A healthy heart

abstract

A healthy heart is the key to good life . The heart is a vital organ of the human body which ensures the effective pumping of blood throughout the circulatory system. Due to our sedentary lives and food habits, the heart is prone to malfunctioning, and heart attack (i. e. coronary artery disease), is one of the primary cause of death [1]. Heart attack is caused by a blockage of the coronary arteries, typically at a site of narrowing (stenosis) caused by atherosclerosis. It is difficult to accurately determine the degree of atherosclerosis in arteries, particularly in the early stages of disease. One method that has been introduced is the intravascular ultrasonic catheter (IVUS), which sends a pulse of sound from a receiver and uses the returned echo to deduce the properties of the arterial tissues.

Doppler Ultrasound is a similar a diagnostic, noninvasive technique which can effectively evaluate the blood flow velocity in the coronary arteries by passing the high frequency ultrasound waves into the blood using a single receiver. Our group has found that an improvement in velocity estimation can be obtained if the returned Doppler ultrasound echo is collected by multiple receivers and the information from those receivers is combined. The research proposed here will use simulation methods to determine the extent to which this same concept can be applied to multiple IVUS receivers.

1. introduction

Doppler ultrasound provides a measure of the velocity distribution of blood throughout the volume of the artery. Because the signal is a superposition of echoes from multiple scatterers, the red blood cells, distributed in space, the signal at the receiver is subject to constructive and destructive interference.

This phenomenon is called “ coherent scattering” and is the primary reason that Doppler ultrasound signals are inherently noisy. The same phenomenon applies to ultrasonic imaging (B-mode imaging), in which the Doppler shift is ignored and only the magnitude of the returned signal is used to form an image of the tissue. However, in B-mode ultrasound, the scatterers are variations in the acoustic impedance of the tissue. Figure 1 shows a typical IVUS image [6]. Whereas it is possible to differentiate between the lumen, media and adventitia, the exact boundaries are difficult to determine as a result of the coherent scattering effects.

2. background

2. 1. Principles of Ultrasonic B-Mode Imaging

B mode imaging is typically used for ultrasound imaging as it facilitates the display of the echoes at various brightness or gray levels corresponding to their amplitude.[ee handbook]

Most B-mode systems in use today

create an image in 0. 1 s or less, so that the image is displayed in real-time for viewing of moving structures,

such as structures in the heart or the fetus moving within the womb. This is not possible with the typical

magnetic resonance or computed tomography system.

Most of these systems use the Doppler principle, but some use time domain detection. In Doppler detection, if the ultrasound is reflected from a target moving at some speed vt toward (away from) the source at an angle

q with respect to the beam axis, the frequency of the transmitted signal f is shifted up (down) by an amount

fD, the Doppler shift, according to the following relation:

(116. 4)

In principle a measurement of fD, when f, c, and q are known, will yield the speed of the target vt. However, it

is often difficult to determine q because the angle the transducer axis makes with a blood vessel, for example,

is often unknown. Even when that angle is known, the flow is not necessarily along the direction of the vessel

at every location and for all times

Two-dimensional B-mode display: Echoes from a transducer, or beam, scanned in one plane displayed as

brightness (or gray scale) versus location for the returned echo to produce a two-dimensional image. Duplex ultrasound: Simultaneous display of speed versus time for a chosen region and the two-dimensional

B-mode image. B-mode display: Returned ultrasound echoes displayed as brightness or gray scale levels corresponding to the amplitude versus depth into the body

fig 6 in devts in cardio vascular ultrasound. pdf—describes input signal used………………………

B -mode (Brightness mode) ultrasound is the most commonly applied ultrasound technique for intracoronary artery visualization. B-mode images are made up of one dimensional signals from transducer crystals aligned in an arrays, which can also be displayed in two dimensional in the form of a sector[August et al]. In This mode of imaging, depth and the brightness are the measure of radial axis and echo intensity.

2. 1. 1. Scattering of sound

echo description fundamental sources of errors

2. 1. 2. Relationship between sound intensity and scattering coefficient

The size of the scattering shadow is called the effective cross-section (s [cm2]) and can be smaller or larger than the geometrical size of the scattering particle (A [cm2]), related by the proportionality constant called the scattering efficiency Qs [dimensionless]:

The scattering coefficient µs [cm-1] describes a medium containing many scattering particles at a concentration described as a volume density s [cm3]. The scattering coefficient is essentially the cross-sectional area per unit volume of medium.

Scattering coefficient: The factor that expresses the attenuation caused by scattering, e. g., of radiant or acoustic energy, during its passage through a medium. Note: The scattering coefficient is usually expressed in units of reciprocal distance. Attenuation: The decrease in intensity of a signal, beam, or wave as a result of absorption of energy and of scattering out of the path to the detector, but not including the reduction due to geometric spreading. [After JP1] Note 1: Attenuation is usually expressed in dB. Note 2: “ Attenuation” is often used as a misnomer for “ attenuation coefficient,” which is expressed in dB per kilometer. Note 3: A distinction must be made as to whether the attenuation is that of signal power

Measurement of the intensity distribution of laser radiation using…

by VV Morozov – 1979 – Related articles – All 4 versions

of interaction between the light and sound, i. e., in which the scattering coefficient would depend linearly on the sound intensity. …

2. 1. 3. Constructive and destructive interference

Sound travels in the form of waves. These waves are associated with frequency and amplitude. From basic laws of physics it is known that intensity is directly proportional to amplitude of the wave which is the discriminating factor between different modes of ultrasound imaging. When these sound waves interact with each other interference occurs. The type of interference is determined by measuring the amplitude of the resultant wave formed by interaction of 2 sound waves. If the amplitude of 2 waves is either positive or negative then the resultant wave has larger amplitude. This phenomenon is known as constructive interference(or in phase interference).

If the interacting waves have opposite amplitude then the resultant wave has a lower amplitude. This phenomenon is known as destructive interference(or out of phase interference). The interference type depends on the difference in distances that each wave has to take.

In this context, if the ultrasonic signals are emitted from a single transmitter and captured from multiple receivers separated by a distance of half the wavelength, then we can observe constructive interference of returned echo amplitudes of the scaterrers locate d in region of interest at one receiver and destructive interference occurring at the second receiver located half the wavelength apart.[cite reference wu thesis book].

2. 1. 4. Signal processing for B-Mode images (e. g. envelope detection)

Intracoronary ultrasonic is done on envelope detection of the sum of (returned) echo signals from each receiver . there are many/three ways of envelope detection. One simpler method of doing it is to perform a full wave rectification on the returned echo followed by a low pass filtering to remove the side lobes of the signal. [rectify/demodulate the signal and process it by passing it through a low pass filter to remove the side lobes of the returned echo]. though this method of envelope detection appears to be simpler, the operating center frequency for each of the returned signal is to be known and possibly tracked from time to time for changes. The second/another yet complex method of envelope detection is using Hilbert transform to get /generate/create a rational /methodical representation of the returned signal from each scatterer at each receiver. The advantage of using this method is that it is independent of the dampening effect present in the returned signal. (i. e. the changes in center frequency of the echo with time). The magnitude obtained from the complex signal is used as the final signal for ultrasonic imaging/next stage of converting into polar plots and plotting it using weighted average method . (refr: sprab12 page 11 and B-mode handbook). quadrature detection can also be used for extracting the envelope of the signal. refrce(high resolution ultrasound)

2. 2. Current implementations of intracoronary imaging

Heart disease can be diagnosed with the aid of Doppler and B-mode ultrasound, where the Doppler method provides a measure of flow rate and B-mode ultrasound provides an image. Generally these techniques, as typically used, do not have the spatial resolution to examine flow in the coronary arteries. Coronary artery geometry is diagnosed by injecting a radio-opaque dye into the artery with a catheter and taking x-ray images. However, this method does not specifically provide the locations of atherosclerotic lesions. It provides the internal geometry of the arterial lumen. Intracoronary Doppler ultrasound is a method in which a Doppler-tipped catheter is inserted into the coronary artery to measure blood velocity. IVUS uses a more complicated catheter that has an array of ultrasound crystals arranged in a ring at the tip of the catheter, and each crystal transmits an ultrasound pulse radially and then receives the returned echo. With multiple crystals, a 2-dimensional image of the cross-section of the arterial lumen can be reconstructed. This technique is currently capable of providing real time cross sectional images in vivo [3].

The main objective of Doppler ultrasound is to extract “ the flow velocity measurements and interpret them in physiologically significant variables “ through assumptions and velocity calculations [2]. The most fundamental quantity we consider is the flow rate as it best describes the extent of perfusion of blood in the region of interest [2] (i. e. a section of the coronary artery). The objective of IVUS is to obtain a mapping of the make-up of the artery as an image. Although IVUS uses several transmitter-receivers, only a single receiver is being used to capture the reflected ultrasound wave and to view the circumferential view of the artery.

The problem involved by using a single receiver is that we miss many precise details about the physiological status of the artery due to its limited view and the obtained images are noisy because of coherent scattering. The possible solution can be to use multiple receivers to look at the region of interest from different angles to get a detailed view. The doctor can get a clear picture of the artery in terms of velocity, flow rate, the size of plaque present inside artery and can treat the patient in a better way.

2. 2. 5. Geometry of the transducers.

We assume transducer as a point size spherical shaped piezoelectric crystal.

Papers: B-mode handbook. pdf

sprab12. pdf page 7

2. 2. 6. Transducer frequency response characteristics.

The phrase frequency response characteristic usually implies a complete description of a system’s sinusoidal steady-state behavior as a function of frequency.

2. 2. 7. Specifics of the transmitted signal

from program

2. 2. 8. Signal conditioning and signal processing

2. 3. Limitations of intracoronary imaging (particularly coherent scattering/scattering from multiple scatterers)

One of the main problems encountered with Doppler and B-Mode ultrasound velocity estimation is coherent scattering of noise. Coherent scattering error is caused by the changes in phase of the reflected echo as the red blood cells enter and leave the sample volume. " This phase depends on the distance of the transmitter to the scatterer and then to the receiver" [5]. The main research objective is to simulate this process of multiple receiver Doppler ultrasound using Matlab simulation software and to see how well it improves the understanding of image quality and clarity. Even the state of art of image is to be observed using the simulations.

paper : basic model of ivus. pdf page 8

Intravascular ultrasonic image quality remains poor due to speckle noise, imaging artifacts and shadowing of parts of vessel wall by calcifications. (Refce: intravascular ultrasound image segmentation.)

2. 4. Previous work done on multiple receivers – independence of coherent scattering noise in Doppler signals when receivers are sufficiently far apart.

Velocity Measurements made/obtained over the region of interest (ROI) in an intracoronary artery have inherited velocity estimation errors due to coherent scattering. One of the methods to reduce these estimated errors is the use of multiple receivers . The echo received from each of the receivers will have some complementary information which not only improves velocity estimation, also contributes in enhancing the image clarity in a B-mode ultrasound image processing. [Jones, Krishnamurthy 2002] Improvement in velocity estimation is observed if returned Doppler ultrasound echo is collected and combined from all the multiple receivers. Most importantly the returned echo signal obtained at each receiver is independent of coherent scattering noise in Doppler signals when receivers are sufficiently far apart. In the case of an intracoronary artery, the RBC's are the major kind of multiple scatterers distributed in artery space. Since the returned echo signal detected/obtained at each of the receiver is a summation of all the echo amplitude signals from multiple scatterers in the region of interest, they are subjected to constructive and destructive interference This way multiple receivers in B-mode can improve the image quality of B -mode intravascular ultrasound (IVUS) images.

Initially a 2 dimensional geometry for the artery would be simulated. The transmitted signal would be generated using by a piezoelectric crystal in an ultrasound in real time applications. But in this proposed research, using Matlab, we first try generating a discrete signal using the pulse generator. Based on the defined parameters such as the frequency, pulse width, amplitude, pulse repetition time, artery geometry[r (?)], angle of transmission of the transmitted and received signal, the image would be extracted. Primarily, the signal from a single scatterer is modeled. In the advanced stages, multiple scatterer signals would be modeled. The following questions would be answered while doing the actual simulation.

Each scatterer is modeled as a point source that reflects the transmitted signal with a set reflectivity. The scatterer does not alter the signal's phase, but alters the amount of power that is returned to the receiver. Each receiver therefore is subjected to a signal that is the sum of returns from all of the scatterers, where it is important to keep track of the phases of the signals from each scatterer so that coherent scattering is adequately accounted for.

The signal at each receiver is rectified and then averaged in time with a moving window to produce a signal that represents scattered power as a function of time. The range, corresponding to the location in the image, is proportional to the delay time of the returned signal.

Each receiver will provide an image, and a composite image will be produced as the average over all of the receivers.

3. 4. 1. Transmitter/Receiver characteristics (transmitted frequency, beam width)

3. 4. 2. Speed of sound

3. 4. 3. Scattering coefficients for (1) Background and (2) Plaques

The fraction of the incident energy reflected or scattered is very small for soft tissues like elastin collagen etc. [ee handbook] The differential Backscattering coefficient/scattering coefficient is the aspect that expresses the attenuation caused by scattering, of acoustic energy, while passing through a medium. The scattering coefficient (µs) is usually expressed in units of reciprocal distance. There certainly lies a difference between the normal aortic intima and various kinds of atherosclerotic plaques. More than 90% of normal vessels usually have scattering coefficients in the range of 15 mm-1 to 36 mm-1 , where as atherosclerotic plaques like the lipid rich blocks, fibrocalcific plaques have scattering coefficients lesser than 20 mm-1 [Levitz, Andersen et al]. The fibrous plaques which constituted elasin , lipids and collagen demonstrated a relatively large variations in terms of scattering coefficient. Out of the three kinds of atherosclerotic plaques, fibrocalcific samples do not show up as sharp regions in any kind of image and hence can be assumed as in homogeneities within the tissue wall having highly scattering coefficient.

3. 4. 4. Random numbers (particle location and scattering coefficients)

3. 5. Signal Analysis (envelope detection)

http://www. mathworks. com/products/demos/shipping/dspblks/dspenvdet. html

Hilbert Transform can be used to generate a time domain envelope. The point is to create a “ rectified” signal that is more suitable for calculating a smooth envelope. In the frequency domain, magnitude data is already all positive, so I don’t know why you’d use Hilbert Transform. To get a spectrum envelope, just average several spectrum frames together. The key then is to choose correct frame size prior to FFT, which should be based on the nature of your data and the sampling rate you are using. Averaging will help your SNR and maybe you can differentiate key frequencies between good and damaged bearings. x = Hilbert (xr) returns a complex helical sequence, sometimes called the analytic signal, from a real data sequence. The analytic signal x = xr + i*xi has a real part, xr, which is the original data, and an imaginary part, xi, which contains the Hilbert transform. The imaginary part is a version of the original real sequence with a 90° phase shift. Sines are therefore transformed to cosines and vice versa. The Hilbert transformed series has the same amplitude and frequency content as the original real data and includes phase information that depends on the phase of the original data. If xr is a matrix, x = Hilbert (xr) operates column wise on the matrix, finding the Hilbert transform of each column. x = Hilbert (xr, n) uses an n point FFT to compute the Hilbert transform. The input data xr is zero-padded or truncated to length n, as appropriate. The Hilbert transform is useful in calculating instantaneous attributes of a time series, especially the amplitude and frequency. The instantaneous amplitude is the amplitude of the complex Hilbert transform; the instantaneous frequency is the time rate of change of the instantaneous phase angle. For a pure sinusoid, the instantaneous amplitude and frequency are constant. The instantaneous phase, however, is a saw tooth, reflecting the way in which the local phase angle varies linearly over a single cycle. For mixtures of sinusoids, the attributes are short term, or local, averages spanning no more than two or three points. Reference [1] describes the Kolmogorov method for minimum phase reconstruction, which involves taking the Hilbert transform of the logarithm of the spectral density of a time series. The toolbox function rceps performs this reconstruction. For a discrete-time analytic signal x, the last half of fft(x) is zero, and the first (DC) and center (Nyquist) elements of fft(x) are purely real. http://dip. sun. ac. za/~weideman/research/mfiles/hilb1. m

function h = hilb1(F, N, b, y)% The function h = hilb1(F, N, b, y) computes the Hilbert transform% of a function F(x) defined on the real line, at specified % values of y (y could be a scalar, vector, or matrix.)

8. bibliography

1. http://www. johnshopkinshealthalerts. com/white_papers/heart_health_ch_wp/digital08_landing. html, last accessed on 02/26/2008, 08: 30 pm.

2. Jones SA, “ Fundamental Sources of error and spectral broadening in Doppler ultrasound signals”, Crc critical reviews in Biomedical Engineering, page(s): 399-483, 1993.
3. Van der Steen AFW, Cespedes EI, de Korte C. L, Carlier S. G , Li W, Mastik F, Lancke C. T, Borsboom J, Lupotti F, Krams R, Sermys P. W, Bom N,” Novel developments in intravascular imaging”, Ultrasonics Symposium Proceedings, IEEE, Volume 2, page(s): 1733 – 1742, 1998.
4. Kumar P and Shoukri MM,” Copula based prediction models: an application to an aortic regurgitation study,” BMC Medical Research Methodology, page 7: 21, 2007.
5. Jones SA and Krishnamurthy K,” Reduction of coherent scattering noise with multiple receiver doppler”, Ultrasound in Med. & Biol., Volume 28, page(s): 647-653, 2002.
6. Zhu H, Oakeson K D, and Friedman M H,” Retrieval of Cardiac Phase from IVUS IVUS Sequences”, Medical Imaging 2003: Ultrasonic Imaging and

   Signal Processing, Volume 5035, 2003 page(s): 1605-7422.