Temporal Processing of High Frame Rate Ultrasound Images

by

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Duke University

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Dissertation submitted in partial fulfillment of the requirements for the degree of Doctor of Philosophy in the Department of Biomedical Engineering in the Graduate School of Duke University

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ABSTRACT

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Abstract

High frame rate ultrasound imaging is a nascent field that holds the potential to bridge the gap between strictly anatomic imaging and functional, or physiologic, imaging. Other currently available functional imaging methods in ultrasound, such as Doppler, ARFI, and SWEI, require transmit-receive sequences different than used in standard B mode or volumetric imaging, requiring a reduction in imaging rate. Using the Duke University Phased Array System, T5, a high frame rate imaging scheme was developed for live acquisition of adult echocardiographic images at rates up to 1000 per second.

It is hypothesized that with higher imaging rates low signal level, rapidly moving blood flowing through vessels can be differentially extracted from B mode or volumetric images and provide enhanced contrast with respect to high signal surrounding or overlapping stationary or slowly moving structures. To achieve this goal, an online method of subtracting sequential images was developed for the T5 system. The origin of the increase in contrast is investigated, and the statistical model of speckle in B mode images is extended to include the changes in brightness in a fixed location in the image field as targets pass through that location. Further studies were directed at increasing the contrast between blood and surrounding tissues by
summation of difference images and considering the limitations of such summing for cyclical blood flow.

It is concluded that:

A) With the T5 system, high images rates are feasible without significant reduction of image resolution or signal to noise.

B) Statistical targets, or speckle, have a pixel-by-pixel linear relationship between successive image frames between changes in brightness and amount of translation up to the diffraction limited resolution in azimuth and proportional to the pulse length in range. Then, brightness within a given pixel may be statistically independent from the brightness within the same pixel depending on the speed of imaging.

C) The point at which the brightness in a given pixel from frame to frame is statistically independent in a given pixel from frame to frame is directly related to the resolution of the configured ultrasound system.

D) Subtracted images of independent speckle patterns are statistically independent with each other and summation of such images within physiological constraints in time will further increase the contrast of blood with respect to vessel walls and surrounding tissue.
E) Within physiologic constraints, time domain processing methods can be used to increase image contrast of rapidly moving targets and suppress the appearance of slowly moving, transiently moving, or stationary targets.
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1. Introduction and Hypotheses

1.1 Introduction

Cardiac disease is the leading cause of morbidity and mortality in the United States [Benjamin 2017]. With advances in pharmaceutical treatment and the availability of more sophisticated implantable devices, the instruments for diagnosis also need to advance to increase success of treatment. Current clinical practice employs ultrasonic imaging systems capable of imaging at maximum rates of 60-100 frames per second for standard adult echocardiographic B-Mode imaging. For the diagnosis of many anatomic based heart failure via mechanisms such as dilation or hypertrophy, these imaging rates may have been sufficient.

It is proposed that for the non-invasive diagnosis of coronary disease, high frame rate imaging is also required. Coronary arteries are small targets, typically of 5 mm diameter, situated within on the surface of the heart which moves dynamically within the thoracic cavity. To image blood flow within coronary arteries will require a method of increasing contrast of low signal blood while simultaneously suppressing high signal myocardial targets. As peak coronary blood flow rates in healthy adults reach 1 m/sec or higher [Ofilli 1993], a method of discriminating between high velocity targets such as coronary blood flow and slower targets such as the myocardium would allow for the visualization of coronary blood flow. Preliminary work in two dimensions will be
explored for the extension to three dimensions for the non-invasive visualization of coronary flow without injected contrast.

For the accurate diagnosis and successful treatment of electromechanically based heart failure, such as in the case of Left Bundle Branch Block (LBBB), these image rates and current diagnostic metrics result in up to 1 in 3 chance of the patient not responding to the echocardiographically indicated treatment [Bax 2005]. Here, it proposed to increase the efficacy of ultrasound based diagnosis by increasing the frame rate of imaging. Also, the ability to image and analyze electromechanical events in the heart requires higher frame rates than currently available. The current American Heart Association/American College of Cardiology consensus for electrical measurements via the electrocardiogram (EKG) is a minimum of 500 samples per second [Kligfield 2007]. This indicates that to be able to correlate mechanical events as measured via echocardiography to electrical events via EKG, a comparable imaging rate of at least 500 frames per second must be achieved.

1.2 Hypotheses

In this work, it is hypothesized that:

A) The Duke Phased Array Ultrasound System, T5, can be configured to acquire live B mode images at rates up to 1000 per second for adult cardiac applications without significant loss of image resolution or signal to noise.
B) Within a given image pixel that contains a speckle target, the change in image brightness is directly proportional to the movement of the speckle target as long as the speckle remains correlated. Beyond the point of correlation, the brightness in a given pixel from one frame to the next will be statistically independent.

C) Two statistically independent speckle patterns may be subtracted to create a new image, and the brightness distribution within the new image be described by the difference of two Rayleigh distributions. As an extension, that if two or more difference images are generated from three or more statistically independent speckle patterns, that the resulting difference images are also statistically independent.

D) Two or more statistically independent difference images may be summed for a continued increase in image contrast, and continued summation will result in the reduction of variation of brightness in the resultant difference image.
2. Background

Given the rates of physiologically significant events in the heart such as the propagation of depolarization events that precede mechanical contraction (1-3 m/sec), images rates under 100 fps are unlikely to allow for the visualization of these rapid events and preclude the ability to describe the propagation of such events. Inadequate temporal sampling is further exacerbated as images are formed by scanning the image field over a given amount of time. A single ultrasound image is formed one B-Mode line at a time, with the acoustic pulse being steering in a different direction for each transmit-receive operation. As a result of this method of image generation, each portion of one image frame has been sampled at different times. This is an additional confounding aspect to B-mode imaging when compared to conventional imaging, e.g. photography, where each portion of the image has been sampled at the same time. In ultrasound imaging, there is a variation in temporal sampling in each part of an image frame.

The scanned nature of ultrasound images introduces several complications when trying to make physiologically relevant measurements from these images. Since in traditional B-Mode imaging, the image field is scanned sequentially from one extreme of the field of view to the other, targets on one side of the image are captured at a significantly different time than at the other side of the image. In the example system specified above, this corresponds to a timing difference of 23 msec from one side of the image to the other. In applications such as measuring mechanical correlates of electrical
depolarization events, which have been measured to propagate at velocities of 1-3 msec, this duration of time will lead to the mischaracterization of these events across the image field. As an example, consider a point target centered in the image field at a range of 10 cm. If this target were to split in half and each half travel in opposite directions, parallel to the transducer face at equal velocities of 2 m/sec, and scanning were to begin at the leftmost edge of the image field at the instant the targets begin moving, these two targets would appear to have moved a different amount within the same image. In this example, the target moving left in the image will be imaged at 9.0 msec into image acquisition and have traveled 15.8 mm to the left of the center. The rightward travelling target will not be imaged until 16.8 msec into image acquisition, 7.8 msec after the other target was imaged. This rightward moving target will appear to have travelled 33.2 mm to the right of the origin, or 18.4 mm further than the leftward moving target even though these targets have identical velocities. Had the image been acquired as a single snapshot of the image field at half the full acquisition interval, i.e. 11.5 msec, both targets would have appeared to travel the same distance of 23.0 mm.

The previous example assumes point targets that are moving very rapidly, at velocities comparable to some of the highest velocity physiologic phenomena, such as the depolarization of the myocardial tissue. It is uncommon to encounter true point targets in echocardiography. Targets can generally be grouped into two categories: large specular, mirror-like structures and or speckle-like structures. Examples of the
former include the mitral and aortic valves as well as vascular lumens such as the aortic root. Statistical targets are tissues or liquids, such as myocardium or blood, that are composed of a large number of very small (i.e. smaller than the system resolution) scattering targets that result in the received echoes appearing as an interference pattern from these sub-resolution scatterers.

The previous example demonstrated the uncertainty of measurements made using a scanned system with point targets, but is also of interest to extend this discussion to targets of finite extent. If for example the target were a bullet traveling through the image field at a high velocity relative to the scanning rate, the appearance of the bullet will depend upon the direction of scanning with respect to the propagation direction of the bullet. In a system scanning left to right, if the bullet also travels left to right through the image plane, the bullet will be distorted to appear larger in the image than it actually is. This is due to the scan direction tracking in the same direction as the target is moving. As the velocity of the bullet increases, it will continually elongate until the scan velocity and target velocity are equal at which point the bullet will be smeared across the entire image field if the scanning is synchronous to the bullet entering the image field. If not synchronous, targets with velocities near or exceeding the scan rate may never be imaged using a scanned system. When target velocities exceed the scanning velocity, spatial aliasing can occur, and in this case, the bullet, if imaged, will appear truncated. In the counter example where the bullet is travelling opposite the
scan direction, the bullet will be truncated, and as target velocity increases to the scan rate, the bullet will truncate down to a point target, if it appears in the image at all.

So far only motion in one dimension has been discussed. The heart, however, is a three dimensional target with complex motion in all dimensions and directions. As an example, the mitral valve moves in multiple directions and when open during diastolic filling exhibits very complex fluttering as blood enters the left ventricle. In a parasternal long axis view, the mitral valve is aligned approximately parallel to the transducer face. Given the length of the valve and typical location in a parasternal long axis view, it may take 10-20 msec to scan along the full length of the valve. As the valve is constantly moving in three dimensions at relatively high velocities, slowly scanned images could display discontinuities in the valve of a healthy patient or could obscure otherwise clinically significant information in the images.

Assuming that imaging frame rates are fast enough to capture and visualize rapidly moving targets or short duration events, the ability to track and derive subsequent physiologically relevant measurements is more complicated. In echocardiography most clinical scanners in use today employ the sector scan format for image formation. If the total width of the scan or field of view (FOV) is $\theta_{\text{max}}$ and the frame rate is FR per second, then the maximum trackable velocity across the field of view parallel to the transducer for a deterministic target such as an air bubble or small objects such as a BB is:
where $R$ is the range of the target in the image. For targets moving only in range, the maximum trackable velocity is $v_{\text{max}, r} = R_{\text{max}} \cdot FR \text{ m/sec}$, where $R_{\text{max}}$ is the maximum range of the sector scan. Of course, the amplitude of the deterministic target must be such that it clearly exceeds the background level. Typically, the limit of detectability in a noisy random environment is reached when the signal $S$ equals the noise level $N$ for a SNR = 1. Usually $S$ and $N$ are root mean square values.

In the case of blood flow where there exists a large number of micron sized formed particles such as red and white blood cells, platelets, etc. or myocardial targets where myocardial cells are smaller than the resolution of the imaging system, the coherent nature of pulse echo imaging results in a three dimensional interference pattern [Bashford 1995]. In two dimensional sector images, this speckle pattern has Rayleigh statistics [Burkhardt 1978, Wagner 1983] and decorrelates rapidly for both motion parallel to the transducer face and in range. In 2D images, Trahey et al [1986, 1988] found that speckle decorrelated for displacements of half the aperture width for motion parallel to the transducer for an extended region about a mid-range, 7 cm transmit focus. If speckle is to be used as a means of tracking moving objects such as blood or myocardium in B-mode images, then a non-zero correlation between image frames is necessary. Supposing a 0.5 correlation value, which corresponds to a 0.2 aperture size motion, is acceptable in terms of tracking sensitivity, then for a typical sector scan, the
maximum trackable velocity parallel to the transducer face at mid-range in a B-mode image would simply be:

\[ v_{max} = 0.2L \cdot FR \text{ mm/sec} \]

where \( L \) is the aperture width associated with scan direction \( \theta \) in mm. As an example, suppose \( L = 20 \text{ mm} \) and \( FR = 30 \text{ /sec} \). Then \( v_{max} = 12 \text{ cm/sec} \), while at a FR of 1000 fps, \( v_{max} \) would be 4 m/sec. Of course, for improved sensitivity in tracking, a higher degree of correlation would be appropriate. It is clear that as the level of correlation decreases, so does the ability to accurately track the random speckle target. For pure range motion, decorrelation is proportional to 50% of the effective pulse length in space. For example, a 4 cycle pulse at 3.5 MHz would span about 1.8 mm in water or tissue so that for reasonable accuracy the maximum range velocity would be approximately \( PL/2 \cdot FR \), or about 2.7 cm/sec at 30 fps, or 90 cm/sec at 1000 fps. All of the above examples assume that images are formed from a single transmit-receive sequence. For scanned images, the complications include all of those mentioned previously.
3. Realization of live, high frame rate B-Mode imaging

3.1 Background

Increasing imaging frame rates in ultrasound has been a research topic of interest since the 1980’s [von Ramm 1984]. As ultrasound systems have advanced to include increased parallelism and increased digital memory storage, a resurgence in high frame rate, two dimensional ultrasonic imaging started in the late 2000’s and has continued since [Yoshiara 2007, Couade 2009, Tanter 2014, Cikes 2014].

3.2 Echocardiographic image formation

Frame rates in cardiac ultrasound are fundamentally limited by the physics of acoustic propagation as well as the design of the ultrasound system. The majority of two and three dimensional imaging in adult echocardiography is tomographic, pulse-echo, brightness mode (B-Mode imaging). To generate an image, an ultrasonic pulse is transmitted from a transducer into the patient, and the same transducer is used to listen for reflected acoustic pulses from within the body. Through several processing stages, the envelope of the received echo is detected and converted into a brightness value in the resultant image. This will generate a single tomographic image line in one direction for each transmitted pulse. Typically, in echocardiography, the 2D images are in a sector scan format while 3D images are scanned over a solid angle in both azimuth and elevation. To generate a two or three dimensional image, pulses are transmitted in different directions and are either steered electronically or mechanically between
subsequent transmitted pulses to interrogate the entire desired field of view. Angular steering can be accomplished by mechanically rotating the transducer or by temporally phasing the transmitted pulses across array elements. In traditional B-Mode imaging, a single pulse is transmitted in one direction, the system listens for reflections for a duration determined by the required maximum range, or depth, of the image, and then the next pulse is transmitted in a different direction. Using this approach, the maximum achievable frame rate (FR) is determined by several factors which will be broken down into greater detail but can be summarized by:

$$FR = \frac{1}{N_{TxRx} \cdot t_{line}}$$

where $t_{line}$ is the amount of time required for one transmit-receive operation and $N_{TxRx}$ is the number of transmit-receive operations required to generate one image.

For all following examples, the system in question will be a two-dimensional, sector scanning, B-Mode imaging system. In this system, the time required for one image line depends on the maximum desired range in the image and the speed of sound, or:

$$t_{line} = \frac{2r_{max}}{c_{tissue}}$$

where the two is required due to the pulse-echo nature of the imaging system. The speed of sound is a physical property of the medium being imaged and cannot be changed. On average for tissues involved in echocardiography, $c_{tissue} \approx 1540$ m/sec. The
maximum range required is dependent upon the size and anatomy of the patient being imaged. For most adult humans, to image the entire heart transthoracically, ranges of 10-15 cm are required. In cases of obese patients, a maximum range of 20 cm or greater may be required; however due to the attenuation of acoustic pulses with propagation, increasing the maximum range beyond 20 cm may not result in any additional information due to the limited signal to noise ratio of the system. For the consideration of imaging rates, the time for the acquisition of a single line is dependent upon the anatomy of interest.

To determine the number of image lines required in an adequately sampled sector scanned image, further anatomic and physical factors must be considered. The minimum number of image lines for an adequately sampled image is:

$$N_{\text{lines}} = \frac{2 \, \text{FOV}}{\theta_r}$$

where FOV is the desired field of view of the image, $\theta_r$ is the resolution limit of the imaging system, and the factor of two satisfies the Nyquist sampling criterion for spatial sampling in the system. The field of view is determined by the anatomy of interest being imaged but is limited on the upper end by the angular response of the transducer used for imaging. Most phased array transducers employed in echocardiography have a practical limit of ±45º due to the angular response of individual array elements. For adult cardiac imaging, typical field of view is 70-90º, as this width is required to image
the majority or entirety of the heart. The Rayleigh resolution limit of an ultrasound system depends on both the size of the transducer and the wavelength of sound used for imaging, which for a rectangular aperture is:

$$\theta_r = \sin^{-1} \frac{\lambda}{D}$$

where, $\theta_r$ is the Rayleigh resolution limit, $\lambda$ is the wavelength of the transmitted pulse, and $D$ is the width of the aperture of the transducer. For adult echocardiography, typical imaging frequencies are in the 2 to 5 MHz range. Lower frequencies are undesirable due to the loss of resolution. Higher frequencies provide higher resolution although offer significantly less penetration as attenuation is frequency dependent. Apertures for transthoracic echocardiographic imaging are limited by anatomic considerations. Acoustic waves propagate through media of similar acoustic impedance but are mostly reflected when encountering an interface between two media of significantly differing impedances. In the thoracic area, blood and soft tissues have similar impedances, while bone (e.g. ribs) has much higher impedance and air (e.g. lungs) has much lower impedance. Acoustic pulses that encounter a tissue/rib or tissue/lung interface will be almost entirely reflected, limiting the useful size of the acoustic window and thus the maximum useful aperture size. For adult echocardiography, typical practical transducer aperture sizes are 15-20 mm in azimuth,
i.e. the scan plane in two dimensional imaging, and 10-15 mm in elevation, i.e. at right angles to the scan plane.

Additional consideration to imaging rates include system reset times between transmission and reception, or turnaround time, as there may need to be time for instructions to be issued to various sub systems in the ultrasound scanner. Depending on system architecture, there may be a fixed delay between successive transmit-receive operations and an additional delay between image frames. Such system delays are inherent in the system architecture and may be difficult to change.

Taking account all anatomic and physical limitations, the maximum frame rate for a conventional ultrasound scanner is:

\[
FR = \frac{1}{(2 \frac{r_{\text{max}}}{c_{\text{Tissue}}} + t_1) \left( \frac{2 \text{FOV}}{\sin^{-1} \frac{\lambda}{D}} \right) + t_2}
\]

where \(t_1\) represents the system delay between successive transmit-receive operations, and \(t_2\) represents the system delay between successive image frames. A typical scanner used for adult echocardiography would have the following specifications: maximum range of 15 cm, FOV of 80º, transmit center frequency of 3 MHz, aperture width of 20 mm, and a turnaround time of 5 \(\mu\)sec each between transmit-receive operations and between frames. These specifications take 23 msec to generate a single image frame, allowing for a maximum possible frame rate of 43 per second.
3.3 Increasing Imaging Rates for Echocardiography

Increasing imaging frame rates in ultrasound has been a topic of research interest since 1984 when von Ramm et al presented the first system capable of imaging at rates up to 240 frames per second using a high speed video camera for image storage. As systems transitioned from analog to digital, the realization of ultrasonic scanners capable of imaging at rates greater than 240 per second was delayed until the mid-2000’s [Yoshiara 2007].

The principle advance that allowed for this increase in acquisition rate in echocardiography was provided by a technique known as explososcanning. Explososcanning is a method developed in the 1980’s where for a single transmitted acoustic pulse, multiple image lines are received simultaneously in parallel [Shattuck 1984]. To accomplish this, a widened transmit beam is used, and the received echoes from each element of the array are piped into multiple delay systems, with a maximum amount of parallelism determined by the total number of delay systems available. In its original implementation, four independent delay systems were used to allow for a 4 times increase in maximum frame rate. In more recent literature, an alternative name has also been used for this technique, Multi-Line Acquisition [D’hooge 2014].

As mentioned, explososcanning does require a wider transmitted acoustic beam, as there will be no reflected sound unless there is acoustic energy transmitted to that
location. Several methods have been proposed for widening the transmitted acoustic beam for the purpose of increasing the amount of parallelism physically viable in receive. Initial studies of exploscanning proposed the reduction of the active transmit aperture [Pavy 1993]. As the system resolution, which is directly related to beamwidth, is inversely proportional to the aperture size, a reduction of the active transmit aperture will naturally widen the transmitted beam. This method, however, sacrifices electronic signal to noise in the resultant image as less acoustic energy is deposited into the body for each transmitted pulse and is spread over a wider beam. An additional method that was used in conjunction with aperture reduction was an unfocused transmit beam [Pavy 1993, Mallart 1992]. Conventional B-Mode imaging uses a transmit beam focused within the image field to maintain as narrow a beam as possible over the entire range to be imaged. By not focusing the transmit beam electronically, a wider beam that focuses at the diffraction determined transition distance will be transmitted. For the transducers typically used in adult echocardiography, the transition distance is beyond the maximum range being imaged, so a relatively wide beam can be achieved over the area being imaged. Even when using the full transmit aperture, an unfocused transmit beam will still result in an overall loss of signal in resultant images as the same amount of acoustic energy remains spread over a wider area as compared to a focused beam.

A third method of widening the transmitted beam is by negatively focusing the transmitted pulse. In this method, elements are timed such that they are activated by a
point target originating behind the transducer face, creating a convex delay profile where the center elements are activated earlier than elements on the edges of the array. In this case, the geometry of the transmitted wave front could be shaped to be “diverging” or circular if diffraction effects did not dominate the beam shape. Negatively focusing can be used to achieve even wider beam widths than simply reducing the aperture or unfocusing the transmit. However, there are considerable tradeoffs when using a negatively focused transmit beam. As in both previous cases, the acoustic energy is spread over an even wider area, resulting in less signal than either of the previous cases. Additionally, the pulse is spatially stretched, resulting in a loss of range resolution and increasing the potential for multi-pathing artifacts and other such artifacts that can reduce signal or increase noise in the image.

Signal to noise could be increased by increasing the transmit amplitude; however, there are potential limitations. As the majority of transducers used in echocardiography are made of a piezoelectric ceramic, increasing transmit voltage may cause the crystal to fatigue over time and can even cause the crystal to break if increased too high. At higher voltages nonlinearities may limit the amount of energy in the passband of the system. Safety regulations also prohibit the maximum amount of acoustic energy that can be deposited at the skin surface for diagnostic ultrasound and increasing transmit amplitudes beyond this level cannot be employed in clinical practice.
While parallelism was first implemented in receive, there can also be parallelism employed in transmit. Multiple transmit beams can be accomplished by temporally serializing multiple transmits or by superimposing multiple transmit profiles simultaneously [Dubberstein 1998, 2000]. Multiple transmit beams allow for narrower beams on each transmit for a given number of explosions (i.e. independent receive delay systems) but introduce other potential artifacts in images. Whether temporally serialized or superimposed, crosstalk between the beams cannot be excluded due to inherent background power levels in array imaging. Multiple frequency beams have been used to reduce this crosstalk [Dubberstein 1998], but attenuation variations between beams adds a level of complexity to the display system. Temporally serializing these transmitted pulses results in a reduction of maximum achievable frame rate and can introduce discontinuities in images if the timing is not controlled with high accuracy. To properly superimpose multiple transmits it may also necessitate apodization of the transmit profiles which will lead to a reduction of transmitted energy and loss of signal in resultant images.

In adult echocardiography there are very rapid, short duration events as well as high propagation velocity waves that are currently of great academic interest [Konofagou 2010, 2012, Provost 2011, 2013, Kanai 2011] and potential clinical significance [Cikes 2014]. With the available T5 hardware, it allows a total of 32 to 1
explosos. The studies presented are aimed at determining beam responses for various focusing schemes to widen the transmit beam using the Duke T5 system.

**3.4 T5 Machine Specifications**

T5, the Duke University Phased Array Scanner, was designed and built over the past 15 years to perform real-time cardiac volumetric scanning at rates up to 40 volumes per second. The high acquisition rate is a result of the custom hardware and parallelism designed into the machine: 512 transmit channels, 1024 receive channels, and 32 to 1 parallel processing per channel in receive. Additionally, T5 is capable of 8:1 near-simultaneous transmits. Each channel is independently programmable and processed individually with no partial summation in the transducer handle as is common in commercial scanners. To accommodate the high data rates achievable with this system, a customized Dell PC is used for data acquisition. This computer is outfitted with three *DALSA X64 FrameGrabber* cards, allowing for high frame rate 15 bit data acquisition via two CameraLink connections and simultaneous acquisition of synchronous EKG and various other data on a third CameraLink connection. In-house software has been developed for to perform the acquisition and display of detected ultrasound data [Kuo 2007]. This software allows for the simultaneous real-time acquisition and display of B-Mode data as well as temporally processed images as well. This display runs in real-time during acquisition and can be slowed during retrospective playback or advanced
frame by frame to assess motion and changes in the images. Various image parameters, such as brightness, gamma and contrast, can be adjusted during playback mode and various temporal and spatial filters applied.

### 3.5 Optimizing Transmit Beam for T5 System

To use the T5 system described above for high frame rate 2D B Mode imaging, the transmit beam needs to be modified to capitalize on the available hardware in the system. For 2D imaging, a 3.5 MHz, 96 element linear array is used. This array is 21 mm in azimuth and 14 mm in elevation. Given this geometry and transmit frequency, the theoretical Rayleigh resolution limit is 1.2°. The total field of view is set to 80°, as this is sufficient to image the majority of adult hearts transthoracically. To insure adequate spatial sampling, image lines are spaced 0.5° apart, necessitating 160 total image lines. Given the receive parallel processing available in the T5 system, one full B Mode image can be generated in five transmit-receive operations.

The two possible schemes considered were a single transmit with the full 32 to 1 parallelism applied and a dual transmit with 16 to 1 parallelism applied to each transmit direction. Preliminary studies compared the tradeoffs between the single and dual transmit approach, and example images of each configuration can be seen in Figure 1. While overall image dynamic range and contrast are higher when using the dual transmit approach, crosstalk between the transmit beams causes undesirable artifacts.
such as the duplication of very bright targets on both halves of the image field. Additionally, when imaging dynamic targets such as the heart, introducing a discontinuity in the temporal sampling in the center of the image was found to significantly reduce image quality.

To improve image quality for imaging at high frame rates with the T5 system, the optimal transmit beam was qualified by three metrics. The first metric is -6 dB beam width averaged over 13 cm in range. The beam width is important as regions that do not have acoustic energy transmitted cannot be imaged. The second metric is peak amplitude with respect to the peak amplitude of a focused transmit beam. The lower transmitted amplitude will result in a loss of signal and reduction in image contrast and dynamic range. The third metric is the ripple in the main beam, or the variation in amplitude across the -6 dB beam width. Minimizing ripple is important to minimize variations in brightness not due to the target being imaged.

3.6 Methods

Three transmit schemes were investigated to achieve a beam at least 16º in width to accommodate 32 image lines spaced at 0.5º, with minimal ripple and lowest average signal loss. To avoid the self-focusing effect (i.e. the natural narrowing of the beam), a beam was generated by negatively focusing at 30 cm, i.e. behind the transducer face. For the transducer used, the transition distance is approximately 30 cm. For comparison, a beam focused at mid-range (7 cm) and a beam focused at 1,000 cm (the unfocused beam)
were also measured. The f-number in receive varied from 1 to 6.5 depending on the depth. No amplitude apodization was implemented in transmit or receive, nor was spatial or temporal compounding employed.

For each transmit scheme, beam plots were measured by clamping the transducer to a ring stand in a water tank. An OndaCorp HGL-0200 hydrophone (Onda Corp., Sunnyvale, CA, USA) was attached to a custom rotational stage with the center of rotation aligned to the center of the array transducer. Peak-to-peak receive voltage from the hydrophone was measured with an oscilloscope (Agilent DSO6054A, Agilent Technologies Inc., Santa Clara, CA, USA) at 1° intervals over a field of view from −35° to +35°. Measurements were made every 2 cm, from 3 cm to 13 cm in range. At each range, the hydrophone’s location was centered on axis in both azimuth and elevation by translating the hydrophone until the maximum receive voltage was measured. The transmit beam was focused at the range being measured during the centering process. Then measurements were made using the three transmit beams. To visualize the appearance of the ultrasound beams as a function of distance from the transducer, we scanned a synthetic sponge with regular cell spacing in a water tank. A B-mode image of the transmit beam pattern was produced by transmitting on axis, at 0°, while receive beamformers were steered in the usual fashion over the sector field of view. For these images, the transducer was manually held at the edge of the sponge. Images were obtained with unfocused, negatively focused, and focused beams and were recorded.
To quantify image quality for high FR imaging, both spatial resolution and image contrast were measured. Spatial resolution was determined by imaging an AIUM standard resolution phantom. This phantom contains wires spaced at 1, 2, 3, and 4 mm apart in the center. The linear array described previously was clamped in a fixed position and the phantom was aligned so that the resolution targets were located near the focal point of the focused transmit beam, at 70 mm depth. Images were acquired with focused, unfocused, and negatively focused transmit beams, acquiring two image lines for each transmitted acoustic pulse to achieve a target FR of 60 Hz as is typical in clinical practice. A fourth image was acquired using the same settings as when scanning at 1000 fps, using a negatively focused transmit beam and receiving 32 images lines for each transmitted pulse. The image contrast ratio was measured by imaging a tissue mimicking phantom (model 040GSE Multi-Purpose Multi-Tissue Ultrasound Phantom, Computerized Imaging Reference Systems Inc., Norfolk, VA, USA). A central region of the phantom that contains a 0.5 cm diameter anechoic void was imaged four times, once for each of the three transmit schemes with two image lines received per transmitted pulse, to achieve a FR of 60 Hz. The fourth time, the phantom was imaged at 1000 fps with a negatively focused transmit and 32 image lines received for each transmit. System gain and compression settings were identical for all four sets of images and were set to levels typically used for adult echocardiography. The average brightness of a 5 by 5 region of samples within the void was calculated over 60 frames; likewise, the average
brightness of an 11 by 11 region of samples in the tissue mimicking region (i.e., speckle pattern) was taken over 60 frames. The ratio of these two values was calculated to give the image contrast ratio.

3.7 Results

For all studies, the array used in vitro was a linear array with 96 elements, 21 mm aperture in the scan plane (azimuth), height of 14 mm (elevation), and center frequency of 3.5 MHz. Beam plots measured for each transmit scheme are shown in Figures 2, 3, and 4. The received voltages are displayed on a decibel scale, using the maximum of all measurements as the reference voltage. This reference voltage at 0 dB in Figure 4 corresponds to the peak voltage measured at the 7 cm range for the focused transmit. The amplitude of each beam plot at different ranges can be measured by using the receive amplitude scale on the y-axis on the left of the 3 cm range plot. The diagonal line indicates the intersection of the ~40 dB amplitude at 0º off axis at each range.

Figure 1: Example B mode images of single transmit on left, dual transmit on right
Figure 2: Beamplot of Unfocused Transmit Beam

Figure 3: Beamplot of Negatively Focused Transmit Beam
Figure 4: Beamplot of Focused Transmit Beam

As can be seen from the beam plots in Figures 2-4, the location of the focal point greatly effects the profile of the transmitted acoustic beam. In the case of the unfocused transmit seen in Figure 2, the beam consistently converges over all measured ranges. As the transducer is diffraction limited, the transmitted beam naturally focuses at the transition distance, which, given the size of the aperture and wavelength transmitted, is approximately 30 cm for this transducer.

The negatively focused transmit in Figure 3 also converges, albeit not as rapidly as the unfocused beam. Over the measured range, the negatively focused beam is consistently the broadest, but at the expense of ripple in the main lobe and reduced overall amplitude. For this case, the negative focus is analogous to placing a diverging
lens on the aperture. A negative focus corresponding to the transition distance reduces the self-focusing effect of the aperture.

For the focused case in Figure 4, the beam has already begun to converge at 3 cm and continues to narrow until the focus is reached at 7 cm. Beyond the focus, the beam diverges. The peak amplitude at each range increases to its peak at 7 cm and then decreases at further ranges as the beam spreads out.

The three transmit beams presented here were compared quantitatively in terms of average beam width, amplitude, and amplitude ripple over the 3 to 13 cm range. The -6 dB beam width averaged over all ranges was found to be $6.8 \pm 5.4^\circ$ for the focused transmit beam, $16.0^\circ \pm 8.3^\circ$ for the unfocused, and $22.6 \pm 11.1^\circ$ for the negatively focused case. The second metric is the average reduction in peak amplitude, at each range, with respect to the maximum amplitude at the 7 cm range for the focused transmit beam. For the focused case, this is a metric of variation in amplitude of the beam over 13 cm, whereas for the other two transmit schemes, this is a measurement of loss of signal with respect to a focused transmit beam. The average reduction was found to be $-8.6 \pm 5.0$ dB for the focused transmit, $-2.7 \pm 1.8$ dB for the unfocused, and $-13.3 \pm 1.5$ dB for the negatively focused beam. The third metric was the ripple in the main beam, or the difference in the maximum and minimum voltages measured in the main lobe of each transmit beam as defined by the -6 dB beamwidth. For the focused transmit, the ripple
was found to be $0.6 \pm 0.9$ dB, for the unfocused $2.7 \pm 0.2$ dB, and for the negatively focused $2.4 \pm 0.4$ dB.

Qualitative differences between the beam profiles can be seen in the beam images in Figure 5. These images show the same relationship between the beams as the measured beam plots. The focused beam converges at 7 cm, where the brightness in the image is at its maximum value. The unfocused beam converges, but does not reach its nature focus (i.e. the transition distance). The negatively focused beam converges as well, although not as quickly as the other two.

![Figure 5](image)

**Figure 5:** Speckle target images demonstrating the relative beam widths of each transmit scheme. (a) unfocused, (b) negative 30 cm focus, and (c) focused at 7 cm.

Images of the resolution target taken with different transmit configurations can be seen in Figure 6. The targets in each image are spaced 1, 2, 3, and 4 mm apart, left to right. For the focused transmit beam in Figure 6(a), all targets are clearly resolved. With this transmit configuration, the system has 1 mm or $0.8^\circ$ lateral resolution. For two other transmit configurations, unfocused in Figure 6(b) and negatively focused in Figure 6(c), the 2 mm spaced targets are clearly resolved whereas the 1 mm spaced targets are not.
The lateral resolution for these imaging configurations is 2 mm, or 1.6º. When increasing the number of image lines received per transmit to 32 to acquire at 1000 fps, the lateral resolution is unaffected, as seen in Figure 6(d). The range resolution was 1 mm in all cases.

![Figure 6: Images of a standard resolution target with (a) focused transmit, (b) unfocused transmit, (c) defocused transmit. The final image (d) was acquired at 1000 frames per second. Wires in this target are spaced 1, 2, 3, and 4 mm apart, left to right.](image)

Images of the tissue-mimicking phantom are shown in Figure 7. The void used to calculate the image contrast ratio can be seen in the upper left quadrant of each image.

The image contrast ratio was calculated to be 25.7 dB for the focused transmit, 11.5 dB for the unfocused, 10.9 dB for the negatively focused transmit. When imaging at 1000 fps, the image contrast ratio was 10.2 dB.

![Figure 7: Images of a tissue mimicking phantom with an anechoic void in the top-left quadrant. Images were acquired with (a) a focused transmit beam, (b) unfocused transmit beam, and (c) defocused transmit beam. The final image (d) was acquired at 1000 frames per second. The anechoic void is 0.5 cm in diameter.](image)
3.8 Discussion

For the high frame rate system presented, the beam characteristic of greatest importance was the average beam width. In this approach, the beam width required for imaging is determined by the receive resolution of the system and the number of parallel image lines desired or available for each transmit beam. As T5 has a hardware limitation of 32 received image lines per transmit and the transducer used has a Rayleigh resolution limit of 1.2° at 3.5 MHz, the ideal beam width for this application would be 16°, assuming that the image is spatially oversampled at 0.5° intervals. Both the unfocused and negatively focused transmit beams satisfy this criterion; however, the unfocused transmit beam narrows significantly with range resulting in uneven signal levels at deeper ranges.

The negative focus of 30 cm was chosen for this imaging scheme because, in diffractive optics, placing a point source at the secondary focus of a diverging lens will produce an optical beam that is a projection of the aperture. This may not be the best location of the negative focus because the resulting beam had an average width of 22°, or 8° wider than necessary. By combining apodization with a different negative focal length, a transmit beam that is confined to a narrower and more consistent width might be produced. This warrants further study. Implementing apodization on the T5 system is a goal for future work.
As the ripple in the main beam was found to be only 2.4 dB, this amount of variation in signal amplitude was deemed acceptable for preliminary clinical imaging due to the signal compression used in hardware processing and brightness transfer functions. Inspection of the beam images in Figure 5 evinces no perceivable difference in brightness laterally across the beams, reflecting the negligible effect of beam ripple on the resultant images. An apodization scheme to minimize ripple in the main beam remains a challenge to be solved in future work.

As seen in Figures 6 and 7, both lateral resolution and image contrast are reduced when using a broadened transmit beam. The overall resolution in the images, however, is not greatly effected as long as receive resolution is maintained. The loss of 13 to 15 dB in image contrast is significant. It should be noted that the phantom used presents a case more extreme than in typical cardiac imaging applications where low contrast regions, such as the cavity of the left ventricle, are larger than 0.5 cm in diameter. Increasing the transmit power would not result in an increase in image contrast in a linear system.

The transmit beam negatively focused at 30 cm was deemed the most appropriate transmit scheme for high frame rate imaging with the T5 system. The ripple was less than 3 dB across the main beam, and the beam was wider than necessary at ranges used in adult echocardiography. For all further high frame rate applications a negative 30 cm focus is the transmit scheme used.
4. Difference Imaging and Speckle Statistics

4.1 Background

Pulse-echo ultrasound imaging as used in adult echocardiography is a coherent imaging modality. One of the hallmarks of coherent imaging systems is the appearance of speckle in the images. Speckle is the mottled appearance that arises in an image due to the interference of reflected signals from a large number of scattering targets that are significantly smaller than the resolution of the imaging system. As speckle arises from an interference pattern and is not an intrinsic property of the target being imaged, it is often considered undesirable and much research has been performed to reduce the appearance of speckle in ultrasound images using methods such as spatial or frequency compounding [Magnin 1982, Trahey 1986]. Others have capitalized on the nature of speckle and used it for tracking moving objects via methods such as speckle or feature tracking [Bohs 1991, Bashford 1996, Kuo 2008]. The nature and description of speckle are outlined and extended to describe the speckle in high frame rate difference images.

4.2 Speckle description in B-Mode images

As speckle arises from an interference phenomenon the description of the brightness and size of speckle has been done using a statistical model based off early work in laser speckle [Goodman]. In B-mode images, it was found that the distribution of brightness values in a well-developed speckle pattern is accurately modeled with a Rayleigh distribution, or
where $x$ is the value of brightness at a given location and $u(x)$ is the unit step function.

Although the brightness values within the image are described statistically, for a stationary target the speckle pattern is deterministic, that is for a stationary target imaged over time images will remain identical and successive images can be subtracted from one another to result in an image containing only background noise. As measured in three dimensions by Bashford et al., the -6 dB extent of speckle both laterally and axially are proportional to the system resolution [Bashford 1995]. It was found that the -6 dB width of speckle was roughly twice the system resolution in each dimension when measured in three dimensions. Nearest neighboring peaks were found to be spaced at three times the transmitted wavelength apart in distance.

Of particular interest has been the behavior of individual or groups of speckles as they move through the image plane. The applications have ranged from the reduction of speckle appearance in an image through spatial or frequency compounding to the tracking of individual speckles via correlation search or other time domain methods. It has been found that speckles do remain correlated for limited distances as they move through the image plane, but the degree of correlation depends upon imaging rates relative to target velocity. For individual speckles undergoing pure lateral motion, speckles decorrelate to a correlation coefficient of 0.5 after a displacement of $0.2L$, where

$$p_B(x) = \frac{x}{\sigma^2} e^{-\frac{x^2}{2\sigma^2}} u(x)$$
$L$ is the aperture size in the lateral dimension [Trahey 1986]. For pure range motion, the decorrelation length is proportional to 50% of the effective pulse length in millimeters. Speckles that have decorrelated cannot be tracked reliably using current methods. In the case of speckle tracking, this enforces a minimum useful frame rate for a given target velocity.

In view of the decorrelation of speckle due to motion, higher frame rates will permit reliable tracking of high speed speckle targets. Under the assumption that 0.5 correlation is adequate, using a 20 mm aperture, a speckle target would have a correlation coefficient of 0.5 after 4 mm of lateral translation. At a frame rate of 30 images per second, this corresponds to a maximum trackable velocity of 12 cm/sec. When imaging at 1000 frames per second, this increases to a maximum trackable velocity of 4 m/sec. Even with an increased threshold of maximum trackable velocity, current methods are limited to tracking well developed and high contrast speckles. For increased imaging rates of interest is the brightness value in a given image pixel or image region as a speckle target or pattern moves across that pixel or region.

**4.3 Difference Imaging**

A major challenge in B mode images has been the detection of low echo amplitude blood flow in vessels surrounded by high echo amplitude tissue. High frame rate imaging permits enhancing contrast between fast moving targets and a slow moving background by subtracting consecutive images or groups of images. As
compared to Doppler, this is a time domain method first used in MTI radar systems and applied to one dimensional situation at Duke by Barnes and Thurstone [1971]. Coupled with high frame rate imaging, a simple time-domain processing method can be applied to ultrasound images. This method, difference imaging, takes the detected brightness values from two sequential images and subtracts on a pixel-by-pixel basis. The absolute value of the difference is stored, and these values are scan converted to generate a difference image. Initial observations of difference images of dynamic targets demonstrated an increase of contrast of high velocity targets and the suppression of low velocity and stationary targets regardless of the brightness of the targets in the B mode images.

With motion of speckle targets in B Mode images, it is expected that for a given location in the image field, speckle will remain correlated for a certain amount of translation, and after enough translation, the speckle pattern in a given region of the image field will be statistically independent from the original. With enough target motion in that image location there were will be two statistically independent speckle patterns at two different points in time. In the event that two statistically independent speckle patterns are subtracted from one another, the resultant distribution of brightness values would be expected to be the correlation of the original distributions, or

$$p_{DB}(x) = \int_{-\infty}^{\infty} \frac{\alpha}{\sigma^2} e^{\frac{-\alpha^2}{2\sigma^2}} \frac{\alpha - x}{\sigma^2} e^{-\frac{(\alpha - x)^2}{2\sigma^2}} u(\alpha - x) d\alpha$$
In addition to the resulting brightness values, the size and distance to nearest neighbor is also of interest. The final point of investigation is to find how much a speckle pattern must move before a statistically independent speckle pattern arises in a given location in the image field.

4.4 Methods

To investigate the behavior of speckle after subtraction, both simulations and B mode images of a tissue mimicking phantom were used. For the simulation of difference brightness distributions, Monte Carlo methods were employed. By subtracting 100,000 random values from two statistically independent Rayleigh distributions, a distribution of difference brightness values was generated. For comparison, two B mode images of statistically independent speckle patterns were subtracted, and histograms were generated for both the brightness values in the original B mode images and in the difference image.

For the remaining analysis of speckle in difference images, the T5 system was used to image a tissue mimicking phantom (CIRS Model 040GSE Multi Purpose Multi-Tissue Ultrasound Phantom). A 3.5 MHz, 96 element linear array with an active aperture of 21 mm in azimuth and 14 mm in elevation was used to generate all images. T5 was programmed to acquire a single B mode image of the phantom after an external manual trigger was pressed. Each acquired image had 160 image lines, with a single image line received for each transmit beam, with an 80º FOV and 0.5º angular spacing.
between image lines. The transducer was clamped to a high-resolution translational stage with the translation in the lateral direction, i.e. parallel to the transducer face. An image was acquired for every 10 µm of translation over a total displacement of 3.0 mm, for a total of 301 images. Four sets of images were acquired in this manner, changing the transmit focus between image sets. The transmit focal lengths were 4.0, 7.0, 10.0, and 12.0 cm. When changing focal length, the transducer was moved in range with respect to the phantom so that the same region of the target was located at the transmit focus regardless of focal length. After acquisition, a speckle region surrounding the transmit focus was chosen with size of 10 x 10 mm of the scan converted image. The size of the region was chosen so that multiple resolution cells were encompassed within the target area. For all frames, the difference was calculated, pixel by pixel, between the first frame and the Nth. The absolute value of the difference was taken for each pixel, and absolute values were summed over this region and normalized by the maximum total summed difference brightness for that set of images. Using the first and last images from the 7.0 cm transmit focus set, difference speckle size, amplitudes, and distance to neighbors was also investigated.

A similar process was repeated to investigate the behavior of targets moving purely in range. Four sets of images were acquired in the same manner as above, except with 10 µm increments of translation purely in the range direction. The transmit focal length was changed for each set of images, using 4.0, 7.0, 10.0, and 12.0 cm as the
transmit focus. A third set of images was acquired changing the pulse length, and thus the range resolution, between image sets. The transmit focus was maintained at 7.0 cm for the third set of images, with the transducer being actively driven with a 2, 4, 6, and 8 cycle pulse for each iteration. The difference brightness as a function of translation was calculated for each configuration in the same way as previously described, using a 10 x 10 mm region centered around the transmit focus.

For initial in vivo testing of high speed difference imaging, a 25 year old male was imaged using the T5 system. The same 3.5 MHz, 96 element linear array was used, and the system was scanning at 360 images per second. Images were acquired at 360 per second in an apical view. Synchronous EKG was acquired for reference.

4.5 Results

The technique presented for use with high frame rate imaging visually enhances contrast between slow and fast moving targets without loss of flow directionality by using speckle as a source of contrast. For two statistically independent speckle patterns, as would arise when a target is moving at high velocity with respect to imaging rates, the nature of speckle in difference images was investigated. The distributions of brightness values from the difference of two statistically independent speckle patterns generated via simulation can be seen in Figure 8 and from B mode images in Figure 9. The dashed lines superimposed are Gaussian distributions with mean of zero and standard deviation of 1.0 for the simulated data and standard deviation of 3800 for the
image data, which was the calculated standard deviation of the distribution. For the
distribution of brightness values in the B mode images, a Rayleigh distribution with $\sigma_R = 5000$ is superimposed to demonstrate deviation from the theoretical distribution of values. The mean value of the Rayleigh distribution is $1.25 \sigma_R$, and when the absolute value is applied to the difference distribution, the average is $0.73 \sigma_R$, yielding an average reduction in brightness of 40% due to subtraction.

Figure 8: Two original Rayleigh distributions (left) and the resulting difference brightness distribution (right). Both original distributions were generated with $\sigma_R=1.0$. The difference brightness distribution has a Gaussian fit superimposed (red dashed line) with $\mu = 0$ and $\sigma_C = 0.93$, calculated from the histogram.
Figure 9: Distributions from B-Mode (left) and DB-Mode (right) images. The original B-Mode distributions are closely approximated by a Rayleigh distribution with $\sigma_R = 5000$, as indicated by the dotted green line. The resulting Difference Brightness distribution is shown with a Gaussian curve overlaid (red dotted line), with $\mu_G = 0$ and $\sigma_G = 3800$

A region of 15 x 15 mm centered about the transmit focus was taken from the first and last images of the 7.0 cm transmit focus set with lateral translation to describe the speckle in difference images. Peaks in the difference image were qualified as being the maximum intensity pixel within a 0.5 mm radius, or 2 image pixels, in all directions. Image regions as well as the difference image of the region are shown in Figure 10.

There were a total of 65 speckles identified by the selection criterion. The distribution of speckle dimensions for the B mode image is shown Figure 11 and the distribution of sizes for the difference images is shown in Figure 12, where the width and height of speckles are defined by the full extent of the -6 dB brightness of the speckles. Of note is the system resolution in each dimension: 1.5 mm laterally and 0.44 mm in range. For the
lateral dimension, the average width was 1.56 mm and standard deviation 0.51 mm in the difference image, while speckles in the original B mode image were 3.9 mm wide on average with a standard deviation of 2.2 mm. In range, the average extent was 0.91 mm and standard deviation 0.31 mm in the difference image, while in the B mode image average axial extent was 2.2 mm with a standard deviation of 1.3 mm. Distance to nearest peak was calculated for all identified features, and the distributions can be seen in Figure 13. In the B mode image, average distance between peaks was 1.35 mm with standard deviation of 0.32 mm. The average distance between peaks was 1.28 mm, and standard deviation was 0.32 mm in the difference image.

**Figure 10**: Example image regions from (a) original position, (b) after 1.5 mm lateral translation, and (c) difference image generated from the two regions.
Figure 11: Distributions of -6 dB speckle sizes as measured from B mode images in lateral dimension (top) and axial dimension (bottom).
Figure 12: Distributions of -6 dB speckle sizes as measured from difference images in lateral dimension (top) and axial dimension (bottom).
Figure 13: Distance to nearest identified neighboring speckle peak for B mode image (top) and difference image (bottom).

The conditions of statistical independence of speckle patterns as they move was investigated. The average difference brightness of a speckle pattern in a given region of the image field as a function of lateral translation can be seen in Figure 14. Each curve represents a different transmit focal length, with the leftmost being the 4.0 cm transmit focus case, and each curve corresponding to increased focal length going left to right: 7.0 cm, 10.0 cm, and 12.0 cm. Of note, the knee of each curve, or the point at which the average difference brightness saturates, is roughly located at the theoretical Rayleigh resolution distance for the transducer used and the focal length. The theoretical
resolution for this array operating at 3.5 MHz is 0.8 mm for a focal length of 4 cm, 1.5 mm at 7 cm, 2.1 mm at 10 cm, and 2.5 mm at 12 cm.

Figure 14: Normalized Average Difference Brightness as a function of lateral translation is shown for 4 different transmit focus locations. The blue curve represents a 4 cm transmit focus, red 7 cm, green 10 cm, cyan 12 cm. Note that the transition from the linear region to saturation region occurs around the theoretical Rayleigh resolution of the system at each range. For example, for the 96 element 3.5 MHz array used, the lateral resolution at 7 cm transmit focus is 1.47 mm, and the theoretical Rayleigh resolution limits for each depth have been indicated on the x-axis.
Figure 15: Normalized Absolute Difference Brightness as a function of range (Axial Translation)

Normalized Absolute Difference Brightness as a function of axial translation is shown for four different transmit focal lengths. The blue curve represents a 4 cm transmit focus, red 7 cm, green 10 cm, and cyan 12 cm. Note that due to constraints of the experimental setup, there was a non-speckle target included in the image at 12 cm range. For axial translation, the transition from the linear to saturation regions is not dependent upon depth.

The average difference brightness as a function of axial translation is shown in Figures 15 and 16. Figure 15 shows the average difference brightness for different focal lengths. Since varying the length of the water path between the transducer and the phantom introduced reverberations that manifested as undesired artifacts in the images, the same portion of the phantom could not be imaged at different focal lengths. This also resulted in the inclusion of a point target at the 12 cm transmit focus, which
explains the rapid increase in difference brightness after only 10 µm of translation. The overall trend indicates that the difference brightness is not a function of transmit focus for axial motion.

Figure 16: Normalized Absolute Difference Brightness as a function of pulse length at a transmit focus of 7 cm is shown. The blue curve represents an active 2 cycle activation pulse, red 4 cycles, green 6 cycles, and cyan 8 cycles. The transition between linear and saturation regions occurs roughly at 1.5 time the theoretical pulse length, and the transition point scales linearly with increased pulse length.

Figure 16 shows the average difference brightness as a function of axial translation for various pulse lengths. The leftmost curve represents a 2 cycle transmit with increasing pulse length left to right: 4 cycles, 6 cycles, and 8 cycles. As the axial
resolution of a pulse echo system is half the pulse length, this corresponds to a theoretical resolution of 0.4 mm for a 2 cycle transmit, 0.8 mm for 4 cycles, 1.2 mm for 6 cycles, and 1.7 mm for 8 cycles. Note that the pulse lengths cited are the duration of the active excitation of the transducer. The pulse length in the medium is longer than the electrical excitation due to the finite bandwidth of the transducer as well as ringdown of the transducer after activation. As a result, the saturation point is roughly 1.5 times the theoretical or excitation pulse length.

A preliminary in vivo difference images of an adult heart is shown in Figure 17. On left is the standard B mode image and on right is the difference image generated from the image on left and the previous frame. The image was acquired during late diastole, as indicated by the EKG on bottom left. During the late diastolic filling period in normal hearts, the myocardium is almost completely stationary while the blood in the ventricle tortuously swirls around the ventricular cavity. In the difference image, the stationary myocardium is almost completely black (i.e. stationary), while the complex pattern of blood flowing in the ventricle can be seen. For comparison, in the standard B mode image, there is little to no visible blood signal in the ventricle while the myocardium is a very bright target.
Figure 17: Example of a high speed B mode (left) and difference (right) of an adult heart in the apical 4 chamber view.

4.6 Discussion

For difference imaging to be a viable visualization technique, of interest is how the brightness of a speckle in a given location in an image changes as the target moves as well as what amount of translation must occur before the speckle pattern in that location is statistically independent from the original. As seen in Figures 8 and 9, statistically independent speckle patterns follow the extension of the statistical model of B mode speckle. Various values of $\sigma_R$ were used for the Rayleigh distributions in simulations, and a linear relationship between the mode of the Rayleigh distribution ($\sigma_R$) and the standard deviation of the resulting Gaussian distribution ($\sigma_G$) was found: $\sigma_G = 0.93 \sigma_R$. 
As seen in Figures 15 and 16, after only one subtraction, a Gaussian approximation is appropriate for modeling the distribution of difference brightness values, both in simulation and in *in vitro* images. Thus, with sufficient translation of targets between images, difference images behave as expected given the extension of models of speckle in B mode images.

Comparison of speckle in B mode images to speckle in difference images yields interesting results. Where in three dimensions, speckle has a -6 dB width roughly proportional to twice the theoretical resolution of the system [Bashford 1995], in difference images the width is proportional the resolution. In range, the speckles as measured in 3D are roughly twice the theoretical resolution of the system; however, as can be seen from Figures 15 and 16, the speckle sizes are about 1.5 times the size of the theoretical. From inspection of Figure 10, difference imaging does sharpen the edges of the speckles and create higher contrast at the edges of features. As all images are linearly normalized in brightness, comparison of the changes in brightness in difference images are not relevant. When measuring distance to nearest speckles, both the average and standard deviation of peak to peak distances in difference images is comparable to that of speckle in B mode images. The average brightness within the speckle pattern is reduced by 40% due to subtraction. While this can be mitigated by changing the display transfer function during image display, subtraction results in a loss of 4.4 dB dynamic range from the original images.
The behavior of regions of speckle undergoing translation, as seen in Figures 14-16, shows that there are two distinct regions: a linear region and a saturation region. The transition between these two regions is found to be related to the theoretical resolution of the imaging system in the relevant direction. In the linear region, brightness and velocity are related in such a way that for slow moving targets a velocity could be calculated from the local changes in brightness values. In the saturation region, speckles are statistically independent at each point in successive images. This implies that this method is not angle dependent in the sense of Doppler techniques, but rather dependent on the system resolution, both spatially and temporally.

Initial *in vivo* studies demonstrated the utility of difference imaging immediately. In the case of late diastolic images when the myocardium is stationary in Figure 17, difference images almost entirely suppressed signal from the myocardium, and by increasing the contrast of the display transfer function low level blood signals became the dominant signal in the difference image. When difference images were played back in slow motion, the complex flow patterns of the blood within the ventricle were clearly visible.

In addition to the resolution dependency, the images will also depend on the scan geometry. In a sector scan, the resolution cell rotates as a function of angle in the image, i.e. the lateral dimension is always orthogonal to the beam direction. For most ultrasound systems, the range resolution is typically greater than lateral resolution, so
that the resolution parallel to the transducer face is greater on the sides of the image than it is in the center. As an example, if a tube parallel to the transducer face with laminar flow is imaged using a sector scan, the effective resolution cell on the edges of the image will be smaller in the flow direction. Because the speckle targets will become independent faster on the edges, a tube with uniform flow will appear brighter on the edges of the image than in the center of the image. This will not be the case for a linear scan, as the resolution is constant parallel and orthogonal to the transducer face. For a linear scan, this tube would have the same brightness along its entire length in the difference image; however, the distribution of brightness may change when the orientation of the tube changes with respect to the transducer.
5. Increasing Contrast and Continuity of Difference Images

5.1 Background

Difference imaging has been shown to be an effective method for increasing image contrast of rapidly moving targets while suppressing slowly moving or stationary targets. One obstacle faced when using this method is the low signal level of blood. As seen in the distribution of brightness values in Figures 8 and 9, when the target is moving rapidly, there is an average reduction of brightness by 40% due to subtraction. This may be overcome in some imaging situations, such as in the case of myocardial tissue, by changing the display transfer function. However, difference imaging inherently reduces the image dynamic range by 4.4 dB. Since this method is primarily aimed at visualizing low intensity targets such as blood within high intensity targets such as the liver or myocardium, it was necessary to investigate how this method may be used to increase the apparent contrast of moving targets.

As an example, consider the problem of imaging blood flow in the coronary arteries. Coronary arterial walls are highly reflective as opposed to the blood signal and the arteries move with the regional motion of heart. As seen in the previous chapter, even small translations will generate a signal in the difference image. Since the changes in brightness displayed in the difference image are proportional to the original brightness of the target, this presents a confounding situation where a bright target moving slowly will appear the same as a low intensity target moving very rapidly. As
seen previously, motion of only 10% of one resolution cell will result in an average signal that is 10% of the original brightness in the B mode image. A target, such as the arterial lumen that is at least 10 times brighter than the blood within the coronary vascular system, only need to move a tenth of a resolution cell, or 120 µm at 7 cm range, before the signal is equal to that of any blood moving faster than a resolution cell per frame. Provided there is a difference in the rate of motion of the coronary arterial walls as compared to the blood velocity, difference imaging may reduce or even eliminate this amplitude problem. To further increase the visibility of the flowing blood, the possibility of summing several high speed difference image frames was investigated. Also, the effect of sharpening of speckles seen earlier is not intuitive for those accustomed to viewing B mode images, so a spatial smoothing of the difference speckle pattern is also desired. If speckle regions are statistically independent in a series of B mode images, the patterns in the difference images generated from these will also be statistically independent. This would allow for the summation of several difference images.

5.2 Methods

To demonstrate the effect of summing multiple difference images, Monte Carlo methods were used. One million differences were taken from two statistically independent Rayleigh distributions, and a histogram of the absolute values was generated. This process was repeated one hundred times, and absolute differences were
cumulatively summed as would be done in the case of difference images. For all histograms beyond the first, brightness values were normalized to a mean value of 1.0, as the display for these images has a limited dynamic range, and brightness values will be normalized for display.

Six B mode images of speckle regions in a tissue mimicking phantom were acquired using the same system configuration as in Chapter 3. Between each image, the transducer was translated 3 mm laterally to insure that statistically independent speckle patterns were imaged. From these, five difference images were generated. Successive summation of difference images was performed, and histograms of image brightness was generated for \( N = 1-5 \) summed difference images. All histograms were linearly normalized to a mean of 1.0.

To demonstrate this approach \textit{in vivo}, a normal human volunteer was imaged at high frame rates with the T5 system, and summed difference images were generated. Images were obtained with an 8 MHz, 96 element 1D array at 1038 frames per second in the neck showing both the jugular and carotid flow simultaneously. Images of a liver in a normal human volunteer were obtained at 726 frames per second using a 3.5 MHz, 96 element array. Finally, a 3D flow phantom (Gammex 420) was scanned using a 1500 element 2D array with T5 programmed to scan a 30° by 30° pyramidal volume at 183 volumes per second at a maximum range of 10 cm. Depending on the nature of the
motion in each target, varying amount of difference images were summed to increase the contrast of moving targets as well as spatially smooth the visualized flow profile.

5.3 Results

The histograms of brightness values for summed difference images can be seen in Figure 18. The topmost histogram is for a single difference image, with each subsequent histogram representing 2, 5, 10, 25, 50, and 100 difference images summed. For a single difference image, the result is as seen in the previous chapter except that all values are positive. For increasing number of images summed, the mean value of brightness increased linearly, and the variance of the distributions increased linearly as expected. As these brightness values are ultimately displayed with limited image dynamic range, all histograms were linearly normalized to a mean of 1.0 for comparison. The resulting effect as it would be seen in a summed difference image is less variation in image brightness for moving targets in the image, with a significant reduction in dark, or zero brightness, areas. Since the mean and variance both scale linearly, this results in a reduction of variation of brightness values across the moving target by a factor of $1/\sqrt{N}$, where $N$ is the number of images summed.

The brightness distribution for summed difference images can be seen in Figure 19. All values are normalized so that the mean of each distribution is 1.0. Before normalization, the means and variance of each distribution scale linearly with number of images summed. After normalization, the stand deviation values were reduced by a
factor of $1/\sqrt{N}$, where $N$ was the number of images summed. For the target being imaged, 1.5 cm was the maximum amount of translation possible before deterministic targets were included in the sample region.

Both standard B mode and summed difference images of an adult human neck can be seen in Figure 19. The left image is the summed difference image, and the standard B mode is on right. The jugular vein is the upper vessel and the carotid artery is the lower. The patient is oriented such that the right side of the image is in the cranial direction. Images were acquired at 1038 per second, and a total of 20 difference images were summed to generate the image on the left, a temporal window of 19.3 msec. Longer summation windows were possible; however, it was observed that flow in the jugular vein reversed (i.e. caudal to cranial) subsequent to atrial contraction as indicated in the EKG. Longer time windows obscured the visualization of this phenomenon.
Figure 18: Summed difference brightness distributions for $N = 1, 2, 5, 10, 25, 50,$ and 100 images. Each number of summed images has been normalized so that the mean brightness for each distribution is 1.0. Note the sharpening of each distribution as the number of images summed increases.
Figure 19: Summed difference brightness distribution for $N = 1$-$5$ summed difference images. All distributions are normalized to an average brightness of 1.0. Note the reduction in the spread of brightness values as the number of summed images increases.
Figure 20: Integrated difference image (left) and B-Mode image of the jugular vein (top) and carotid artery (bottom) of a healthy 29 yo male on right. Images were acquired at 1038 per second, and 20 DB-Mode images (19.3 msec) were summed to generate the right hand sector. Flow direction is visually detectible in the two vessels during slowed playback.
Figure 21: Integrated difference image (left) and B-Mode image of the liver of a healthy 29 yo male (right). Images were acquired at 726 per second, and 50 DB-Mode images (68.9 msec) were summed to generate the right hand sector. Note the increased contrast of the vasculature in the integrated difference image, as well as the appearance of small vessels branching off of the main trunk. The small vessels are measured 1 mm in diameter in the image. They may be smaller.

The summed difference image and B mode of a human liver can be seen in Figure 20. Images were acquired at 726 per second, and a total of 50 differences were summed to generate the left hand sector, for a time window of 68.9 msec. Note the increased contrast of the vasculature in the difference image as well as the appearance of vessels otherwise obscured in the original B mode image. As measured in the image these smallest vessels visualized in the image are approximately 1 mm in diameter. However, the actual vessels may be smaller in size as 1 mm is approximately the
resolution of the system. Of note was the pulsatile nature of the blood flow in the vessels which was obscured with temporal summation windows longer than 100 msec.

The summed difference volume and standard volume rendering of a self-contained ultrasound flow phantom is shown in Figure 22. If the standard volume rendering is oriented properly, the flow channel can be seen but only at the intersection of the flow channel and the edge of the pyramid. In the difference volume, the static regions of the phantom are suppressed and the flow channel can be visualized regardless of the orientation of the pyramid or view direction. For the summed difference volume on left, 25 difference volumes were summed for a total time window of 136.6 msec time window. With the current system and display software, acquisition, rendering, and temporal processing of the difference volume can be performed live in real time.
Figure 22: Integrated difference volume (left) and standard volume rendering (right) and of a self-contained flow phantom. If the volume is manipulated properly, the flow channel can only be visualized in the standard volume rendering at the edge of the sector. In the difference volume, the static media around the flow channel is suppressed, and the moving fluid can be seen in all possible views.

5.4 Discussion

With initial observations of in vivo differences images, it was noted that rapidly moving targets did increase in contrast, yet many motion artifacts manifested in images that obscured the visibility of low intensity, high velocity targets. The sources of these artifacts were either global motion in the image field which enhanced the contrast in the entire image field or the small motion of significantly brighter targets that led to the obfuscation of much lower signals. Global motion arose due to many sources, exacerbated by the manner of clinical scanning. In standard clinical ultrasound imaging, the transducer is hand-held and the patient is often lying in an unnatural manner such as slightly inclined to one side. Sources of motion include patient respiration, motion of
the transducer, operator respiration, and even operator pulse rate. As the sensitivity of
difference imaging is very high to any motion, these artifacts may be difficult to avoid
on a frame by frame basis. By summing multiple difference images, these effects are
reduced.

As demonstrated in Figure 18, increasing the number of difference images
summed has great benefit for the visualization of rapidly moving targets. The overall
brightness increases linearly with the number of summed images, and the variance of
brightness values. The decreased variance of brightness as well as the shift away from
lower brightness values creates a more uniform appearance in space, thus the
appearance of structured vasculature such as seen in Figures 20 and 21.

As predicted from the statistical model, two difference images generated from
three statistically independent speckle patterns will yield two statistically independent
difference images. Summation of multiple difference images will result in a linear
increase in both average brightness and variance of brightness, as predicted by the
Central Limit Theorem. As ultrasound images are displayed with limited dynamic
range, linear compression of summed difference images will result in a $1/\sqrt{N}$ reduction
in the standard deviation of brightness values in the summed difference image. This
yields a more uniform appearance of moving targets in the summed difference images
and a reduction of low brightness values for moving targets.
For low intensity scatterers such as blood, the decreased variance also results in much higher visibility due to the entire vessel being higher contrast rather than a mottled collection of moving speckles. This effect presents itself for motion that is above the saturation value of difference images for the entire duration of the summation window. As noted previously, direction of flow will also be preserved and visible if the direction is uniform for the duration of the summation window.

If the motion in the images is constant and of infinite duration, an infinite number of images could be summed and a uniform flow field visualized. However, in medical imaging, motion and flows are either transient or cyclic in nature, thus the time window over which summation can be performed without the loss of information is limited by the physiology of the organ being imaged. For peripheral vasculature, patient heart rate becomes a limiting factor. If the summation window is as long as a typical heartbeat, 0.5 second in pediatrics or 1.0 second in adults, vasculature will be clearly visualized but the details of blood flow over the cardiac cycle will not be visible. For adult imaging, summing windows greater than 50 msec have the potential to obscure complex flow patterns that arise due to vascular obstruction or cardiac abnormalities. It was observed that while imaging an adult neck as the volunteer performed a Valsalva maneuver, complex, tortuous flow patterns were visible in the jugular vein with the summation window was 20 msec, but the flow pattern became significantly more uniform when the window was increased to 50 msec or greater.
Additionally, for natural variations in blood flow, such as the reversal of venous flow subsequent to atrial contraction, 20 msec was the maximum duration of summation before flow reversal was obscured. When imaging at 1000 frames per second, this limits summation to 20 frames, yielding a maximum decrease in variance by a factor of 4.5.

For imaging of vasculature in other organs, such as the liver or brain, where flow is pulsatile yet not as variable, longer summation windows may be used. For the hepatic images shown in Figure 21, a summation window of almost 70 msec was used, and the pulsatile nature of flow was still clearly appreciable in the images. Longer windows may be used; however other sources of motion, primarily patient or operator, become more common and potentially will obscure flow details of interest. In effect, summation acts a filter on the duration of motion, suppressing lower duration motion while increasing the contrast of and reducing variations in brightness for longer duration flow or motion.

For cardiac imaging the possibility of summing difference images becomes significantly more complex, depending upon the target of interest. For myocardial motion, myocardial targets are inherently high intensity and will remain bright in difference images. Summing difference images reduces the effective temporal resolution of the images, and beyond 5-10 msec, the fine temporal details of high velocity events such as the electrical depolarization of the myocardium and the resultant propagation of the onset of mechanical contraction will be lost. Since signal levels for myocardial
targets are relatively high, summation is unnecessary and often detrimental to the visualization of rapid events. For imaging of blood in the heart and coronary system, however, the suppression of myocardial targets is desirable. As mentioned, the heart is a very dynamic organ and is in motion for approximately two thirds of the cardiac cycle. This presents difficulties for visualizing blood flow both within the ventricle and in the coronary system. As the signal from blood is typically an order of magnitude lower than that of myocardial targets, long summation windows are desired for the visualization of coronary blood flow. For an average adult with a heart rate of 60 beats per minute, the myocardium will remain relatively stationary for 300 msec at the end of diastole. Peak coronary flow occurs at the end of diastole, with peak flow rates measured up to 1 m/sec [Ofilli 1993]. Ideally a window of 200 msec could be used to image coronary flow, giving a signal improvement by a factor of 14 imaging at 1000 frames per second; however, the duration of peak blood flow is very brief, occurring in under 50 msec overlapping with myocardial systole. Imaging at faster rates holds the possibility of more images being summed during the window and increasing the visibility of coronary blood flow in a non-invasive imaging method without the use of an injectable contrast agent. Additional complications with imaging coronary flow using two dimensional imaging techniques arise due to the heart constantly moving while the coronary vasculature is maximally 5 mm in diameter. Due to motion of the heart, the coronary vessels will constantly move in and out of the imaging plane which
indicates the need to move to three dimensional imaging for the potential of non-invasive, non-contrast imaging of coronary vasculature.

5.5 Conclusions

With high enough imaging rates, 1000 per second or above, difference images in vivo have been shown to increase the contrast of high velocity, low intensity targets moving in a high intensity stationary or slow moving background. Depending on the physiology of the target organ, multiple difference images may be summed to yield higher contrast of moving targets, more uniform brightness distribution of moving targets which results in spatial smoothing of the flow field, and for suppression of shorter duration events that are not of interest. As this visualization method can be performed retrospectively on stored B mode images, various parameters such as the summation window can be varied to find the ideal settings for a given set of images. Extension of this method to include more appropriate temporal filters is a plan for future work.
6. Discussions and Conclusions

6.1 System Configuration for High Speed Imaging

For the given capabilities of the T5 ultrasound scanner, a transmit-receive configuration was found that was an adequate compromise between maximizing the usage of available receive parallelism while maintaining a beam that is of consistent amplitude across the image field with minimal loss of signal. As increasing receive parallelism necessitates a widened transmit beam, and thus a reduction in overall amplitude across the wave front, other methods such as multiple simultaneous transmit beams remain an interest in future studies. Additionally, as high frame rate imaging is extended to volumetric imaging, a more thorough understanding of the interactions between the tradeoffs in beam width, amplitude, and ripple are desirable.

6.2 Speckle Statistics

The description of speckle in B mode images been known and verified over the years. The behavior the brightness in a given image location when a speckle target moves through that location follows the natural extension of that statistical description. As expected, for a certain amount of target movement, the brightness changes in an image pixel is linearly related to the amount of translation of the target as well as the original brightness of the target in the B mode image. The point at which the brightness within the given image pixel is no longer related to translation, or the saturation point, is directly related to the diffraction limited system resolution laterally; specifically, once
the speckle target has moved by one resolution cell, the difference in brightness between sequential frames becomes a statistical phenomenon. In range, the saturation point is proportional to the excitation pulse length. Due to this behavior, two successive images may be subtracted and the absolute value of the difference displayed as an image as a method of contrast enhancement for rapidly moving speckle targets and contrast reduction for slow moving or stationary targets. As this method is related to system resolution, it is not angle-dependent in the sense of Doppler methods and can readily be extended from two to three dimensions. If a truly angle independent method of flow visualization is desired, the system imaging system could be configured to have a spherically symmetric spatial resolution, eliminating all angle dependence of difference imaging.

6.3 In vivo applications

For the visualization of rapidly moving, low intensity targets like blood, a single difference image can increase contrast for adequate visualization of flow; however, overall signal is inherently reduced by a factor of 40% due to subtraction in addition to the spatial sharpening of the speckles in the difference image. This loss in signal and increase in mottled appearance can be overcome by summing multiple difference images. Summation provides the additional advantage of reducing the variance of brightness values in the image after the display transfer function has been applied and reducing the number of low brightness pixels within the flow field or moving target.
The benefit of summation increases by a factor of \( \sqrt{N} \), and since there are physiologic temporal limitations to the useful time window for summation, the target organ must be considered when determining the duration of the time window. The spatial smoothing of the flow field due to temporal processing is an interesting result.

### 6.4 Directions for Future Work

#### 6.4.1 Beam formation for high speed volumetric imaging

The design and construction of the next generation of volumetric ultrasound scanners for high volume rate scanners will present many difficulties. Of great importance will be the transmit scheme employed to capitalize on the increased parallelism that is required to acquire volumetric images at rates of 200-400 per second or greater. Extension of methods known to be effective in two dimensions can be extended to three dimensions. Other methods such as multiple parallel transmit directions need to be revisit for three dimensional imaging, and other methods such as apodization also tested.

#### 6.4.2 Further exploration of temporal processing techniques

It has been shown that by applying simple time domain processing to B mode images, low velocity, high contrast targets can be suppressed while the contrast of high velocity targets can be increased. Furthermore, by applying time domain filters to the difference images, spatial variation in brightness can be reduced to produced higher contrast images that not only highlight flow more clearly but also preserve flow
direction. In its current implementation, the sensitivity of this method is dependent upon scan geometry and system resolution. One possible direction for future work is to configure the imaging system to have spherically symmetric resolution to have a completely angle independent method of flow detection and visualization.

Of more practical importance in future work is to further refine filtering functions to target the visualization of blood flow in the coronary vasculature. More complex time domain filters can be designed to specifically remove myocardial motion before the nonlinear step of the absolute value during the generation of difference images and need to be explored further. Additionally, the effects of the spatial distribution of targets achieved by applying filters after the difference images have been generated need to be investigated. All of the above need extension to three dimensions to see their potential of temporal processing for the detection and visualization of coronary flow.

6.5 Conclusions

This work has shown that with high frame imaging, time domain processing can be used to both increase the overall contrast and reduce the speckle appearance of moving targets in ultrasound images. Specifically, it was proven that:

A) With the T5 system, high images rates are feasible without significant reduction of image resolution or signal to noise.
B) Statistical targets, or speckle, have a pixel-by-pixel linear relationship between successive image frames between changes in brightness and amount of translation up to the diffraction limited resolution in azimuth and proportional to the pulse length in range. Then, brightness within a given pixel is statistically independent from the brightness with the same pixel in temporally subsequent images.

C) The point at which the brightness in a given pixel from frame to frame is statistically independent in a given pixel from frame to frame is directly related to the resolution of the configured ultrasound system.

D) Subtracted images of independent speckle patterns are statistically independent with each other and summation of such images within physiological constraints in time will further increase the contrast of blood with respect to vessel walls and surrounding tissue.

E) Within physiologic constraints, time domain processing methods can be used to increase image contrast of rapidly moving targets and suppress the appearance of slowly moving, transiently moving, or stationary targets.
References


Biography

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Born: September 5, 1984, Nashville, TN, USA

Degrees Earned:

BSE, Biomedical Engineering, Duke University, December 2006

Published Articles:


Patents:


### Honors:

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<td>2009</td>
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<td>2012</td>
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