1.1General review

Medical images are usually corrupted by noise in its acquisition and Transmission. The main objective of Image denoising techniques is necessary to remove such noises while retaining as much as possible the important signal features.

Ultrasonic imaging is a widely used medical imaging procedure because it is economical, comparatively safe, transferable, and adaptable. Though, one of its main shortcomings is the poor quality of images, which are affected by speckle noise. The existence of speckle is unattractive since it disgrace image quality and it affects the tasks of individual interpretation and diagnosis.

Accordingly, speckle filtering is a central pre-processing step for feature extraction, analysis, and recognition from medical imagery measurements. Previously a number of filters have been proposed for speckle mitigation. An appropriate method for speckle reduction is one which enhances the signal to noise ratio while conserving the edges and lines in the image. [12]

1.2 Problem of the statement

The usefulness of ultrasound imaging is degrading by the presence of signal dependent noise known as speckle. This noise is correlated multiplicative noise, that different from other types of noise because related to the signal and should be processed and removing without affecting important image features.

1.3 General objective

Give an overview about speckle noise, how to generate, has properties, and what the effectiveness of it on the ultrasound image.

1.4 Specific objectives

A- Learning about types of speckle reduction techniques in ultrasound imaging.

B- To carry out a comparative evaluation of despeckling filtering based on image quality evaluation metrics.

C- Proposed new method as a despeckle filter based on hybrid techniques.

1.5 Methodology

Images from The Children`s Hospital of Philadelphia database of fetal ultrasound image, and IBE Tech (Giza.Egypt) database of ultrasound image including liver and vagina. In the quantitative study, add speckle noise with different variance on ultrasound images and using a most importantly techniques to removing that noise.

A- **Modified Hybrid Median Filter(MHMF)**

This proposed technique is the modified version of the hybrid median filter. It works on the sub windows similar to hybrid median filter. By applied the median filter and max filter on noisy image.

B- **SRAD and Hybrid Median Filter**

The proposed filter is hybrid technique of despeckling which based on method that applied the SRAD filter and applied the hybrid median filter on sub band obtained from wavelet decomposition of noisy image then applied the total variation filter on resulted image.

Then the quality evaluation metrics was found from all methods to compare the performance of those filters.

1.6 thesis layout

The layout of this thesis consist of six chapters there are: chapter one include introduction, while chapter two involve theoretical background, literature review in chapter three , in chapter four materials and methodology, however in chapter five the results and discussion were viewed , finally chapter six is conclusion and future work.

2.1 Waves

There are two types of waves: Transverse waves: these waves are perpendicular to the direction of energy transfer, e.g., violin string .Longitudinal waves: these waves are parallel to the direction of energy transfer, e.g., a pulse from a piston in a cylinder, sound waves. [1]

2.2 Sound waves

Sound wave propagate by longitudinal motion compression/expansion) but not transverse motion (side-to-side) Can be modeled as weights connected by springs.[2]

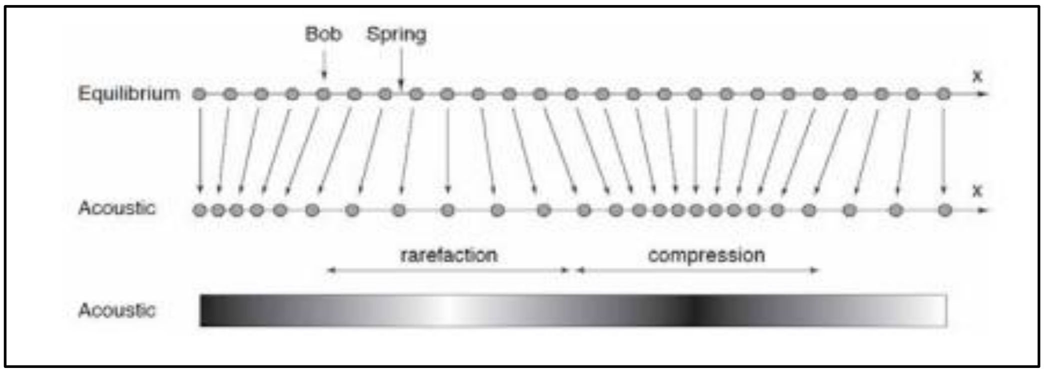


Figure2.1: Sound wave propagate [2]

The measuring of longitudinal waves in two ways Distance: the wave length Frequency: how many times per second the compression peak occurs at a point in space.

Frequency (**f**) and wavelength (**λ**) are related by the speed of sound in the medium: $V=f\ \lambda$ Generally speaking, V is related to the compressibility of the medium, slower in gasses, faster in liquids, and fastest in solids.[1]

2.2.1 Categories of sound

Infrasound (subsonic) below 20Hz

Audible sound 20-20,000Hz

Ultrasound above 20,000Hz

Non-diagnostic medical applications <1MHz

Medical diagnostic ultrasound >1MH.[3]

2.3 What is ultrasound?

Ultrasound or ultraSonography is a medical imaging technique that uses high frequency sound waves and their echoes. Known as a(**pulse echo technique**). The technique is similar to the echolocation used by bats and dolphins, as well as SONAR used by submarines etc. In Ultrasound, the following events happen

1. The ultrasound machine transmits high-frequency (1 to 12 megahertz) sound pulses into the body using a probe.
2. The sound waves travel into the body and hit a boundary between tissues (e.g. between fluid and soft tissue, soft tissue and bone).
3. Some of the sound waves reflect back to the probe, while some travel on further until they reach another boundary and then reflect back to the probe.
4. The reflected waves are detected by the probe and relayed to the machine.
5. The machine calculates the distance from the probe to the tissue or organ (boundaries) using the speed of sound in tissue (1540 m/s) and the time of the each echo's return (usually on the order of millionths of a second).
6. The machine displays the distances and intensities of the echoes on the screen, forming a two dimensional image. [3]

2.3.1 Types of ultrasound waves

Place with acoustic waves. Therefore, there will often be made references to optics.

There are two types of waves that are relevant. They can both be visualized in 2D with a square acrylic water tank placed on an overhead projector:

A. The plane wave which can be observed by shortly lifting one side of the container.

B. The spherical wave, which can be visualized by letting a drop of water fall into the surface of the water.

When the plane wave is created at one side of the water tank, one also is able to observe the reflection from the other side of the tank. The wave is reflected exactly as a light beam from a mirror or a billiard ball bouncing off the barrier of the table.

The spherical wave, that on the other hand, originates from a point source and propagates in all directions; it creates a complex pattern when reflected from the four sides of the tank [4]

2.4 Ultrasound's interaction with the medium

The interaction between the medium and the ultrasound emitted into the medium can be described by the following phenomena:

The echo that travels back to the transducer and thus gives information about the medium is due to two phenomena: reflection and scattering. Reflection can be thought of as when a billiard ball bounces off the barrier of the table, where the angle of reflection is identical to the angle of incidence. Scattering (spreading) can be thought of, when one shines strong light on the tip of a needle: light is scattered in all directions.[4]

In acoustics, reflection and scattering is taking place when the emitted pulse is travelling through the interface between two media of different acoustic properties, as when hitting the interface of an object with different acoustic properties.

Specifically, reflection is taking place when the interface is large relative to the wavelength (e.g. between blood and intima in a large vessel). Scattering is taking place when the interface is small relative to the wavelength (e.g. red blood cell).

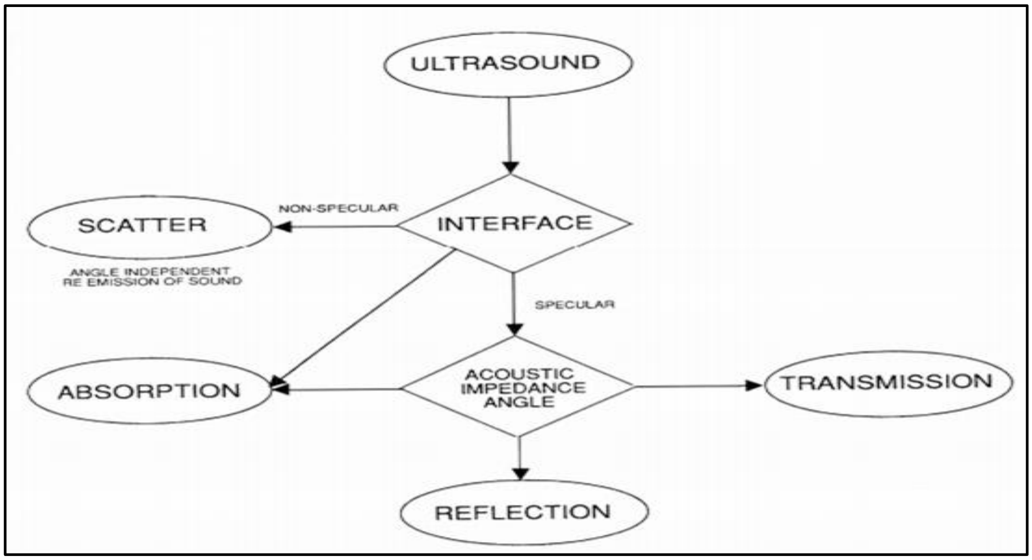
The abstraction of a billiard ball is not complete, however: In medical ultrasound, when reflection is taking place, typically only a (small) part of the wave is reflected. The remaining part is transmitted through the interface. This transmitted wave will nearly always be refracted, thus typically propagating in another direction.

The only exception is when the wave impinges perpendicular on a large planar interface: The reflected part of the wave is reflected back in exactly the same direction as it came from (like with a billiard ball) and the refracted wave propagates in the same way as the incident wave.

Reflection and scattering can happen at the same time, for instance, if the larger planar interface is rough. The smoother, the more it resembles pure reflection (if it is completely smooth, specular reflection takes place). The rougher, the more it resembles scattering.

When the emitted pulse travels through the medium, some of the acoustic (mechanical) energy is converted to heat by a process called Absorption. Of course, also the echoes undergo absorption.

Finally, the loss in intensity of the forward propagating acoustic pulse due to reflection, refraction, scattering and absorption is under one named attenuation.[4]



Figur2.2: Interaction of Ultrasound with Tissue[1]

2.4.1 Reflection

When a plane wave impinges on a plane, infinitely large, interface between two media of different acoustic properties, reflection and refraction occurs meaning that part of the wave is reflected and part of the wave is refracted. The wave thus continues its propagation, but in a new direction.

In the human body, approximate reflection can be observed atthe interface between blood and the intima of large vessel walls or at the interface between urine and the bladder wall.[4]

2.4.2 Scattering

While reflection takes place at interfaces of infinite size, scattering takes place at small objects with dimensions much smaller than the wavelength. Just as before, the specific acoustic impedance of the small object must be different from the surrounding medium. The scattered wave will be more or less spherical, and thus propagate in all directions, including the direction towards the transducer. The latter is denoted backscattering. [4]

Biologically, scattering can be observed in most tissue and especially blood, where the red blood cells are the predominant cells. They have a diameter of about 7 μm, much smaller than the wavelength of clinical ultrasound. [4]

2.4.3 Absorption

Absorption is the conversion of acoustic energy into heat. The mechanisms of absorption are not fully understood, but relate, among other things, to the friction loss in the springs, mentioned in Subsection pure absorption can be observed by sending ultrasound through a viscous liquid such as oil.[4]

2.4.4 Attenuation

The loss of intensity (or energy) of the forward propagating wave due to reflection, refraction, scattering and absorption is denoted attenuation. The intensity is a measure of the power through a given cross-section; thus the units are$\mathrm{W/m2}$. It can be calculated as the product between particle velocity and pressure:

$$\mathrm{I} = \rho \mathrm{u} = \frac{\rho^2}{\mathrm{z}} \qquad \textbf{(2.1)}$$

Where Z is the specific acoustic impedance of the medium . If I (0) is the intensity of the pressure wave at some reference point in space and I (x) is the intensity at a point x further along the propagation direction then the attenuation of the acoustic pressure wave can be written as:

$$\mathrm{I(x)} = \mathrm{I(0)e^{-\alpha x}} \qquad (\mathbf{2.2})$$

Where α(in units of m^{-1}) is the attenuation coefficient. αdepends on the tissue type (and for some tissue types like muscle, also on the orientation of the tissue fibers) and is approximately proportional with frequency. [4]

2.5 Imaging Techniques

The echo principle forms the basis of all of the commonly used diagnostic ultrasound techniques. These are:

US A-mode

US B-mode

US M-mode

Doppler techniques

US A-mode (amplitude modulation) is a one-dimensional technique.

The echoes received are displayed on a screen as vertical deflections this technique is rarely used today except for measurements.

US B-mode (brightness modulation) is a technique in which the echo amplitude is depicted as dots of different brightness (gray scale). It is mostly used as a two-dimensional B-scan to form a two-dimensional ultrasound image by multiple ultrasound beams, arranged successively in one plane. The images are built up by mechanically or electronically regulated scanning in a fraction of a second. The image rate of more than 15 per second enables an impression of “permanent” imaging during the examination (real time).

US M-mode (also sometimes referred to as TM-scan) is a way to display motion, e.g. of parts of the heart. The echoes produced by a stationary ultrasound beam are recorded over time, continuously .

Doppler techniques use the Doppler Effect as a further source of information: if the ultrasound waves are reflected by an interface moving towards the transducer or away from it, the reflected frequency will be higher or lower respectively than the transmitted frequency. The difference between the emitted and received frequencies is proportional to the speed of the moving relector. This phenomenon is called the Doppler Effect, and the difference is called the Doppler frequency or Doppler shift.

An unsharp masking filter was suggested in which the smoothing level is adjusted depending on the statistics of log compressed images

The above mentioned filters have difficulty in removing speckle near or on image edges recently proposed filter utilizing short line segments in different angular orientations and selecting the orientation that is most likely to represent a line in the image.

This technique poses a tradeoff between effective line enhancement and speckle Reduction. [5]

2.6 Ultrasound imaging system

Ultrasound is a widely used medical imaging modality. The use of ultrasound has expanded enormously over the last two decades, largely due to the fact that it is safe, allows real-time visualization of moving structures, suitable for many clinical applications, and is relatively inexpensive. However, like all imaging modalities, ultrasound is still subject to a number of inherent artifacts that compromise image quality and impair diagnostic utility.[17]

The construction of ultrasound B-mode image involves capturing the echo signal returned from tissue at the surface of piezoelectric crystal transducers. These transducers convert the ultrasonic RF mechanical wave into electrical signal. Convex ultrasound probes collect the echo from tissue in a radial form.

Each group of transducers is simultaneously activated to look at a certain spatial direction from which they generate a raw line signal (stick) to be used later for raster image construction. These sticks are then demodulated and logarithmically compressed to reduce their dynamic range to suit the commercial display devices. The final Cartesian image is constructed from the sampled sticks in a process called scan conversion.

Speckle reduction techniques can be applied on envelope detected data, log compressed data or on scan converted data. However, slightly different results will be produced for each data. In the compression stage some useful information about the imaged object may be deteriorated or even lost. However, any processing which works with envelope detected data has more information at its disposal and preserves more useful information.

Compared to processing the scan converted image, envelope detected data has fewer pixels and thus incurs lower computational cost.

For optimum result envelope detected data processing is preferred because some information that lost after the compression stage cannot be recovered by working with log compressed data or the scan converted image. However, the real time speckle reduction methods are applied on the scan converted image, since the scan converted image is always accessible where most commercial ultrasound systems do not output the envelope detected or log compressed data.[6]

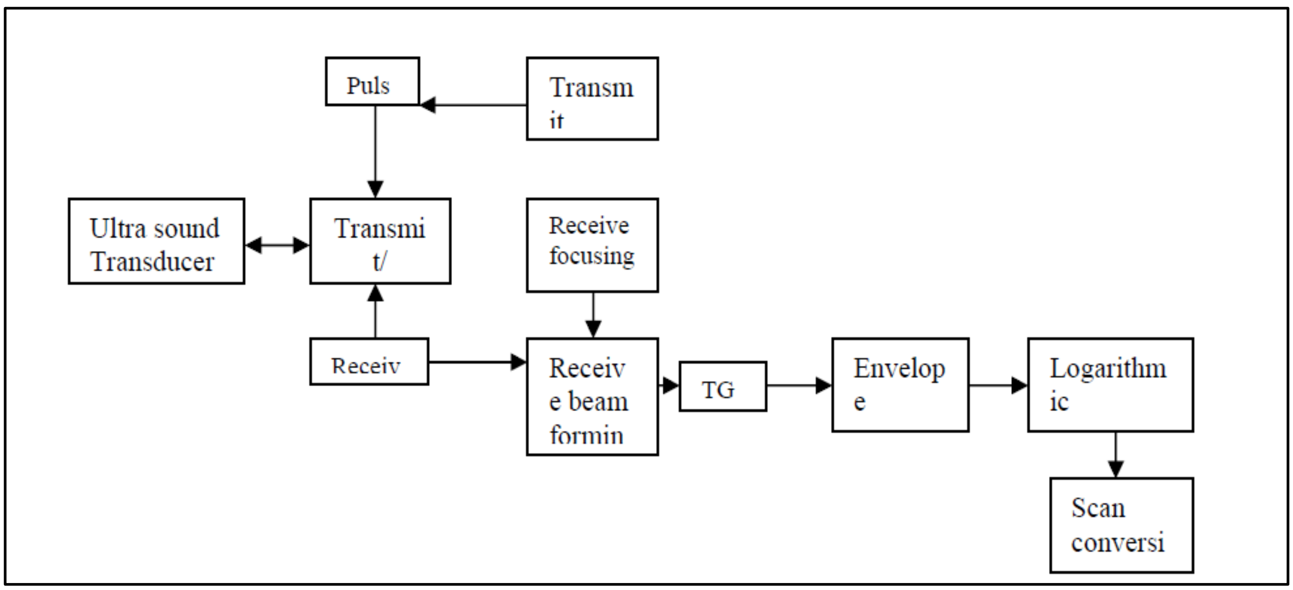


Figure 2.3: Block diagram of ultrasound imaging system. [6]

Comparative study

2.7 Introduction

Different types of noise. For example, the x-ray images are often corrupted by Poisson noise, while the ultrasound images are affected by Speckle noise. Speckle is a complex phenomenon, which degrades image quality with a back scattered wave appearance which originates from many microscopic diffused reflections that passing through internal organs and makes it more difficult for the observer to discriminate fine detail of the images in diagnostic examinations [18].Thus, denoising or reducing these speckle noise from a noisy image has become the predominant step in medical image processing.

2.8 Speckle noise in ultrasound imaging

Speckle is a form of locally correlated multiplicative noise that corrupts medical ultrasound imaging making visual observation difficult ,Speckle in US B-scans is seen as a granular structure which is caused by the constructive and destructive coherent interference of back scattered echoes from the scatters that are typically much smaller than the spatial resolution of medical ultrasound system. [8]

Speckle is not truly noise in the typical engineering sense since its texture often carries useful information about the image being viewed the speckle are essential information to track features, many cases the speckle noise deteriorates the image quality, degrades the fine details and edge definition.[8]

It also limits the contrast resolution, limiting the detectability of small, low contrast lesions in body. Speckle is always considered as a primary source of medical ultrasound imaging noise, and it should be filtered out without affecting important features of the image. [8]

2.9 Physical Properties and the Pattern of Speckle Noise

The speckle pattern, which is visible as the typical light, and dark spots the image is composed of, results from destructive interference of ultrasound waves scattered from different sites. The nature of speckle has been a major subject of investigation. When a fixed rigid object is scanned twice under exactly the same conditions, one obtains identical speckle patterns. Although of random appearance, speckle is not random in the same sense as electrical noise. However, if the same object is scanned under slightly different conditions, say, with a different transducer aperture, pulse length, or transducer angulations, the speckle patterns change. [8]

The most popular model adopted in the literature to explain the effects that occur when a tissue is insinuated is illustrated in Figure 3.1, where a tissue may be modeled as a sound absorbing medium containing scatters, which scatter the sound waves. These scatters arise from in homogeneity and structures approximately equal to or smaller in size than the wavelength of the ultrasound, such as tissue parenchyma, where there are changes in acoustic impedance over a microscopic level within the tissue. Tissue particles that are relatively small in relation to the wavelength (i.e., blood cells), and particles with differing impedance that lie very close to one another, cause scattering or speckling..

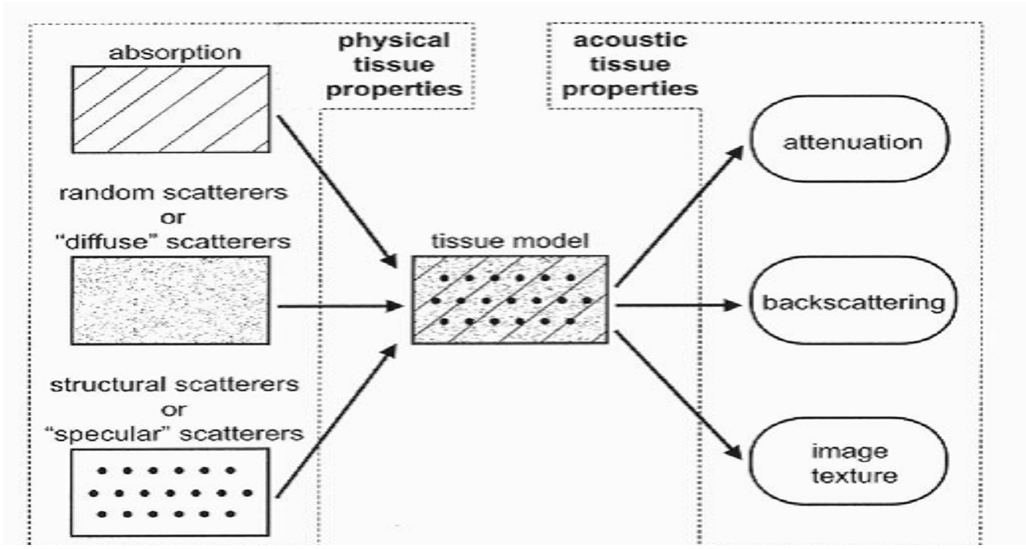


Figure 2.4 the usual tissue model in ultrasound imaging [3]

Absorption of the ultrasound tissue is an additional factor to scattering and refraction, responsible for pulse energy loss. The process of energy loss involving absorption, reflection, and scattering is referred to as attenuation, which increases with depth and frequency. Because a higher frequency of ultrasound results in increased absorption, the consequence is a decrease in the depth of visualization. [6]

The nature of the speckle pattern can be categorized into one of three classes according to the number of scatters per resolution cell or the so called scatter number density (SND), spatial distribution and the characteristics of the imaging system itself. These classes are described as follows:

1. FFS (Fully formed speckle) pattern, which occurs when many fine randomly distributed scattering sites exist within the resolution cell of the pulse-echo system. In this case, the amplitude of the backscattered signal can be modeled as a Rayleigh distributed random variable with a constant SNR of 1.92. Under such conditions, the textural features of the speckle pattern represent a multivariate

signature of the imaging instrument and its point spread function. Blood cells are typical examples of this type of scatterers.

2. Non randomly distributed with long-range order (NRLR). Examples of this type are the lobules in liver parenchyma. It contributes a coherent or specular backscattered intensity that is in itself spatially varying. Due to the correlation between scatterers, the effective number of scatterers is finite. This situation can be modeled by the K-distribution. This type is associated with SNR below 1.92. It can also be modeled by the Nakagami distribution.

3. Non randomly distributed with short-range order (NRSR). Examples of this type include organ surfaces and blood vessels. When a spatially invariant coherent structure is present within the random scatterer region, the probability density function (PDF) of the backscattered signals becomes close to the Rician distribution. This class is associated with SNR above 1.92.[6]

2.10 Need for despeckling

Speckle is considered as the dominant source of noise in ultrasound imaging and should be processed without affecting important image features, certain speckle diagnostic information and should be retained.

The main purposes for speckle reduction in medical ultrasound imaging are:

1. To improve the human interpretation of ultrasound images – speckle reduction makes an ultrasound image cleaner with clearer boundaries.

2. Despeckling is a preprocess step for many ultrasound image processing tasks such as segmentation and registration – speckle reduction improves the speed and accuracy of automatic and semiautomatic segmentation & registration. [6]

2.11 Speckle reduction methods

Speckle filtering methods over the years, several techniques have been proposed to despeckle ultrasound images. There are two major classifications of speckle reduction filters namely compounding method and post acquisition method [18]

2.11.1 Compounding methods

In this method a series of ultrasound images of the same target are acquired from different scan directions and with different transducer frequencies or under different strains. Then the images are averaged to form a composite image. The compounding method can improve the target detectability but they suffer from degrade spatial resolution and increased system complexity. [6]

2.11.2 Post acquisition methods

This method do not require many hardware modification .The post acquisition image processing technique falls under two categories (1) Single scale spatial filtering (2) Multi scale Methods.[6]

2.11.2.1 Single scale spatial filtering Methods

A speckle reduction filter that changes the amount of smoothing according to the ratio of local variance to local mean was developed .in that method smoothing is increased in homogeneous region where speckle is fully developed and reduced or even avoided in other regions to preserve details.

An unsharp masking filter was suggested in which the smoothing level is adjusted depending on the statistics of log compressed images. The above mentioned filters have difficulty in removing speckle near or on image edges.

Recently proposed filter utilizing short line segments in different angular orientations and selecting the orientation that is most likely to represent a line in the image .This technique poses a trade-off between effective line enhancement and speckle reduction.

Numbers of Region growing based spatial filtering methods have been proposed. In these methods it is assumed that pixels that have similar gray level and connectivity are related and likely to belong to the same object or region. After all pixels are allocated to different groups, spatial filtering is performed based on the local statistics of adaptive regions whose sizes and shapes are determined by the information content of the image .The main difficulty in applying region growing based methods is how to design appropriate similarity criteria for region growing. Different types of filters are used in the application of despeckling in ultrasound imaging. The most commonly used types of filters are:

A. ***Mean Filter* -** It is simple and intuitive filter. It does not remove speckle noise at whole but reduces at some extend. It works on average basis that is the centre pixel is replaced by the average of the all pixels. Hence this filter gives blurring effect to the images, so it is least satisfactory method to remove speckle noise as it results in loss of details.[19]

B. ***Median Filters*** it is non linear filter. It gives quite better result than the mean filter. Here center pixel is replaced by the median value of all pixels and hence produces less blurring. Due to this nature it is used to reduce impulsive speckle noise. Advantage is it preserves the edges. Disadvantage is extra time needed for computation of the median value for sorting N pixels, the temporal complexity is O (N log N). Median filter follows algorithm as follows: 1. Take a 3 × 3 (or 5×5 etc.) region centered around the pixel (i, j). 2. Sort the intensity values of the pixels in the region into ascending order 3. Select the middle value as the new value of pixel (i, j).[19]

2.11.2.2 Multi scale methods

Several multi scale methods based on wavelet and pyramid have been proposed for speckle reduction in ultrasound imaging. It classified to:

2.11.2.2.1 Wavelet based speckle reduction methods

The wavelet based speckle reduction method usually include

(1) Logarithmic transformation.

(2) Wavelet transformation.

(3) Modification of noisy co efficient using shrinkage function.

(4) Invert wavelet transform.

(5) Exponential transformation. This method can be classified into three groups:

1. ***Thresholding methods*** **-** The wavelet coefficients smaller than the predefined threshold are regarded as contributed by noise and then removed. The thresholding techniques have difficulty in determining an appropriate threshold.

2. ***Bayesian estimation methods*** **–** This Method approximates the noise free signal based on the distribution model of noise free signal and that of noise. Thus, reasonable distribution models are crucial to the successful application of these techniques to medical ultrasound imaging.

3. ***Coefficients correlation methods* -** This is an undecimated or over complete wavelet domain denoising method which utilizes the correlation of useful wavelet coefficients across scales. However this method does not rely on the exact prior knowledge of the noise distribution and this method is more flexible and robust compared to other wavelet based methods. [6]

2.11.2.2.2 Pyramid based speckle reduction methods

Pyramid transform has also been used for reducing speckle. Approximation and interpolation filters in pyramid transform have low pass properties so that pyramid transform does not require quadrature mirror filters unlike sub band decomposition in wavelet transform.

A ratio laplacian pyramid was introduced by considering the multiplicative nature of speckle. This method extended the conventional Kaun filter to multi scale domain by processing the interscale layers of the ratio laplacian pyramid. But this method differs from the need to estimate the noise variance in each interscale layers.

A speckle reduction method based on non linear diffusion filtering of band pass ultrasound images in the laplacian pyramid domain has been proposed which effectively suppresses the speckle while preserving edges and detailed features.[6]

2.12 Speckle noise modeling

To be able to derive an efficient despeckle filter, a speckle noise model is needed. The speckle noise model for ultrasound images may be approximated as multiplicative. The signal at the output of the receiver demodulation module of the ultrasound imaging system may be defined as

$$Y_{i,j} = x_{i,j}\, n_{i,j} + a_{i,j}. \tag{2.3}$$

where $Y_{i,j}$ represents the noisy pixel in the middle of the moving window, $x_{i,j}$ represents the noise free pixel, $n_{i,j}$ and $a_{i,j}$ represent the multiplicative and additive noise, respectively, and i, j are the indices of the spatial locations that belong in the 2D space of real numbers, i, j Î R2.

Despeckling is based on estimating the true intensity $x_{i,j}$ as a function of the intensity of the pixel $Y_{i,j}$ and some local statistics calculated on a neighborhood of this pixel.

the histogram of amplitudes within the resolution cells of the envelope-detected RF signal backscattered from a uniform area with a sufficiently high scatter density has a Rayleigh distribution with mean proportional to the standard deviation s (with m/s= 1.91). This implies that speckle could be modeled as multiplicative noise.

However, the signal processing stages inside the scanner modify the statistics of the original signal, i.e., the logarithmic compression. The logarithmic compression is used to adjust the large echo dynamic range (50–70 dB) to the number of bits (usually 8) of the digitization module in the scan converter. More specifically, logarithmic compression affects the high-intensity tail of the Rayleigh and Rician probability density functions more than the low-intensity part. As a result, the speckle noise becomes very close to the white Gaussian noise corresponding to the uncompressed Rayleigh signal. In particular, it should be noted that speckle is no longer multiplicative in the sense that, on homogeneous regions, where $x_{i,j}$ can be assumed constant, the mean is proportional to the variance (m » s 2) rather than the standard deviation (m » s). In this respect, the speckle index C will be for the log-compressed ultrasound images, i.e., C = s 2/m.

Referring back to Eq. (2.3), since the effect of the additive noise is considerably smaller compared with that of the multiplicative noise, it may be written as

$$Y_{i,j} \approx x_{i,j}\, n_{i,j} \tag{2.4}$$

Thus, the logarithmic compression transforms the model in Eq. (2.4) into the classical signal in the additive noise form as

$$\log(Y_{i,j}) = \log(x_{i,j}) + \log(n_{i,j}) \tag{2.5}$$

and

$$g_{i,j} = f_{i,j} + n\, l_{i,j}. \tag{2.6}$$

For the rest of the book, the term $\log(Y_{i,j})$, which is the observed pixel on the ultrasound image display after logarithmic compression, is denoted as $g_{i,j}$, and the terms $\log(x_{i,j})$ and $\log(n_{i,j})$, which are the noise-free pixel and the noise component after logarithmic compression, are denoted as $f_{i,j}$ and $n\, l_{i,j}$, respectively [see Eq. (2.6)].[8]

2.12 Despeckling filter

In this section several despeckling techniques such as Median, hybrid median filter, Modified Hybrid Median Filter, geometric filtering, linear scaling filter, Anisotropic diffusion filtering, speckle reducing Anisotropic diffusion filtering, wavelet filter are discussed. [9]

2.12.1 Nonlinear filtering

Non linear filtering is based on non linear operation involving the pixels in a neighborhood .for example, letting the center pixel in the moving window be equal to the maximum pixel in its neighborhood is anon linear filtering operation.[8]

2.12.1.1 Median filter

It is a spatial domain filter. A median filter generally smoothens the image to reduce noise and at the same time it preserves edges. It replaces the middle pixel in the window with the median-value of its neighbors. This filter does not create new pixel value. Instead it chooses the median value which is selected from the neighborhood. This will not affect other pixels significantly. Hence this filter preserves the edges, this filter is relatively slow, even with fast sorting algorithms such as quick sort. The median filter does not blur the contour of the objects.

2.12.1.2 Hybrid median filter

The hybrid median filter is another modification of median filter. This filter is also called as corner preserving median filter is a three-step ranking operation. In a 5X5 pixel neighborhood, pixels can be ranked in two different groups as shown in fig2.4

The median values of the 45̊ neighbors forming an "X" and the 90̊ neighbors forming a "+" are compared with the central pixel and the median value of that set is then saved as the new pixel value.

The three step ranking operation does not impose a serious computational penalty as in the case of median filter. Each of the ranking operations is for a much smaller number of values than used in a square region of the same size. For example, the 5 pixel wide neighborhood used in the examples contains either 25 (in the square neighborhood) which must be ranked in the traditional method. In the hybrid method, each of the two groups contains only 9 pixels, and the final comparison involves only three values. Even with the additional logic and manipulation of values, the hybrid method is faster than the conventional median. This median filter overcomes the tendency of median and truncated median filters to erase lines which are narrower than the half width of the neighborhood and to round corners [10]

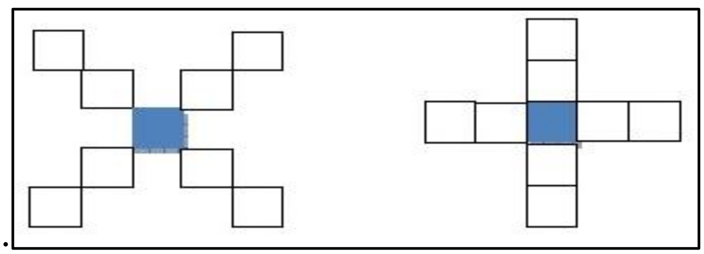


Figure 2.5Diagram of neighborhood pixels used in the

Hybrid Median Filter.[10]

2.12.1.3 Geometric filtering

The concept of the geometric filtering is that speckle appears in the image as narrow walls and valleys. The geometric filter, through iterative repetition, gradually tears down the narrow walls (bright edges) and fills up the narrow valleys (dark edges), thus smearing the weak edges that need to be preserved.

The geometric filtering uses a nonlinear noise reduction technique. It compares the intensity of the central pixel in a 3 × 3 neighborhood with those of its eight neighbors and, based upon the neighborhood pixel intensities, it increments or decrements the intensity of the central pixel such that it becomes more representative of its surroundings.[11]

Can see that although the result obtained by D, given poor performance for removing the speckle noise, it is lead to increasing the contrast significantly of the image.

2.12.1.4 Linear scaling filter (DsFca, DsFls)

The linear scaling gray level (DsFca) filters despeckle the image through linear scaling of the gray-level values. In a window of [5*5] pixels, compute the mean of all pixels whose difference in the gray level with the intensity $g_{i,j}$ (the middle pixel in the moving window) is lower than or equal to a given threshold ϑ . Assign this value to the gray level $g_{i,j}$ with $\vartheta = \alpha * g_{max}$, where g_{max} is the maximum gray level of the image and $\alpha = [0,1]$, Best results can be obtained with $\alpha = 0{,}1$.

The linear scaling (DsFls)has high degree of blurring and was affect on gray level because In a window of [5*5] pixels it is compute the mean of all pixels how's difference in the gray level with the intensity (the middle pixel in the moving window) is lower than or equal to a given threshold.[8]

2.12.2 Diffusion Filtering

Diffusion filters remove noise from an image by modifying the image via solving a partial differential equation (PDE). The smoothing is carried out, depending on the image edges and their directions. [11]

2.12.2.1 Anisotropic diffusion filtering

Anisotropic diffusion is an efficient, nonlinear technique for simultaneously performing contrast enhancement and noise reduction. It smoothes homogeneous image regions but retains image edges without requiring any information from the image power spectrum. Thus, it may be applied directly to logarithmic-compressed images. Modifying the image according to this linear isotropic diffusion equation is equivalent to filtering the image with a Gaussian filter.

Diffusion is a physical process for balancing concentration changes. In image processing the image intensity can be seen as concentration.

The noise can be modeled as little concentration in homogeneity. This in homogeneity could be smoothened by diffusion.

Consider applying the isotropic diffusion equation given by $d\, g_{i,j}, t\, /dt = \mathrm{div}\, (d\nabla g)$ using the original noisy image $g_{i,j}, t=0$ as the initial condition, where $g_{i,j}, t=0$ is an image in the continuous domain, $_{i,j}$ specifies spatial position, t is an artificial time parameter, d is the diffusion constant, and ∇g is the image gradient. [11]

2.12.2.2 Speckle reducing anisotropic diffusion (SRAD)

Anisotropic Diffusion is a nonlinear smoothing filter which uses a variable conductance term that controls the contrast of the edges that influence the diffusion. This filter has the ability to preserve edges, while smoothing the rest of the image to reduce noise. The anisotropic diffusion has been used by several researchers in image restoration and image recovery . SRAD is an edge-sensitive Partial Differential Equation (PDE) anisotropic diffusion approach to reduce speckle noise in images.

The anisotropic filtering in SRAD simplifies image features to improve image segmentation and smoothes the image in homogeneous area

While preserving edges and enhances them. It reduces blocking artifacts by deleting small edges amplified by homomorphic filtering.

SRAD equation for an image u is given by the Equation.

$$SRAD (u') = ut+1 = ut + (\Delta t/4) \text{ div } (g (ICOV (u')) x\Delta u' \qquad (2.7)$$

Where t is the diffusion time index, f't is the time step responsible for the convergence rate of the diffusion process (Normally in the range 0.05 to 0.25), g(.) is the diffusionΔ function and is given by equations :[12]

$$G (ICOV (u')) = e-(P) \qquad (2.8)$$

$$P = \frac{\left[\frac{ICOV\ (u')}{q^t}\right]^2}{1+(q^t)^2} \qquad (2.9)$$

2.12.3 Wavelet filtering

Wavelet filtering exploits the decomposition of the image into the wavelet basis and zeros out the wavelet coefficients to despeckle the image. Wavelets are simply mathematical functions and these functions analyze data according to scale or resolution. We use a processing which is carried out without implementing very complex transform. It consists of eliminating certain frequencies in order to eliminate any existing noise. Since we know that in an image HH, LH and HL components contain most of the noise. We can eliminate noise by eliminating those components. This does not mean that all noise present in the image is eliminated. Some details in the image may also be lost. [9]

The wavelet techniques are widely used in the image processing, such as the image compression, image de-noising. It has been shown that its performance of image processing is better than the methods based on other linear transformation.

The wavelet de-noising method decomposes the image into the wavelet basis and shrinks the wavelet coefficients in order to despeckle the image. From the noisy image, global soft threshold coefficients are calculated for every decomposition level. After the thresholding, the image is reconstructed by inverse wavelet transforming and the despeckled image is derived. After the wavelet transformation, the signal energy will only concentrate on several wavelet coefficients and the majority of the coefficients will become zeros. It has been proved that the simple wavelet de-noising methods could provide a almost optimal request to the polynomial piecewise signals. The errors of the estimation [7]

$$E\,[\|X - \dot{X}\|]\,\hat{}2/N \text{ is the same order of } O\,(\log\hat{}2\ N/N). \tag{2.10}$$

2.12.4 Total Variation

Total variation denoising (TVD) is an approach for noise reduction developed so as to preserve sharp edges in the underlying signal. Unlike a conventional low-pass filter, TV denoising is defined in terms of an optimization problem. The output of the TV denoising 'filter' is obtained by minimizing a particular cost function.

$$U=f-P_{GA}(f) \tag{2.11}$$

Where is the noisy image, U is the image we want to restore from f-P_{GA}(f) is the orthogonal projection of f on GA and the space G is proposed by Meyer for modeling oscillating patterns .[20]

The TV filter is now considered to be among the most successful methods for image restoration and edge enhancement, mainly, because of its capability of filtering out the noise without blurring or distorting the most universal and crucial features of images – edges.[20]

2.13 Limitation of despeckle filtering techniques

Despeckling is always a tradeoff between noise suppression and loss of information, which is something that experts are very concerned about. It is, therefore, desirable to keep as much important information as possible. The majority of speckle reduction techniques have certain limitations that can be briefly summarized as follows.

They are sensitive to the size and the shape of the window. The use of different window sizes greatly affects the quality of the processed images. If the window is too large, over smoothing will occur, subtle details of the image will be lost in the filtering process, and edges will be blurred.

On the other hand, a small window will decrease the smoothing capability of the filter and will not reduce the speckle noise, thus making the filter not effective.

In homogenous areas, the larger the window size, the more efficient the filter in reducing the speckle noise. In heterogeneous areas, the smaller the window size, the more it is possible to keep subtle image details unchanged. Our experiments showed that a [3*3] window size is a fairly good choice.

Some of the despeckle methods based on window approaches require thresholds to be used in the filtering process, which have to be empirically estimated. There are a number of thresholds introduced in the literature, which include gradient thresholding, soft or hard thresholds, nonlinear thresholds, and wavelet thresholds. The inappropriate choice of a threshold may lead to average filtering and noisy boundaries, thus leaving the sharp features unfiltered.

3-Most of the existing despeckle filters do not enhance the edges, but they only inhibit smoothing near the edges. When an edge is contained in the filtering window, the coefficient of variation will be high, and smoothing will be inhibited. Therefore, speckle in the neighborhood of an edge will remain after filtering. They are not directional in the sense that in the presence of an edge, all smoothing is precluded. Instead of inhibiting smoothing in directions perpendicular to the edge, smoothing in directions parallel to the edge is allowed.

Different evaluation criteria for evaluating the performance of despeckle filtering are used by different studies. Although most of the studies use quantitative criteria like the MSE and the speckle index (C), there are additional quantitative criteria like texture analysis and classification, image quality evaluation metrics, and visual assessment by experts that could be investigated. [8]

2.14 Image quality evaluation metrics

Objective evaluation of the image quality on ultrasound images is a comprehensive task due to the relatively low image quality compared to other imaging techniques. It is desirable to objectively determine the quality of ultrasound images since quantification of the quality removes the subjective evaluation which can lead to varying results.

Differences between the original, $g_{i,j}$, and the despeckled $f_{i,j}$, images were evaluated using image quality evaluation metrics.

$$\text{MSE}=\frac{1}{MN}\sum_{i=1}^{M}\sum_{j=1}^{N}(g_{i,j}-f_{i,j})^2 \qquad (2.12)$$

This measures the quality change between the original and processed image in an MxN window.

The root mean square error (RMSE), which is the square root of the squared error averaged over an MxN window:

$$\text{RMSE}=\sqrt{\frac{1}{MN}\sum_{i=1}^{M}\sum_{j=1}^{N}(g_{i,j}-f_{i,j})^2}. \qquad (2.13)$$

The signal-to-noise ratio (SNR) is given by:

$$\mathrm{SNR}= 10\log_{10}\frac{\sum_{i=1}^{M}\sum_{j=1}^{N}\left(g_{i,j}{}^{2}+f_{i,j}{}^{2}\right)}{\sum_{i=1}^{M}\sum_{j=1}^{N}(g_{i,j}-f_{i,j})^{2}} \tag{2.14}$$

The peak SNR (PSNR) is computed using:

$$\mathrm{PSNR}=10\log_{10}\frac{\mathrm{MSE}}{g_{\max}^{2}} \tag{2.15}$$

Where $g_{\max}^{2}$ is the maximum intensity in the unfiltered image.

The PSNR is higher for a better-transformed image and lower for a poorly transformed image. It measures image fidelity, which is how closely the despeckled image resembles the original image.

The structural similarity index between two images is given by:

$$\mathrm{SSIM}=\frac{\left(2\bar{g}\bar{f}+c_{1}\right)\left(2\sigma_{gf}+c_{2}\right)}{\left(\bar{g}^{2}+\bar{f}^{2}+c_{1}\right)\left(\sigma_{g}{}^{2}+\sigma_{f}{}^{2}+c_{2}\right)},\ -1<\mathrm{SSIM}<1, \tag{2.16}$$

Where $c_1 = 0.01dr$ and $c_2 = 0.03dr$, with $dr = 255$ representing the dynamic range of the ultrasound images. The range of values for the SSIM lies between −1, for a bad and 1 for a good similarity between the original and despeckled images, respectively. It is computed, for a sliding window of size 8 × 8 without overlapping. [19]

Literature review

Hussien Ali Hussien , Software Implementation Of Hybrid Median Filter (2013), In this paper the algorithm of hybrid median filter is presented and then software is implemented using Matlab package.

Jyoti Jaybhay and Rajveer Shastri , a study of speckle noise reduction filters (June 2015) , Different filters have been developed as Mean and Median filters, Srad filter. This paper re views filters used to remove speckle noise.

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Nishtha Attlas Dr. Sheifali Gupta, Wavelet Based Techniques for Speckle Noise Reduction in Ultrasound Images, (February 2014),This paper presents a review of various techniques for reduction of speckle noise in ultrasound images.

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Christos P. Loizou and Constantinos S. Pattichis , Despeckle Filtering Algorithms and Software for Ultrasound Imaging,(2008). The goal for this book is to introduce the theoretical background (equations), the algorithmic steps, and the MATLAB™ code for the following group of despeckles filters: linear filtering, nonlinear filtering, anisotropic diffusion filtering and wavelet filtering.

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S.Kalaivani Narayanan and R.S.D.Wahidabanu, A View on Despeckling in Ultrasound Imaging, (September 2009) , the objective of this paper is to give an overview about types of speckle reduction techniques in ultrasound imaging.

R.vanithamani , G.umamaheswari, M.ezhilarasi , Modified Hybrid Median Filter for Effective Speckle Reduction in Ultrasound Images, This paper proposes a statistical filter, which is a modified version of Hybrid Median Filter for speckle Reduction.

Ehsan Nadernejad, Mohammad Reza Karami , Despeckle Filtering in Medical Ultrasound Imaging, (2009), this paper proposes filtering techniques for the removal of speckle noise from the image.

4. Materials and Methodology

This chapter explain the materials and steps of hybrid technique which it is improvement of the disadvantage of hybrid filter and SRAD filter in using as despeckling filter and better for preserving the image texture

4.1 First proposed method Modified Hybrid Median Filter (MHMF)

This proposed filter is the modified version of the hybrid median filter explained below. It works on the sub windows similar to hybrid median filter . By applied the median filter for the result of mean filter for the value of the 45̊ neighbor's forming an "X" and the mean filter for the result of median filter for the for the value of the 90̊ neighbor's forming a "+"and the max value of that set is then saved as the new pixel value.

Algorithm:-

Step1: Find the median for the pixels marked as *R* then applied mean on the resulted pixels in the 5x5 window (A).

step2: : Find the mean for the pixels marked as *D* then applied median on the resulted pixels in the 5x5 window (B).

step3: Finally compute M_1

$M_1 = max\ (C, A, B)$.

Step4: filter value $y_{i,j}=M_1$

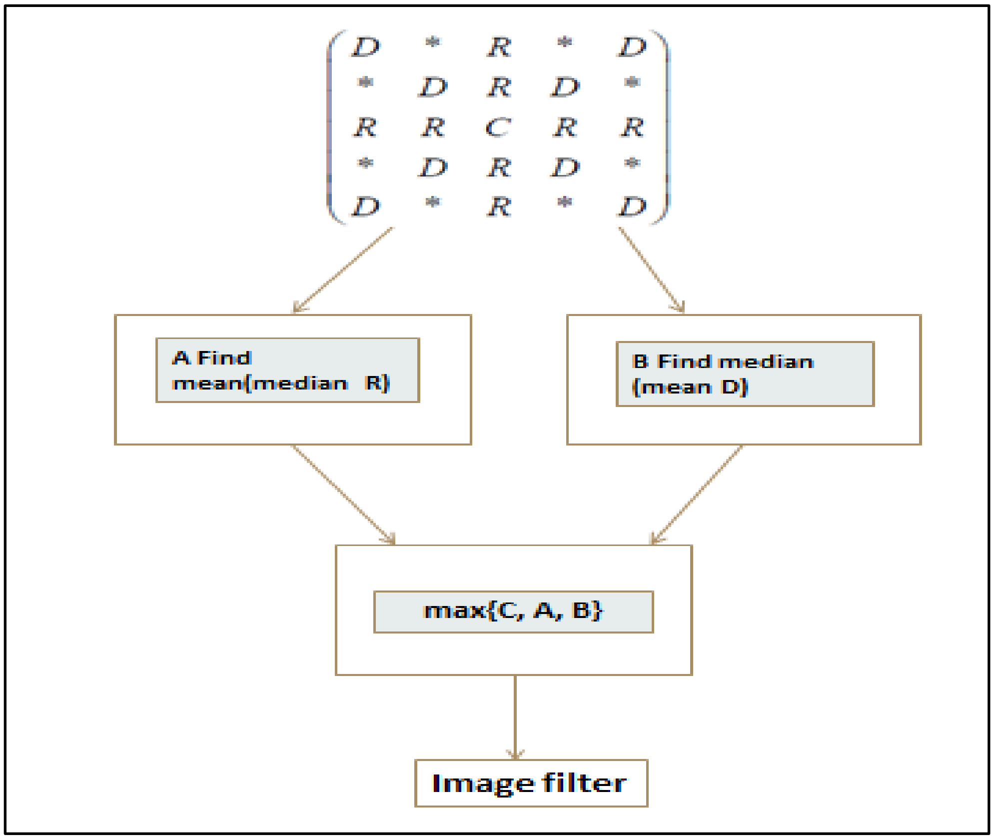


Figure 4.1: Algorithm of modified Hybrid Median Filter

4.2 Second proposed method (SRAD hybrid median filter)

4.2.1 Wavelet transforms

Wavelet transform (WT) represents an image as a sum of wavelet functions (wavelets) with different locations and scales. Any decomposition of an image into wavelets involves a pair of waveforms: one to represent the high frequencies corresponding to the detailed parts of an image (wavelet function ψ) and one for the low frequencies or smooth parts of an image (scaling function Ø).[14]

Wavelet analysis represents the next logical step: a windowing technique with variable-sized regions. Wavelet analysis allows the use of long time intervals where we want more precise low-frequency information, and shorter regions where we want high-frequency information. [14]

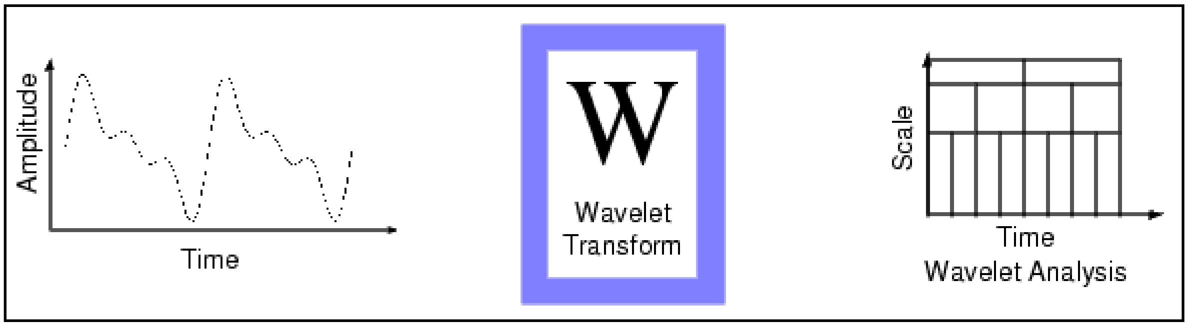


Figure 4.2: Wavelet Transform on a signal [14]

Wavelet filtering exploits the decomposition of the image into the wavelet basis and zeros out the wavelet coefficients to despeckle the image. Wavelets are simply mathematical functions and these functions analyze data according to scale or resolution.

We use a processing which is carried out without implementing very complex transform. It consists of eliminating certain frequencies in order to eliminate any existing noise. Since we know that in an image HH, LH and HL components contain most of the noise. We can eliminate noise by eliminating those components. This does not mean that all noise present in the image is eliminated. Some details in the image may also be lost. [14]

4.2.2 Wavelet Decomposition

The multiscale wavelet analysis has a very useful property of space and scale localization. It has variety significant applications in signal processing problems such as image coding and image de-noising.

The principle of the wavelet decomposition is to decompose the original raw particle image into several components: one low-resolution and high resolution, it called approximation low-pass filter and details High-pass filter. The noise is mainly appeared in the details.

In practice, multi-resolution investigation is carried out using 4 channel filter banks composed of a low-pass and a high-pass filter and each filter bank is then sampled at a half rate of the previous frequency.

By repeating this process, it is possible to obtain wavelet transform of any order. The down sampling procedure keeps the scaling parameter at constant throughout the successive wavelet transforms so that it benefits for simple computer implementation. In the case of an image, the filtering is implemented in a separable way by filtering the lines and columns. the discrete wavelet transform (DWT) of an image consists of a four frequency sub-band for each level of decomposition (LL,HL,LH,HH) as shown in. [15]

LL1	HL1
LH1	HH1

Figure 4.3: One level image decomposition by using DWT

The HH sub-band gives the diagonal information of the US image; the HL sub-band gives the horizontal features while the LH sub-band represents the vertical structures of the US image. The LL sub-band is the low-resolution residual consisting of low frequency components.[13]

The basic Procedure for all thresholding method is as follows:

1 · Calculate the DWT of the image.

2· Threshold the wavelet coefficients. (Threshold may be universal or sub band adaptive)

3 · Compute the IDWT to get the denoised estimate.

There are two thresholding functions frequently used, i.e. a hard threshold, a soft threshold. The hard-thresholding is described as:

$$\eta 1\,(w) = wI\,(|\,w\,| > T) \tag{5.1}$$

Where w is a wavelet coefficient, T is the threshold. The Soft-thresholding function is described as:

$$\eta 2\,(w) = (w - \mathrm{sgn}\,(w)\,T)\,I\,(|\,w\,| > T) \tag{5.2}$$

Where sgn(x) is the sign function of x. The soft-thresholding rule is chosen over hard-thresholding, as for as speckle (multiplicative nature) removal is concerned a preprocessing step consisting of a logarithmic transform is performed to separate the noise from the original image. Then different wavelet shrinkage approaches are employed. The different methods of wavelet threshold denoising differ only in the selection of the threshold. [16]

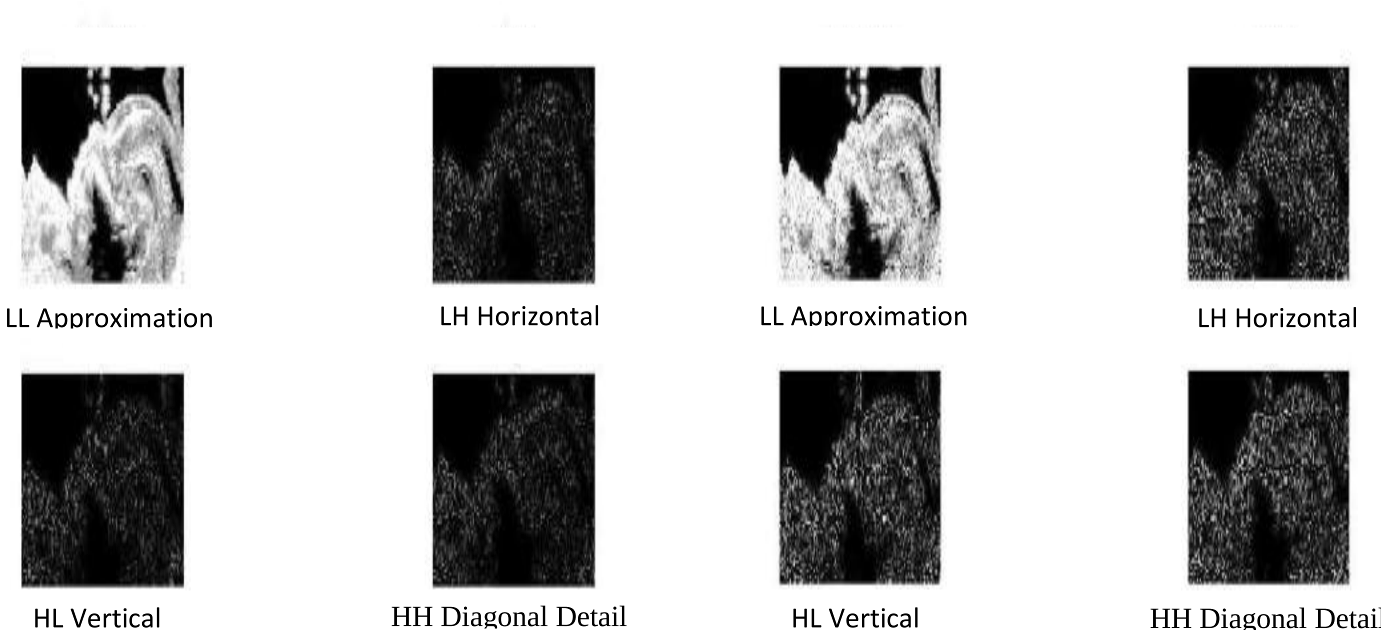


Figure 5.4: Wavelet decomposition of the fetal image noise =0.05 and 0.5, respectively

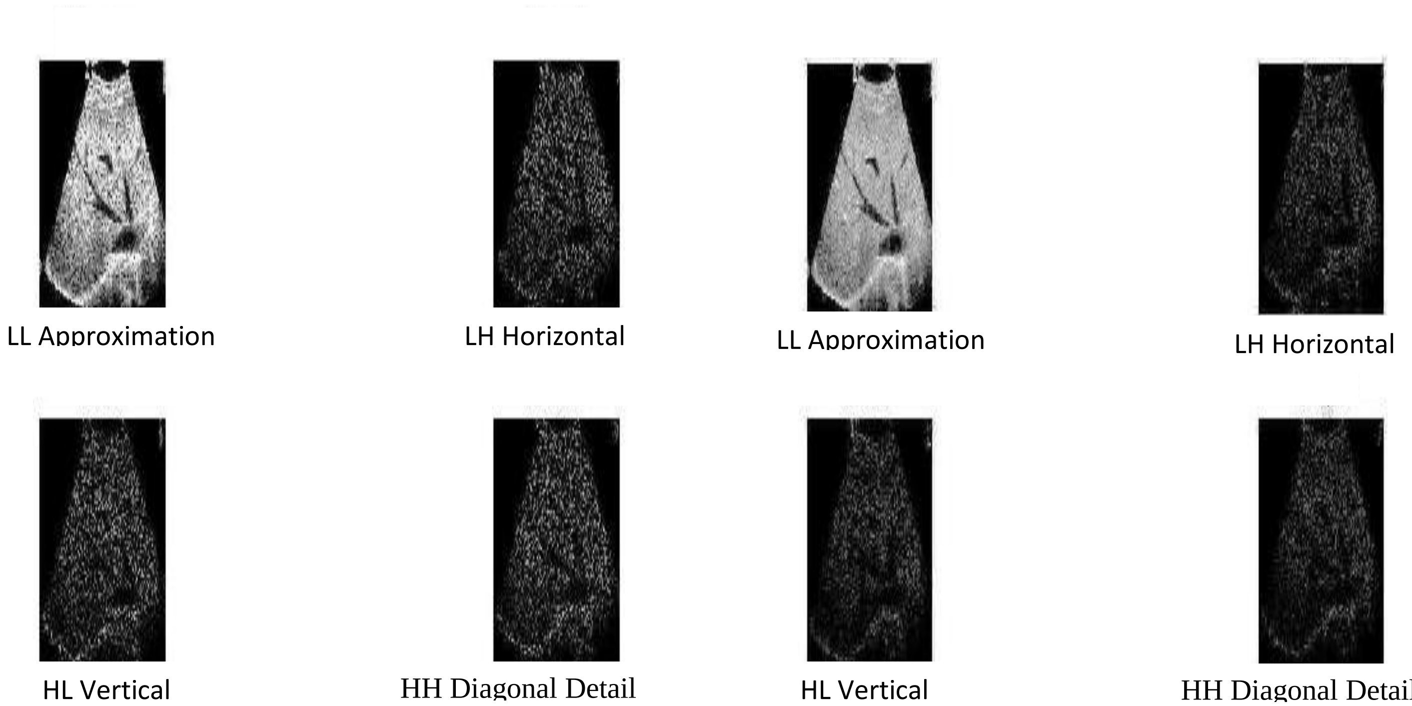


Figure 4.5: Wavelet decomposition of the liver image noise =0.05 and 0.5, respectively

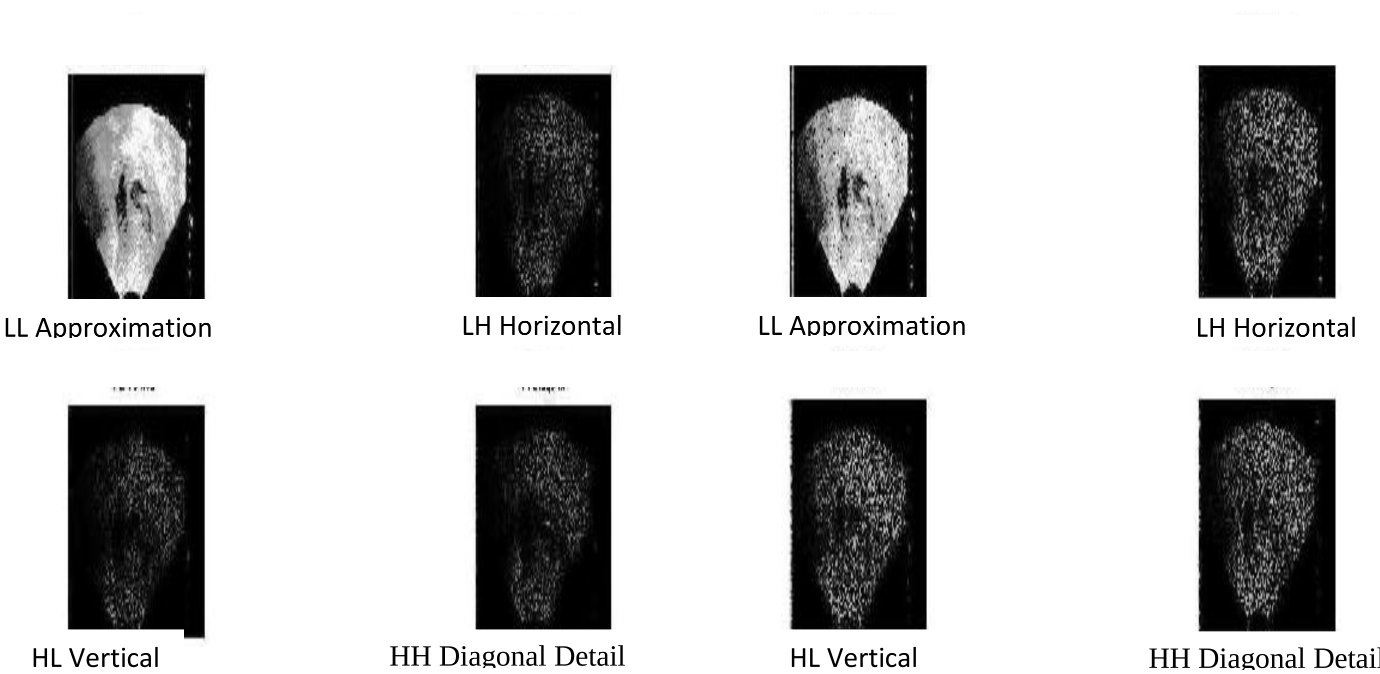


Figure 4.6: Wavelet decomposition of the vagina image noise =0.05 and 0.5, respectively

Above figures 4.4, 4.5, 4.6. Shows the low frequency subband LL and the high frequency subbands of horizontal, vertical and diagonal directions, respectively. We can observe that most of the noise and texture are concentrated in the three high frequency subbands.

In this technique the LL subband was denoised by SRAD filter without losing texture information. The noise in the three high frequency subbands were extracted from the textures by hybrid median filter, after that image reconstructed using inverse wavelet transform to that four subband. This technique was been called SRAD hybrid median technique (SRAD HMF).

4.2.3 Wavelet Decomposition Based Speckle Reduction Method for Ultrasound Images by Using Speckle Reducing Anisotropic Diffusion and Hybrid Median:

The proposed multiscale wavelet decomposition based SRAD described below. The proposed approach aims to improve the US images quality both subjective visualization and auto-segmentation applications. The Proposed approach contains three steps:

Step 1: The first-level wavelet decomposition is performed. In this processing implementation, the speckle noise and important feature detail are present in sub-band.

Step 2: Apply SRAD filter to LL_1 sub band and Hybrid Median Filter (HMF) to LH_1 sub band, HL_1 sub band, and HH_1 sub band.

Step 3: For the US image reconstruct, inverse 2D DWT is preformed.

Step 4: applied total variation on the output image.

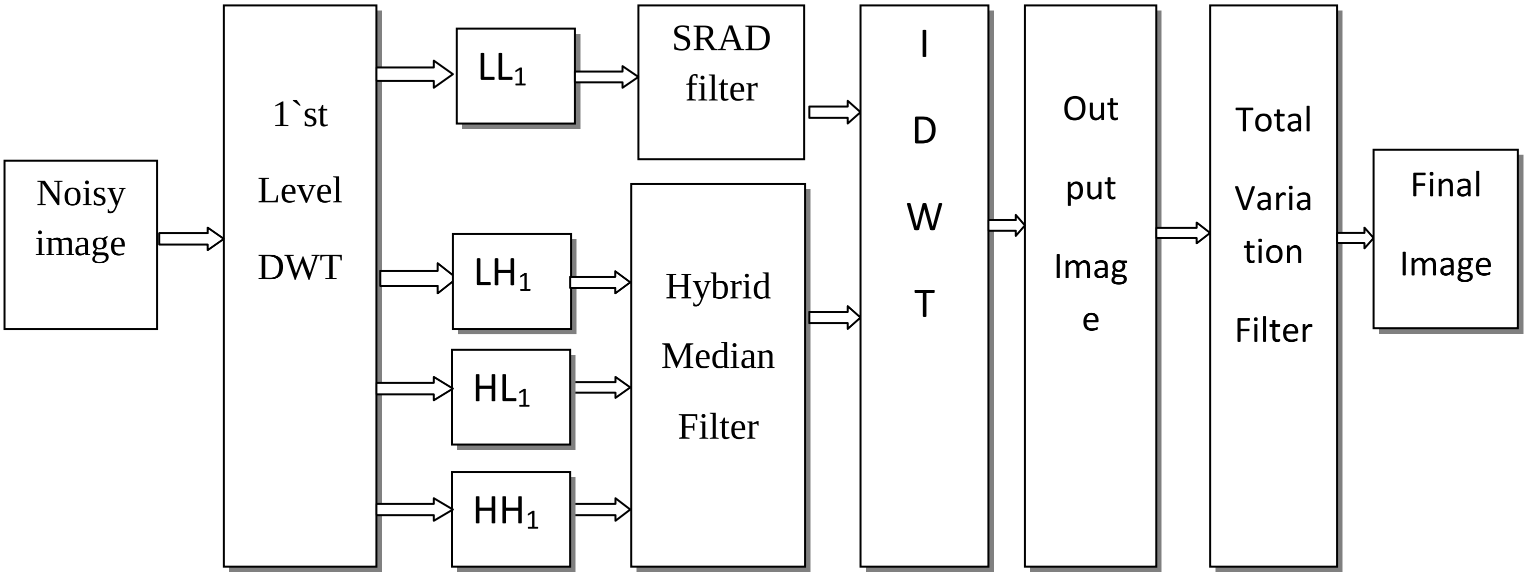


Figure 4.7: Block diagram of the proposed method (SRAD HMD)

5.1 Experimental results

The modified hybrid median and SRAD hybrid median filter techniques has been implemented in the MATLAB environment. Various US B-scan images from the Children`s Hospital of Philadelphia database of fetal ultrasound image, and IBE Tech (Giza.Egypt) database of ultrasound image including liver and vagina. And artificially corrupted by speckle noise (multiplication noise) with variance σn = 0,05 and 0.5 using the MATLAB command "imnoise (image, ‘speckle’, 0.05 or 0.5)”.

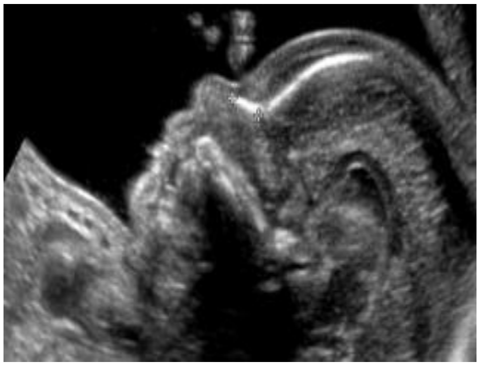
To estimate the performance of the (MHMF) and (SRAD new) techniques. eight standard filters namely: linear scaling gray level filter(DsFca),geometric despeckle filter(DsFg4d), linear scaling(DsFls), speckle reducing anisotropic diffusion(srad), median filter(Med),hybrid median filter(HMF), wavelet filter and total variation despeckle filter(TV) have been implemented in the same US images with both variance value.

To quantify the performance improvements of the speckle reduction method various measures may be used. The commonly preferred measures are root mean squared error (RMSE), signal to noise ratio (SNR), peak signal to noise ratio(PSNR) and structural similarity index(SSIM), which have been calculated from the denoised US images and are found in the literatures. The PSNR and SNR is higher for a better-transformed image and lower for a poorly transformed image, on the contray in RMSE. Whilst the range of values for the SSIM lies between -1, for bad and 1 for good similarity between the original and despeckled images.

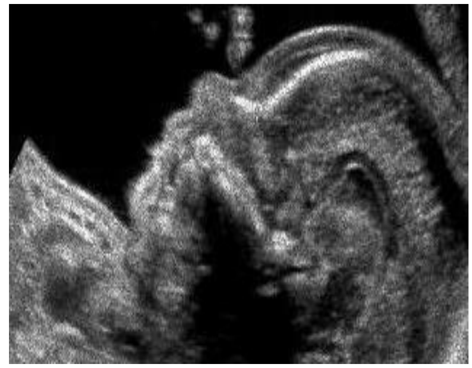
In this chapter the differences between the original, and the despeckled images were evaluated using image quality evaluation metrics. The following measures, which are easy to compute and have clear physical meaning.

The test results of US B-scan images namely fetal, liver, vagina with the two variance value of multiplication noise (σn= 0.5 and 0.05) given in figures 6.1, 6.3, 6.5, 6.7, 6.9, 6.11, 6.13, 6.15, 6.17, 6.19, 6.21 Also the performance image quality evaluation metrics calculated from the denoised image of the different filters are summarized in Tables from 6.1 to 6.12 for comparison.

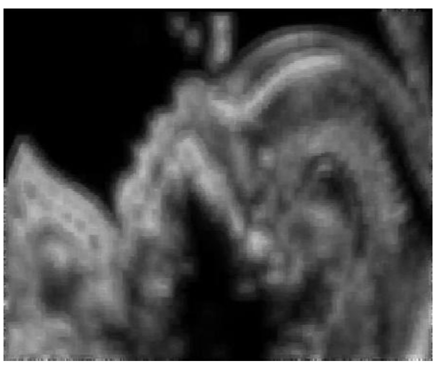
5.2 First Proposed method(modified hybrid median filter)



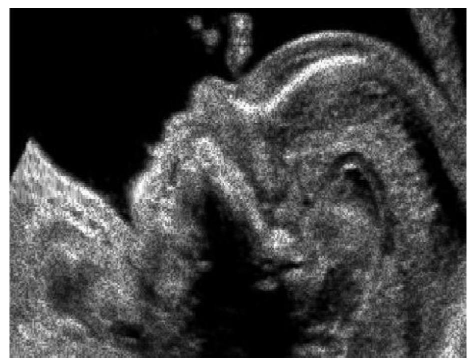
(a) Original image



(b) Noisy image

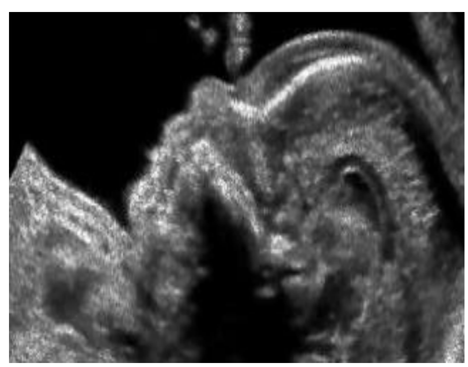


(c) Linear scaling gray level filter(DsFca)

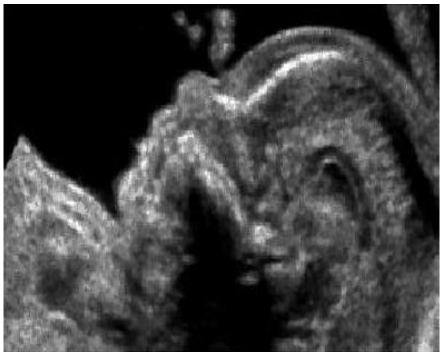


(d) Geometric filter (DsFgf4d)

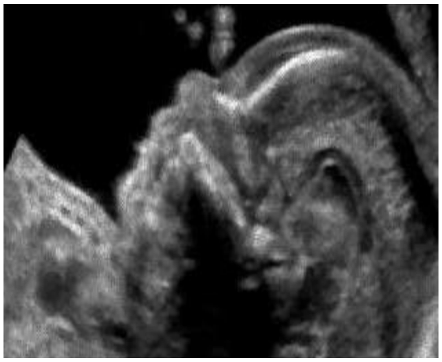
(e) Linear scaling filter(DsFls)



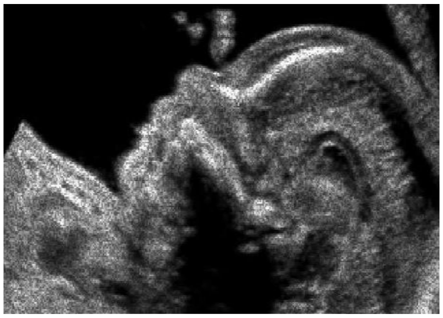
(f) Speckle reducing anisotropic diffusion filter



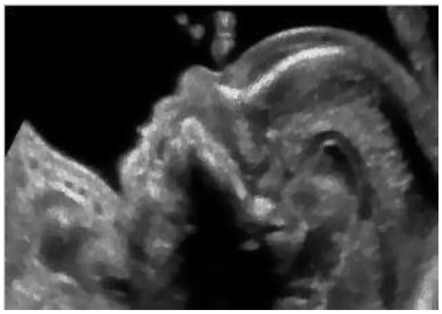
(g) Median filter (med)



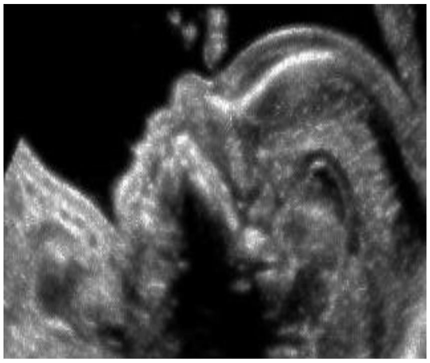
(h) Hybrid median filter (hmf)



(i) Wavelet filter



(j) Total variation



(k) Modified hybrid median filter (mhmf)

Figure 5.1 : Results of fetal despeckled by various filter on multiplication noise (σn=0.05)

Table5.1: Image quality evaluation metrics computed for the **fetal (σn =0.05)** at statistical measurement of PSNR, SNR and SSIM for different filter types and for MHMF.

Filter type	Image quality evaluation metrics		
	SNR	PNSR	SSIM
DsFca filter	**20.6551**	**45.1205**	0.6166
DsFgf4d Filter	20.6543	43.9144	0.7875
DsFls Filter	20.6156	44.3001	0.5398
SRAD filter	20.6249	45.0864	0.6153
Med	20.5665	**45.1205**	**0.7915**
HMF	20.5603	45.0344	0.7381
Wavelet	20.6166	45.0864	**0.8219**
TV	**21.5404**	44.8482	0.7692
MHMF	**20.7966**	**45.1035**	**0.8035**

Bold number indicates the best values.

* signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR), structural similarity index (SSIM).

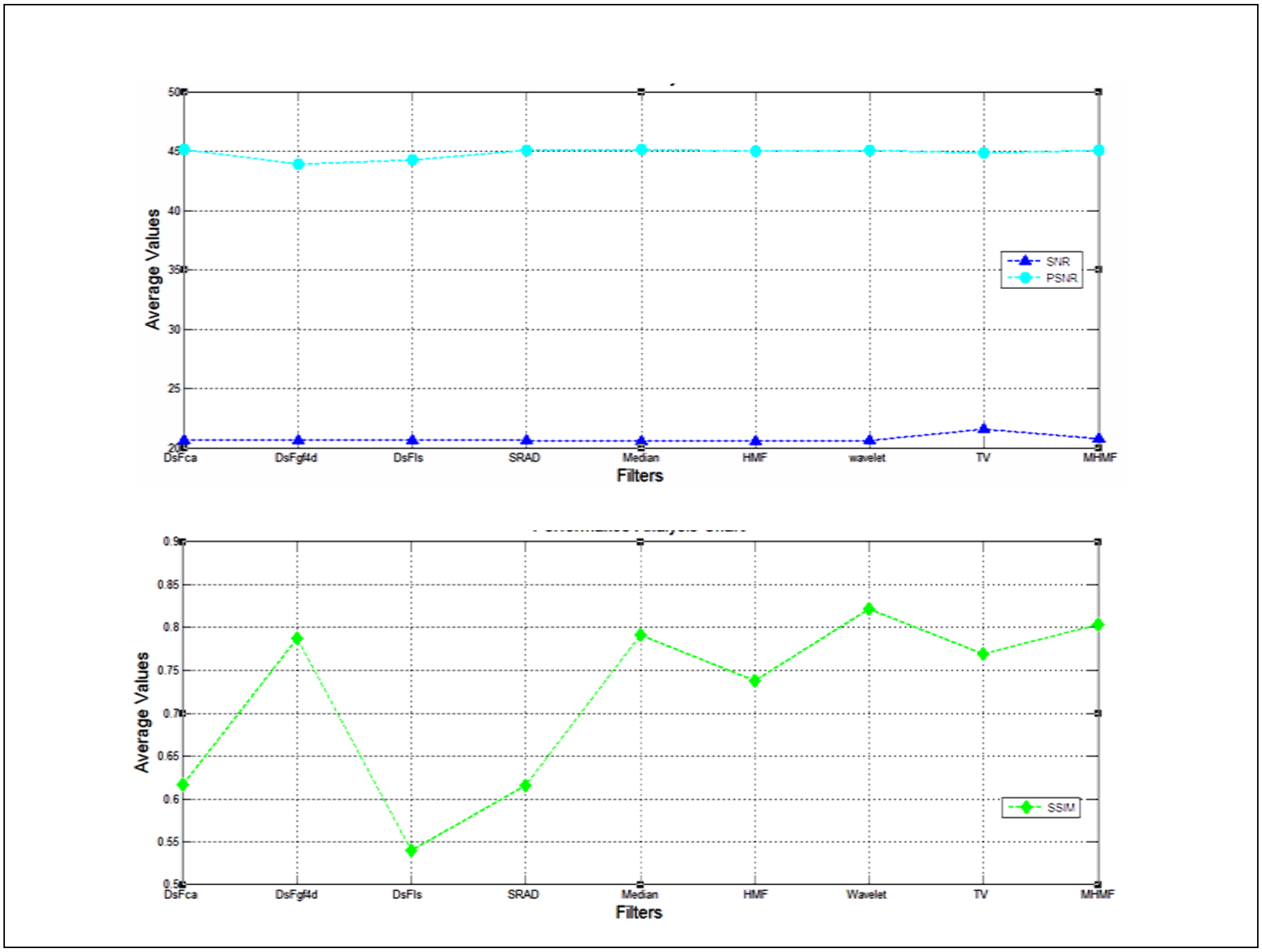
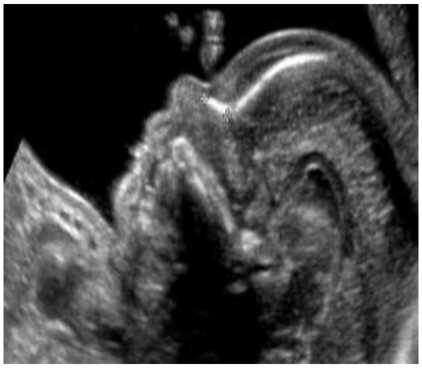
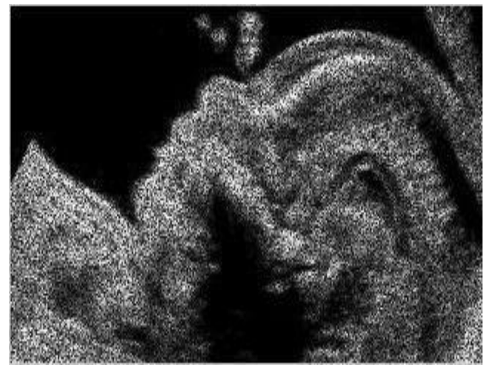


Figure 5.2: Performance analysis graph to image quality evaluation metric for fetal image

(noise**σn** =0.05).

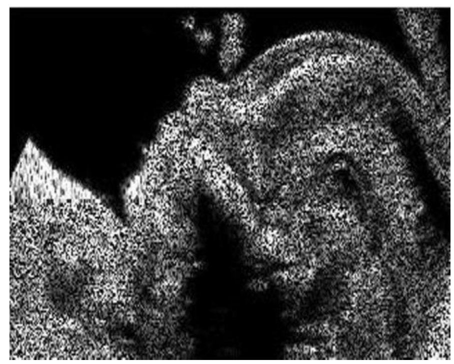


(a) Original image

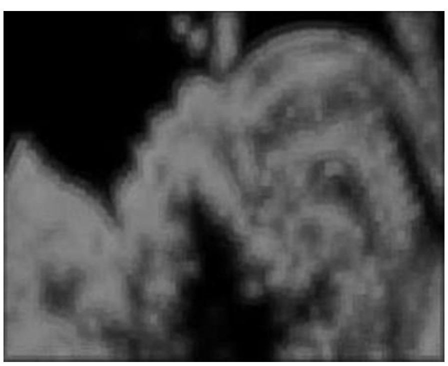


(b) Noisy image

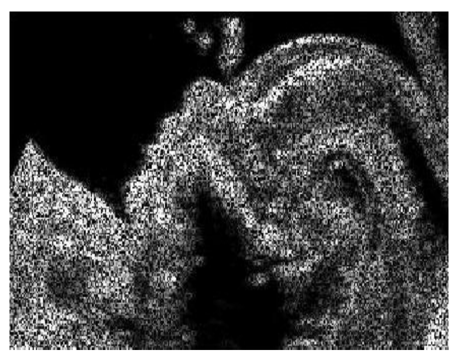
(c) Linear scaling gray level filter(DsFca)



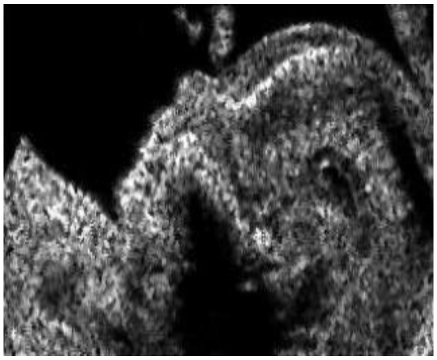
(d) Geometric filter (DsFgf4d)



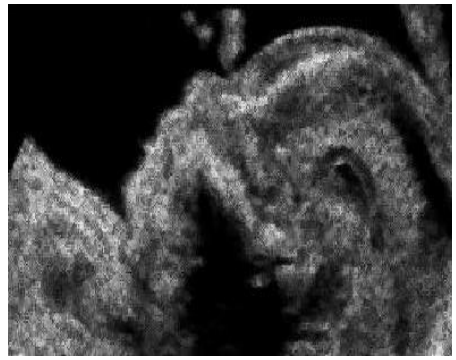
(e) Linear scaling filter(DsFls)



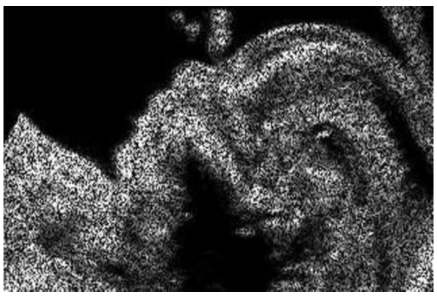
(f) Speckle reducing anisotropic diffusion filter



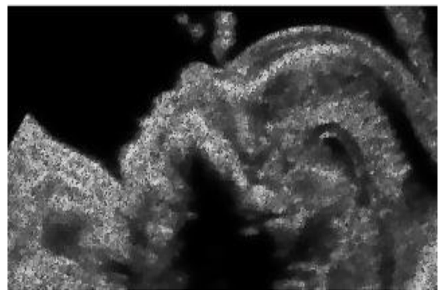
(g) Median filter (med)



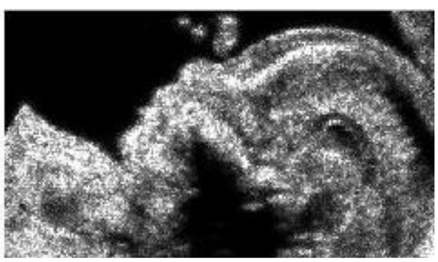
(h) Hybrid median filter (hmf)



(i) Wavelet filter



(j) Total variation



(k) Modified hybrid median filter (mhmf)

Figure 5.3 : Results of fetal despeckled by various filter on

Multiplication noise (σn=0.5)

Table5.2: Image quality evaluation metrics computed for the fetal (σn =0.5)at statistical measurement of PSNR ,SNR and SSIM for different filter types and for MHMF.

Filter type	Image quality evaluation metrics		
	SNR	PNSR	SSIM
DsFca filter	20.6325	44.2420	0.5416
DsFgf4d Filter	**20.7903**	**45.1205**	0.4853
DsFls Filter	20.6139	43.2688	0.4828
SRAD filter	**20.7420**	**45.1205**	0.4460
Med	20.6081	45.0864	0.4096
HMF	20.5071	44.7929	**0.6562**
Wavelet	20.6931	45.1205	0.4721
TV	20.5571	44.3921	**0.6263**
MHMF	**21.4016**	**45.1205**	**0.5577**

Bold number indicates the best values.

* signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR),structural similarity index(SSIM).

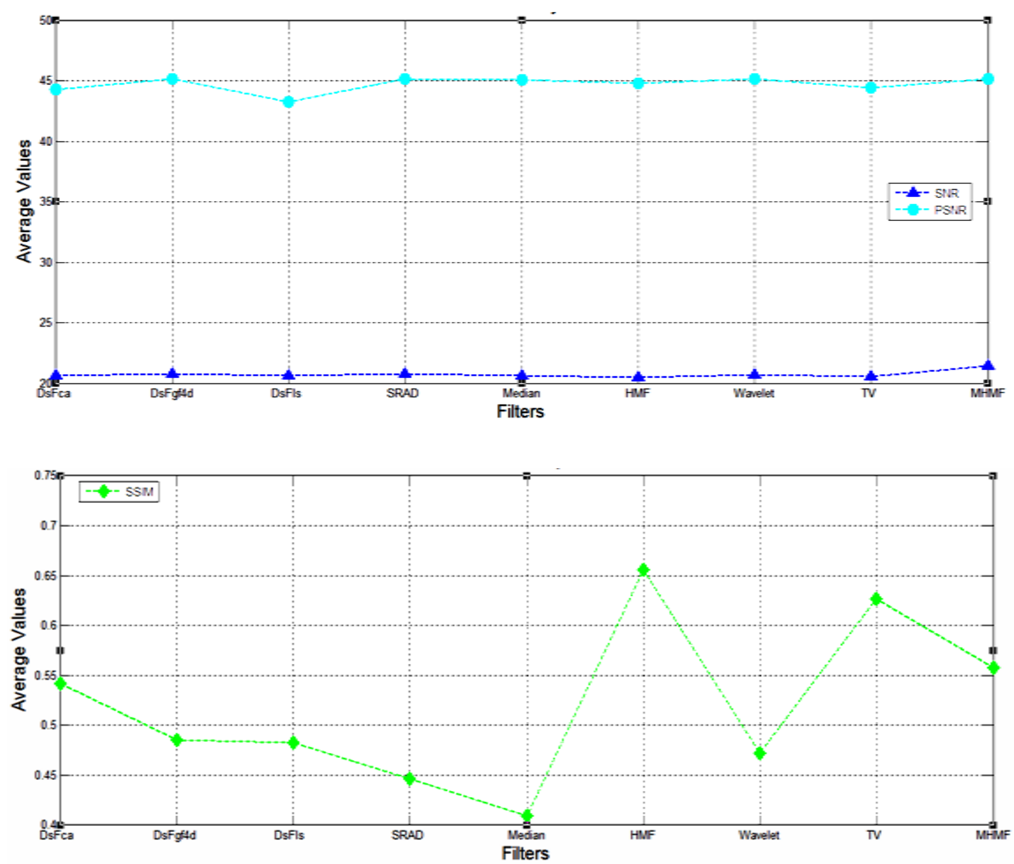


Figure 5.4: Performance analysis graph to image quality evaluation metric for fetal image

(noise**σn** =0.5).

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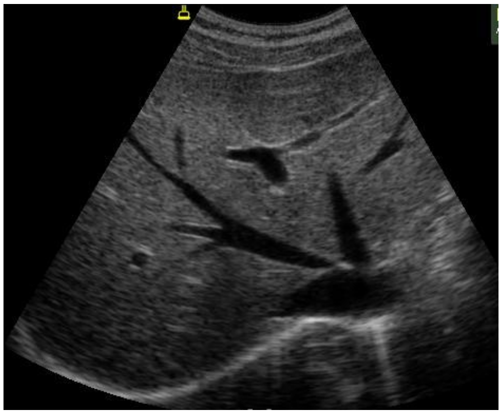
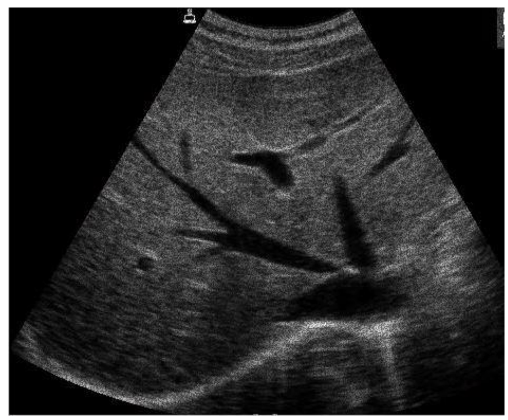


(a) Original image

(b) Noisy image

(c) Linear scaling gray level filter(DsFca)



(d) Geometric filter (DsFgf4d)

(e) Linear scaling filter(DsFls)

(f) Speckle reducing anisotropic diffusion filter

(g) Median filter (med)

(h) Hybrid median filter

(i) Wavelet filter

(j) Total variation

(k) Modified hybrid median filter (mhmf)

Figure 5.5 : Results of liver despeckled by various filter on multiplication noise (σn=0.05)

Table5.3: Image quality evaluation metrics computed for the **liver (σn =0.05)**at statistical measurement of PSNR, SNR and SSIM for different filter types and for MHMF.

Filter name	Image Quality Evaluation Metrics		
	SNR	**PSNR**	**SSIM**
DsFca filter	18.5023	44.0741	0.6992
DsFgf4d Filter	**18.5293**	**45.0013**	0.7334
DsFls Filter	18.5289	42.2283	0.6493
SRAD filter	18.3719	44.3935	0.6703
Median filter	18.3650	43.2634	0.7840
HMF	18.3718	43.5002	**0.7940**
Wavelet filter	**18.5480**	**44.6616**	0.7510
TV	18.5006	44.1924	**0.8327**
MHMF	**18.6209**	**44.3024**	**0.8682**

Bold number indicates the best values.

* signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR),structural similarity index(SSIM).

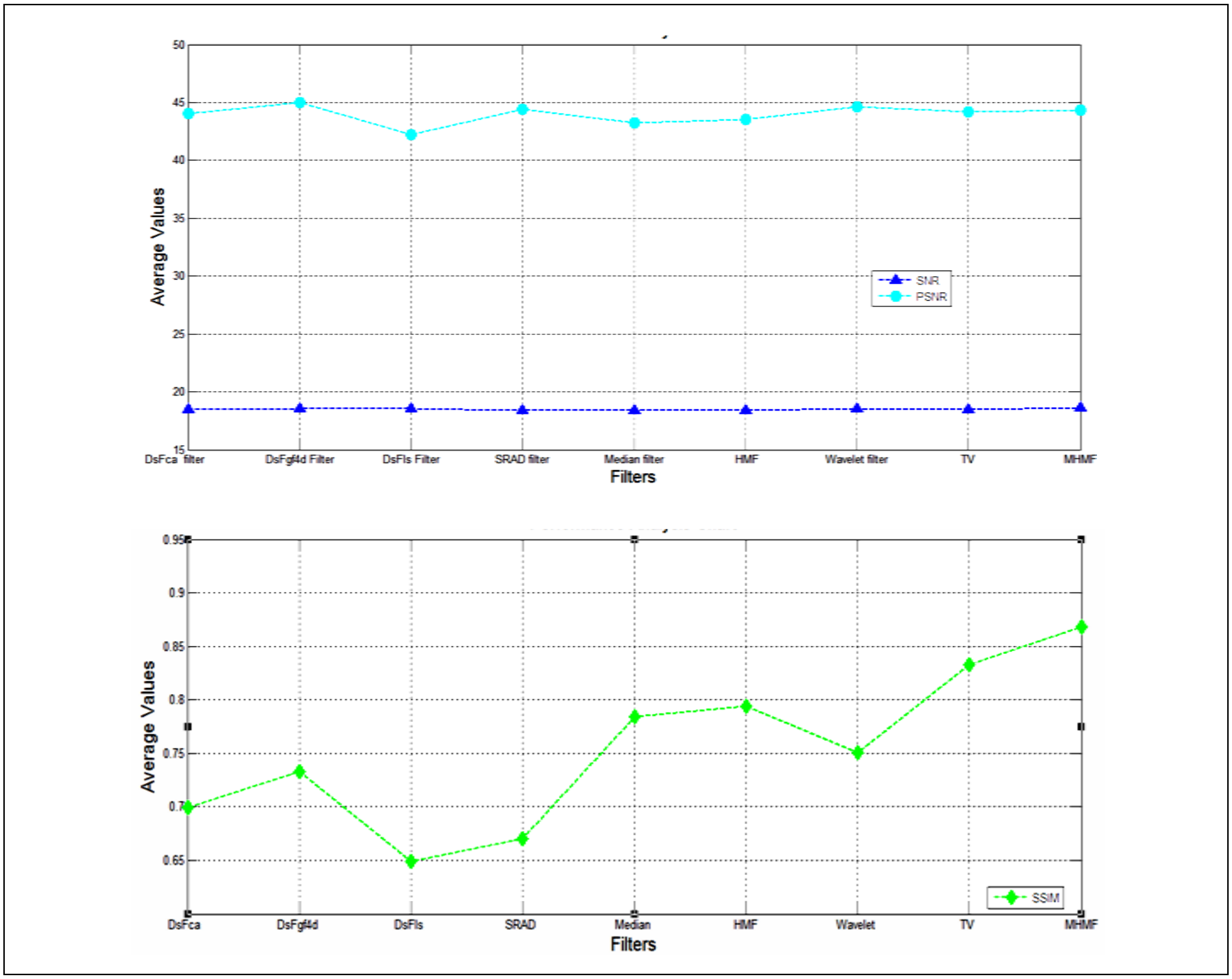
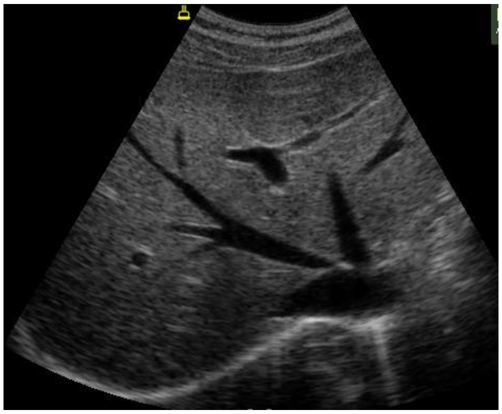


Figure 5.6: Performance analysis graph to image quality evaluation metric for liver image

(noise**σn** =0.05).



(a) Original image

(b) Noisy image

(c) Linear scaling gray level filter(DsFca)

(d) Geometric filter (DsFgf4d)

(e) Linear scaling filter(DsFls)

(f) Speckle reducing
anisotropic diffusion filter

(g) Median filter (med)

(h) Hybrid median filter

(i) Wavelet filter

(j) Total variation

(k) Modified hybrid median filter (mhmf)

Figure 5.7: Results of liver despeckled by various filter on multiplication noise (σn=0. 5)

Table5.4: Image quality evaluation metrics computed for the **liver (σn=0.5)**at statistical measurement ofPSNR, SNR and SSIM for different filter types and for MHMF.

Filter name	Image Quality Evaluation Metrics		
	SNR	**PSNR**	**SSIM**
DsFca filter	**18.8367**	44.8471	0.6500
DsFgf4d Filter	18.8436	**44.9843**	0.4590
DsFls Filter	18.8035	42.1681	0.6356
SRAD filter	18.5798	44.8652	0.6392
Median filter	18.4127	44.3486	0.6412
HMF	18.4653	43.1833	**0.6697**
Wavelet filter	**18.8328**	**44.5987**	0.4934
TV	18.3942	43.9810	**0.6858**
MHMF	**19.0237**	**44.8653**	**0.6727**

Bold number indicates the best values.

* signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR),structural similarity index(SSIM).

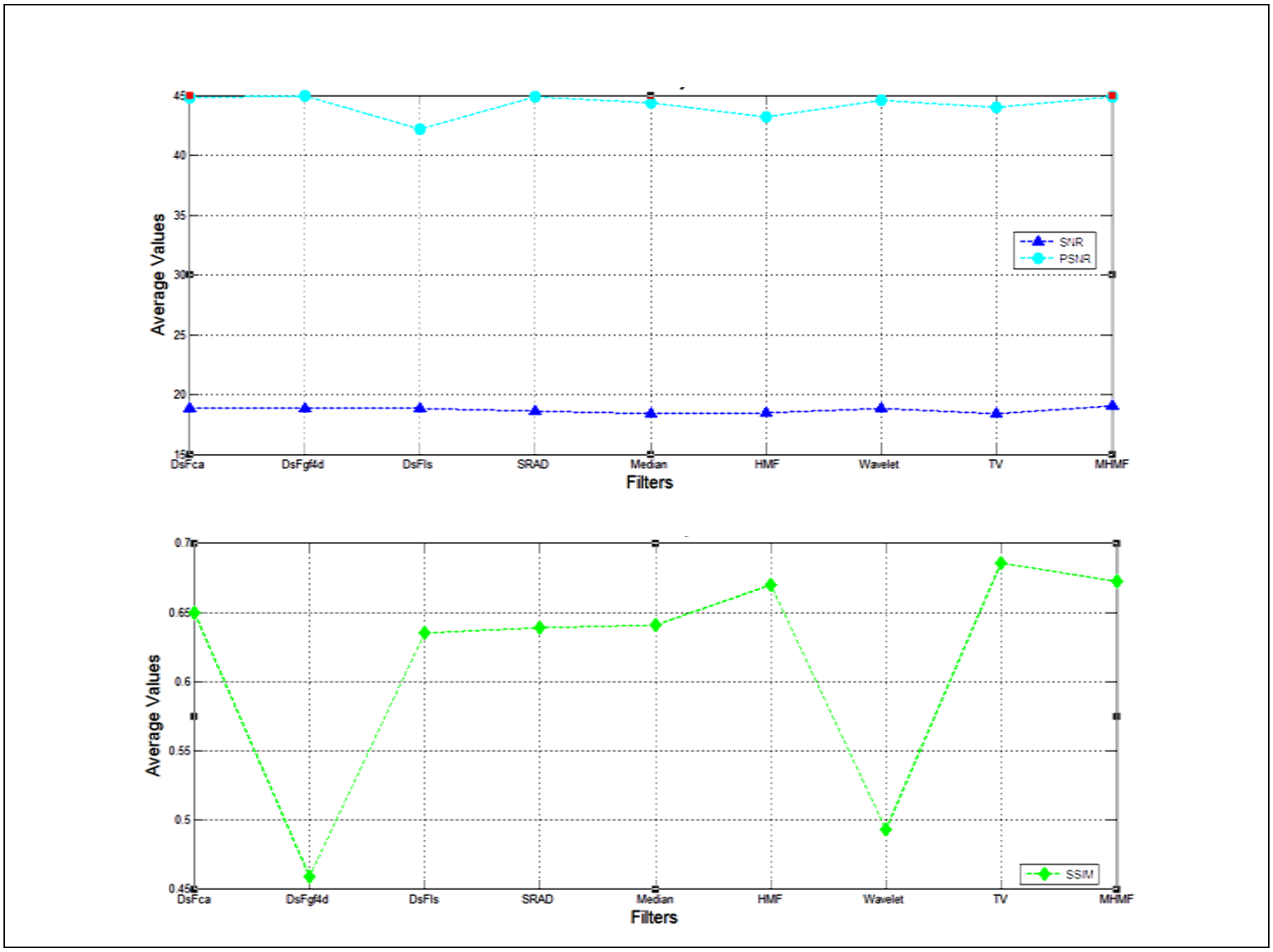
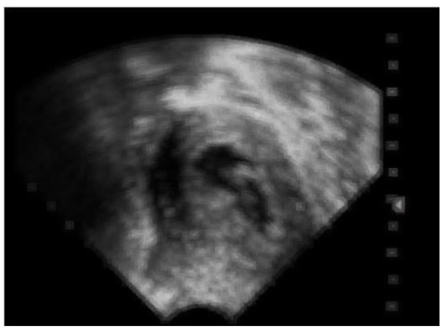


Figure 5.8: Performance analysis graph to image quality evaluation metric for liver image

(noise**σn** =0.5).

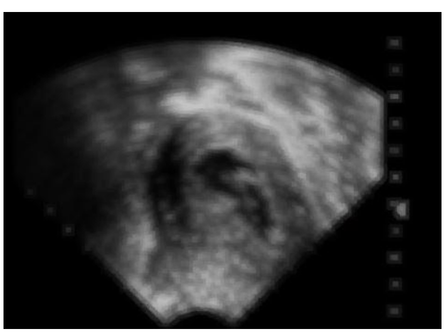
(a) Original image

(b) Noisy image

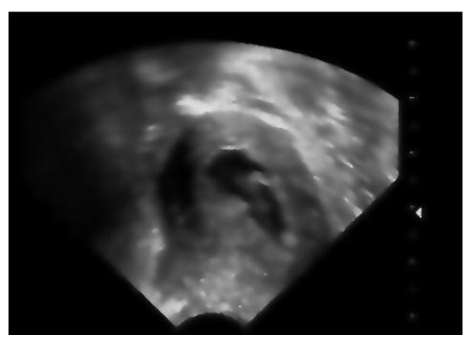


(c) Linear scaling gray level filter(DsFca)

(d) Geometric filter (DsFgf4d)



(e) Linear scaling filter(DsFls)

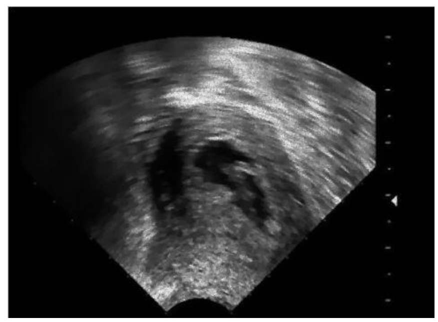


(f) Speckle reducing anisotropic diffusion filter

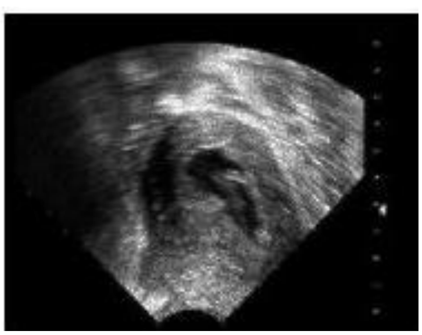
(g) Median filter (med)

(h) Hybrid median filter (hmf)

(i) Wavelet filter



(j) Total variation



(k) Modified hybrid median filter (mhmf)

Figure 5.9 : Results of vagina despeckled by various filter on multiplicationnoise (σn=0.05)

Table5.5: Image quality evaluation metrics computed for the **vagina (σn =0.05)**at statistical measurement of PSNR ,SNR and SSIM for different filter types and for MHMF.

Filter type	Image quality evaluation metrics		
	SNR	PNSR	SSIM
DsFca filter	**19.8433**	44.0225	0.7176
DsFgf4d Filter	**19.8585**	**44.9663**	0.8021
DsFls Filter	19.8044	44.0225	0.6631
SRAD filter	19.7652	44.1712	0.7290
Median filter	19.7186	44.7186	**0.8248**
HMF	19.7508	44.5044	**0.8711**
Wavelet filter	19.7948	**45.0013**	0.7552
TV	19.7841	44.8482	0.7545
MHMF	**19.9543**	**45.1205**	**0.8203**

Bold number indicates the best values.

* signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR),structural similarity index(SSIM).

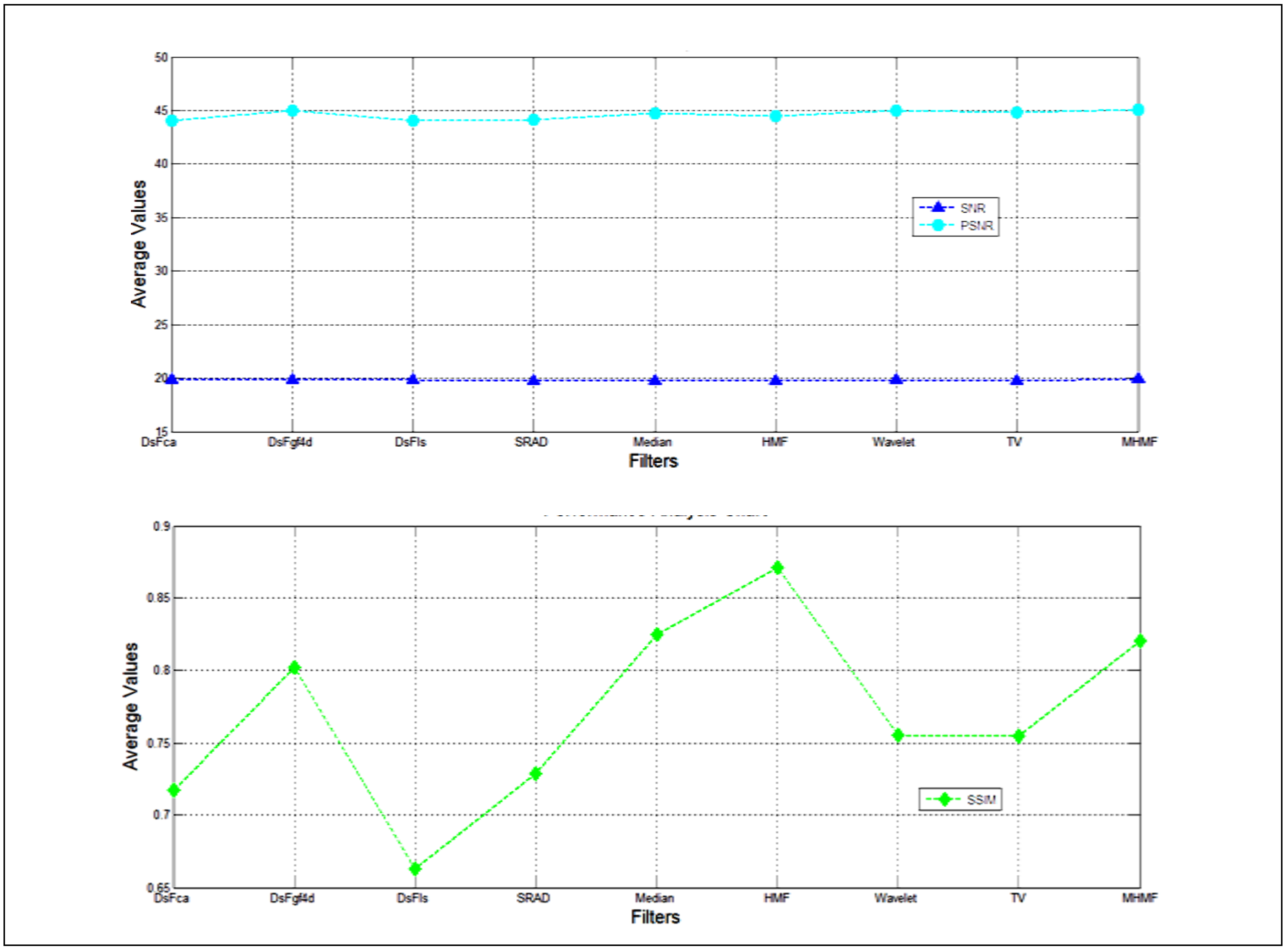
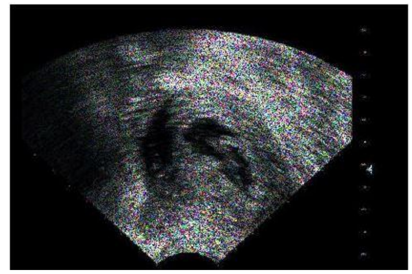


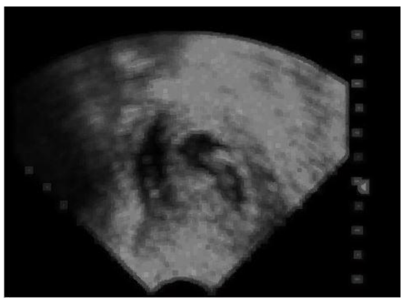
Figure 5.10: Performance analysis graph to image quality evaluation metric for vagina image

(noise**σn** =0.05).

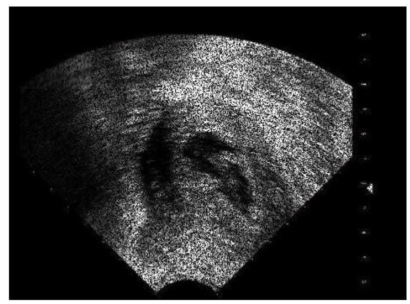
(a) Original image



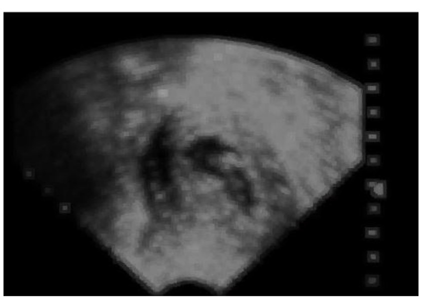
(b) Noisy image



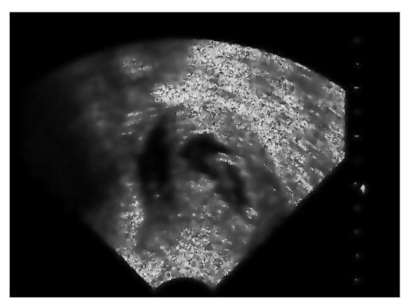
(c) Linear scaling gray level filter(DsFca)



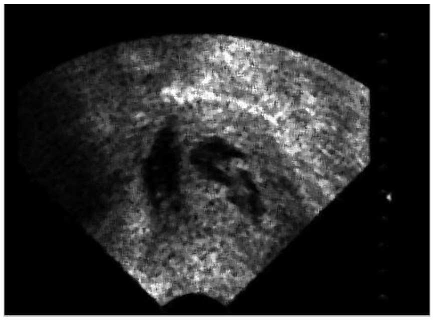
(d) Geometric filter (DsFgf4d)



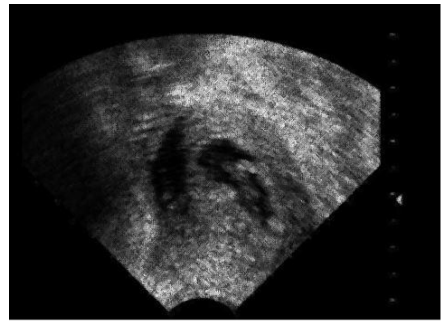
(e) Linear scaling filter(DsFls)



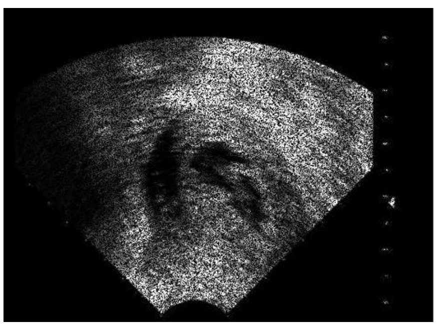
(f) Speckle reducing
anisotropic diffusion filter



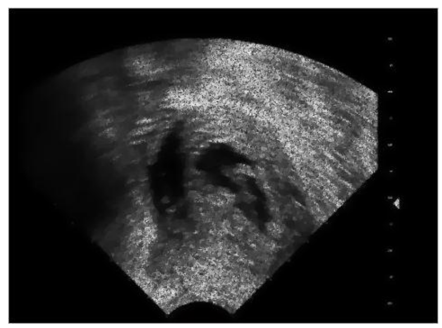
(g) Median filter (med)



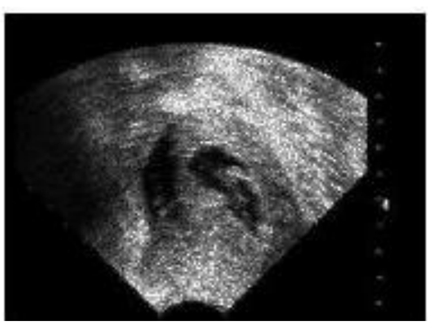
(h) Hybrid median filter (hmf)



(i) Wavelet filter



(j) Total variation



(k) Modified hybrid median filter (mhmf)

Figure 5.11: Results of vagina despeckled by various filter on multiplicationnoise (σn=0.5)

Table5.6: Image quality evaluation metrics computed for the **vagina (σn =0.5)**at statistical measurement of PSNR, SNR and SSIM for different filter types and for MHMF.

Filter type	Image quality evaluation metrics		
	SNR	**PNSR**	**SSIM**
DsFca filter	**19.7474**	43.2458	0.6630
DsFgf4d Filter	**19.9703**	**45.1205**	0.5722
DsFls Filter	19.7273	43.3246	0.6327
SRAD filter	19.9009	44.8454	**0.7370**
Median filter	19.7837	45.1035	0.6938
HMF	19.6831	44.8110	**0.7457**
Wavelet filter	19.9619	**45.1205**	0.5771
Tv	19.6362	43.7631	0.6941
MHMF	**20.2149**	**45.0694**	**0.6775**

Bold number indicates the best values.

* signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR),structural similarity index(SSIM).

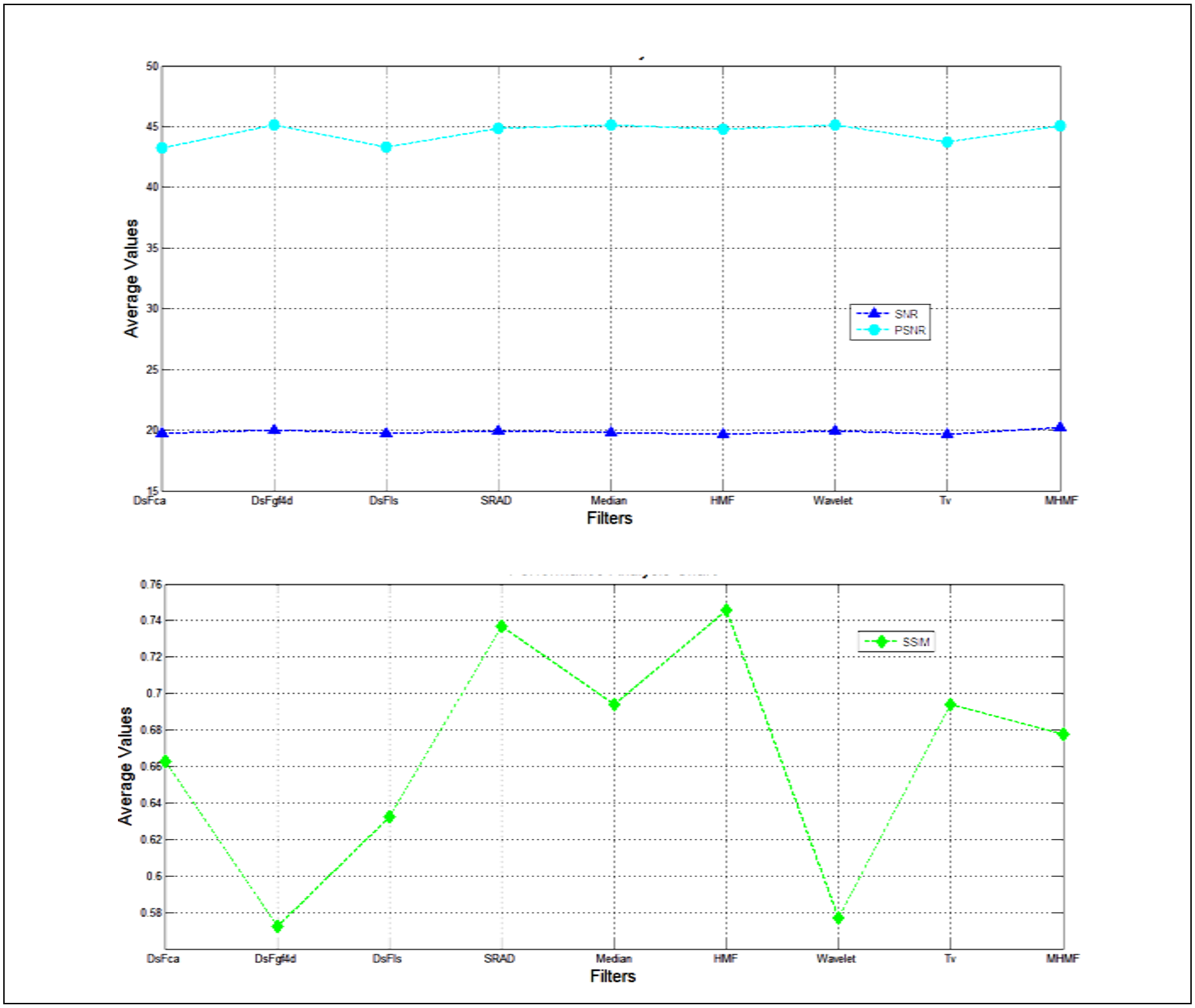


Figure 5.12: Performance analysis graph to image quality evaluation metric for vagina image

(noise**σn** =0.5).

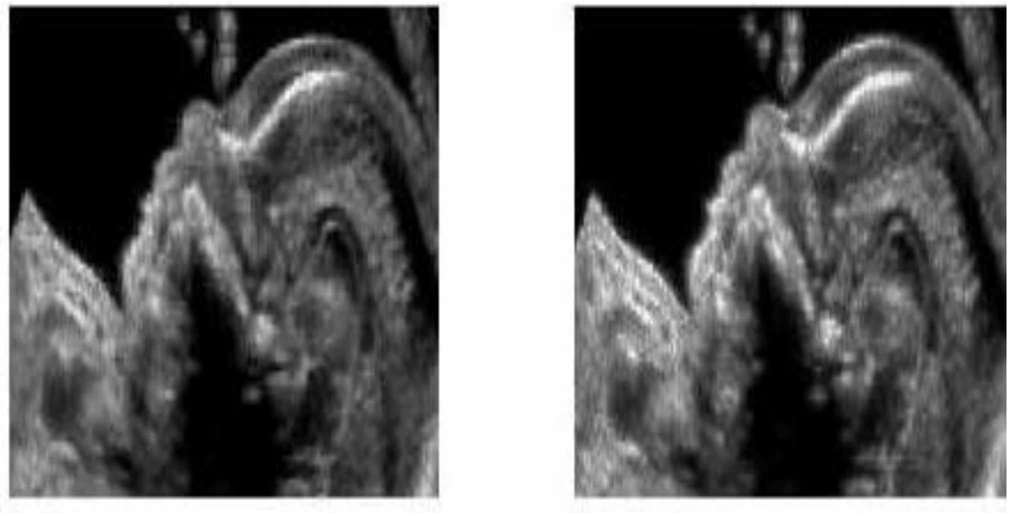
Most importantly,adespeckle filtering analysis and evaluation framework is proposed for selecting the most appropriate filter or filters for the images under investigation. The filters can be further developed and evaluated at a larger scale, texture analysis, image quality evaluation metrics, and visual evaluation by experts.

From figures 5.1,5.2,5.3,5.4,5.5,5.6 show an ultrasound image (a) with noisy (b) and the despeckled images. In(c) can see that,the linear scaling gray level filter(DsFca) has high degree of blurring and was affect on gray level, because it is compute the mean of all pixels whose difference in the gray level with the intensity(the middle pixel in the moving window) is lower than or equal to a given threshold, (d) although the result obtained by geometric despeckle filter(DsFgf4d) given poor performance for removing the speckle noise from the ultrasound image, it is lead to increasing the contrast significantly of the image. (e) Show the result obtained by liner scaling (DsFls*)*filter scales the pixel intensities by finding the maximum and the minimum gray-level values in every moving window, and then replaces the middle pixel with the average of them also give blurred image. (f) show the result of speckle reducing anisotropic diffusion filtering(srad), it is better for preserves the edges as a comparison with the other despeckle filtering techniques and subjectively has good result, and referred to evaluated metrics, it was also given bad results, (g) show the result obtained bymedian despeckle filter, which don't able to remove the speckle and produced blurred edges in the filtered image .figure(h)showThe result of hybrid median filter(hmf) that given better edge preserving characteristics than normal median filter, (i)the result through wavelet despeckle filtering perceived that it's moderate in order of variance decreasing but execute to decrease the contrast,(J)show the result obtained by total variation despeckle filter(TV)methods.We see that most of the unwanted details haven't been removed efficiently, whilst preserving important details such as edges.

From table 5.1,5.2,5.3,5.4,5.5,5.6 tabulates the image quality evaluation contains the metric result of filters under study, The best visual results were obtained for the filters DsFgf4d, wavelet, DsFca and TV because with higher SNR and PSNR and lower RMSE and Best values for the SSlM, but visually, smoothed the image. Loosing subtle details are been observed.

From table 5.1,5.3,5.7,5.9 show that the modified hybrid median filter has best performances but in other tables has result fallback, that indicate to the performance of despeckled filters are depended on image's features and quantity of speckle noise which applied on image.

By modify the hybrid median filter,this gives better edge preserving characteristics than hybrid median filter, and give less blurred image , and increase the brightness of image by taking the max value, as shown in the image quality metrics the result is better than normal hybrid median .



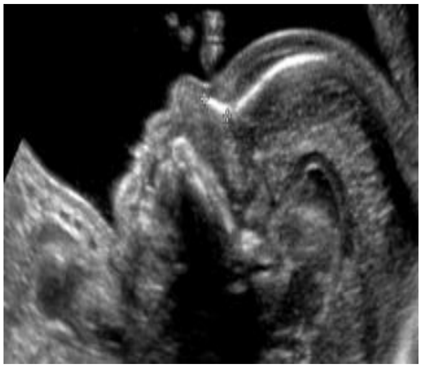
(a) (b)

Figure 5.13 (a),(b) images filtered by hybrid median filter and modified hybrid median filter, respectively from speckled fetal image with variance (σn=0.05).

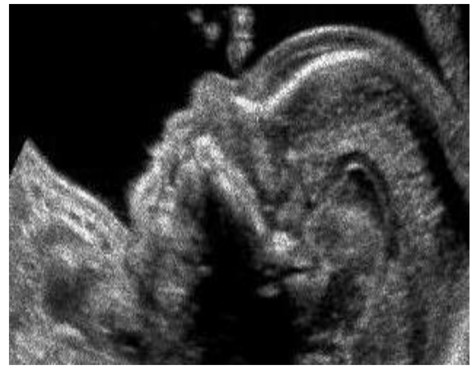
(a) (b)

Figure 5.14 (a),(b) images filtered by hybrid median filter and modified hybrid median filter, respectively from speckled liver image with variance (σn=0.5).

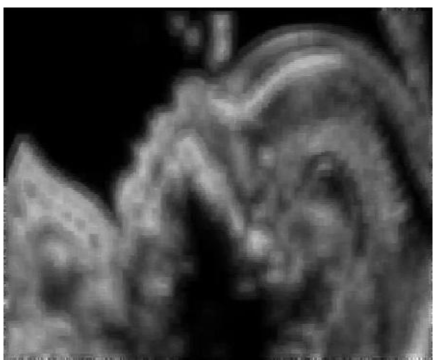
5.3 Second proposed method (SRAD HMF)



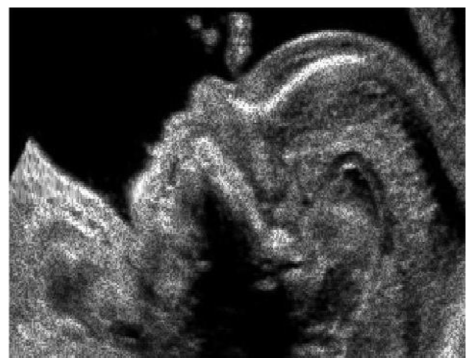
(a) Original image



(b) Noisy image

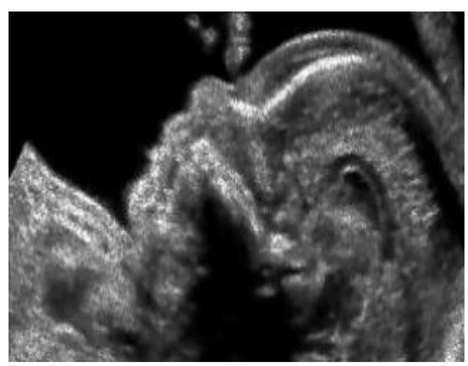


(c) Linear scaling gray level filter(DsFca)

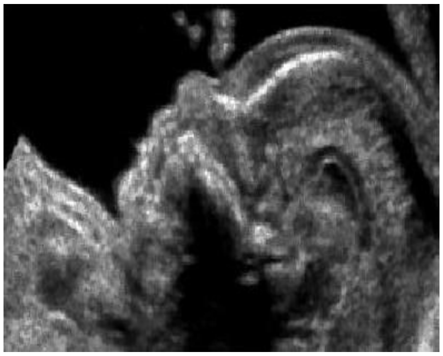


(d) Geometric filter (DsFgf4d)

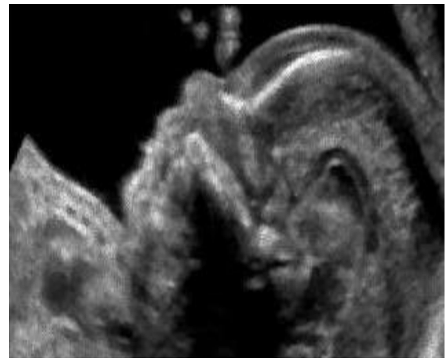
(e) Linear scaling filter(DsFls)



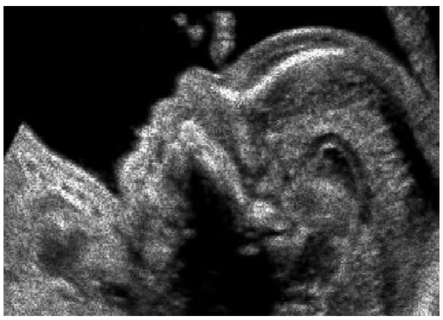
(f) Speckle reducing anisotropic diffusion filter



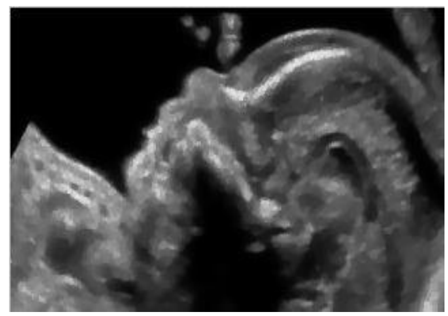
(g) Median filter (med)



(h) Hybrid median filter (hmf)



(i) Wavelet filter



(j) Total variation

(k) Srad hybrid median filter
(SRAD new)

Figure 5.15: Results of fetal despeckled by various filter on multiplicationnoise (σn=0.05)

Table 5.7: Image quality evaluation metrics computed for the **fetal (σn =0.05)**at statistical measurement of RMSE, PSNR ,SNR and SSIM for different filter types and for SRAD HMF .

Filter type	Image quality evaluation metrics			
	RMES	SNR	PNSR	SSIM
DsFca filter	17.4916	**20.6551**	**45.1205**	0.6166
DsFgf4d Filter	14.7985	20.6543	43.9144	0.7875
DsFls Filter	26.0022	20.6156	44.3001	0.5398
SRAD filter	21.1283	20.6249	45.0864	0.6153
Med	13.2533	20.5665	**45.1205**	**0.7915**
HMF	**10.6843**	20.5603	45.0344	0.7381
Wavelet	**12.4262**	20.6166	45.0864	**0.8219**
TV	21.5404	**21.5404**	44.8482	0.7692
Srad new	**10.5559**	**20.6644**	**45.1205**	**0.8192**

Bold number indicates the best values.

*Root mean square error (RMSE), signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR),structural similarity index(SSIM).

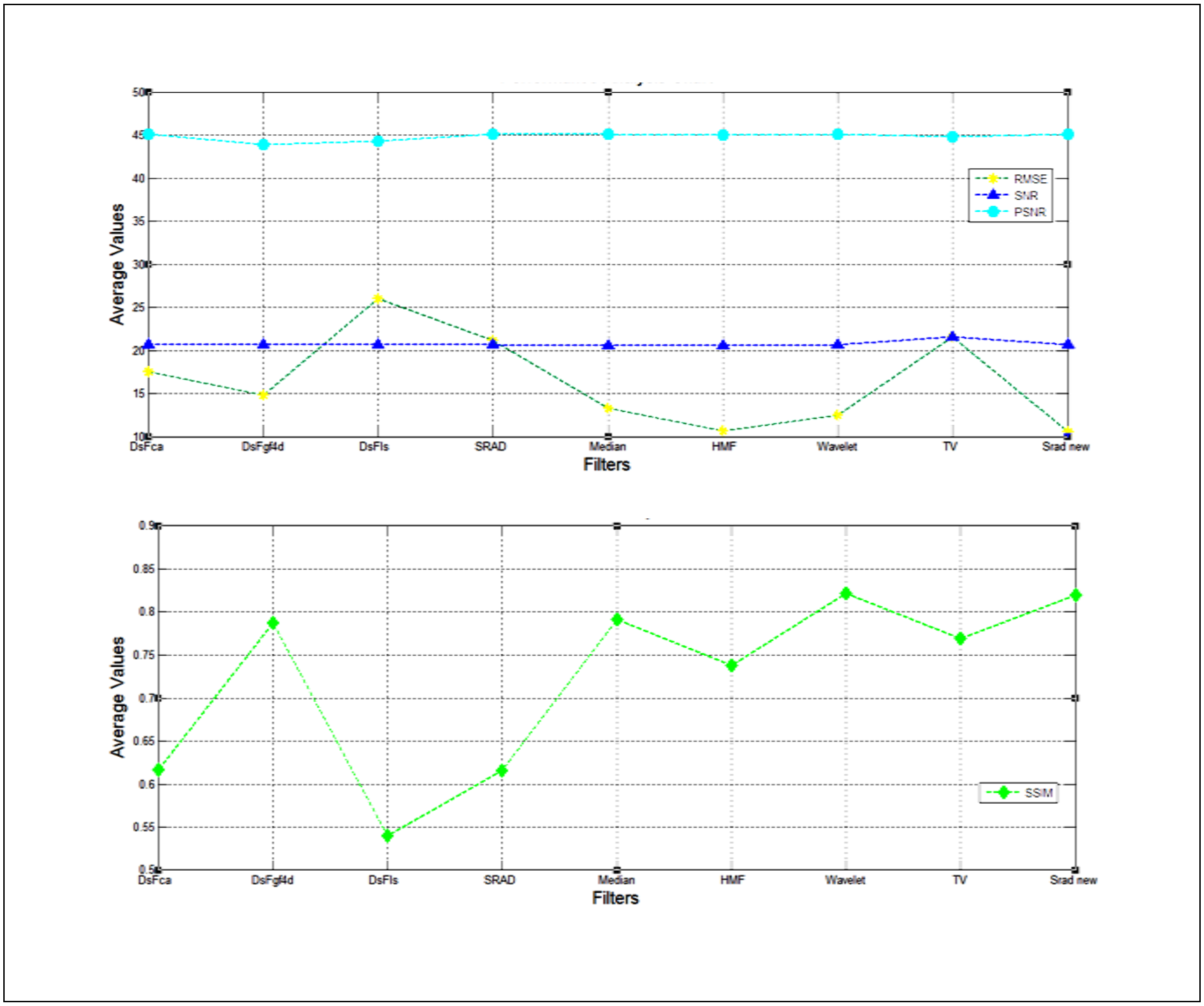
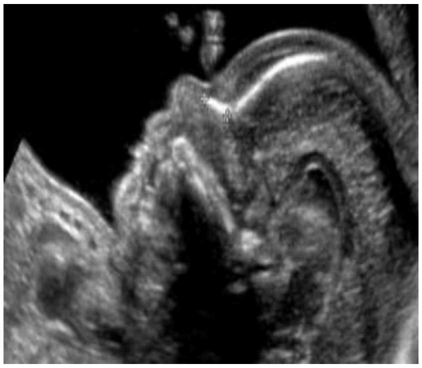
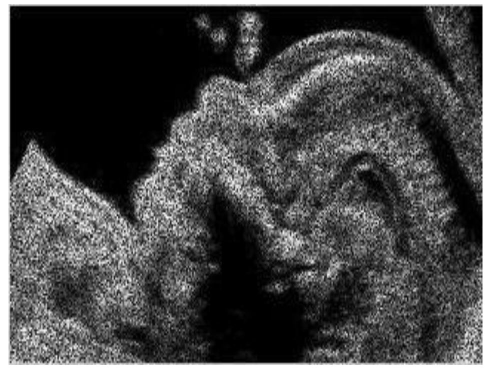


Figure 5.16: Performance analysis graph to image quality evaluation metric for fetal image (noise**σn** =0.05).

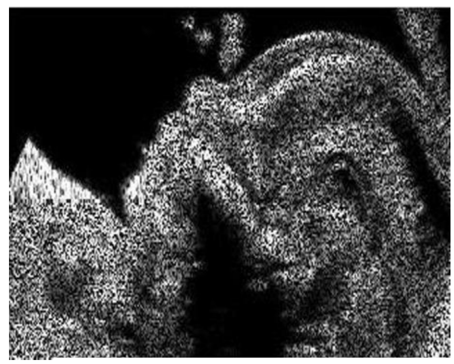


(a) Original image

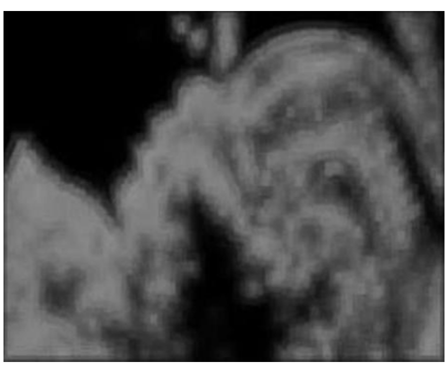


(b) Noisy image

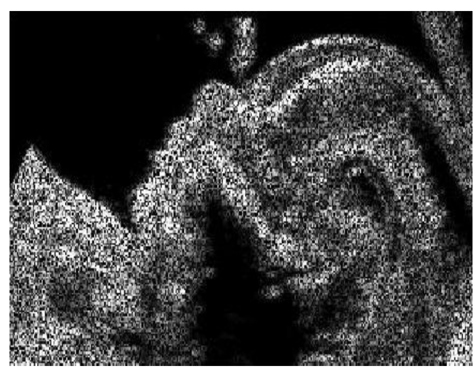
(c) Linear scaling gray level filter(DsFca)



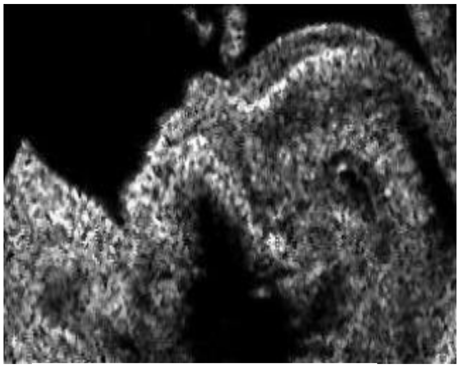
(d) Geometric filter (DsFgf4d)



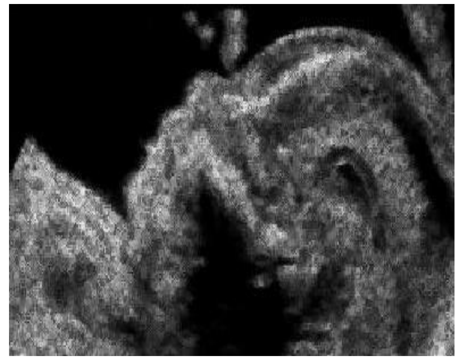
(e) Linear scaling filter(DsFls)



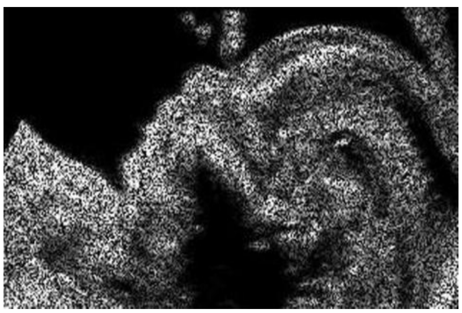
(f) Speckle reducing anisotropic diffusion filter



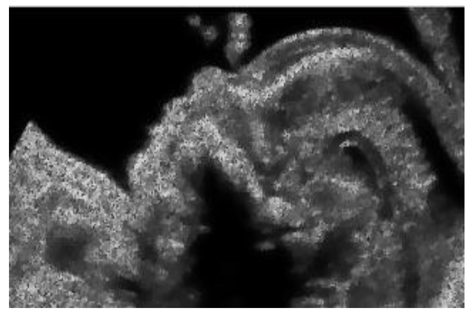
(g) Median filter (med)



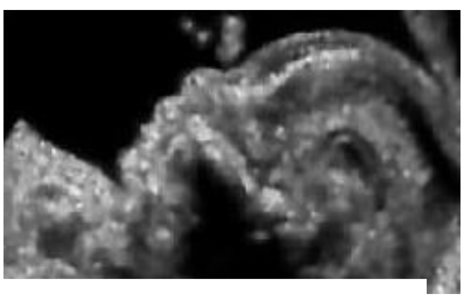
(h) Hybrid median filter (hmf)



(i) Wavelet filter



(j) Total variation



(k) Srad hybrid median filter
(SRAD new)

Figure5.17 : Results of fetal despeckled by various filter on

multiplicationnoise (σn=0.5)

Table5.8: Image quality evaluation metrics computed for the **fetal (σn =0. 5)**at statistical measurement of RMSE, PSNR ,SNR and SSIM for different filter types and for SRAD HMF.

Filter type	Image quality evaluation metrics			
	RMES	SNR	PNSR	SSIM
DsFca filter	**24.4291**	20.6325	44.2420	0.5416
DsFgf4d Filter	37.5317	**20.7903**	**45.1205**	0.4853
DsFls Filter	26.4144	20.6139	43.2688	0.4828
SRAD filter	38.4544	**20.7420**	**45.1205**	0.4460
Med	31.5228	20.6081	45.0864	0.4096
HMF	**18.6728**	20.5071	44.7929	**0.6562**
Wavelet	35.2916	20.6931	45.1205	0.4721
TV	25.3902	20.5571	44.3921	**0.6263**
Srad new	**17.2906**	**20.7651**	**45.1205**	**0.6842**

Bold number indicates the best values.

*Root mean square error (RMSE), signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR),structural similarity index(SSIM).

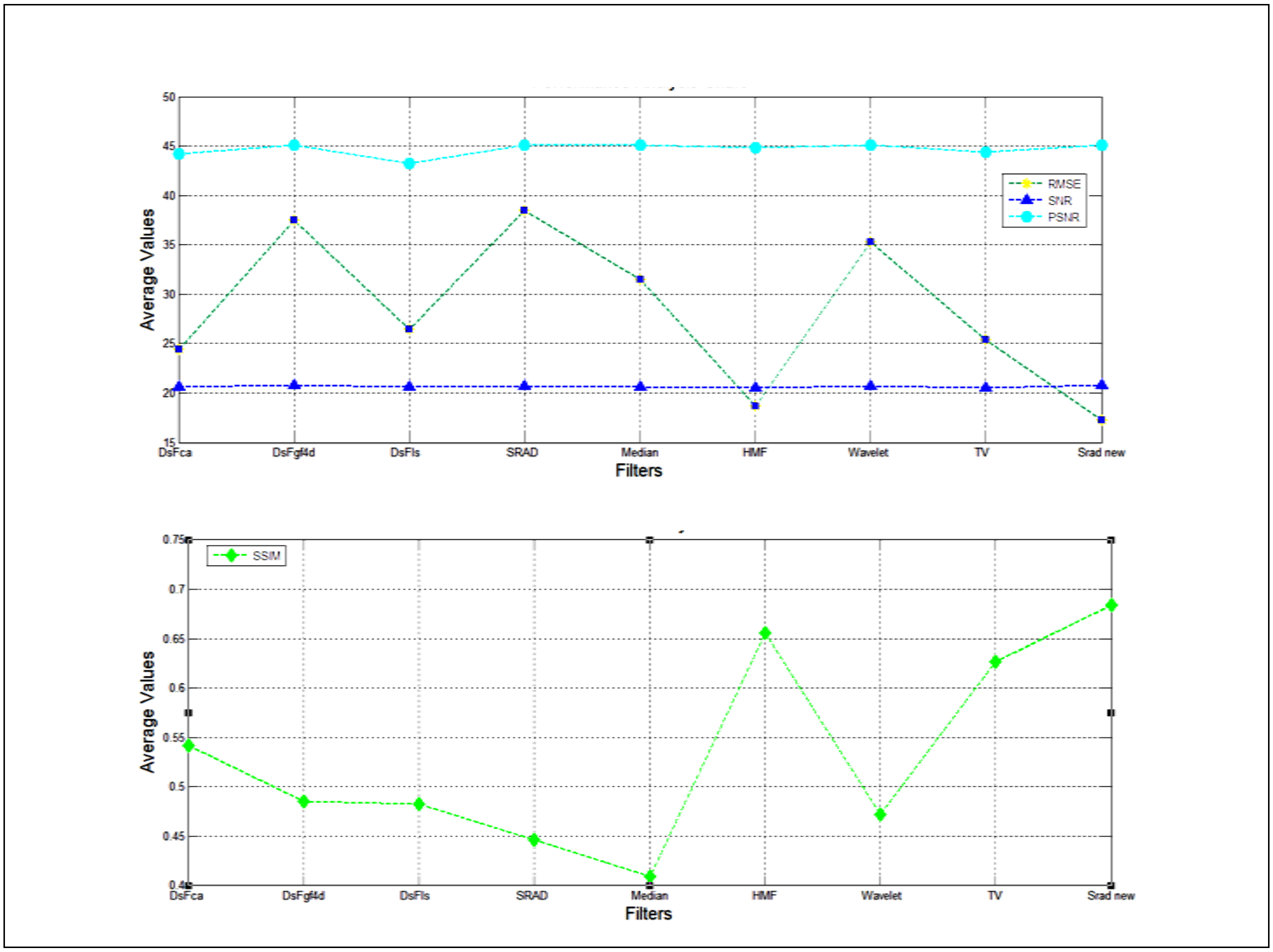
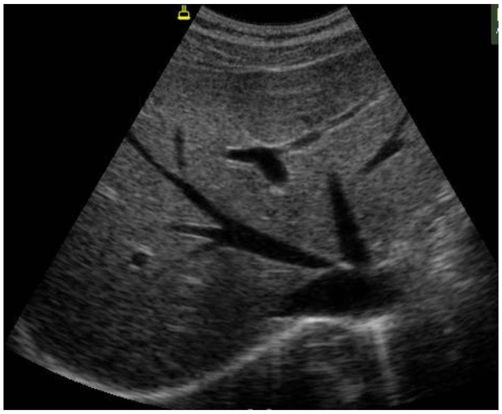


Figure 5.18: Performance analysis graph to image quality evaluation metric for fetal image (noise**σn** =0.5).



(a) Original image

(b) Noisy image

(c) Linear scaling gray level filter(DsFca)

(d) Geometric filter (DsFgf4d)

(e) Linear scaling filter(DsFls)

(f) Speckle reducing
anisotropic diffusion filter

(g) Median filter (med)

(h) Hybrid median filter (hmf)

(i) Wavelet filter

(j) Total variation

(k) Srad hybrid median filter
(SRAD new)

Figure 5.19 : Results of liver despeckled by various filter on multiplication noise (σn=0.05).

Table5.9: Image quality evaluation metrics computed for the **liver (σn =0.05)**at statistical measurement of RMSE, PSNR ,SNR and SSIM for different filter types and for SRADHMF .

Filter name	Image Quality Evaluation Metrics			
	RMSE	**SNR**	**PSNR**	**SSIM**
DsFca filter	11.0212	18.5023	44.0741	0.6992
DsFgf4d Filter	17.6988	**18.5293**	**45.0013**	0.7334
DsFls Filter	15.4541	**18.5289**	42.2283	0.6493
SRAD filter	**10.0797**	18.3719	44.3935	0.6703
Median filter	12.7495	18.3650	43.2634	0.7840
HMF	**9.4148**	18.3718	43.5002	**0.7940**
Wavelet filter	19.5141	18.5480	**44.6616**	0.7510
TV	17.6283	18.5006	44.1924	**0.8327**
Srad new	**7.6240**	**18.4401**	**44.7206**	**0.8099**

Bold number indicates the best values.

*Root mean square error (RMSE), signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR),structural similarity index(SSIM).

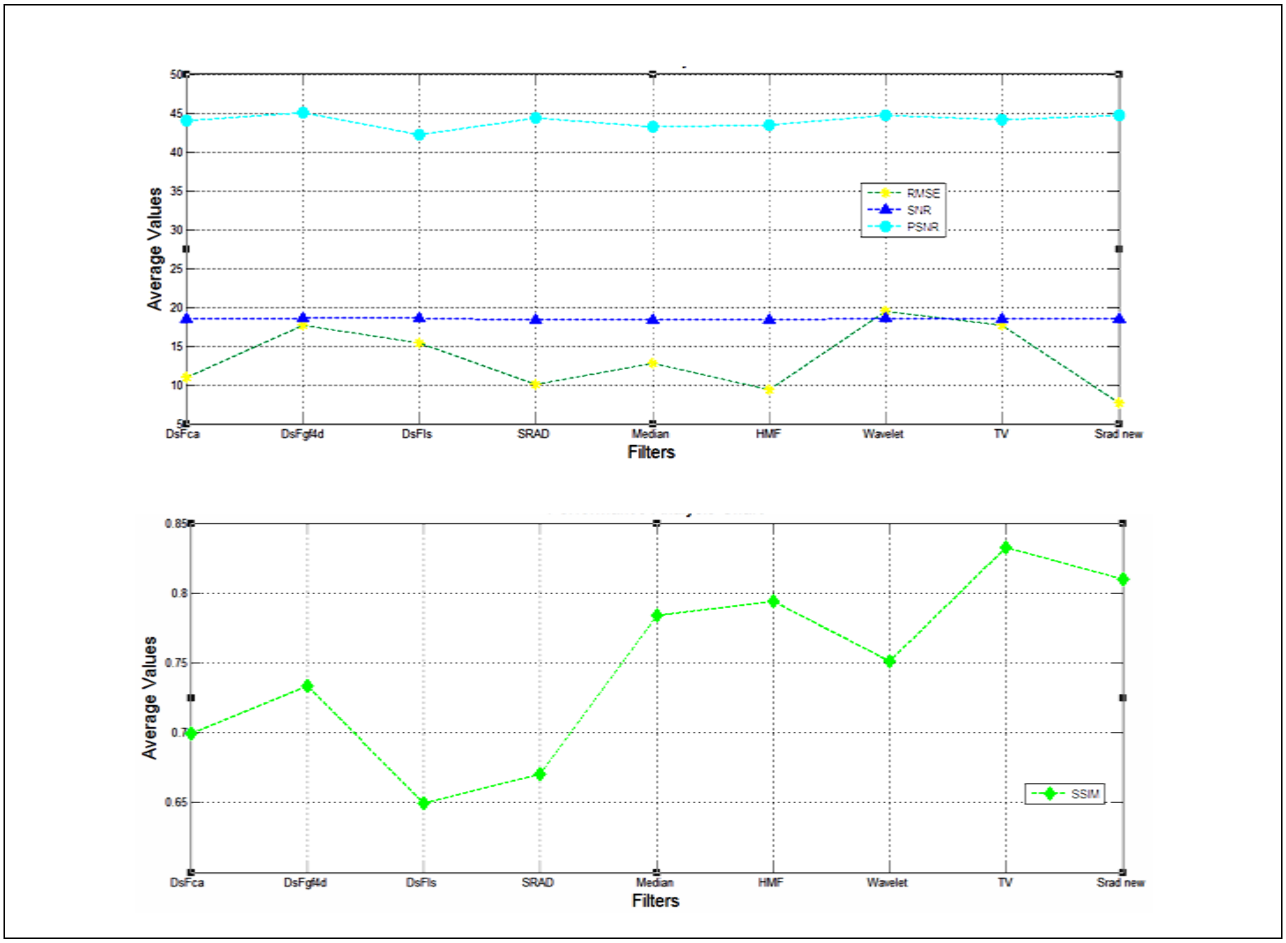
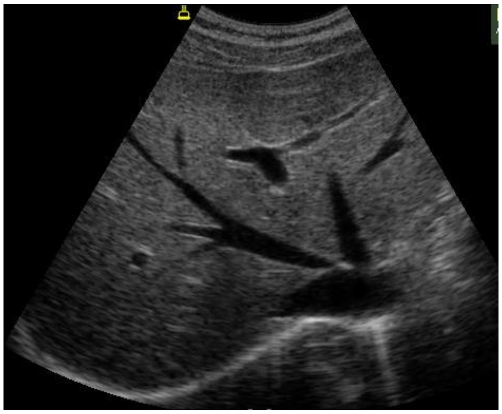


Figure 5.20 Performance analysis graph to image quality evaluation metric for liver image

(noise**σn** =0.05).



(a) Original image

(b) Noisy image

(c) Linear scaling gray level filter(DsFca)

(d) Geometric filter (DsFgf4d)

(e) Linear scaling filter(DsFls)

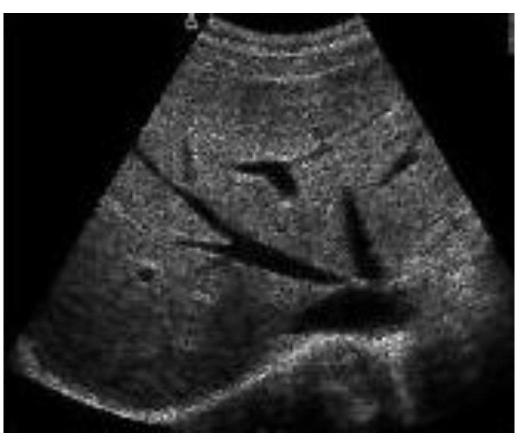
(f) Speckle reducing anisotropic diffusion filter

(g) Median filter (med)

(h) Hybrid median filter

(i) Wavelet filter

(j) Total variation



(k) Srad hybrid median filter (SRAD new)

Figure 5.21 : Results of liver despeckled by various filter on multiplication noise (σn=0.5)

Table5.10: Image quality evaluation metrics computed for the **liver (σn =0.5)**at statistical measurement of RMSE, PSNR ,SNR and SSIM for different filter types and for SRAD HMF .

Filter name	Image Quality Evaluation Metrics			
	RMSE	**SNR**	**PSNR**	**SSIM**
DsFca filter	20.7081	**18.8367**	44.8471	0.6500
DsFgf4d Filter	32.5679	**18.8436**	**44.9843**	0.4590
DsFls Filter	18.6283	18.8035	42.1681	0.6356
SRAD filter	**16.3678**	18.5798	**44.8652**	0.6392
Median filter	19.5775	18.4127	44.3486	0.6412
HMF	18.9317	18.4653	43.1833	**0.6697**
Wavelet filter	31.7211	18.8328	44.5987	0.4934
TV	**11.9094**	18.3942	43.9810	**0.6858**
Srad new	**15.5099**	**18.6402**	**44.8652**	**0.6644**

Bold number indicates the best values.

*Root mean square error (RMSE), signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR),structural similarity index(SSIM).

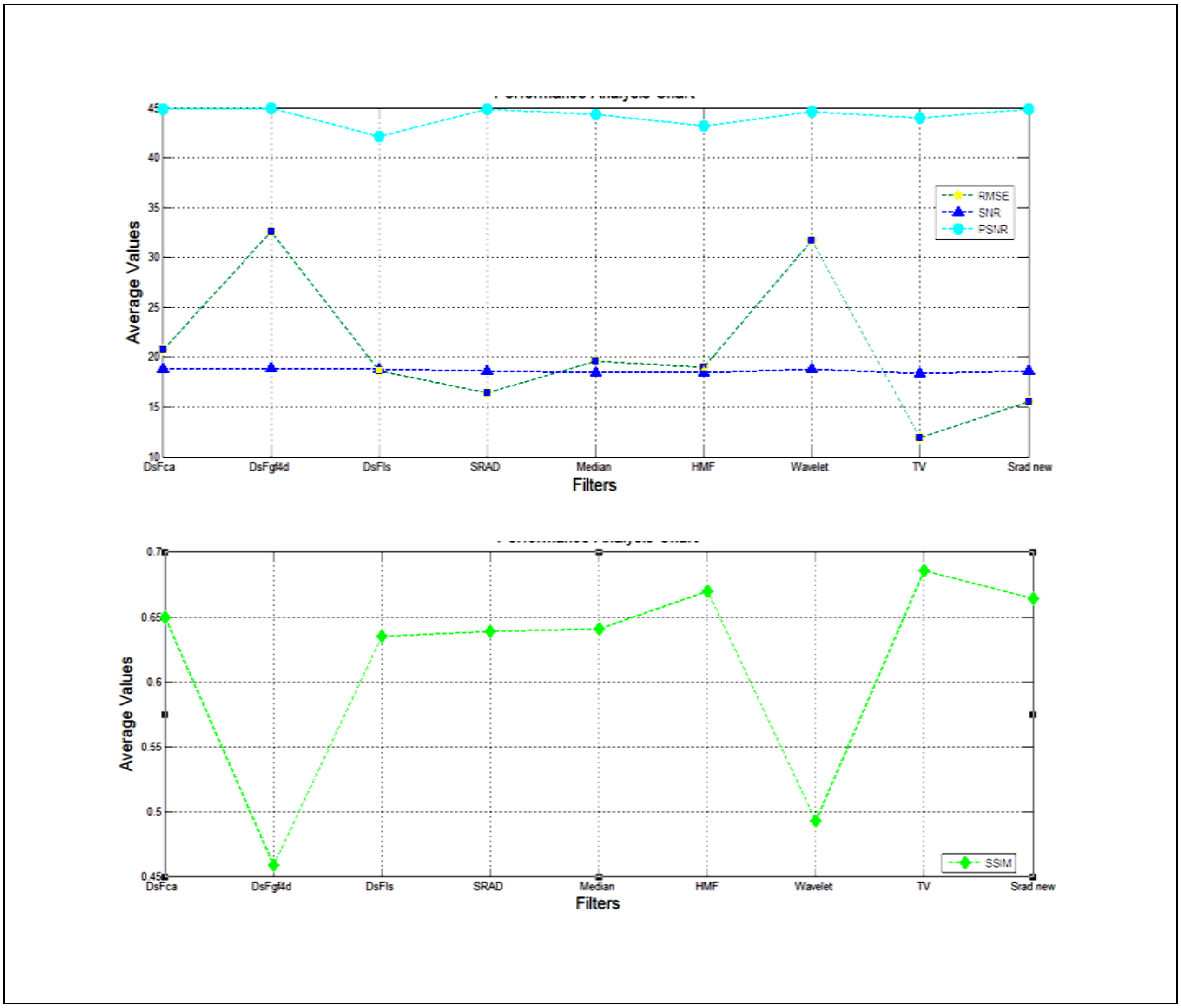
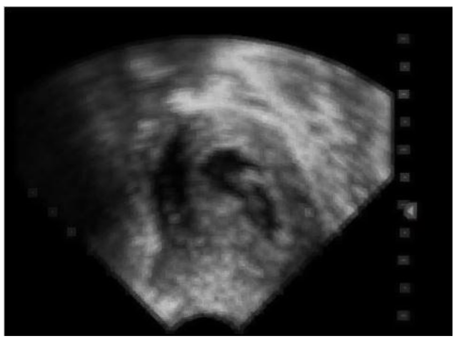


Figure 5.22: Performance analysis graph to image quality evaluation metric for liver image

(noise**σn** =0.5).

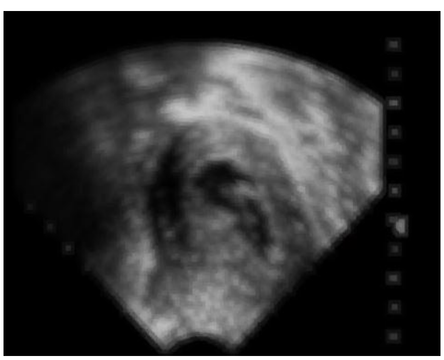
(a) Original image

(b) Noisy image

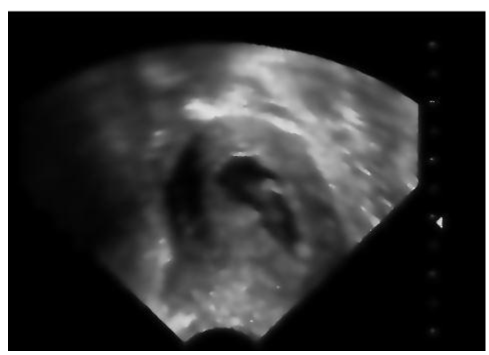


(c) Linear scaling gray level filter(DsFca)

(d) Geometric filter (DsFgf4d)



(e) Linear scaling filter(DsFls)

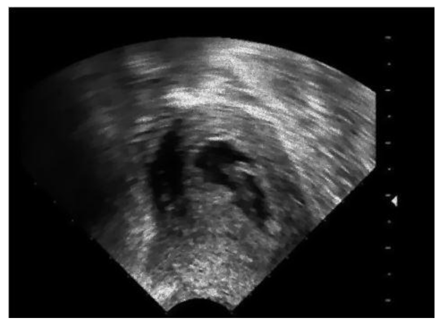


(f) Speckle reducing anisotropic diffusion filter

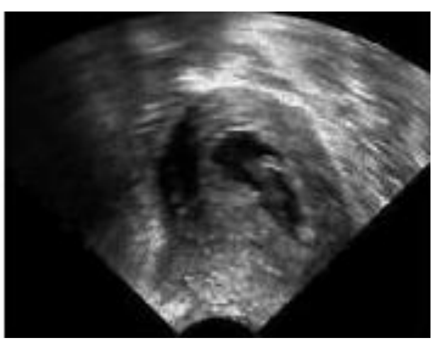
(g) Median filter (med)

(h) Hybrid median filter (hmf)

(i) Wavelet filter



(j) Total variation



(k) Srad hybrid median filter
(SRAD new)

Figure 5.22: Results of vagina despeckled by various filter on multiplication noise (σn=0.05)

Table5.11: Image quality evaluation metrics computed for the **vagina (σn =0.05)**at statistical measurement of RMSE, PSNR ,SNR and SSIM for different filter types and for SRAD HMF .

Filter type	**Image quality evaluation metrics**			
	RMES	**SNR**	**PNSR**	**SSIM**
DsFca filter	19.7348	19.8433	44.0225	0.7176
DsFgf4d Filter	24.2321	**19.8585**	**44.9663**	0.8021
DsFls Filter	**15.6629**	19.8044	44.0225	0.6631
SRAD filter	**16.0393**	19.7652	44.1712	0.7290
Median	16.3242	19.7186	44.7186	**0.8248**
HMF	16.6171	19.7508	44.5044	**0.8711**
Wavelet	16.2647	19.7948	**45.0013**	0.7552
TV	19.7348	**19.8433**	44.0225	0.7176
Srad new	**13.0861**	**20.4837**	**45.0522**	**0.7865**

Bold number indicates the best values.

*Root mean square error (RMSE), signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR),structural similarity index(SSIM).

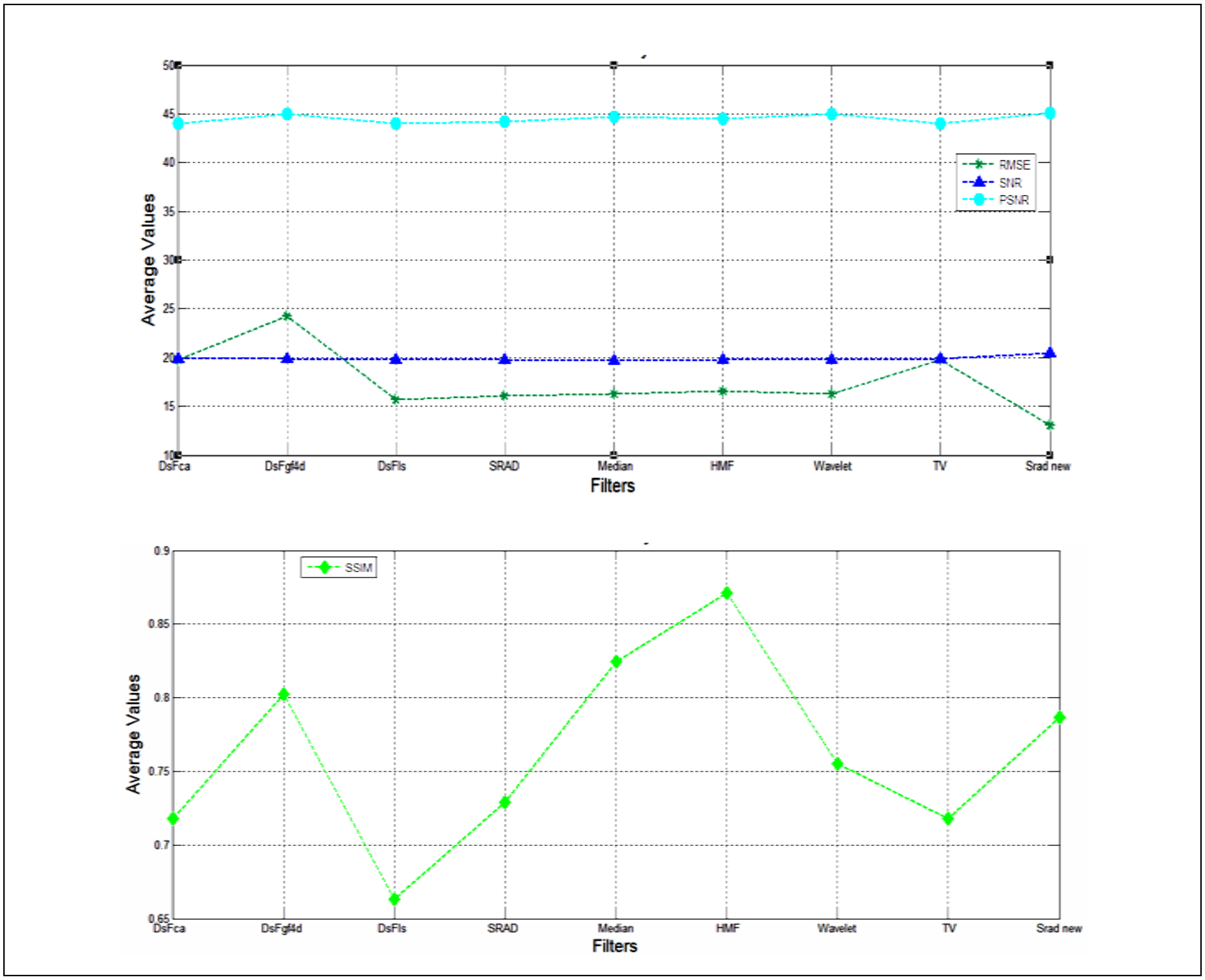
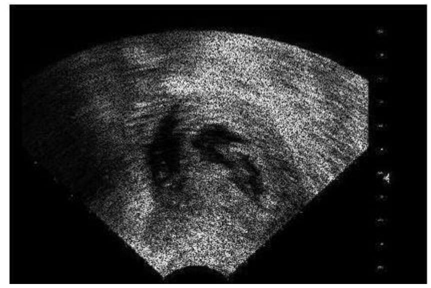


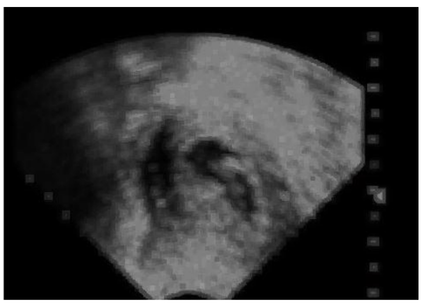
Figure 5.24: Performance analysis graph to image quality evaluation metric for vagina image

(noise**σn** =0.05).

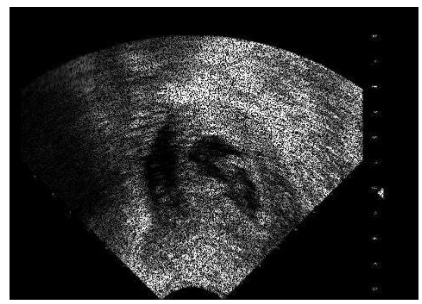
(a) Original image



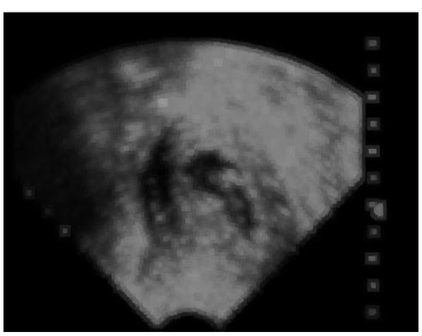
(b) Noisy image



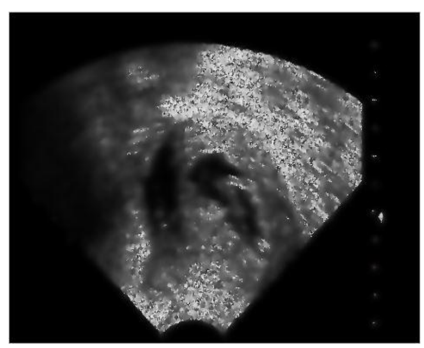
(c) Linear scaling gray level filter(DsFca)



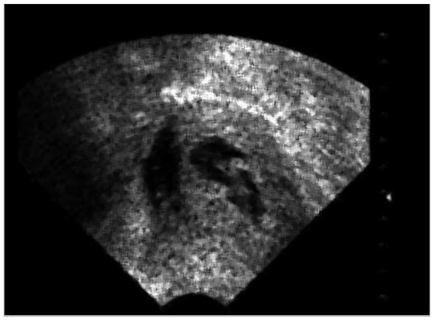
(d) Geometric filter (DsFgf4d)



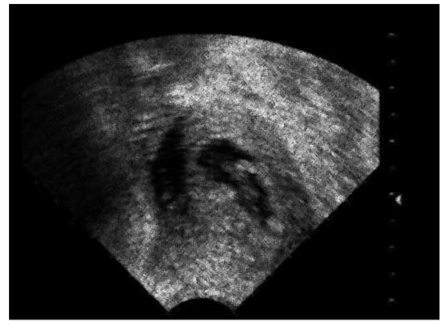
(e) Linear scaling filter(DsFls)



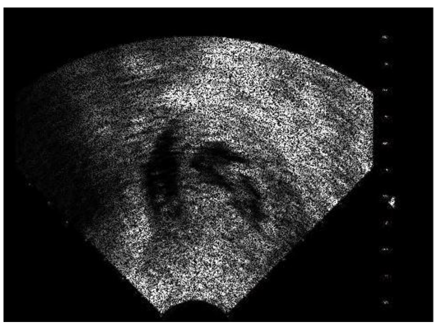
(f) Speckle reducing anisotropic diffusion filter



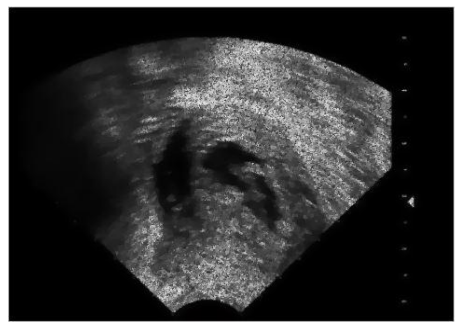
(g) Median filter (med)



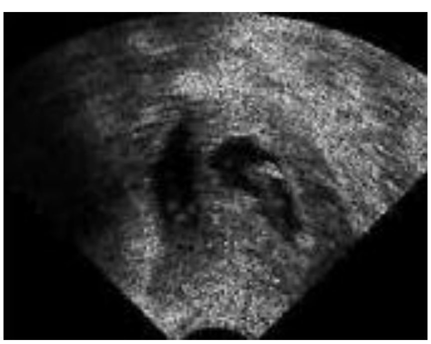
(h) Hybrid median filter (hmf)



(i) Wavelet filter



(j) Total variation



(k) Srad hybrid median filter (SRAD new)

Figure 5.25 : Results of vagina despeckled by various filter on multiplication noise (σn=0.5)

Table5.12: Image quality evaluation metrics computed for the **vagina (σn =0.5)**at statistical measurement of RMSE, PSNR, SNR and SSIM for different filter types and for SRADHMF .

Filter type	**Image quality evaluation metrics**			
	RMES	**SNR**	**PNSR**	**SSIM**
DsFca filter	24.3510	19.7474	43.2458	0.6630
DsFgf4d Filter	37.9514	**19.9703**	**45.1205**	0.5722
DsFls Filter	23.0833	19.7273	43.3246	0.6327
SRAD filter	26.1770	19.9009	44.8454	**0.7370**
Med	27.2703	19.7837	45.1035	0.6938
HMF	**19.2268**	19.6831	44.8110	**0.7457**
wavelet	37.8085	**19.9619**	**45.1205**	0.5771
TV	**20.3000**	19.6362	43.7631	0.6941
Srad new	**25.4869**	**20.1490**	**44.8984**	**0.5267**

Bold number indicates the best values.

*Root mean square error (RMSE), signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR),structural similarity index(SSIM).

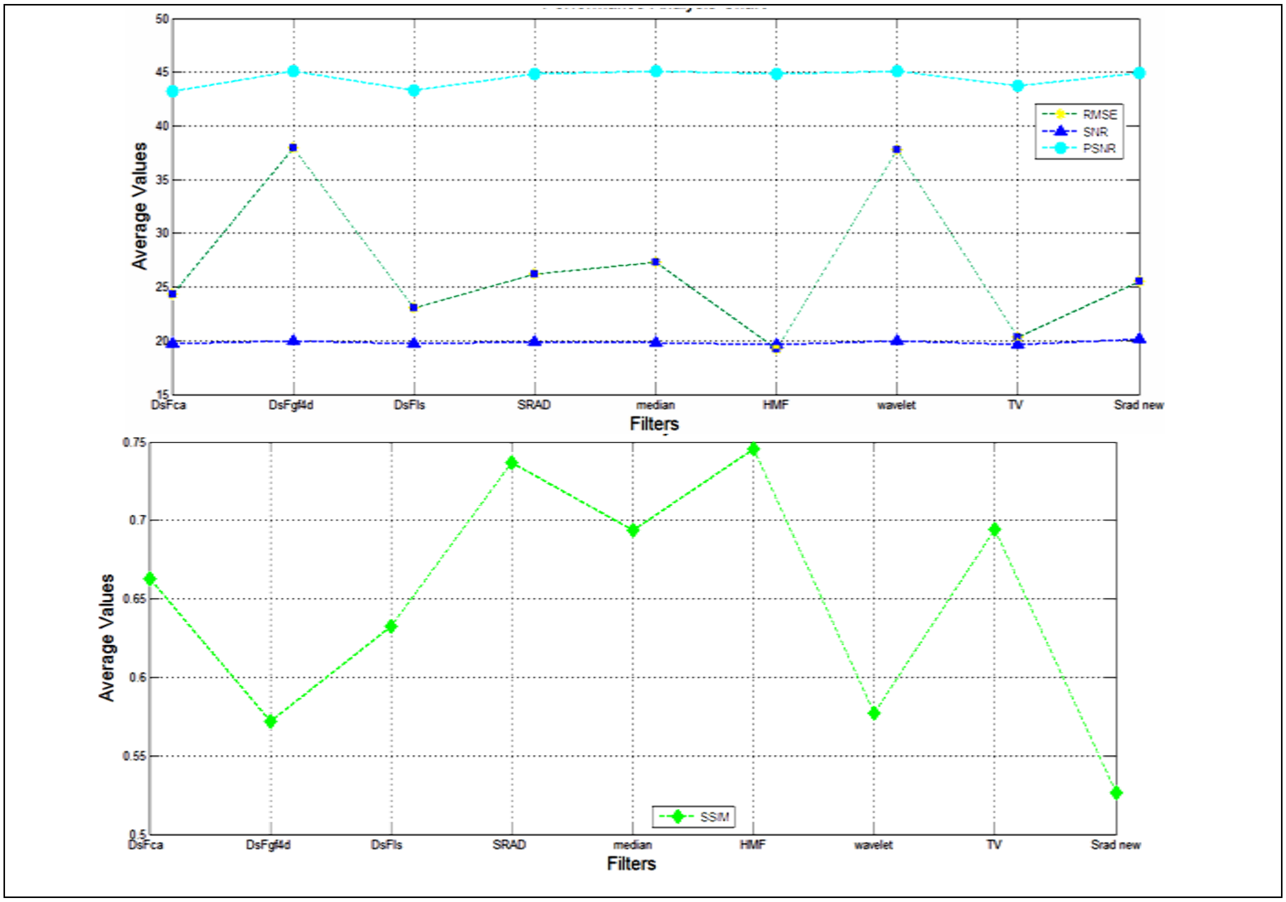


Figure 5.26: Performance analysis graph to image quality evaluation metric for vagina image (noise**σ** =0.5).

Most importantly, adespeckle filtering analysis and evaluation framework is proposed for selecting the most appropriate filter or filters for the images under investigation. The filters can be further developed and evaluated at a larger scale, texture analysis, image quality evaluation metrics, and visual evaluation by experts.

From figures 5.15, 5.17, 5.19, 5.21, 5.23, 5.25, show an ultrasound image (a) with noisy (b) and the despeckled images. In(c) can see that, the linear scaling gray level filter(DsFca) has high degree of blurring and was affect on gray level, because it is compute the mean of all pixels whose difference in the gray level with the intensity(the middle pixel in the moving window) is lower than or equal to a given threshold, (d) although the result obtained by geometric despeckle filter(DsFgf4d) given poor performance for removing the speckle noise from the ultrasound image, it is lead to increasing the contrast significantly of the image. (e) Show the result obtained by liner scaling (DsFls*)* filter scales the pixel intensities by finding the maximum and the minimum gray-level values in every moving window, and then replaces the middle pixel with the average of them also give blurred image. (f) show the result of speckle reducing anisotropic diffusion filtering(srad), it is better for preserves the edges as a comparison with the other despeckle filtering techniques and subjectively has good result, and referred to evaluated metrics, it was also given bad results, (g) show the result obtained by median despeckle filter, which don't able to remove the speckle and produced blurred edges in the filtered image .figure(h)show The result of hybrid median filter(hmf) that given better edge preserving characteristics than normal median filter, (i)the result through wavelet despeckle filtering perceived that it's moderate in order of variance decreasing but execute to decrease the contrast,(J)show the result obtained by total variation despeckle filter(TV)methods. We see that most of the unwanted details haven’t been removed efficiently, whilst preserving important details such as edges.

From table5.7, 5.8, 5.9, 5.10, 5.11, 5.12, tabulates the image quality evaluation contains the metric result of filters under study, The best visual results were obtained for the filters DsFgf4d, wavelet, DsFca and TV because with higher SNR and PSNR and lower RMSE and Best values for the SSlM, but visually, smoothed the image. Loosing subtle details are been observed.

From table 5.13,5.15,5.17,5.19,5.21 show that the srad hybrid median filter has best performances but in other tables has result fallback, that indicate to the performance of despeckled filters are depended on image's features and quantity of speckle noise which applied on image.

Using SRAD hybrid median filter method has been applied to improve performance of both speckle reduction and edge preservation. Although these SRAD methods could improve the speckle reduction and edge preservation, the low-contrast edges were still blurred with speckle that remain in the high-intensity region, this solved by using the hybrid median filter to decrease the blurring and reduce speckle noise.

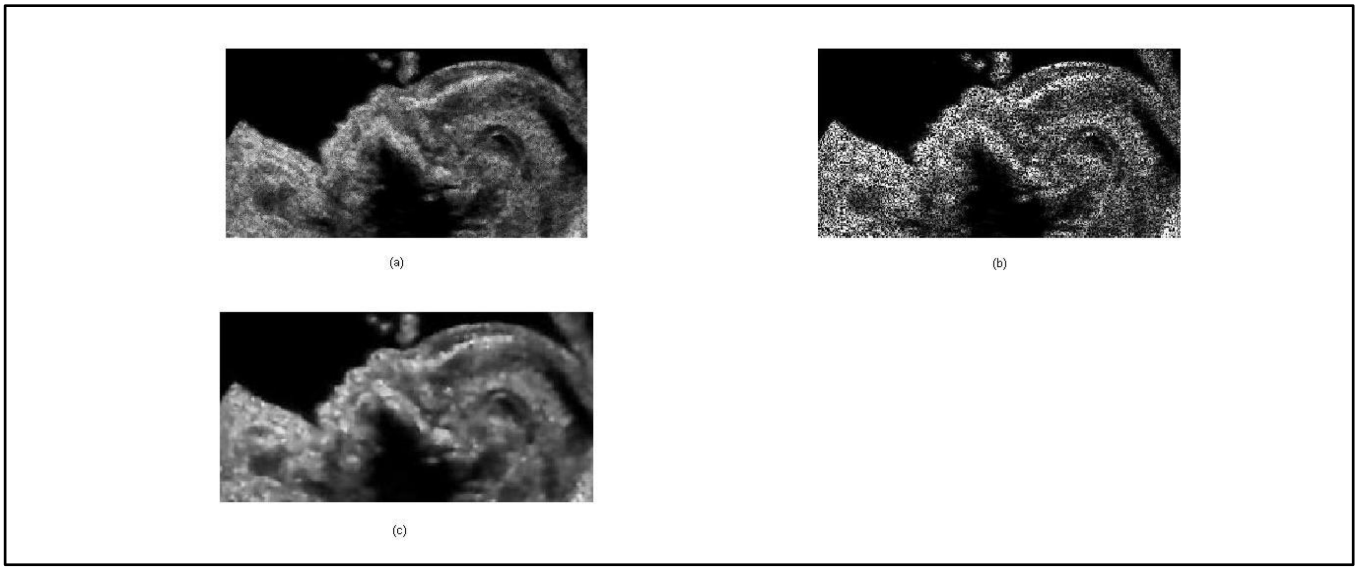


Figure 5.27 (a),(b),(c) images filtered by hybrid median filter and SRAD and SRAD hybrid median filter respectively from speckled fetal image with variance (σn=0.5).

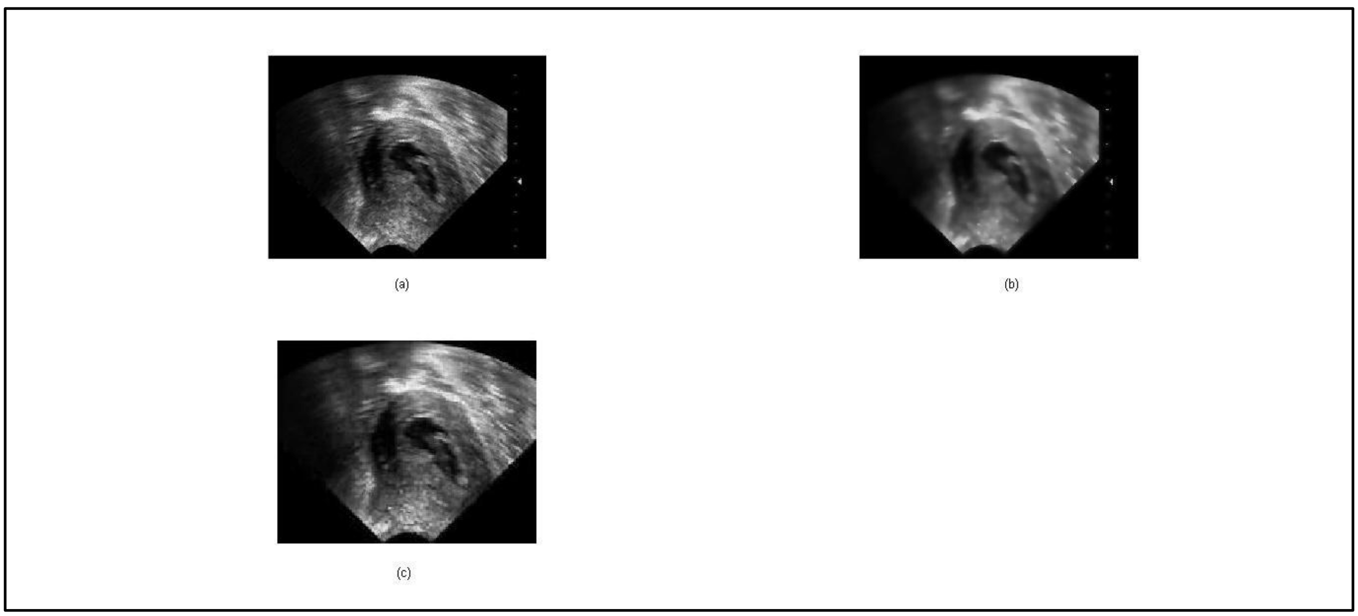


Figure 5.28 (a),(b),(c) images filtered by hybrid median filter and SRAD and SRAD hybrid median filter respectively from speckled vagina image with variance (σn=0.05).

The both proposed method modified hybrid median filter (mhmf) and SRAD hybrid median filter(SRAD new) takes full advantage of combine and modify filters to reduce speckle noise .Experimental results not only to enhancement of those filters but to obtain filters which capable to get a good result referred to quality evaluation metric. while, subjectively, can be used in diagnostic and therapeutic terms.

6.1 Conclusion

Ultrasound imaging is a widely used and safe medical diagnostic technique, due to its noninvasive nature, low cost, capability of forming real time imaging, and the continuing improvements in image quality.[6]

in diagnosis of diseases Ultrasonic devices are frequently used by healthcare professionals. The main problem during diagnosis is the distortion of visual signals obtained which is due to the consequence of the coherent of nature of the wave transmitted. These distortions are termed as speckle N Arbitration between the perpetuation of useful diagnostic information and noise suppression must be treasured in medical images. [16]

The present study focuses on proposing a technique to reduce speckle noise from ultrasonic devices. [12] he results achieved from the other speckle noise reduction techniques demonstrate its higher performance for speckle reduction. [16]

Here there is two proposed method for speckle noise reduction in ultrasound image, first proposed method is (modified hybrid median filter) we modify the hybrid median filter to get best result than the normal one.

The optimization of second proposed method "SRAD hybrid median lechnique" is obtained (SRAD new) algorithm. With the join SRAD,with hybrid median technique have demonstrated more robust estimation and more flexibility over other filters. In the evaluation in several image applications including image interpolation and impulsive noise reduction, both quantitative and qualitative comparison showed that the SRAD hybrid median filter exhibit improved performance and merit further attention.

In this project. The both proposed method modified hybrid median filter (mhmf) and SRAD hybrid median filter(SRAD new) takes full advantage of combine and modify filters to reduce speckle noise .Experimental results not only to enhancement of those filters but to obtain filters which capable to get a good result referred to quality evaluation metric. while, subjectively, can be used in diagnostic and therapeutic terms .

6.2 Recommendation

1- For SRAD hybrid median filter, the hybrid median filter can be changed by modified hybrid filter proposed in this thesis.

2- Use edge detection methods on SRAD hybrid median filter, to detect and measure the ability of this filter to preserve image edges.

3- Use Edge Preservation Factor (EPF) as on of Image Quality Evaluation Metrics to evaluate ability of the filter edge preservation.

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