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(54) **SYSTEM AND METHOD FOR PHASE INVERSION ULTRASONIC IMAGING**

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(22) Filed: **Aug. 21, 2002**

(51) **Int. Cl.**
A61B 8/00   (2006.01)
(52) **U.S. Cl.** **600/443**
(58) **Field of Classification Search** 600/437, 600/443, 447, 454–458; 73/625–626
See application file for complete search history.

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(57) **ABSTRACT**

Ultrasonic imaging utilizing multiple sets of transmit pulses differing in amplitude, frequency, phase, and/or pulse width is disclosed. One embodiment has phase differences between the k transmit signal as

$$\frac{360}{k} \text{ degrees}$$

providing for constructive interference of the kth order harmonic pulse, while an amplitude modulation of each transmit profile is constant between sets. These sets of pulses are transmitted into media of interest and received echoes from these pulses are combined to form art averaged signal. The averaged pulses represent the net common mode signal received from each of the transmit sets. This combined signal set is used to reconstruct an ultrasound image based on broad beam reconstruction methodology.

10 Claims, 6 Drawing Sheets
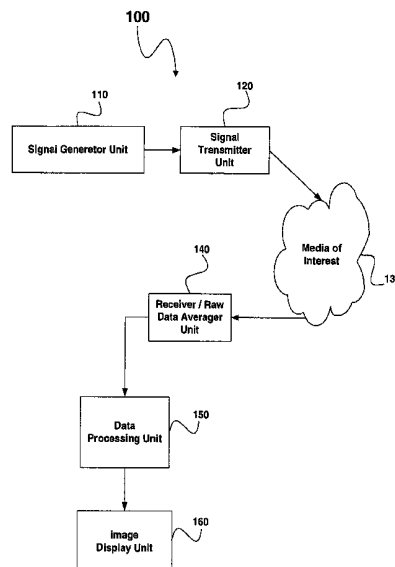
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100

110

| Signal Generator Unit |

120

| Signal Transmitter Unit |

Media of Interest

130

140

| Receiver / Raw Data Averager Unit |

150

| Data Processing Unit |

160

| Image Display Unit |

FIG. 1

$$A(n,t) \cdot e^{j[\varpi_o(t,n)t+\theta_i+\Theta(n)]}$$

to 120

$$A(n,t)$$

$$e^{j[\varpi_o(t,n)t+\theta_i+\Theta(n)]}$$

110

FIG. 2

from
110

310 — Delay
(Optional)

320 ~ Channel Gain
(Optional)

330 ~ Power
Amplifier

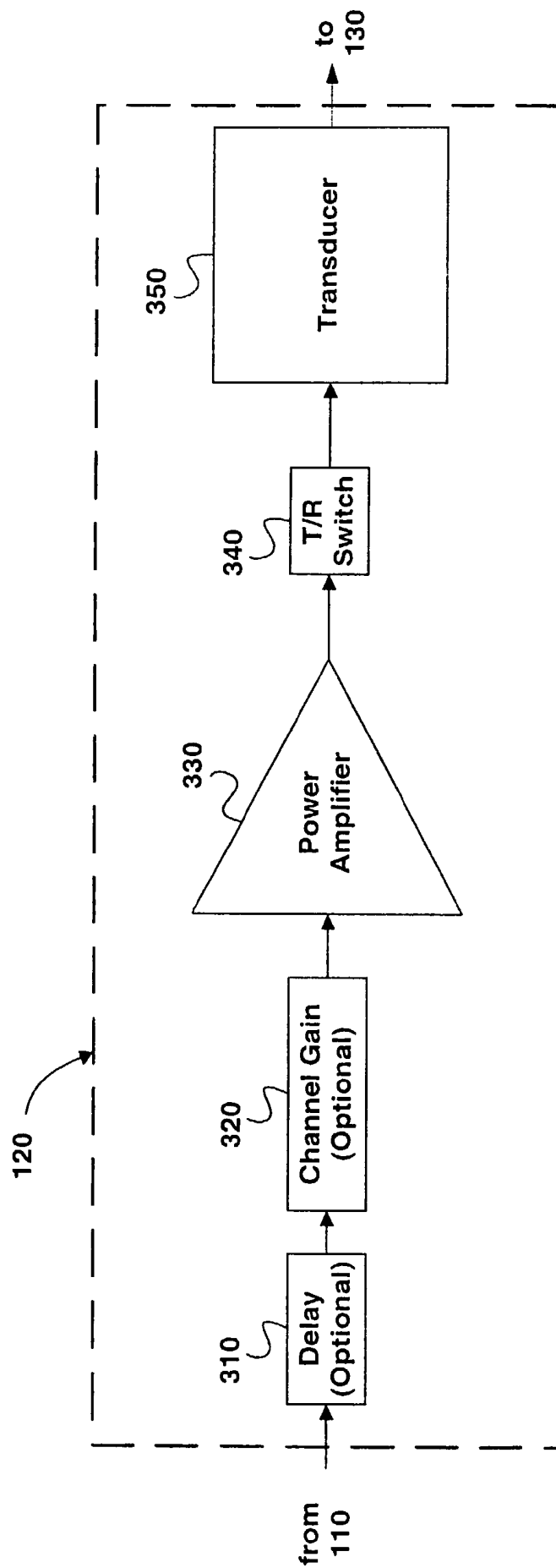340 ~ T/R
Switch

350 ~ Transducer
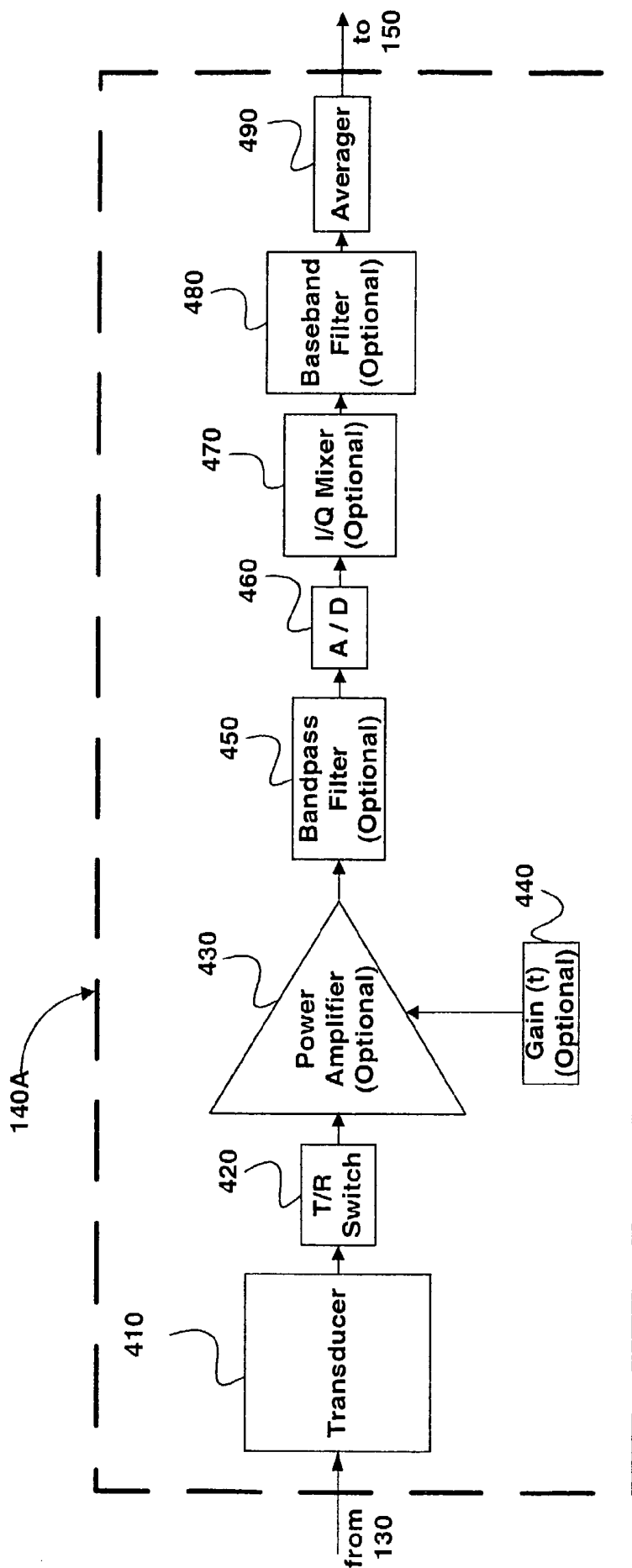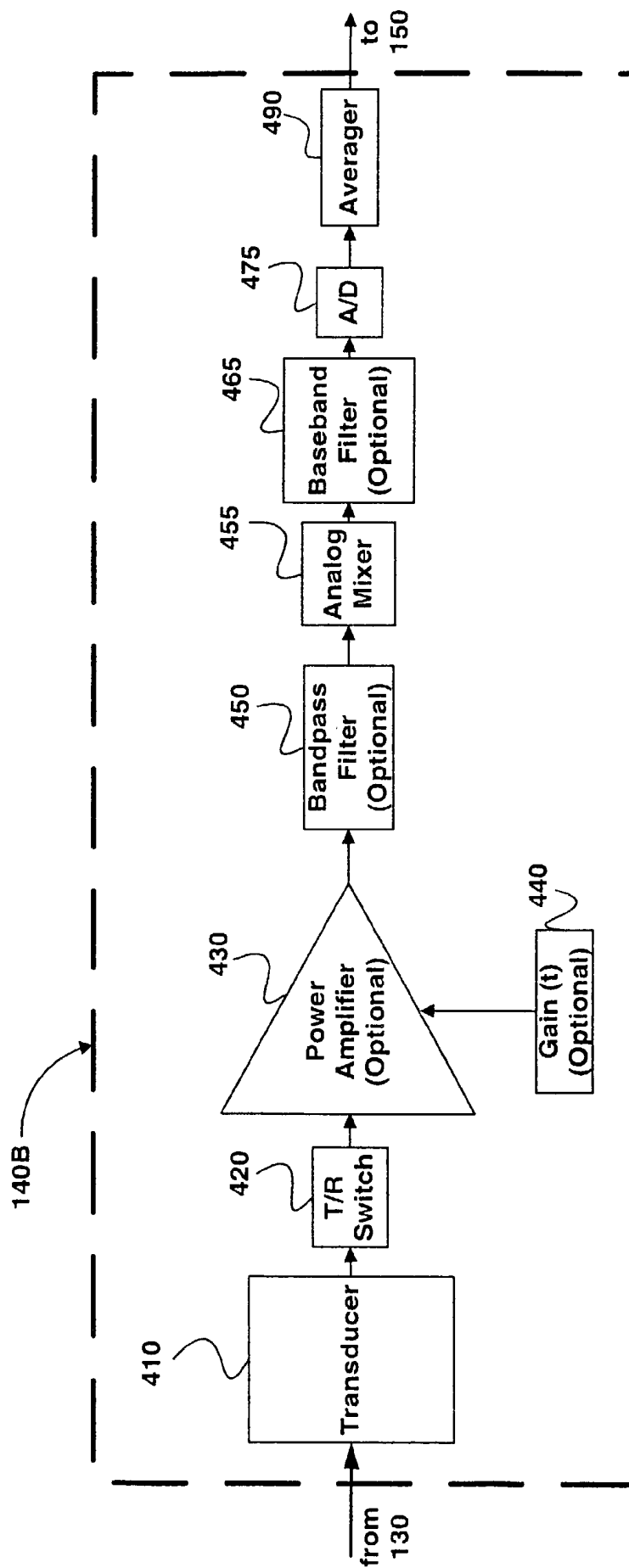
120

to
130

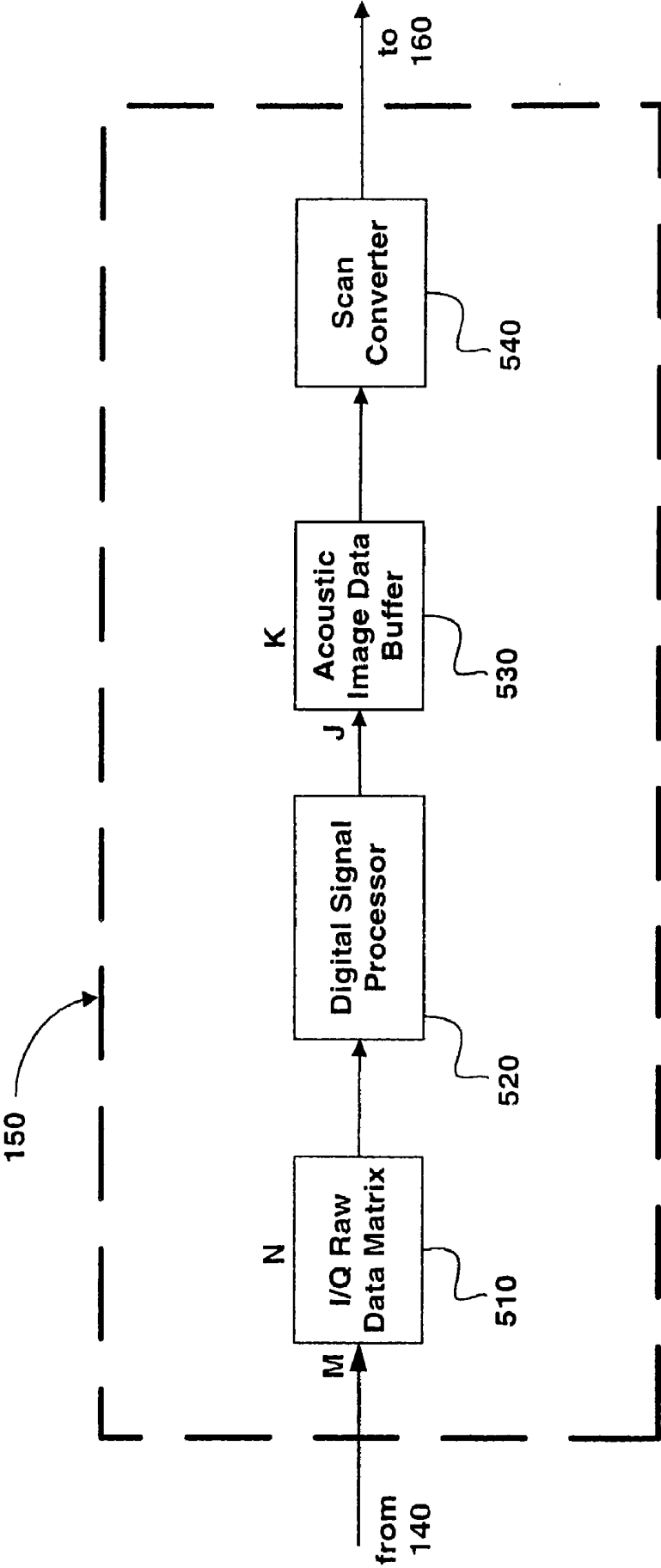**FIG. 3**

FIG. 4A

FIG. 4B

FIG. 5

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SYSTEM AND METHOD FOR PHASE INVERSION ULTRASONIC IMAGING

CROSS-REFERENCE TO RELATED APPLICATIONS

This application is a divisional and claims the priority benefit of U.S. patent application Ser. No. 09/872,541, filed May 31, 2001, and entitled "System for Phase Inversion Ultrasonic Imaging," which is now U.S. Pat. No. 6,866,631, the disclosure of which is incorporated herein by reference. This application is related to U.S. patent application Ser. No. 10/772,926 filed Feb. 4, 2004 and also entitled "System for Phase Inversion Ultrasonic Imaging."

BACKGROUND OF THE INVENTION

1. Field of the Invention

The present invention relates to ultrasound imaging, and more particularly, to a method and apparatus for improving and enhancing ultrasound images.

2. Description of the Background Art

Ultrasonic imaging is frequently used for a variety of diagnostic procedures due to its non-invasive nature, low cost, and fast response time. These qualities are especially true in medical fields where the added benefit is reducing or eliminating a patient's exposure to radiation. Typically, ultrasound imaging is accomplished by 1) generating and directing an ultrasonic beam into media under investigation; and 2) observing any resulting waves that are reflected back from dissimilar tissues and tissue boundaries within that area. The resulting waves are received as signals. These received signals are then post-processed and imaged on a screen by plotting a spot whose intensity is proportional to the amplitude of a reflected beam from a given location. Determination of location is based upon a known transmission and re-radiation rate after the ultrasonic wave is pulsed into the media under investigation.

Typically, an ultrasonic signal transmitted into the media under investigation includes a burst of sinusoidal waves of a given waveform. These sinusoidal waves are applied to a transducer and form a transmitted signal. The transmitted signal is typically in the range of 40 kHz to 50 MHz, but more commonly, in the range of 40 kHz to 1 MHz. As the transmitted signal interacts with tissue layers and boundaries between layers, the ultrasonic signal is modified by being scattered, resonated, attenuated, reflected, or transmitted.

Media under investigation are often a non-linear media such as those commonly found in the human body. Non-linear media produce harmonic frequencies in the echoed signals. These additional frequency components continue to re-radiate through and, in turn, reflect off or interact with other structures. A portion of the reflected (or echoed) signals propagates back to a receiving transducer.

Fundamental and harmonic frequencies that are impinged upon a receiving transducer includes the full signal, which must then be further processed to eliminate noise and extraneous components. The receiving transducer may be the same as a transmitting transducer, or can be completely independent. When the same transducer is used, a transmit/ receive (T/R) switch connects the transducer to either the transmitter electronics or the receiver post-processing electronics. The receiving transducer accepts the echo signal plus any generated noise and furnishes these to a portion of the post-processing electronics known as a beam former. Beam formers reject noise and have either an adaptive or fixed configuration. Adaptive beam formers are designed to

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reject variable directional noise sources by monitoring the noise field and adjusting internal parameters to minimize the background noise. Fixed beam formers are designed to reject isotropic noise and take advantage of the directional property of the reflected signal.

Ultimately, ultrasonic images of the human body are a product of harmonic imaging. Harmonic imaging is generally associated with either visualization of tissue boundaries and densities of different media, or imaging contrast agents at harmonic frequencies. Contrast agents are typically fluid filled micro-spheres that resonate at ultrasonic frequencies. Such agents are injected into the blood stream and are carried to various parts of the body. Once these agents are pulsed at ultrasonic frequencies, harmonic echo-locator signals are generated due to the resonance produced within the micro-spheres.

While ultrasonic procedures have a distinct number of advantages over other types of diagnostic techniques, prior art methods and systems have noise problems that make it difficult to determine the exact location and proper interpretation of the received signal. Various forms of averaging techniques have been employed to reduce the noise, but averaging alone is ineffective in locating images of interest between tissues with similar densities. (Echoed signals from tissues with similar densities will indicate a uniform mass with indistinct boundaries. Averaging won't help in this situation.) These interpretation difficulties are exacerbated by the fact that many tissues in the human body have similar densities. Therefore, a method and system are needed that can effectively overcome the stated difficulties while not negating the positive benefits of ultrasound imaging systems in general.

SUMMARY OF THE INVENTION

The present invention is a system and method for generating enhanced ultrasonic images. The invention utilizes multiple ultrasonic pulses that are transmitted in an alternating fashion into media of interest. These media being imaged may be a human body or some other linear and/or non-linear media. The ultrasonic pulses are modulated in a way that may vary in amplitude, frequency, phase, or pulse width. Each set of ultrasonic pulses is out-of-phase with other ultrasonic pulses by

$$\frac{360°}{k},$$

where k is the number of pulse sets in the pulse sequence for a given transducer element number, n. An out-of-phase condition is a property when waveforms are of the same frequency but do not have corresponding intensity values at the same instant. The echo signals generated by the non-linear media interacting with these out-of-phase signals are measured and appropriately combined.

The present invention is based on the observation that many types of media scatter sound in a non-linear manner. With an ultrasonic imaging system based on linear-scattering media, the return signal is a time-shifted, amplitude-scaled version of the incident signal. Non-linear scattering media produce signals that cannot be produced by simple time-shifts, scaling, or summation of the signal incident to a scattering site. The phase of an ultrasound wave reflected from the boundary of a non-linear medium is altered in a manner that depends on the phase of the incident sound

pulse. For example, consider the special case of two ultrasound pulses (k=2), where the phase difference between the two transmitted sound pulses differ by

$$\frac{360°}{k} = \frac{360°}{2} \text{ or } 180 \text{ degrees.}$$

If the scattering site were purely linear then the received signal from each of the transmitted pulses would be the inverse of each other. These inverse signals, when averaged, would have a sum of zero. If, however, there are signals generated from a non-linear process within the media, then these signals will not be the inverse of each other and, consequently, will not sum to zero.

This non-linear property can be used to construct a system that looks at the non-linear regions within given media. For example, in one embodiment of this system, an image area is formed from three different sets of transmitted signals each differing in phase by 120 degrees. The linear reflections generated by these k=3 sets of excitation pulses will cancel each other, while the kth order non-linear components will not. This pulse cancellation allows an averaged set of raw received data, F(n,t), to be produced. A three-dimensional averaged set of raw received data, F(n,m,t), may also be produced where m is an element from a two-dimensional, or n×m transducer array. The data functionality is dependent on the channel (or transducer) number and time. This single set of averaged data can then be used to re-construct an image area. The image area reconstructed would represent the information generated from the third, or k^{th}, harmonic, generated from the scattering sites within the media.

BRIEF DESCRIPTION OF DRAWINGS

FIG. **1** is a schematic block diagram of an embodiment of an ultrasound imaging system using the present invention;

FIG. **2** shows an exemplary modification of one of a potential plurality of waveforms being modified;

FIG. **3** shows an embodiment of a signal transmitter unit prior to signal delivery to media of interest;

FIG. **4A** shows an embodiment of a receiver and raw data averager unit;

FIG. **4B** shows an alternative embodiment of a receiver and raw data averager unit; and

FIG. **5** shows an embodiment of a data processing unit.

DETAILED DESCRIPTION OF THE INVENTION

The present invention relates to the field of ultrasound imaging. This invention utilizes broad beam technology to perform image extraction of the non-linear elements of media under Investigation. These media will hereinafter be referred to as media of interest. Broad beam technology defines an area under investigation at a given point in time, which is in contrast with a system utilizing a focused beam.

FIG. **1** is a block diagram of an embodiment of an ultrasound imaging system using the present invention. Imaging system **100** includes at least one signal generator unit **110**, at least one signal transmitter unit **120**, media of interest **130** to be imaged, at least one receiver and raw data averager unit **140** to capture signals received from the media of interest **130**, and a data processing unit **150** for taking the averaged received signals and producing an area of image formation on an image display unit **160**.

A signal generator unit **110** drives circuitry for a signal transmitter unit **120**. The signal transmitter unit is shown and described in greater detail in FIG. **3**.

A signal transmitter unit **120** transmits pulsed sets of ultrasonic energy into the media of interest **130**. Echoes received from the media of interest **130** are stored in the receiver and raw data averager unit **140**. Subsequent out-of-phase signals from the signal generator unit **110** pass through the signal transmitter unit **120**, and are converted into pulsed sets of ultrasonic energy that travel to the media of interest **130**. The media of interest **130** modifies the pulsed sets of ultrasonic energy. These modified pulsed sets of ultrasonic energy are received and averaged by the receiver and raw data averager unit **140**. Data from these received pulsed sets are averaged in a data set as a function of channel number, n, and time, t. The averaged data sets are processed by the data processing unit **150** and displayed on an image display unit **160**.

FIG. **2** shows an exemplary modification of one of a potential plurality of waveforms being modified. This FIG. **2** example demonstrates how a signal generator unit **110** may modify a generated signal. For example, a stated modulation may be in the form of any combination of varying an amplitude, frequency, or pulse width of an unaffected input signal. These modulated signals will additionally vary in phase for a given pulse set. To produce a modulation, an envelope function, A(n,t), may be convolved with a sine wave, depicted by $e^{j[\bar{\omega}_n(t,n)t+\theta_r+\theta(n)]}$ giving the final waveform A(n,t) $e^{j[\bar{\omega}_n(t,n)t+\theta_r+\theta(n)]}$. In this waveform notation, n is the transducer element number, and i is a given pulse index (e.g., if a second harmonic is utilized, k=2, then i=1 . . . 2). The phase varies for different pulse sequences within a given pulse set and is indicated by the θ_r notation.

To illustrate the concept of phase variation, take an example where k is three. In this example, each pulse within a pulse set is varied in phase by

$$\frac{360°}{k} = \frac{360°}{2} \text{ or } 120°.$$

A first pulse is generated with a 0° phase orientation, a second pulse is 120° out-of-phase with the first pulse, and a last pulse in the pulse set is 240° out-of-phase with respect to the first pulse. After the first pulse is transmitted and received, the second pulse is transmitted and received, and so on through the sequence. All information is tracked so that fundamental frequencies can be summed and eliminated, leaving primarily only harmonically generated echoes. Recall that harmonically generated echoes are produced by non-linear media.

Further, as an example, an envelope function, A(n,t), may be a Gaussian waveform. The transmitted signal may additionally be modulated as a chirped waveform (i.e., swept-frequency modulation, a Fourier transform of which is still centered around the fundamental with a broader dispersion). Optionally, a digital waveform generator could be used in place of the convolution method shown in FIG. **2**.

FIG. **3** shows an embodiment of a signal transmitter unit **120** prior to signal delivery to media of interest. The signal transmitter unit **120** includes at least a power amplifier **330**, a transmit/receive switch **340**, and a first transducer **350**. Optionally, a signal transmitter unit **120** may further include a delay circuit **310**. The delay circuit **310** may be an analog or digital delay. Also, optionally, the signal transmitter unit **120** may include a channel gain unit **320** to drive the power

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amplifier **330** as a function of channel number and time. Additionally, the signal or pulse may be pulse-width modulated (not shown) to conserve power. Power conservation can become crucial in field applications of the system where battery power may be utilized.

FIG. **4A** shows one embodiment of a receiver and raw data averager unit **140**. A second transducer **410** receives pulsed sets modified by media of interest **130**. These received pulsed sets are transformed from ultrasonic energy into an electrical signal by the second transducer **410**. A second transmit/receive switch **420** may be used to couple the electrical signals to the appropriate circuitry. In one embodiment, a second transducer **410** and a second transmit/receive switch **420** may be coincident or analogous units to the first transducer **350** and first transmit/receive switch **340** shown in FIG. **3**. A second power amplifier **430** may be added and controlled as a function of time by gain control unit **440**. The output of power amplifier **430** sends an amplified signal to an optional bandpass filter **450**. The bandpass filter **450** may be used, among other things, to reduce or eliminate extraneous noise. FIGS. **4A** and **4B** share the same component layout up to and including bandpass filter **450**, where thereafter they diverge thereby exemplifying alternate embodiments.

The electrical signal of the FIG. **4A** embodiment is coupled to a first analog-to-digital (A/D) converter **460**, and may continue into an optional in-phase and quadrature (I/Q) mixer **470**, which produces a single side-band signal, optional first baseband filter **480**, and to an averager **490**. The optional first baseband filter acts to reduce or eliminate any fundamental frequency from the signals received from the original pulse sets, leaving primarily harmonically generated signals. One purpose of the averager is to provide a point-by-point arithmetic average of the received electrical signals. Mathematically, this arithmetic average may be expressed as

$$F(n, t) = \sum_{i=1}^{k} \frac{R_i(n, t)}{k},$$

where the received signal, R_i, is summed for each element of the transmit cycle as a function of channel number and time to correlate with the original transmitted pulse, i. All other components in the signal path are of types commonly known to one of ordinary skill in the art.

FIG. **4B** shows an alternative embodiment of a receiver and raw data averager unit. Recall FIGS. **4A** and **4B** share the same component layout through and including the optional bandpass filter **450**. From the point of this optional bandpass filter **450**, the signal of the FIG. **4B** embodiment is further coupled to an analog mixer **455**, an optional second baseband filter **465**, a second analog-to-digital converter **475**, and an averager **490**.

FIG. **5** shows an embodiment of the data processing unit **150**. Here, data processing unit **150** receives averaged data from the receiver and raw data averager unit **140**. The averaged data are input to the data processing unit **150** and received at I/Q raw data matrix **510**, which stores the averaged data in an M×N area array, where M is the number of samples (1 to 10,000 samples is an exemplary number) and N is the number of elements×2 (both in-phase and quadrature). These averaged data are fed into a digital signal processor (DSP) **520**, which reconstructs the raw data into

an area of acoustic image. An exemplary reconstruction equation may take the form of

$$I(r, \varphi) = \sum_{i=1}^{k} a_i(r, \varphi) \cdot e^{j\theta_i(r,\varphi)} \cdot F[i, t_i(r, \varphi)].$$

In this equation a_i indicates an aperture function, r refers to a radial distance from a transducer center at a given angle ϕ, and the function F is an averaged set of raw received data. The digital signal processor **520** functions could be achieved in any number of ways, For example, in an alternative embodiment, a properly designed application-specific integrated circuit (ASIC) could be used in place of the digital signal processor **520**. These converted data in polar coordinates are saved in an acoustic image data buffer **530** in a J×K matrix (where J is the number of range samples and K is the number of angular samples). At this point, the data are still a function of a distance, r, from the transducer at a given angle, ϕ. This could also be accomplished in a Cartesian coordinate system. The acoustic image data buffer **530** allows the data to be stored until needed by scan converter **540**. The $I(r,\phi)$ image data are converted into a reconstructed image in Cartesian coordinate data I(x,y) through the use of an r-ϕ scan converter **540**. An r-ϕ scan converter is well known in the art and typically converts two-dimensional data from polar to Cartesian coordinates by means of the conversion, x=r cos(ϕ) and y=r sin(ϕ).

Output from the data processing unit **150** produces an image area I(x,y) corresponding to an area irradiated by pulsed sets of ultrasonic energy. These converted I(x,y) data may be displayed on image display unit **160**. Image display unit **160** may be any visual display such as, but not limited to, a computer monitor, flat-panel or liquid-crystal display, cathode-ray tube (CRT), or the like.

From the description of the preferred embodiments of the process and apparatus set forth supra, it will be apparent to one of ordinary skill in the art that variations and additions to the embodiments can be made without departing from the principles of the present invention. For example, it could be easy to envision a system whereby an entire three-dimensional (3D) volume could be displayed at once as opposed to a two-dimensional area. This three-dimensional embodiment may be accomplished by holography or some other means. It would be an obvious extrapolation from the tenets of the two-dimensional system presented herein to construct a three-dimensional apparatus.

What is claimed is:

1. A method for performing ultrasonic imaging, comprising:

generating at least two out-of-phase pulses;

converting the at least two out-of-phase pulses into at least two out-of-phase acoustical pulses;

transmitting the at least two out-of-phase acoustical pulses into media of interest;

receiving at least two modified acoustical pulses from the media of interest;

averaging the at least two modified acoustical pulses to generate a set of averaged raw received data, wherein the averaged raw received data is a function of channel number and time; and

constructing image data from the set of averaged raw received data.

2. The method of claim **1**, wherein the at least two out-of-phase pulses are modulated by a change in amplitude.

3. The method of claim **1**, wherein the at least two out-of-phase pulses are modulated by a change in frequency.

4. The method of claim **1**, wherein the at least two out-of-phase pulses are modulated by a change in pulse width.

5. The method of claim **1**, wherein the two out-of-phase pulses are convolved with an envelope function to produce the at least two out-of-phase acoustical pulses.

6. The method of claim **5**, wherein the envelope function is a Gaussian waveform.

7. The method of claim **5**, wherein the envelope function is a chirped waveform.

8. The method of claim **1**, wherein a phase of the at least two out-of-phase pulses vary by 360 degrees divided by an integral denominator that is equal to an integral number of the at least two out-of-phase pulses.

9. The method of claim **1**, wherein the averaging of the at least two modified acoustical pulses is a point-by-point arithmetic average for an element of a transmit cycle as a function of a channel number and a time to correlate back to the at least two acoustical out-of-phase pulses.

10. A method for performing ultrasonic imaging, comprising:

generating at least two out-of-phase pulses;

convolving the at least two out-of-phase pulses with an envelope function;

converting the at least two out-of-phase pulses into at least two out-of-phase acoustical pulses;

transmitting the at least two out-of-phase acoustical pulses into media of interest;

receiving at least two generated return pulses from the media of interest;

averaging data from the at least two generated return pulses as a function of channel number and time;

constructing image data from the data averaged from the at least two generated return pulses; and

displaying the image data on an image display unit.

*   *   *   *   *

UNITED STATES PATENT AND TRADEMARK OFFICE
CERTIFICATE OF CORRECTION

PATENT NO.   : 7,226,416 B2   Page 1 of 1
APPLICATION NO. : 10/226843
DATED   : June 5, 2007
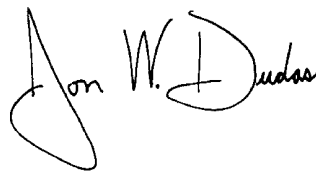INVENTOR(S)   : Glen McLaughlin and Ting-Lan Ji

It is certified that error appears in the above-identified patent and that said Letters Patent is hereby corrected as shown below:

Title Pg, Item (54) Title, and col. 1, lines 1 & 2 should read as follows:

--Phase Inversion Ultrasonic Imaging--

Signed and Sealed this

Sixteenth Day of September, 2008

JON W. DUDAS
Director of the United States Patent and Trademark Office

专利名称(译)	用于相位反转超声成像的系统和方法		
公开(公告)号	US7226416	公开(公告)日	2007-06-05
申请号	US10/226843	申请日	2002-08-21
[标]申请(专利权)人(译)	MCLAUGHLIN GLEN 姬廷局域网		
申请(专利权)人(译)	MCLAUGHLIN GLEN 姬廷-LAN		
当前申请(专利权)人(译)	深圳迈瑞生物医疗电子股份有限公司.		
[标]发明人	MCLAUGHLIN GLEN JI TING LAN		
发明人	MCLAUGHLIN, GLEN JI, TING-LAN		
IPC分类号	A61B8/00 G01N29/44 G01S7/52 G01S15/89		
CPC分类号	G01S7/52038 G01S15/8963 G01S15/8959 G01S7/52077 G01S7/52026 G01S15/104 A61B5/02 A61B5/7257 A61B8/00 A61B8/485 A61B8/488 A61B8/5207 A61B8/5223 A61B8/5238		
其他公开文献	US20030004414A1		
外部链接	Espacenet USPTO		

摘要(译)

公开了利用幅度，频率，相位和/或脉冲宽度不同的多组发射脉冲的超声成像。一个实施例具有k个发射信号之间的相位差 360kdegrees 提供k阶谐波脉冲的相长干涉，同时每组发射轮廓的幅度调制是恒定的。这些脉冲组被传输到感兴趣的介质中，并且来自这些脉冲的接收回波被组合以形成艺术平均信号。平均脉冲表示从每个发射组接收的净共模信号。该组合信号集用于基于宽波束重建方法重建超声图像。