last decade. The problem of THI is that the frequency bandwidth of the fundamental and the harmonic components often overlap. The spectral overlap causes degradation of spatial resolution and the undesirable artifact. One possible solution to this problem is the extraction of harmonic component by pulse inversion. But for even little movement of reflectors in the region of interest may create residual fundamental components with extracted harmonic components. The residual fundamental component has a bad effect on tissue harmonic image. To obtain both sufficient SNR and spatial resolution, a combination of THI and coded excitation, has been proposed by Misaridis & Jensen (2005). But there still remains the problem of the spectral overlap.

Contrast harmonic imaging techniques are also in use today. But they are applicable only for those conduits, which are not usually visible clearly in US image and can be imaged by injecting micro-bubble. It is also cost effective since costly gas-filled micro-bubbles are used.

It is mentioned that hardware-based solution of resolution enhancement such as increasing frequency and increasing the number of transducer array elements are not good solutions. For example, increasing the frequency decreases the capability of imaging higher depth of body. It decreases the scope of diagnostic applications. Increase of transducer array elements will decrease the beam-width and hence enhance the lateral resolution but it increases the size of the probe, cost and complexity of the system. It is reported that limited number of sensors causes point spread function (PSF) of the point source to be blurred. It limits the resolution of the image. Super-resolution (SR) is an algorithmic way (or software centric approach) whereby multi-frame motion is used to overcome the inherent resolution limitations of a low resolution camera system (Chowdhuri, 2002). Considering this fact, one can apply the software centric super-resolution technique to the ultrasound imaging system to enhance the resolution of ultrasound image.

In super-resolution reconstruction, a high-resolution image is formed from several sub-pixel shifted low resolution (LR) observations (i.e. low-resolution frames). Each low resolution observation potentially contains novel information about the desired high resolution image. Super-resolution combines the low resolution frames into a high resolution grid by including the new information from all the low resolution frames. A brief introduction to super-resolution with a few popular techniques is presented in the literature review in Chapter 2.

Few attempts are taken to apply super-resolution algorithms to ultrasound images to enhance the spatial resolution of the ultrasound image. Spatial resolution enhancement of ultrasound image based on artificial neural network was proposed by Carotenuto et al.

(2002). Though the method is fruitful to improve the point spread function (PSF) with the help of a trained artificial neural network, it cannot restore the information lost during sampling process. Optical flow-based iterative back-projection algorithm was proposed by Zhang et al., (2010). This method is applied to scan-converted ultrasound image. Super-resolution reconstruction based on maximum a-posteriori (MAP) approach with transformation information was proposed by Dai et al., (2009). This method also used to scan-converted ultrasound image. Super resolution on the ultrasound scan data is not reported till date. There is a scope of using super-resolution technique to the ultrasound scan data.

Another nuisance in ultrasound imaging is speckle noise. Speckle is caused by constructive and destructive interferences of wavelets scattered by tissue components as they are received by the transducer (Wagner et al. 1983). It is similar phenomenon that occurs in case of laser imaging due to due scattering of coherent laser from rough surface. Speckle is the most important factor that limits the contrast resolution in the ultrasound imaging. It limits the detectibility of small and low-contrast lesions and causes the difficulties in interpreting the ultrasound image. Sometimes it becomes very difficult for an expert to draw a right conclusion from an ultrasound image corrupted with speckle noise.

Comprehensive analysis of statistics of the speckle noise was given by Goodman (1976) Burckhardt, (1978) and Wagner et al., (1983). It was found that the linear filtering is not at all an optimal tool for suppressing speckle noise because it tends to suppress the noise at the expense of image details. Preservation of anatomical contents and edges in an ultrasound image is also an important criterion. Therefore, filtering the ultrasound image consists of two opposite actions. Speckle should be suppressed by smoothing operation in a homogeneous region of the image in one side; the anatomic details and the edges should be protected in other sides. Hence, it needs at least a near optimal solution for the speckle suppression problems.

A large number works are done in the field of speckle suppression. The most widely cited and applied speckle filters are Lee (Lee, 1980), Kuan (Kuan et al., 1987), Frost (Frost et al., 1982), median and adaptive weighted median filters (Loupas et al., 1989). Besides these, other speckle reduction filters are also available in the literature. These include homomorphic filter (Solbo and Eltoft, 2004; Saniie et al., 1989), geometric filter (Busse et al., 1995), anisotropic diffusion filters (Perona and Malik, 1990; Weickert et al., 1998; Jin et al., 2000; Black et al., 1998; Abd-Elmoniem et al., 2002), speckle reducing anisotropic diffusion filters or SRAD (Yongjian and Scott, 2002), Oriented speckle reducing anisotropic diffusion, Γ-map filters (Krissian and Fedrij, 2007), wavelet based denoising (Chang et al., 2000) etc. Lee, Frost and Kuan filters are adaptive linear filters which consider multiplicative speckle noise model and make use of local statistics. Lee and Frost filters have the same structure and Kuan filter is the generalization of Lee filters. However, the adaptive linear filters adapt the filter co-efficients based on the local statistics of the data within a fixed window. The window size for these filters determines the speckle suppression and the edge preservation capability. Large window has better smoothing capability (i.e. speckle suppression capability) and less edge preservation capability whereas small window has more edge preservation capability and less smoothing capability. Park et al., (1999) proposed adaptive windowing techniques for these filters. Whatever may be the fact these linear adaptive filters are unable to suppress speckle noise near high contrast edges.

Non-linear filter class has been proven to be very useful for suppressing impulsive type of noise. Median based filters are the most common non-linear filters having two most important intrinsic properties: edge preservation and efficient noise suppression with robustness against impulsive noise. These properties make these filters popular in image processing applications since edge preservation is essential in this field. The pure median filter is not, however, a perfect filtering operation, nor it is very flexible (Yin et al., 1996). Edge jitter (Bovik et al., 1987) and blotching (Bovik, 1987) may occur since the median filter uses only the rank-order information of the input data within a filter window discarding the spatial or temporal order information. The edge jitter and blotching effect in the median filter may reduce its efficiency of image details preservation. Weighted median filters utilize both rank-order and spatial/temporal-order information to improve the performance of the median filter. The weighted median filters put more emphasis on the center weights. This increases the edge preservation capability of the weighted median filter at the expense of noise suppression capability. Adaptive weighted median filter (AWMF) weighted median type properties with adjustable smoothing. Geomatric filtering algorithm (Busse et al., 1995) in a non-linear iterative algorithm which increment or decrement the pixel values in a neighborhood based on their relative values. Homomorphic filtering is similar to the logarithmic point operations used in histogram enhancement, where dominant bright pixels are de-emphasized (Loizou and Pattichis, 2008). Anisotropic diffusion filters, speckle reducing anisotropic diffusion filters or SRAD and Oriented speckle reducing anisotropic diffusion filters are based on anisotropic diffusion. They simultaneously perform noise reduction and contrast enhancement. The wavelet denoising filters utilize multi-scale filtering algorithms. They make use of realistic distribution of wavelet co-efficients (Donoho, 1995; Zong et al., 1998; Hao et al., 1999; Chang et al., 2000; Achim et al., 2001) where useful wavelet co-efficients are utilized. The useful co-efficients are determined by using shrinkage functions and thresholding techniques. The difference between different wavelet denoising depends on the choice of shrinkage functions and the thresholding techniques.

1.5 Problem Definition

Objective of this research is to enhance the quality of ultrasound image so that it becomes possible for an expert to detect smaller lesions from ultrasound image. The quality of an ultrasound image can be enhanced to such a level by increasing the resolution of the image and by reducing the speckle noise without blurring the edges of the image. The work is, therefore, carried out with following steps:

1. Developing a software-based super-resolution reconstruction technique, which may be performed on the scan data acquired from the ultrasound system. The scan data is the output of the logarithmic amplifier of the ultrasound system.
2. Study the performance and robustness of the proposed SR reconstruction in presence of noise.
3. Developing a new framework of speckle noise reduction, where speckle reduction is performed along with scan conversion. It leads better speckle suppression without increasing complexity. New framework of speckle noise suppression is applicable along with proposed SR reconstruction.
4. Developing a new speckle noise filter that is able to protect edges as well as reduce the speckle noise as far as possible. The filter is adapted to the proposed new framework of speckle reduction and the proposed super-resolution reconstruction technique.

Unlike other existing conventional ultrasound super-resolution techniques which are applied to scan converted ultrasound image, the proposed super-resolution technique is performed on the ultrasound scan data. Hence this is one kind of super-resolution scan conversion. In the conventional ultrasound super-resolution technique, SR is performed on the scan converted data. It is found that the proposed SR technique performs better than the conventional B-mode LR and conventional ultrasound SR reconstruction techniques in terms of quality metrics (defined in chapter 3). The proposed SR shows better object detectibility than that of Conventional B-mode LR and conventional SR reconstruction. Simulation and measurements shows that many small objects may be missed from B-mode LR images. Though no object is missed in the simulated phantom images for conventional SR reconstruction, the contrast of small objects are very poor and may be missed in presence of noise. The image quality of proposed SR reconstructed image at the higher depth is also better than that of the other two cases. The stability and the robustness of the proposed SR algorithm are verified through simulation. It is observed that the proposed SR is more stable and robust with respect to other two conventional methods in presence of noise. To study the performance, the proposed SR reconstruction technique is performed on the scan lines of simulated phantom images as well as on the scan lines of an ultrasound simulated liver image (using Field II software). The conventional SR is performed on the scan converted phantom and ultrasound simulated liver images. Both the cases show the superior performance of the proposed SR reconstruction algorithm over the other two conventional techniques.

The thesis also reports the simulation and the results of the new paradigms of the speckle reduction. Conventionally, the speckle reduction algorithms are applied on the scan converted ultrasound images. But speckle filtering algorithms may be applied either on the scan data before scan conversion or along with scan conversion. Though the quality of the image is better when the speckle reduction algorithms are applied before scan conversion, it increases the computational load. But it is possible to achieve better performance without increasing the computational load by applying the speckle filtering algorithms along with scan conversion. The new paradigms can also be applied with proposed SR reconstruction algorithm. The qualitative and quantitative measurements are done on the simulated images for all the paradigms. The numerical values of the quality metrics confirm the superiority of the new paradigms over the old one.

Finally, a high pass adaptive weighted median filter is proposed for reducing speckle noise from the ultrasound image. The proposed filter adaptively reduces the noise from the homogeneous region of the image and preserves both the positive and negative-slope edges of the objects in the image. The performance of the proposed filter is compared with four popular linear and non-linear filters. The numerical value assures the superiority of the proposed filter over the other. The proposed filter can be applied with the new paradigms and the proposed SR reconstruction techniques.

1.6 Layout of the thesis

A brief overview of the work carried out in this thesis is organized as follows:

Chapter 1 presents a brief introduction to ultrasound imaging system, the motivation and the objective of the thesis work.

Chapter 2 presents a brief review of the reported literature on the ultrasound resolution enhancement techniques and a review of speckle reduction techniques.

Chapter 3 presents a proposed super-resolution technique, which is applied to the data acquired in polar co-ordinate. Performance of the proposed algorithm is verified qualitatively and quantitatively by employing quality metrics. Detectibility of small objects is reported.

Chapter 4 presents the robustness of the proposed algorithm in presence of noise is demonstrated.

Chapter 5 presents a new framework of speckle reduction. The performance of the new framework is quantified with the help of the quality metrics. The new framework is adapted to the proposed super-resolution technique.

Chapter 6 presents a proposed adaptive weighted median filter that admits negative weights. The performance of the proposed filter is compared with other popular filtering techniques. The filter is applied to the super-resolution technique with new framework of speckle reduction.

Finally, **Chapter 7** is equipped with the summaries and important conclusions drawn from the thesis work. The thesis is concluded after suggesting the scope of future work.