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Speckle Noise Reduction in Ultrasound Images by Using Hybrid DWT Technique

الحد من ضوضاء الرقطة في صور الموجات فوق الصوتية باستخدام تقنية هجينة DWT

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الآية

قال تعالى:

(ألم تَرَ أنَّ الله أنزلَ من السَّماء ماءً فأخرجنا به ثمراتٍ مُخْتلفاً ألوانُها ومن الجبالِ جُدَدٌ بيضٌ وحمرٌ مختلفٌ ألوانُها وغرابيبُ سُودٌ ومن النَّاسِ والدَّوابِّ والأنعامِ مُختلفٌ ألوانُهُ كذلك إنَّما يَخشى الله من عبادهِ العلماءُ إنَّ الله عزيزٌ غفو)

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Dedication

I dedicate this work to those who taught me success and patience to those who did not spare me anything...my Father, and to those who taught me and suffered the difficulties of what I am in and when the worries surround me , I swim in the sea of tenderness to ease my pain ..my Mother , and my brothers and my Friends who stood with me ,and then to each of those who taught me characters became a light bulb lights the way in front of me .

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ABBREVITION

US	Ultrasound
US B-mode	Ultrasound Brightness mode
US A-mode	Ultrasound Amplitude mode
US M-mode	Ultrasound Motion mode
SND	Scatter Number Density
FFS	Fully Formed Speckle pattern
NRLR	Non Randomly distributed with Long-Range order
NRSR	Non Randomly distributed with Short-Range order
PDF	Probability Density Function
DsFgf4d	Geometric Filter
DsFls	Linear scaling gray level filter
SRAD	Speckle reduction anisotropic diffusion
PDE	Partial Differential Equation
TV	Total Variation Filters

MSE	Mean Square Error
RMSE	Root Mean Square Error
SNR	Signal to Noise Ratio
PSNR	Peak Signal to Noise Ratio
SSIM	Structural Similarity Index
DWT	Discreet Wavelet Transform
IDWT	Inverse Discreet Wavelet Transform

ABSTRACT

ULTRASOUND (US) imaging application in medicine and other fields is enormous. It has several advantages over other medical imaging modalities. The use of ultrasound in diagnosis is well established because of its noninvasive nature, portable, accurate, low cost imaging modality, capability of forming real time imaging and continuing improvement in image quality. However the usefulness of ultrasound imaging is degraded by the presence of signal dependant noise know as speckle nose. It is well-known that speckle is a multiplicative noise that degrades the visual evaluation in ultrasound imaging decrease the human ability to identify pathological from normal tissue because it depends on the structure of the image tissue and various imaging parameters .The main purpose for speckle reduction in medical ultrasound imaging to improve the image into keep in mind that certain speckle contains diagnostic information and should be retained.

The objective of this thesis is to give an overview about types of speckle reduction techniques in ultrasound imaging and to present a new technique of speckle reduction.

A new speckle reduction method of medical ultrasound images was proposed: the discrete wavelet transforms (DWT).

It has been found that quality evaluation metrics the proposed methods performed better than all other methods while the structural details and results preserved edges and features in a better way than other despeckling filters.

المستخلص

تستخدم الموجات الصوتية في التصوير الطبي وغيرها من المجالات هائلة. لديها العديد من المزايا على طرائق التصوير الطبي الأخرى. ثبوت استخدام الموجات فوق الصوتية في التشخيص أيضا بسبب طبيعة موسع لها، ومحمولة ودقيقة، وانخفاض تكاليف التصوير ، والقدرة على تشكيل التصوير في الوقت الحقيقي والتحسن المستمر في جودة الصورة. ومع ذلك فإن فائدة التصوير بالموجات فوق الصوتية قد تدهور بسبب وجود إشارة تعتمد الضوضاء تعرف باسم ضوضاء رقطة. ومن المعروف أن البقع هو الضوضاء المضاعف التي تحط تقييم البصرية في مجال التصوير بالموجات فوق الصوتية يقلل من قدرة الإنسان على تحديد الانسجةالمريضة من الأنسجة الطبيعية لأنها تعتمد على بنية الأنسجة صورة والمعلمات التصوير المختلفة.
الهدف الرئيسي للحد من البقع في مجال التصوير الطبي بالموجات فوق الصوتية لتحسين الصورة في نضع في اعتبارنا أن بعض البقع يحتوي على معلومات تشخيصية ويجب الاحتفاظ بها.

والهدف من هذه الرسالة هو لإعطاء لمحة عامة عن أنواع من تقنيات للحد من البقع في مجال التصوير بالموجات فوق الصوتية، وتقديم تقنية جديدة للحد من البقع.
تم اقتراح طريقة جديدة للحد من رقطة التصوير بالموجات فوق الصوتية الطبية: تحويل المويجات المنفصلة .
وقد وجد أن الطرق المقترحة هي الأفضل أداء من كل الطرق الأخرى من حيث مقاييس نقييم جودة الصورة ,لانها تحافظ على تفاصيل الصورة و النتائج الهيكلية للحواف والملامح بشكل افضل مقارنة مع الطرق الاخرى لمرشحات ازالة الرقطة.

Chapter one

Introduction

1.1General review

In medical imaging modalities, ultrasound imaging has been considered to be noninvasive and most prevalent diagnostic tool for imaging organs and soft tissue structures of the human body. This is often preferred due to its non-ionizing radiations with low cost. But this imaging has major disadvantage of having Speckle. Speckle in ultrasound imaging is caused by the interference of energy from randomly distributed scatters, too small to be resolved by imaging system. The presence of speckle results degradation in image quality and makes it difficult for human interpretation and diagnosis. The intent of speckle reduction is to remove the distracting speckle pattern without reducing the detail in the ultrasound image. [6]

Speckle noise is introduced as a major noise factor, which limits image resolution and hinters further image processing analysis in ultrasound images. We then introduce different despeckle filtering techniques that may be applied as a preprocessing step for denoising of ultrasound images. [9]

1.2 Problem of the Statement

Speckle noise is the primary factor that limits the contrast resolution in diagnostic ultrasound imaging, thereby limiting the detectability of small low-contrast lesions and making the ultrasound images generally difficult for the no specialist to interpret, Because of the speckle presence.

1.3 General objective

- Reduction the speckle noise (multiplicative noise) that present in ultrasound images.
- Study about despeckle techniques, used eight filters that applied on the chosen images.
- Evaluation of despeckle filtering based on image quality evaluation matrices.
-

1.4 Specific objective

- Speckle noise reduction in ultrasound images using DWT technique.
- Proposed new method as despeckle filter based on DWT 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.
Then the quality evaluation metrics was found from all methods to compare the performance of those filters.

DWT technique &total variation filter

This proposed technique is the discrete wavelet transform (DWT) with total variation & wavelet filters.

The dwt decomposes input image into four component namely LL, HL, LH and HH where the first letter corresponds to applying either a low pass frequency operation or high pass frequency operation to the row, and the second letter refers to the filter applied to the columns, the lowest resolution level LL consist of the approximation part of the original image ,filtered by use total variation filter. The remaining three resolution levels consist of the detail parts and give the vertical high (LH), horizontal high (HL), and high (HH) frequencies , the three parts filtered by wavelet filter.

1.6 thesis layout

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

Chapter two

Theoretical Background

2.1 waves

Mechanical Waves are waves which propagate through a material medium (solid, liquid, or gas) at a wave speed which depends on the elastic and inertial properties of that medium. There are two basic types of wave motion for mechanical waves: **longitudinal** waves and **transverse** waves. The animations below demonstrate both types of wave and illustrate the difference between the motion of the wave and the motion of the particles in the medium through which the wave is travelling .[5]

2.2 Sound waves

The generation of a sound wave requires not only vibration, but also an elastic medium in which the disturbance created by that vibration can be transmitted. To say that air is an elastic medium means that air, like all other matter, tends to return to its original shape after it is deformed through the application of a force. The prototypical example of an object that exhibits this kind of restoring force is a spring. To understand the mechanism underlying sound propagation, it is useful to think of air as consisting of collection of particles that are connected to one another by springs, with the springs representing the restoring forces associated with the elasticity of the medium. [4]

2.3 frequencies of sound waves

The range of human hearing in the young is approximately 20 Hz to 20 kHz—the higher number tends to decrease with age (as do many other things). It may be quite normal for a 60-year-old to hear a maximum of 16,000 Hz .[3]

2.4 About ultrasound

Ultrasound imaging uses sound waves to produce pictures of the inside of the body. It is used to help diagnose the causes of pain, swelling and infection in the body's internal organs and to examine a baby in pregnant women and the brain and hips in infants. It's also used to help guide biopsies, diagnose heart conditions, and assess damage after a heart attack. Ultrasound is safe, noninvasive, and does not use ionizing radiation.[2]

This procedure requires little to no special preparation. Your doctor will instruct you on how to prepare, including whether you should refrain from eating or drinking beforehand. Leave jewelry at home and wear loose, comfortable clothing. You may be asked to wear a gown.[2]

Ultrasound is safe and painless, and produces pictures of the inside of the body using sound waves. Ultrasound imaging, also called ultrasound scanning or Sonography, involves the use of a small transducer (probe) and ultrasound gel placed directly on the skin. High-frequency sound waves are transmitted from the probe through the gel into the body. The transducer collects the sounds that bounce back and a computer then uses those sound waves to create an image. Ultrasound examinations do not use ionizing radiation (as used in x-rays), thus there is no radiation exposure to the patient. Because ultrasound images are captured in real-time, they can show the structure and movement of the body's internal organs, as well as blood flowing through blood vessels.[2]

Ultrasound imaging is a noninvasive medical test that helps physicians diagnose and treat medical conditions.

Conventional ultrasound displays the images in thin, flat sections of the body. Advancements in ultrasound technology include three-dimensional (3-D) ultrasound that formats the sound wave data into 3-D images.

A Doppler ultrasound study may be part of an ultrasound examination. Doppler ultrasound**,** also called color Doppler ultrasonography, is a special ultrasound technique that allows the physician to see and evaluate blood flow through arteries and veins in the abdomen, arms, legs, neck and/or brain (in infants and children) or within various body organs such as the liver or kidneys.[2]

There are three types of Doppler ultrasound:

- Color Doppler uses a computer to convert Doppler measurements into an array of colors to show the speed and direction of blood flow through a blood vessel.
- Power Doppler is a newer technique that is more sensitive than color Doppler and capable of providing greater detail of blood flow,

especially when blood flow is little or minimal. Power Doppler, however, does not help the radiologist determine the direction of blood flow, which may be important in some situations.

- Spectral Doppler displays blood flow measurements graphically, in terms of the distance traveled per unit of time, rather than as a color picture. It can also convert blood flow information into a distinctive sound that can be heard with every heartbeat.[2]

2.5 Ultrasound interaction with the medium

The interaction between the medium and the ultrasound emitted into the medium can be described by the following phenomena: The echoes that travel back to the transducer and thus give 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 (Danish: *spredning*) can be thought of, when one shines strong light on the tip of a needle: light is scattered in all directions. 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).[4]

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.[4]

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 .[4]

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

2.5.1 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. To describe this quantitatively, the *specific acoustic impedance*, *z*, is introduced. In a homogeneous medium it is defined as the ratio of pressure to particle velocity in a progressing plane wave, and can be shown to be the product of the physical density, ρ, and acoustic propagation velocity c of the medium. [4]

2.5.2 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 *back*scattering. [4]

The scattering from particles much less than a wavelength is normally referred to as *Rayleigh* scattering. The intensity of the scattered wave increases with frequency to the power of four.
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 cm, much smaller than the wavelength of clinical ultrasound. [4]

.2.5.3 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 2. More details on this can be found in the literature.
Pure absorption can be observed by sending ultrasound through a viscous liquid such as oil. [4]

2.5.4 Attenuation

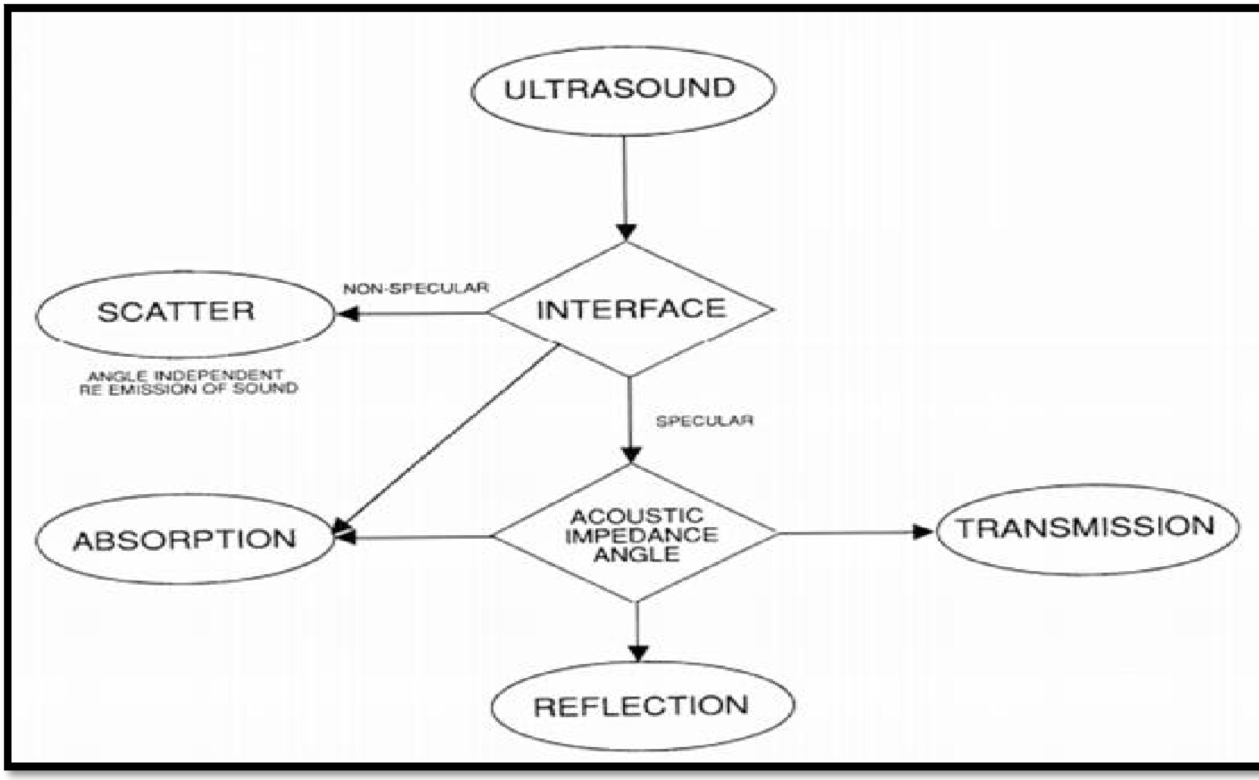
The loss of intensity (or energy) of the forward propagating wave due to reflection, refraction, scattering and absorption is denoted attenuation. The intensity is a measure of the power through a given cross-section; thus the units are W/m2. It can be calculated as the product between particle velocity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

$$I = Pu = p2/z \qquad (2.1)$$

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:

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

Where α (in units of m-1) is the attenuation coefficient. α depends 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]



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

2.6 modes of ultrasound

The two main scanning modes are A- and B-modes. Other modes used are M mode, duplex ultrasound, color-coded ultrasound, and power Doppler ultrasound, which will be briefly introduced below.

A-mode refers to amplitude mode scanning, which is mainly of historical interest. In this mode, the strength of the detected echo signal is measured and displayed as a continuous signal in one direction. A-mode is a line, with strong reflections being represented as an increase in the signal amplitude. This scanning technique has the limitation that the recorded signal is 1D with limited anatomical information. A mode is no longer used, especially for the assessment of cardiovascular disease. Its use is restricted to very special uses such as in ophthalmology to perform very accurate measurements of distance.[9]

B-mode refers to the brightness mode. In B-mode, echoes are displayed as a 2D gray scale image. The amplitude of the returning echoes is represented as dots (pixels) of an image with different gray values. Advances in B-mode ultrasound have resulted in improved anatomic definition, which has enabled plaque characterization .[9]

M-mode is used in cardiology, and it is actually an A-scan plotted against time. The result is the display of consecutive lines plotted against time. Using this mode, detailed information may be obtained about various cardiac dimensions and also the accurate timing of vascular motion.[9]

D-mode (D=Doppler) this imaging mode is based on the Doppler Effect, i.e. change in frequency (Doppler shift) caused by the reciprocal movement of the sound generator and the observer. Diagnostic ultrasound uses the change in frequency of ultrasound signal backscattered from red blood cells. The frequency of the reflected ultrasound wave increases or decreases according to the direction of blood flow in relation to the transducer. [9]

2.7 ultrasound imaging system

Figure 2.2 shows a functional block diagram of an ultrasound imaging system. 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 .[10]

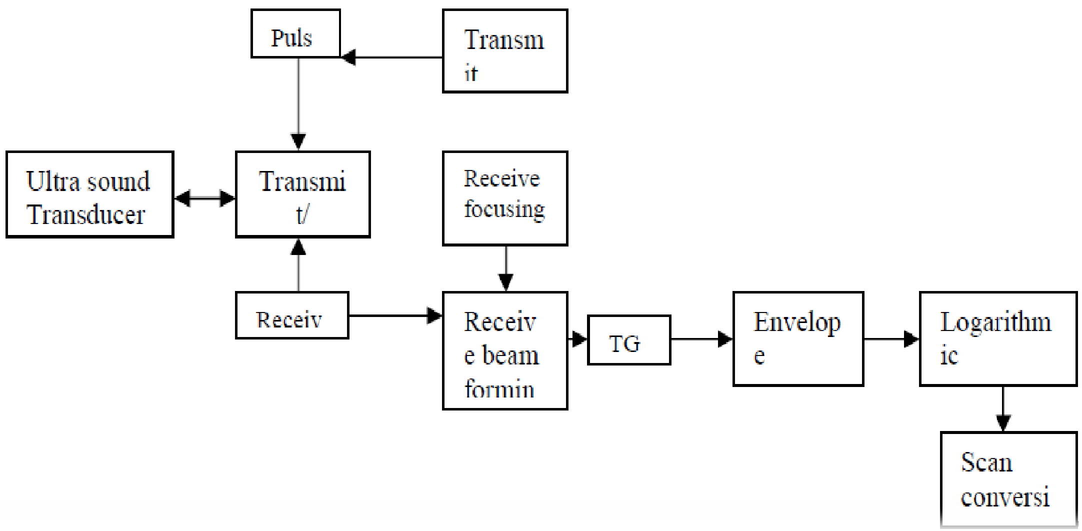


Figure 2.2 Block diagram of Ultrasound Imaging System. [10]

Chapter three

Literature review

3.1 Literature review

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.

S.Kalaivani Narayanan, R.S.D.Wahidabanu,"aview on despeckling in ultrasound imaging", September 2009,the objective of this paper is to give an overview about type of speckle reduction techniques in ultrasound imageing.

Jyoti Jay hay and Raj veer 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.

Rupinder Kaur*, Prabhpreet Kaur, Speckle Noise Reduction in Ultrasound Images (March 2014), various filters are used to reduce the speckle noise and to enhance the quality of the image. This work focuses on the study of various filters used for image enhancement.

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.

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.

Jaspreet kaur, Rajneet kaur,image denoising for speckle noise reduction in ultrasound images using DWT technique, (june 2013),this paper present image denoising and speckle noise reduction modle using different wavelets and combination of wiener filter along with deconvoluation filters.

Chapter four

Comparative study

4.1Introduction

In medical image processing, for Ultrasound images, the noise can restrain information which is valuable for the Medical practitioner. Ultrasonic devices are frequently used by healthcare professionals. The use of ultrasound imaging in medical diagnosis is well established because of its noninvasive nature, low cost, capability of forming real time imaging and continuing improvement in image quality. The main problem during diagnosis is the distortion of visual signals .These distortions are termed as 'Speckle Noise', this makes the image unclear. In the medical literature, speckle noise is referred as "texture". The success of ultrasonic examination depends on the image quality which is usually retarded due to speckle noise .Therefore, noise reduction is very important. In medical image processing, image denoising has become a very essential exercise all through the diagnose.

4.2 Speckle noise in ultrasound imaging

Speckle noise as a major factor limiting visual perception and processing of ultrasound [and synthetic aperture radar (SAR) images]. A mathematical speckle model for ultrasound images is introduced, where the statistics of speckle noise are presented, taking into consideration the log compression of the ultrasound image, which is performed to match the image into the display device. Based on this speckle model, a number of despeckling techniques .[9]

Noise and artifacts can cause signal and image degradation for many medical image modalities. Different image modalities exhibit distinct types of degradation. Images formed with coherent energy, such as ultrasound, suffer from speckle noise. Image degradation can have a significant impact on image quality and, thus, affect human interpretation and the accuracy of computer-assisted methods. Poor image quality often makes feature extraction, analysis, recognition, and quantitative measurements problematic and unreliable. Therefore, image despeckling is a very important task, which motivated a significant number of studies in medical imaging . [9]

The use of ultrasound in the diagnosis and the assessment of arterial disease is well established because of its noninvasive nature, its low cost, and the continuing improvements in image quality Speckle is a form of locally correlated multiplicative noise that corrupts medical ultrasound imaging making visual observation difficult. [9]

4.3 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. The most popular model adopted in the literature to explain the effects that occur when tissue is insulated is illustrated in Figure, where a tissue may be modeled as sound absorbing Medium containing stutterers, 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. 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. [9]

The nature of the speckle pattern can be categorized into one of three classes according to the number of scatterers per resolution cell or the so called scatterer 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 scattered region, the probability density function (PDF) of the backscattered signals becomes close to the Rican distribution .[9]

4.4 Need for despeckling

Image quality is important when evaluating ultrasound images for the assessment of the degree of diseases, or when transferring images through a telemedicine channel, and/or in other image processing tasks and it effect by speckle noise. Thus, speckle is considered as the dominant source of noise in ultrasound imaging and should be processed without affecting important image features. [6]

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

1. **1.** To improve the human interpretation of ultrasound image – speckle reduction makes an ultrasound image cleaner with clearer boundaries.
2. **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 and registration. [6]

4.5 Speckle reduction methods

Several techniques have been proposed for despeckling in medical ultrasound imaging. In This section we present the classification and theoretical overview of existing despeckling Techniques. [6]

4.5.1 Compounding Methods

Number of papers have been proposed based on compounding technique .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 delectability but they suffer from degrade spatial resolution and increased system complexity. [6]

4.5.2 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) Multiscale Methods .[6]

4.5.2.1 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.
1. 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.
2. 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.

3. 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[6].

4.5.2.2 Multi scale Methods

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

1. Wavelet based speckle reduction methods

The wavelet based speckle reduction method usually includes logarithmic transformation, wavelet transformation, modification of noisy coefficient using shrinkage function, invert wavelet transform and exponential transformation. This method can be classified into three groups:

- **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. [6]
- **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. [6]
- **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. [7]

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. [7]

4.6 Speckle models

Although the existing despeckling filters are termed as edge and feature preserving filters some major limitation exists:

1. The filters are sensitive to the noise components.

2. Noise attenuation is not sufficient especially in the smooth and background areas.

3. The existing filters do not enhance edges but they only inhibit smoothing near edges Thus, effective despeckling requires an accurate statistical model of ultrasound signals. A generalized model of the speckle imaging can be written as

$$g = fm + n \qquad (4.1)$$

Let g denote the observed signal, m, n the multiplicative and additive components of noise respectively introduced by the acquisition process and f the original signal without noise. Generally the effect of additive noise is very small compared to multiplicative noise, so the simplified noise model:

$$g \approx fm \qquad (4.2)$$

Thus the logarithmic compression transforms the model in (4.2) into the classical signal in additive noise form as:

$$\log g = \log f + \log m \qquad (4.3)$$

The statistics of speckle noise can be categorized into different classes according to number of stutterers per resolution cell called scattered number density (SND). In the case of many fine randomly distributed stutterers per resolution cell (>10) the speckle can be modeled by a Rayleigh distribution with a constant SNR of 1.92.When the scattered densities are smaller a generalized version of Rayleigh distribution called the K-distribution can be used. For high SNR the Rician model can be used, and also for lower SNR the speckle can be modeled using Homodyne K-distribution. [7]

4.7 speckling filters

In order to be able to derive an efficient despeckle filter, a speckle noise model is needed. The speckle noise model may be approximated as multiplicative, if the envelope signal received at the output of the beam former of the ultrasound imaging system is captured before logarithmic compression. Logarithmic compression is applied to the envelope-detected echo signal in order to fit it in the display range.[6]

Speckle filtering consists of moving a kernel over each pixel in the image and applying a mathematical calculation using the pixel values under the kernel and replacing the central pixel with the calculated value. The kernel is moved along the image one pixel at a time until the entire image has been covered. By applying the filter a smoothing effect is achieved and the visual appearance of the speckle is reduced. [6]
An appropriate method for speckle reduction is one which enhances the signal-to noise ratio while conserving the edges and lines in the image. Filtering techniques are used as preface action before segmentation and classification. In literature many techniques have been studied for speckle noise reduction. [6]

4.7.1 Local Statistics Filtering

Most of the techniques for speckle reduction filtering in the literature use local statistics. Their working principle may be described by a weighted average calculation using sub region statistics to estimate statistical measures over different pixel windows varying from 3×3 up to 15×15. All these techniques assume that the speckle noise model has a multiplicative form, it is two kinds:

1. First Order Statistics Filtering *(lsmv) Mean and variance local statistics despeckle filter* this filter using the first order statistics such as the variance and the mean of the neighborhood. [6]

2. Homogeneous Mask Area Filtering: The *(lsminsc) Minimum speckle index homogeneous mask despeckle filters* a 2-D filter operating in a 5×5 pixel neighborhood by searching for the most homogenous neighborhood area around each pixel, using a 3 × 3 subset window. The middle pixel of the 5×5 neighborhood is substituted with the average gray level of the 3×3 mask with the smallest speckle index, C, where C, represents the variance over mean of the 3×3 window. All these techniques assume that the speckle noise model has a multiplicative form. [6]

4.7.2 Median Filtering

The filter median is a simple nonlinear operator that replaces the middle pixel in the window with the median-value of its neighbors. The moving window for the median filter was 7 × 7.it is a particularly effective to removes pulse or spike noises. The main problem of the median filter is its high computational cost for sorting N pixels.

4.7.3Hybrid 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 fig4.1The median values of the 45 neighbors forming an “X” and the 90 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. [6]

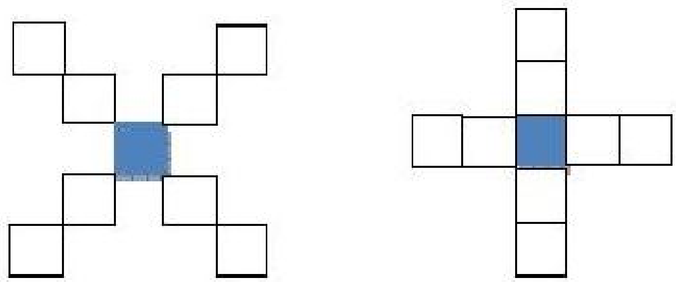


Figure 4.1: Diagram of neighborhood pixels used in the Hybrid Median Filter [6].

4.7.4 Linear Scaling Filter (DsFca, DsFls)

- The DsFca filter despeckles 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 *gi*, *j* (the middle pixel in the moving window) is lower than or equal to a given threshold J. Assign this value to the gray level *gi*, *j* with □ = α $*g_{max}$, where g_{max} is the maximum gray level of the image and α = [0, 1] . [9]
- 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.[9]

- The *DsFlecasort* filter takes *k* points of a pixel neighborhood, which are closest to the gray level of the image at point *gi*, *j* (the middle point in the moving window), including *gi*, *j*. It then assigns the mean value of these points to the pixel *gi*, *j* (usually, *N* = 9 in a 3 ´ 3 window, where *k* = 6) . [9]
- The *DsFls* filter scales the pixel intensities by finding the maximum g_{max} and the minimum g_{min} gray-level values in every moving window, and then replaces the middle pixel with. [9]

$$f_{i,j} = \frac{g_{max} + g_{min}}{2} \qquad (4.4)$$

4.7.5 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 *(gf4d)* geometric despeckle filter investigated in this study uses a nonlinear noise reduction technique.[6]

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 surrounding. [6]

4.7.6 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. [13]

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. 28
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. [14]

E [||X- □||] ˆ2/N is the same order of O (logˆ2 N/N). (4.5)

4.7. 7 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 + (Δt/4) div (g (ICOV (u')) xΔ u' (4.6)

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 diffusion Δ function and is given by equations: [15]

G (ICOV (u')) =e-(P) (4.7)

$$P=\frac{\frac{[\mathrm{ICOV}(u')]^2}{qt}}{1+(qt)^2} \qquad (4.8)$$

4.7.8 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.

$$U=f-PGA\ (f) \qquad (4.9)$$

Where is the noisy image, U is the image we want to restore from f-PGA(f) is the orthogonal projection of f on GA and the space G is proposed by Meyer for modeling oscillating patterns. [16]

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. [16]

4.8 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.

The root mean square error (RMSE), which is the square root of the squared error averaged over an MxN window:[17]

$$\text{RMSE}=\sqrt{\frac{1}{MN}\sum_{I=1}^{M}\sum_{J=1}^{N}(g_{i,j}\_f_{i,j})^2} \qquad (4.10)$$

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

$$\text{SNR}=10\log_{10}\frac{\sum_{i=1}^{M}\sum_{j=1}^{N}(g_{i,j}^2+f_{i,j}^2)}{\sum_{i=1}^{M}\sum_{j=1}^{N}(g_{i,j-}f_{i,j})^2} \qquad (4.11)$$

The peak SNR (PSNR) is computed using:

$$\text{PSNR}=10\log_{10}\frac{MSE}{g_{max}^2} \qquad (4.12)$$

Whereg_{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:[17]

$$\text{SSIM}=\frac{(2f+c1)(2\sigma_{gf}+c2)}{(^2+f^2+c1)(\sigma_g^2+\sigma_f^2+c2)} \qquad (4.13)$$

$$-1< SSIM < 1$$

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. [17]

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

Chapter five
Material and methodology

5.1 proposed method 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. [16]

5.2.1 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).[11]

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. [11]

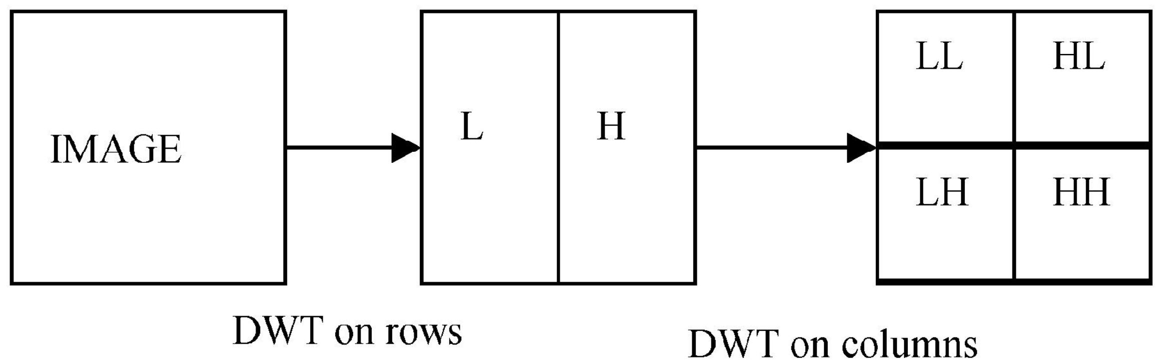
Wavelet filtering exploits the decomposition of the image into the wavelet basis and zeros out the wavelet coefficients to despeckle the image. Wavelets are simply mathematical functions and these functions analyze data according to scale or resolution. [11]

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. [11]

5.2.2 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 [11].



Figuer5.1: DWTdecompositoin of image [12]

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. [11]

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.

5.3The proposed method algorithm

Step1: input the original image.
Step2: adding speckle noise.
Step3: implement the wavelet decomposition(DWT).
Step4: as result of decomposition we get (approximation level (LL), horizontal level (hl), vertical level (LH), diagonal level (HH).
Step5: apply the TV filter to the LL image, and Wavelet filter to the LH, HL, and HH.

Step6: implement the wavelet reconstruction (IDWT).
Step7: we obtain the final image, and calculated the quality matrixes.

5.4 The proposed system (Methodology)

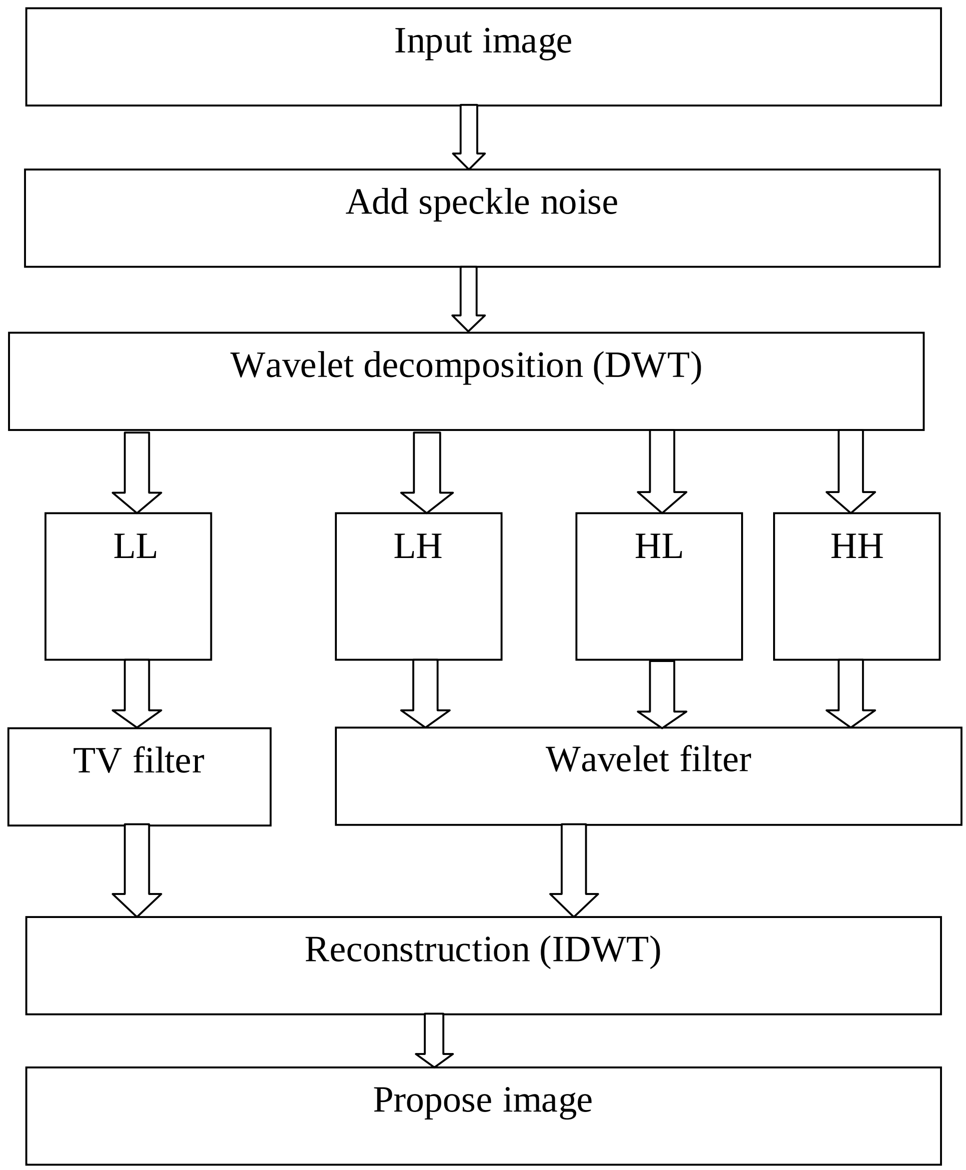


Figure 5.2.: Block diagram of the proposed method (DWT)

Chapter six

Results and discussion

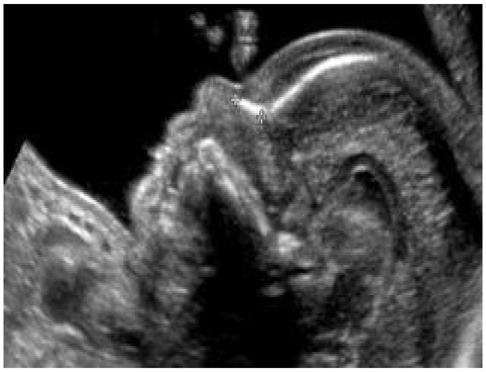
6.1 Experimental results

The DWT technique 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 σ_n= 0.05 and 0.5 using the MATLAB command "imnoise (image, „speckle 0.05 or 0.5)”.

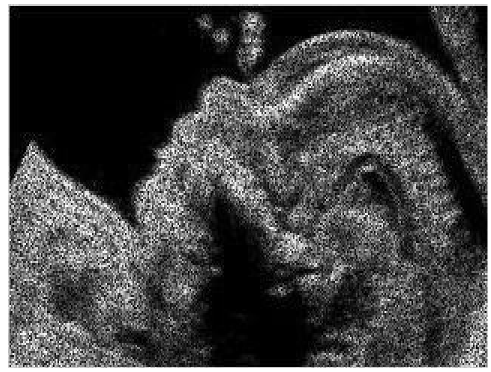
To estimate the performance of the DWT technique. eight standard filters namely: linear scaling gray level filter(DsFca),geometric despeckle filter(DsFg4d), linear scaling(DsFls), speckle reducing anisotropic diffusion(srad), median filter(Med),hybrid median filter(HMF), wavelet filter and total variation 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. 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 are higher for a better-transformed image and lower for a poorly transformed image, on the contrary 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.

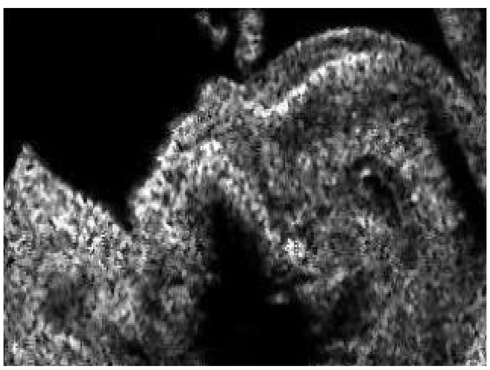
6.2 Proposed method (Discrete wavelet transform)



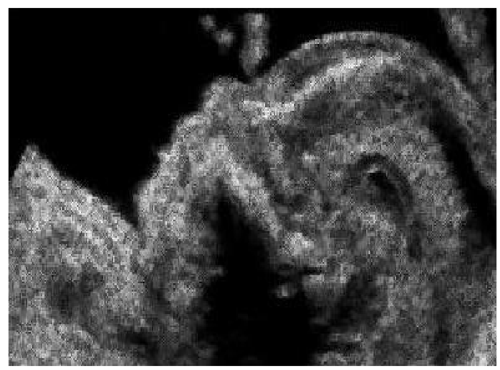
(a) Original image



(b) noisy image



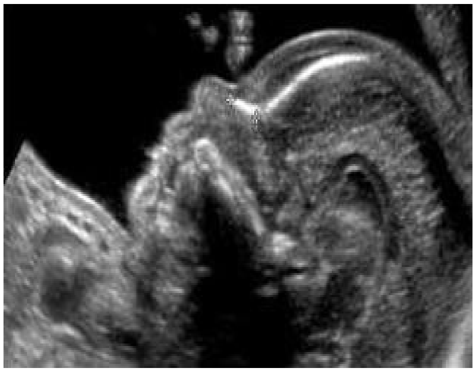
(C) median filter



(d)hybrid median filter

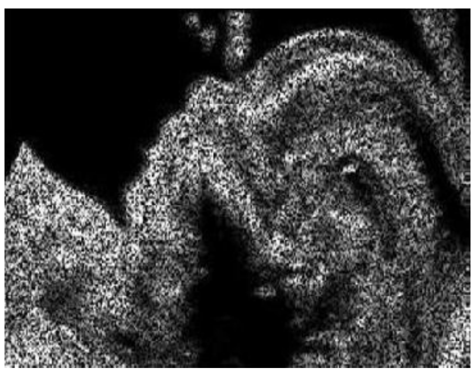
(e) Linear scaling gray level filter (DsFca)

(f) Linear scaling filter (DsFls)

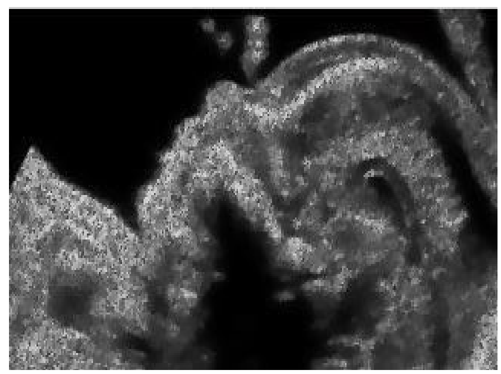


(g) anisotropic diffusion filter (SRAD)

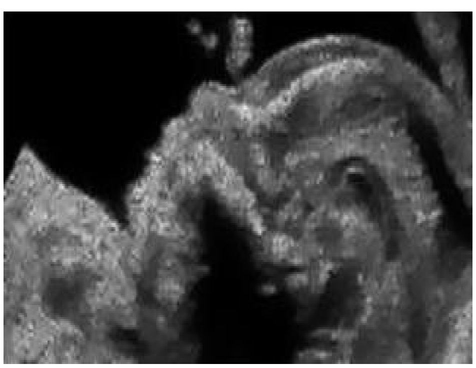
(h) Geometric filter (DsFgf4d)



(i) Wavelet filter



(j) Total variation filter



(k) proposed image

Figure 6.1: Results of fetal despeckled by various filter on multiplication noise (σ_n=0.5)

Table6.1: Image quality evaluation metrics computed for the **fetal (σ_n =05)** at statistical measurement of PSNR, SNR ,SSIM and RMSE for different filter types and for DWT.

Filter type	Image quality evaluation metrics			
	SNR	**PSNR**	**SSIM**	**RMSE**
Median filter	20.6081	45.0864	0.4096	17.0715
Hybrid median filter	20.5071	44.7929	**0.6562**	**20.7139**
Linear scaling gray level filter (DsFca)	20.6325	44.2420	0.5416	25.4192
Linear scaling filter (DsFls)	20.6139	43.2688	0.4828	26.4144
anisotropic diffusion filter (SRAD)	**20.7420**	45.1205	0.4460	38.4544
Geometric filter (DsFgf4d)	**20.7903**	**45.1205**	0.4853	51.8178
Wavelet filter	20.6931	**45.1206**	0.4721	51.0748
Total variation filter(TV)	20.5571	44.3921	**0.6263**	16.2310
Proposed	**23.2697**	**45.1003**	**0.7291**	**14.7212**

Bold number indicates the best values.
* Signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR), structural similarity index (SSIM), Route mean square error (RMSE).

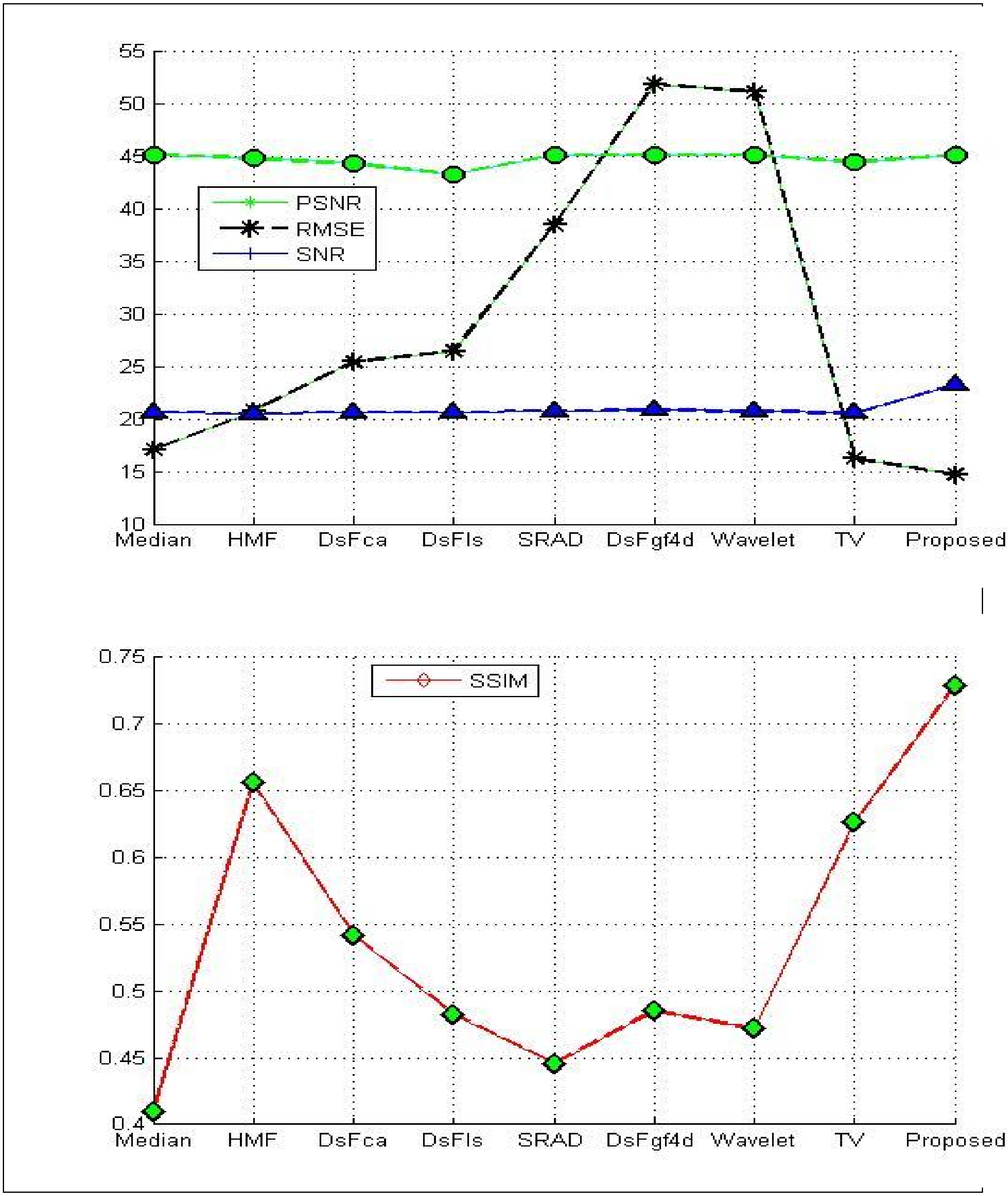


Figure 6.2: Performance analysis graph to image quality evaluation metric for fetal image (noise σ_n= =0.5).

(a) Original image

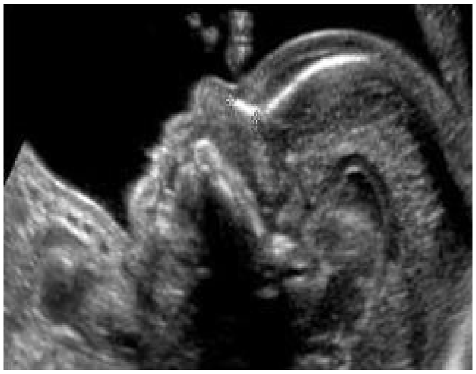
(b) noisy image

(C) median filter

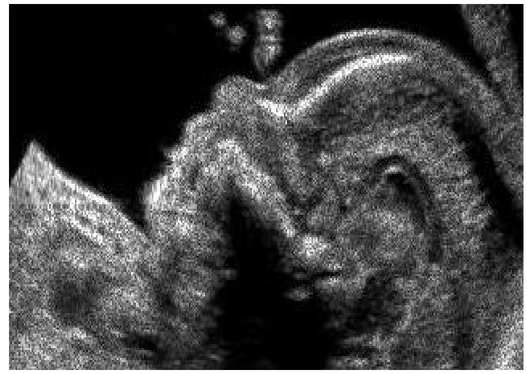
(d)hybrid median filter

(e) Linear scaling gray level filter (DsFca)

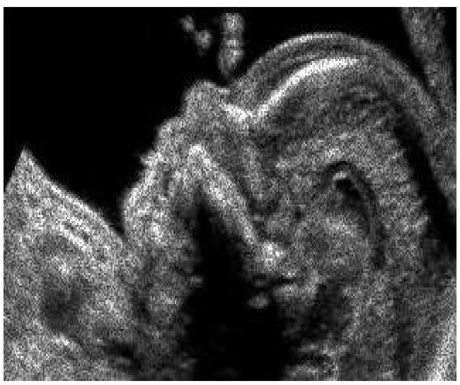
(f) Linear scaling filter (DsFls)



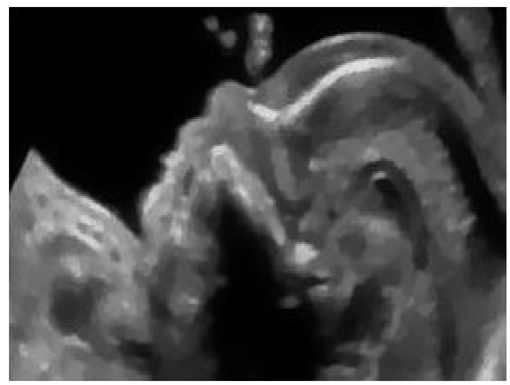
(g) anisotropic diffusion filter (SRAD)



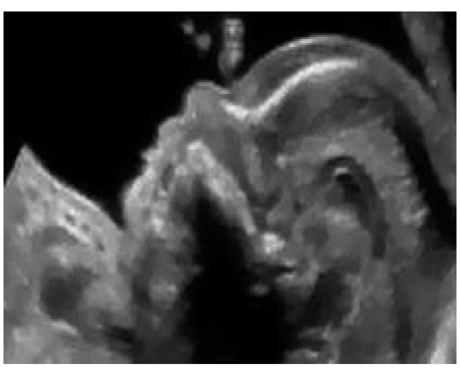
(h) Geometric filter (DsFgf4d)



(i)Wavelet filter



(j) Total variation filter



(k) proposed image

Figure 6.3: Results of fetal despeckled by various filter on multiplication noise (σ_n=0.05).

Table6.2: Image quality evaluation metrics computed for the **fetal (σ_n =0.05)** at
statistical measurement of PSNR, SNR ,SSIM and RMSE for different filter types and for DWT.

Filter type	**Image quality evaluation metrics**			
	SNR	**PSNR**	**SSIM**	**RMSE**
Median filter	20.5665	**45.1202**	**0.7915**	**10.5374**
Hybrid median filter	20.5603	45.0344	0.7381	**10.6843**
Linear scaling gray level filter (DsFca)	20.6551	**45.1203**	0.6166	17.1294
Linear scaling filter (DsFls)	20.6156	44.3001	0.5398	26.0022
anisotropic diffusion filter (SRAD)	20.6249	45.0864	0.6153	21.1283
Geometric filter (DsFgf4d)	**20.6543**	43.9144	0.7875	19.9313
Wavelet filter	20.6166	45.0864	0.8219	18.4718
Total variation filter(TV)	**21.5404**	44.8482	0.7692	10.4686
Proposed	**23.2426**	**45.1205**	**0.7849**	**9.9609**

Bold number indicates the best values.
* Signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR), structural similarity index (SSIM), Route mean square error (RMSE).

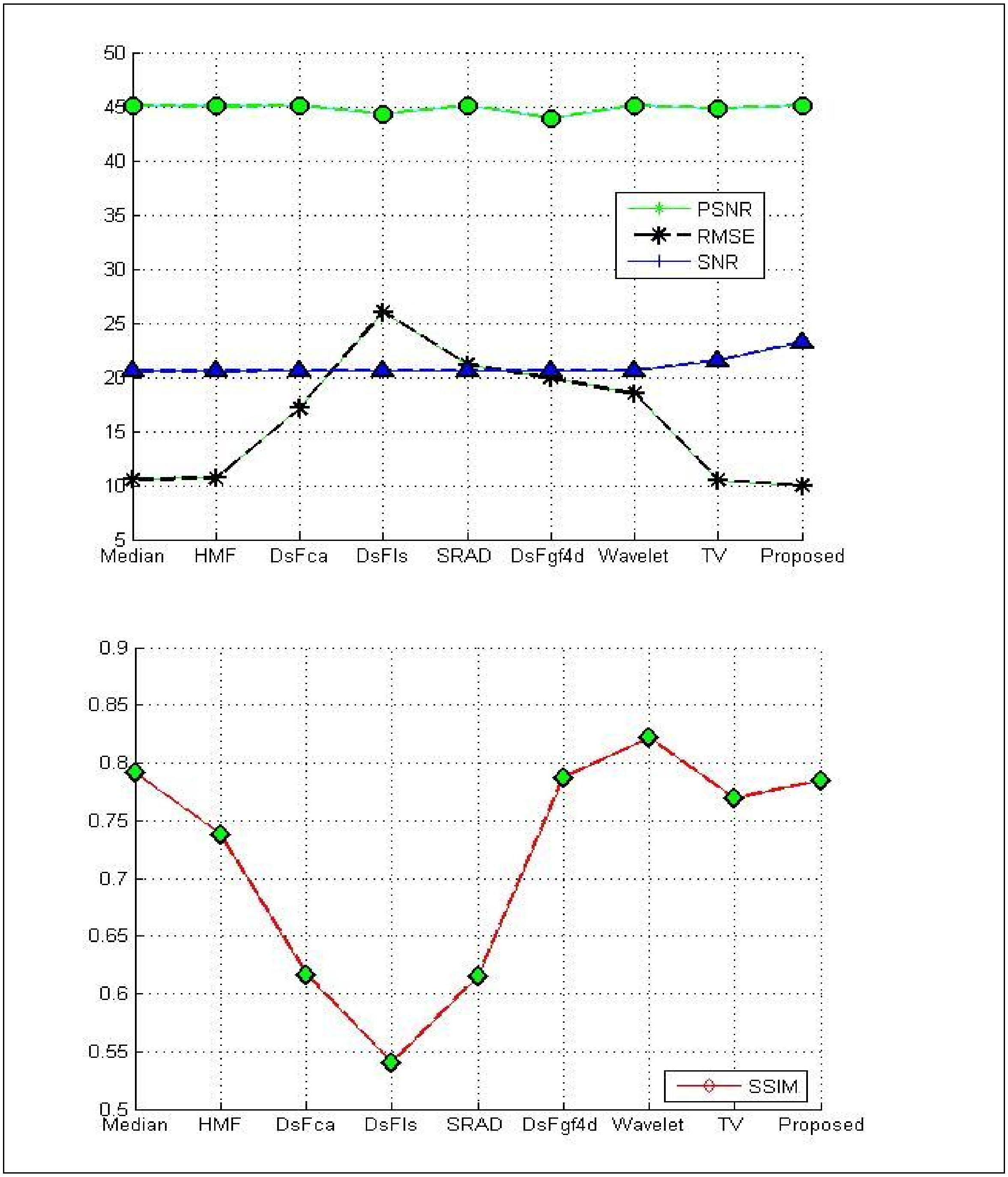
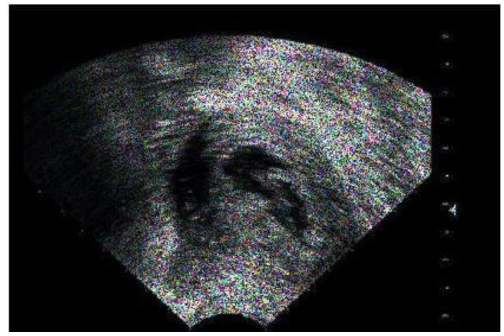
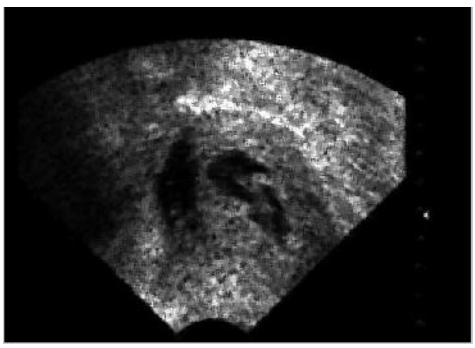


Figure 6.4: Performance analysis graph to image quality evaluation metric for fetal image (noiseσ_n =0.05).

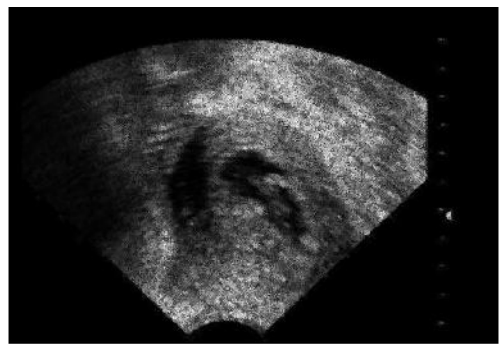
(a) Original image



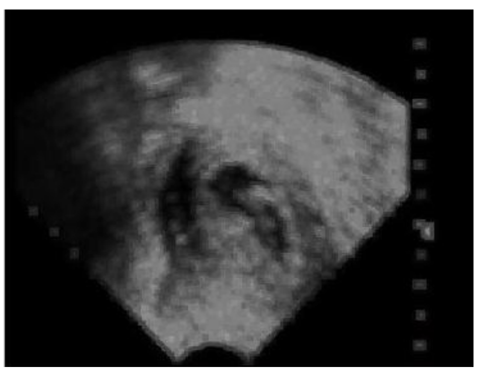
(b) noisy image



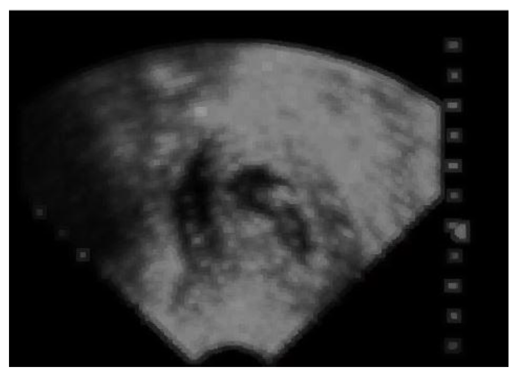
(C) median filter



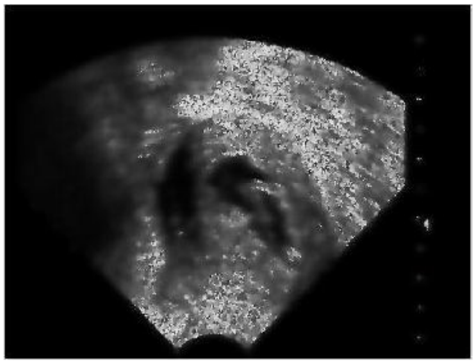
(d)hybrid median filter



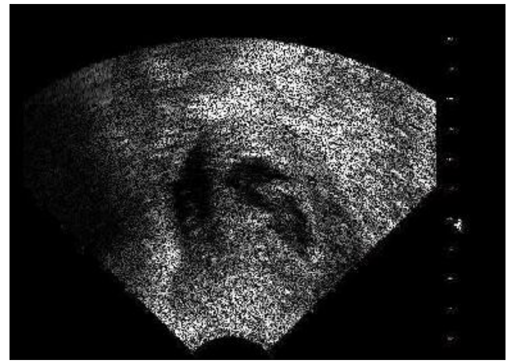
(e) Linear scaling gray level filter (DsFca)



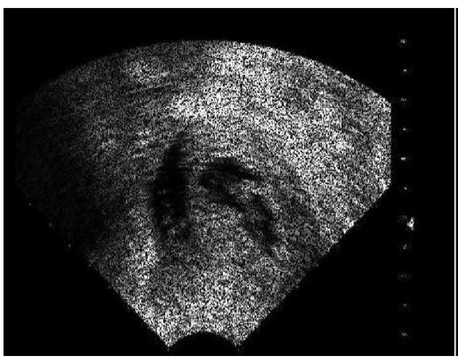
(f) Linear scaling filter (DsFls)



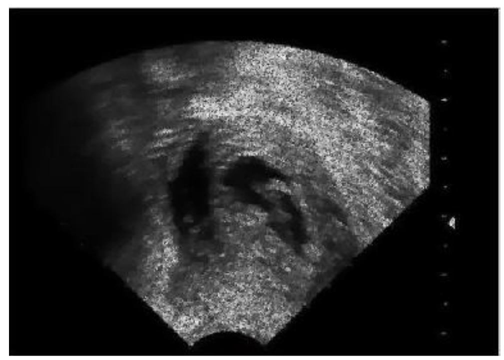
(g) anisotropic diffusion filter (SRAD)



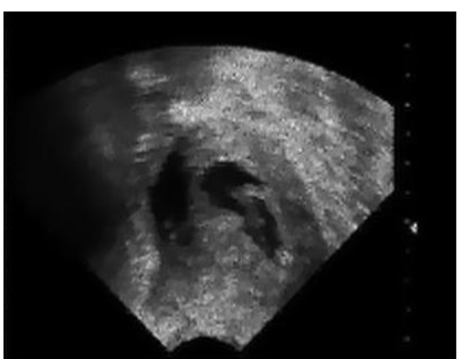
(h) Geometric filter (DsFgf4d)



(i) Wavelet filter



(j) Total variation filter



(k)proposed image

Figure 6. 5: Results of vagina despeckled by various filter on multiplication noise (σ_n=0.5).

Table6.3: Image quality evaluation metrics computed for the **vegan (σ_n =0.5)** at statistical measurement of PSNR, SNR ,SSIM and RMSE for different filter types and for DWT.

Filter type	**Image quality evaluation metrics**			
	SNR	**PSNR**	**SSIM**	**RMSE**
Median filter	19.7837	45.1035	**0.6938**	**27.7439**
Hybrid median filter	19.6831	44.8110	**0.7457**	**30.1765**
Linear scaling gray level filter (DsFca)	19.7474	43.2458	0.6630	29.7006
Linear scaling filter (DsFls)	19.7273	43.3246	0.6327	23.0833
anisotropic diffusion filter (SRAD)	19.9009	44.8454	0.7370	26.1770
Geometric filter (DsFgf4d)	**19.9703**	**45.1205**	0.5722	49.4903
Wavelet filter	**19.9619**	**45.1205**	0.5771	37.8085
Total variation filter(TV)	19.6362	43.7631	**0.6941**	**20.3000**
Proposed	**22.8013**	**45.1003**	**0.6863**	**15.9925**

Bold number indicates the best values.

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

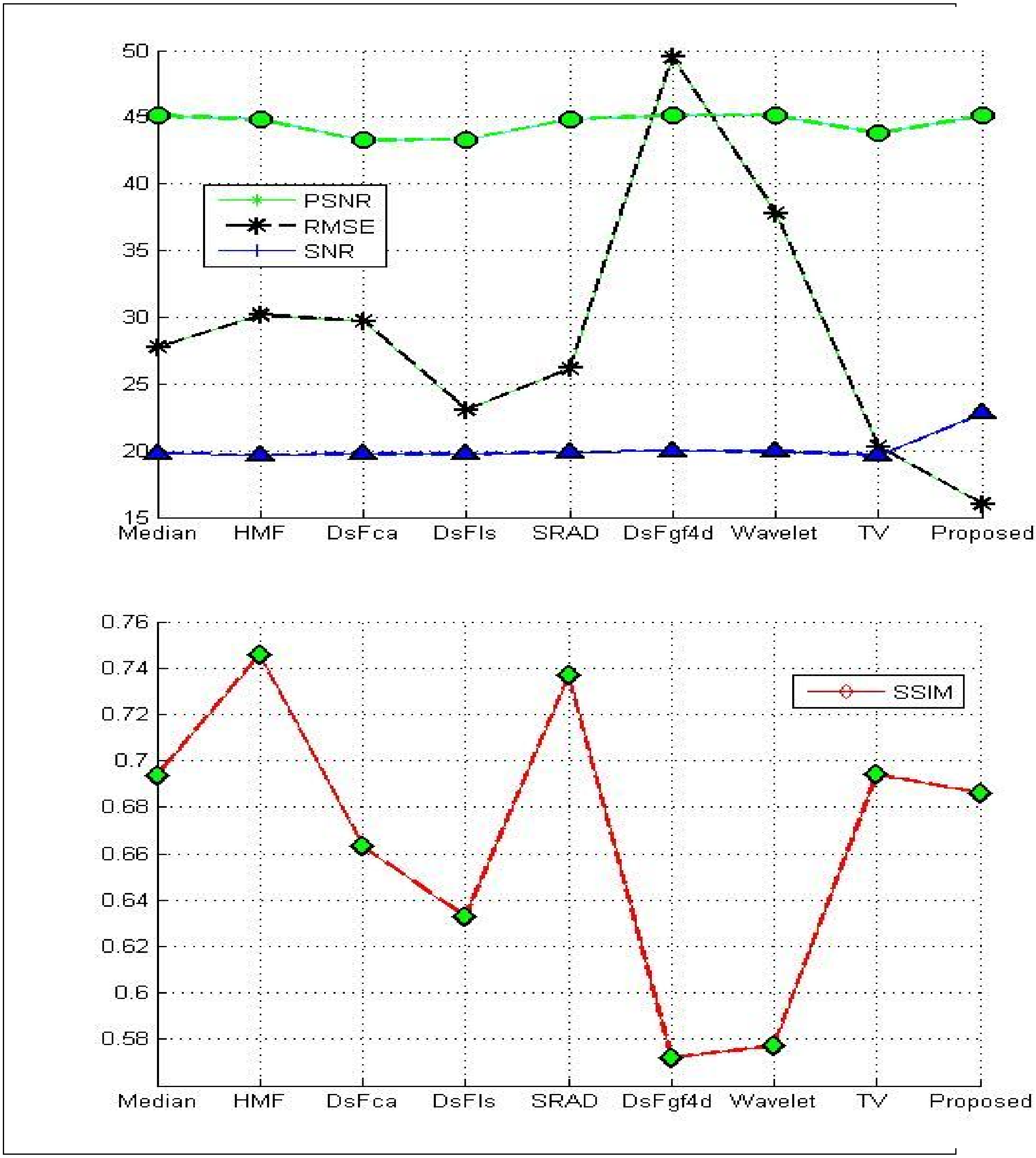
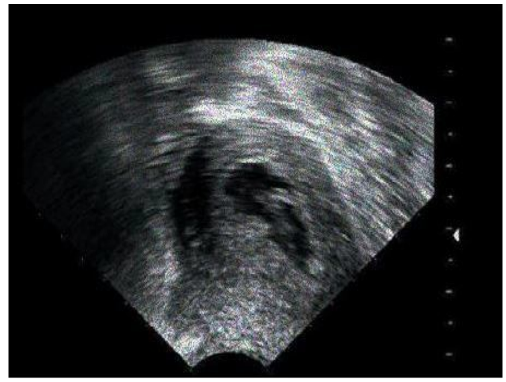
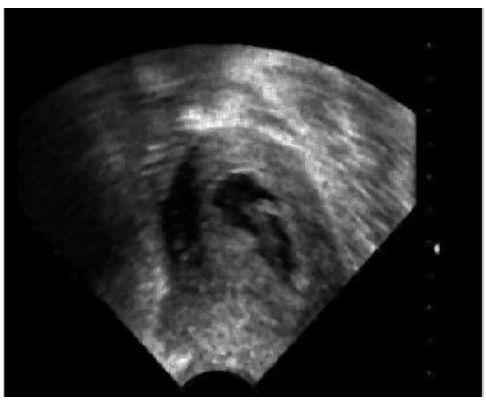


Figure 6.6: Performance analysis graph to image quality evaluation metric for vegan image (noiseσ_n =0.5).

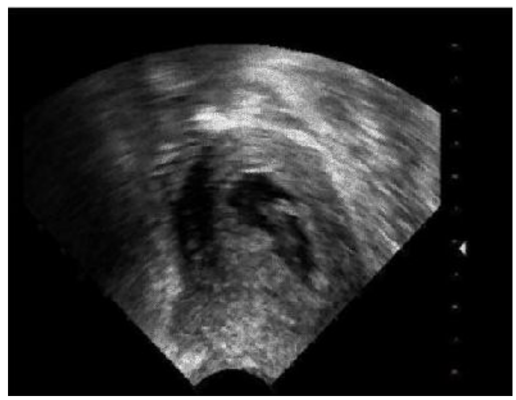
(a) Original image



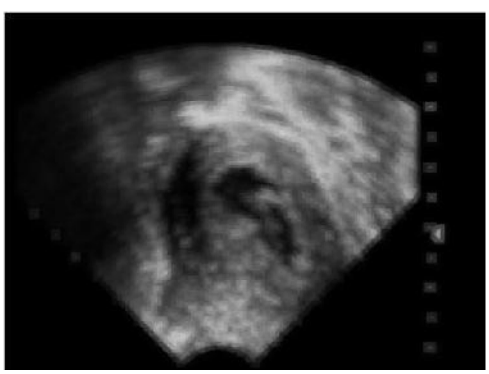
(b) noisy image



(C) median filter

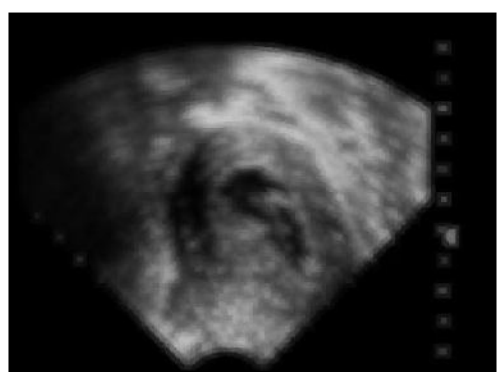


(d) hybrid median filter

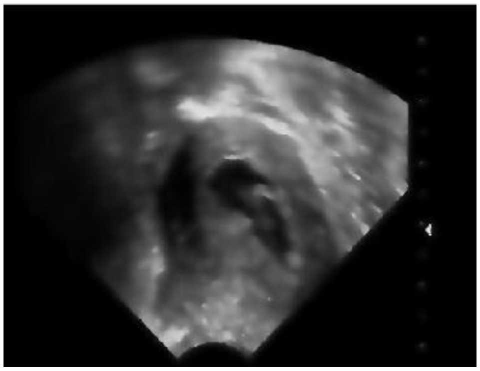


(e) Linear scaling gray level

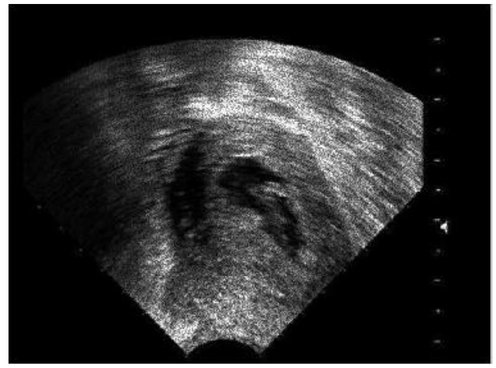
filter (DsFca)



(f) Linear scaling filter (DsFls)



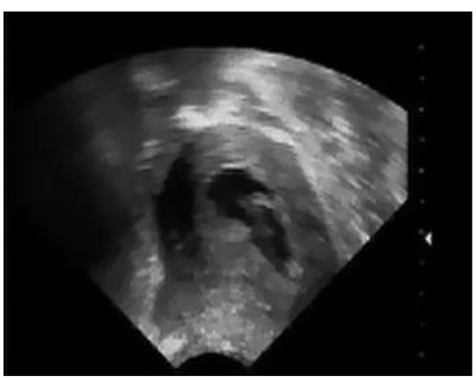
(g) anisotropic diffusion filter (SRAD)



(h) Geometric filter (DsFgf4d)

(i) Wavelet filter

(j) total variation



(k) proposed image

Figure 6.7: Results of vagina despeckled by various filter on multiplication noise (σ_n=0.05).

Table6.4: Image quality evaluation metrics computed for the **vagina (σ_n =0.05)** at statistical measurement of PSNR, SNR ,SSIM and RMSE for different filter types and for DWT.

Filter type	**Image quality evaluation metrics**			
	SNR	**PSNR**	**SSIM**	**RMSE**
Median filter	19.7186	44.7186	**0.8248**	**26.0706**
Hybrid median filter	19.7508	44.5044	**0.8711**	**26.2550**
Linear scaling gray level filter (DsFca)	19.8433	44.0225	0.7176	25.9500
Linear scaling filter (DsFls)	19.8044	44.0225	0.6631	15.6629
anisotropic diffusion filter (SRAD)	19.7652	44.1712	0.7290	16.0393
Geometric filter (DsFgf4d)	19.8585	44.9663	**0.8021**	**31.3622**
Wavelet filter	19.7948	45.0013	0.7552	16.2647
Total variation filter(TV)	19.8433	44.0225	0.7176	19.7348
Proposed	**22.7887**	**45.1205**	0.7156	12.4810

Bold number indicates the best values.

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

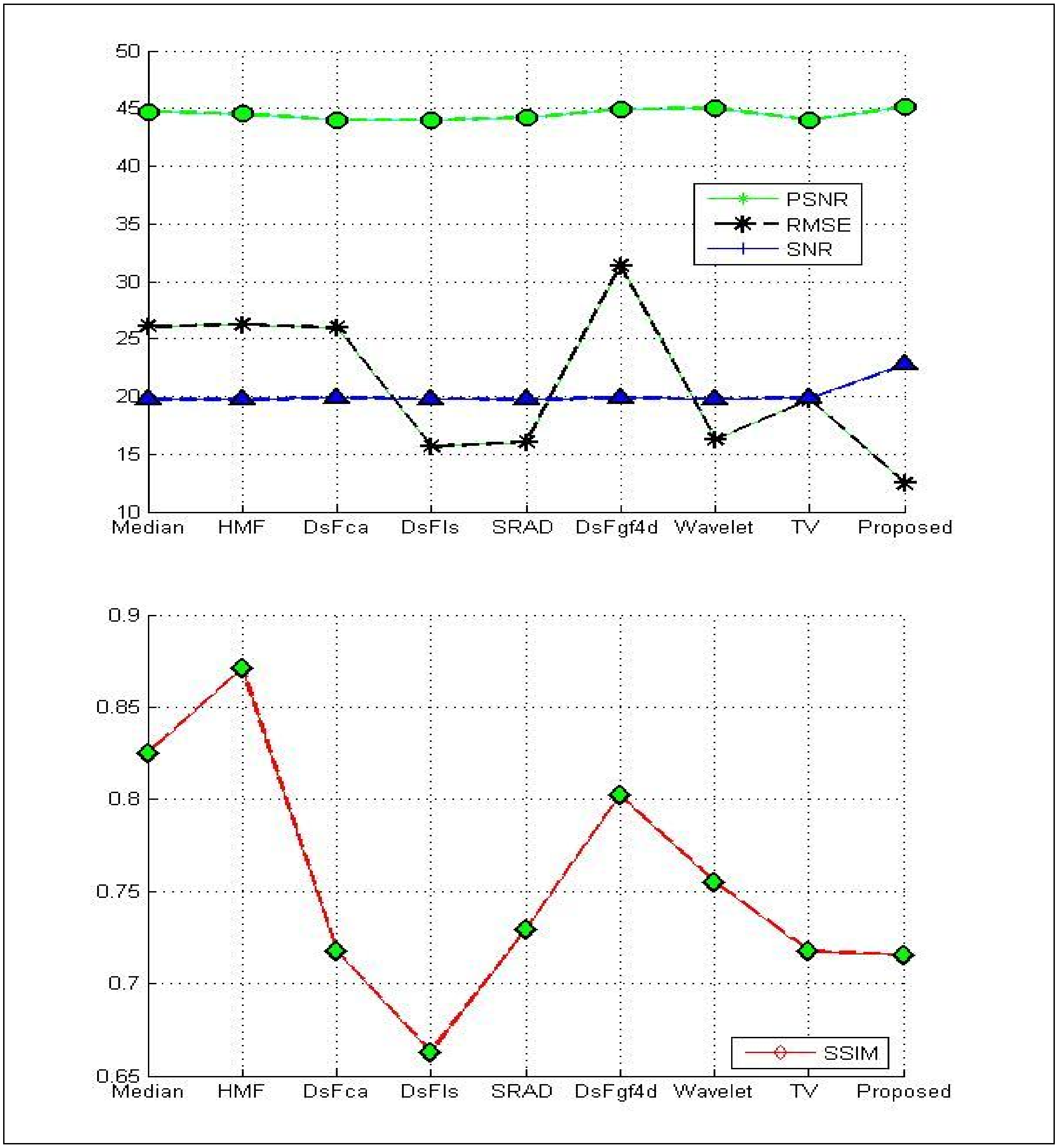


Figure 6.8: Performance analysis graph to image quality evaluation metric for vagina image (noise σ_n =0.05).

(a) Original image

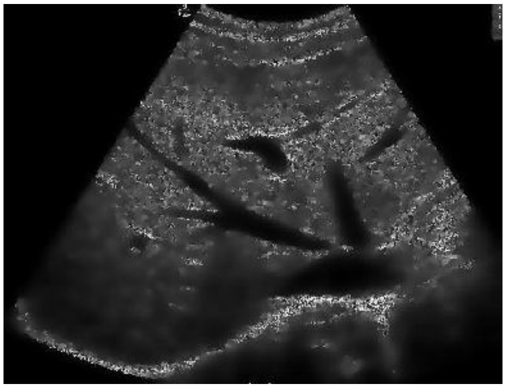
(b) noisy image

(C) median filter

(d)hybrid median filter

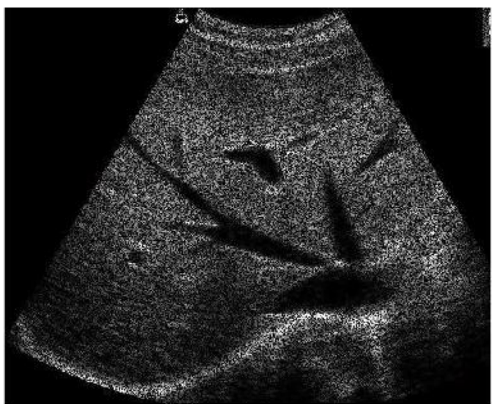
(e) Linear scaling gray level filter (DsFca)

(f) Linear scaling filter (DsFls)



(g) anisotropic diffusion filter (SRAD)

(h) Geometric filter (DsFgf4d)



(i) Wavelet filter

(j) total variation

(k)proposed image

Figure 6.9 : Results of liver despeckled by various filter on multiplication noise (σ_n=0.5).

Table6.5: Image quality evaluation metrics computed for the **liver (σ_n =0.5)** at statistical measurement of PSNR, SNR ,SSIM and RMSE for different filter types and for DWT.

Filter type	**Image quality evaluation metrics**			
	SNR	**PSNR**	**SSIM**	**RMSE**
Median filter	18.4127	44.3486	0.6412	17.0583
Hybrid median filter	18.4653	43.1833	**0.6697**	**18.9768**
Linear scaling gray level filter (DsFca)	**18.8367**	44.8471	0.6500	20.0375
Linear scaling filter (DsFls)	18.8035	42.1681	0.6356	18.6283
anisotropic diffusion filter (SRAD)	18.5798	**44.8652**	0.6392	16.3678
Geometric filter (DsFgf4d)	18.8436	**44.9843**	0.4590	37.8562
Wavelet filter	**18.8328**	44.5987	0.4934	36.8196
Total variation filter(TV)	18.3942	43.9810	**0.6858**	**16.6862**
Proposed	**22.3304**	**45.0694**	**0.7534**	**8.4782**

Bold number indicates the best values.

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

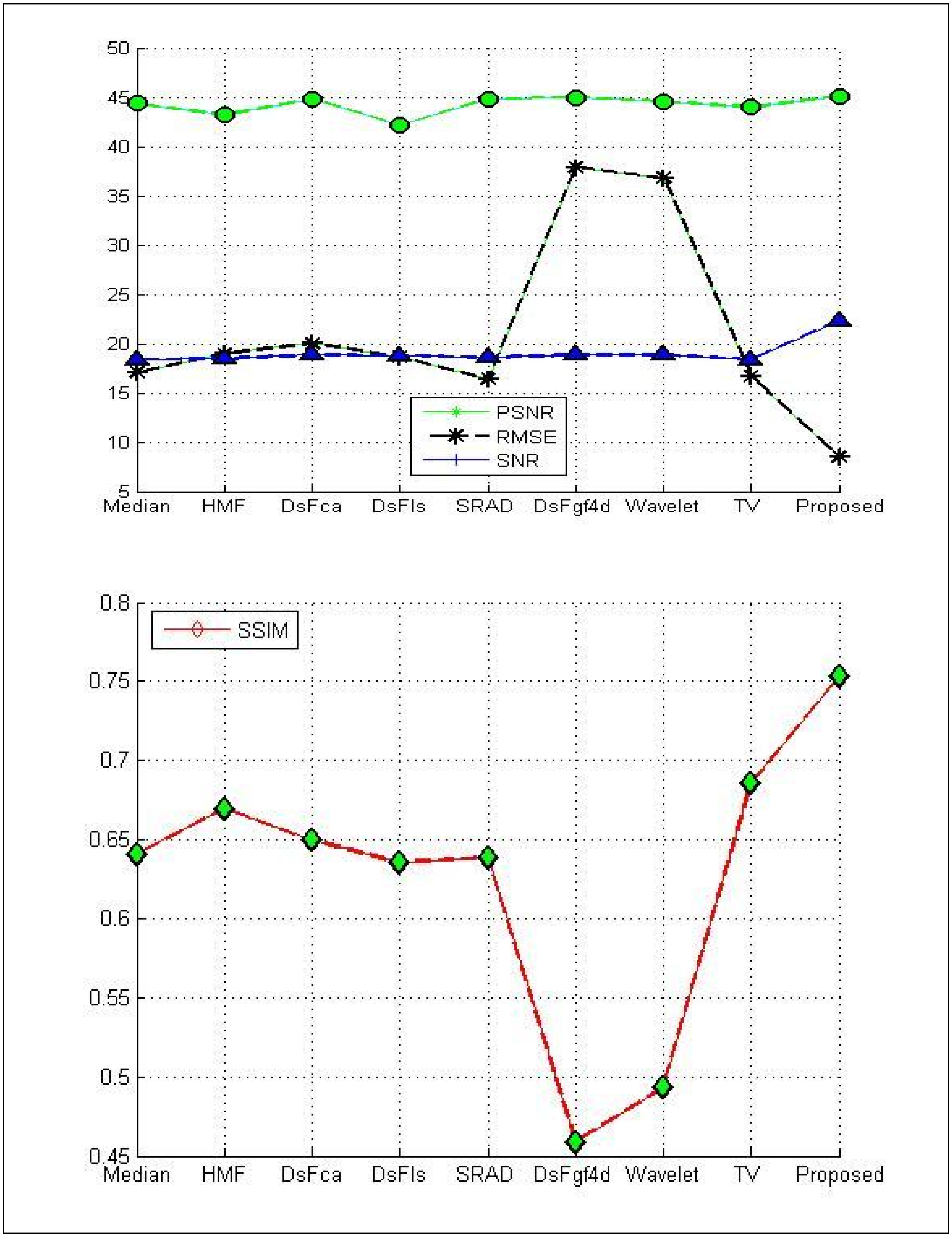


Figure 6.10: Performance analysis graph to image quality evaluation metric for liver image (noise σ_n =0.5).

(a) Original image

(b) noisy image

(C) median filter

(d)hybrid median filter

(e) Linear scaling gray level filter (DsFca)

(f) Linear scaling filter (DsFls)

(g) anisotropic diffusion filter (SRAD)

(h) Geometric filter (DsFgf4d)

(i) Wavelet filter

(j) total variation

(k)proposed image

Figure 6.11 : Results of liver despeckled by various filter on multiplication noise (σ_n=0.05).

Table6.5: Image quality evaluation metrics computed for the **liver (σ_n =0.05)** at statistical measurement of PSNR, SNR ,SSIM and RMSE for different filter types and for DWT.

Filter type	**Image quality evaluation metrics**			
	SNR	**PSNR**	**SSIM**	**RMSE**
Median filter	18.3650	43.2634	**0.7840**	**15.7980**
Hybrid median filter	18.3718	43.5002	**0.7940**	**15.7456**
Linear scaling gray level filter (DsFca)	18.5023	44.0741	0.6992	17.0529
Linear scaling filter (DsFls)	18.5289	42.2283	0.6493	15.4541
anisotropic diffusion filter (SRAD)	18.3719	44.3935	0.6703	10.0797
Geometric filter (DsFgf4d)	**18.5293**	**45.0013**	0.7334	20.0683
Wavelet filter	**18.5480**	**44.6616**	0.7510	19.0555
Total variation filter(TV)	18.5006	44.1924	**0.8327**	**16.1117**
Proposed	**22.3274**	**45.0694**	0.7504	7.9254

Bold number indicates the best values.
* Signal-to-noise ratio (SNR), peak-to-noise ratio (PSNR), structural similarity index (SSIM), Route mean square error (RMSE).

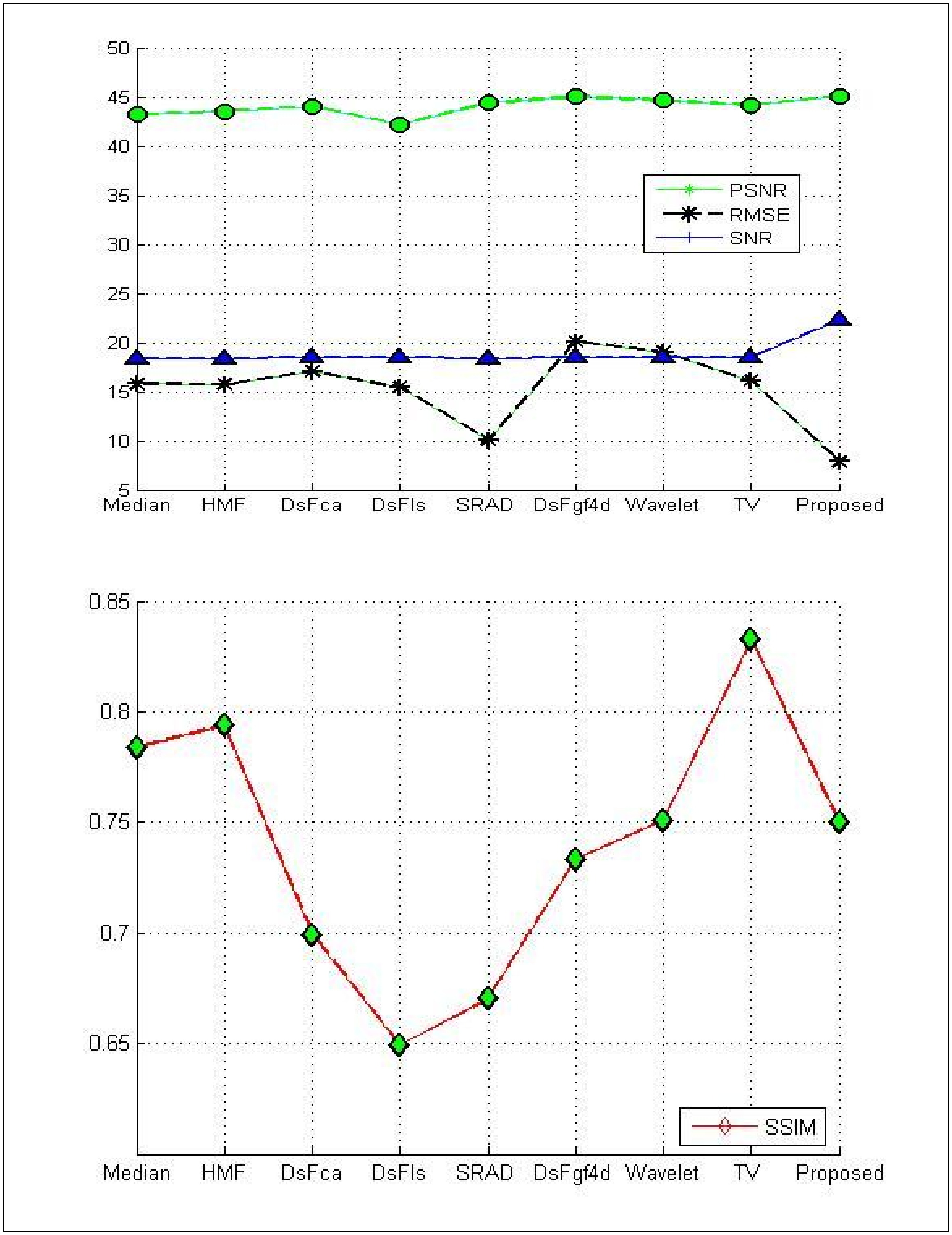


Figure 6.12: Performance analysis graph to image quality evaluation metric for liver image (noise σ_n =0.05).

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 6.1, 6.3 6.5, 6.7, 6.9, 6.11 show an ultrasound image (a) with noisy (b) and the despeckled images. In(e) 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, (h) 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. (f) 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. (g) 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, (c) 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(d)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 69

Despeckle filter (TV),(k)proposed method. We see that most of the unwanted details haven't been removed efficiently, whilst preserving important details such as edges. From table 6.1, 6.2, 6.3, 6.4, 6.5, 6.6 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 Best values for the SSlM, but visually, smoothed the image. Loosing subtle details are been observed. From table 6.1 6.3 6.7 6.9 show that the total variation 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 total variation filter, this gives better edge preserving characteristics than other 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 other filter .

Chapter seven

Conclusion and recommendation

7.1 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. [10]

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. [18]

The present study focuses on proposing a technique to reduce speckle noise from ultrasonic devices. he results achieved from the other speckle noise reduction techniques demonstrate its higher performance for speckle reduction. [18]

The proposed method to reduce noise is discrete wavelet transform (DWT), The DWT decomposes input image into four component namely LL, HL, LH and HH where the first letter corresponds to applying either a low pass frequency operation or high pass frequency operation to the row, and the second letter refers to the filter applied to the columns, the lowest resolution level LL consist of the approximation part of the original image, filtered by use total variation filter. The remaining three resolution levels consist of the detail parts and give the vertical high (LH), horizontal high (HL), and high (HH) frequencies, the three parts filtered by wavelet filter.

7.2 Recommendation

1- Wavelet filter in the proposed method, can be changed by hybrid median filter in this thesis.

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

3-from the result that calculated by quality evaluation matrices ,we recommended to use the highest score of SSIM apply on real cardiac ultrasound images.

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