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

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5,140,558 A 8/1992 Harrison, Jr. et al.
5,224,481 A 7/1993 Ishihara et al.
5,233,993 A 8/1993 Kawano
5,241,473 A 8/1993 Ishihara et al.
5,255,683 A 10/1993 Monaghan
5,278,757 A 1/1994 Hctor et al.
5,291,090 A 3/1994 Dias

(Continued)

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A61B 8/00 (2006.01)

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702/48, 49, 66, 67, 70-76

See application file for complete search history.

(56) **References Cited**

U.S. PATENT DOCUMENTS

4,572,203 A 2/1986 Feinstein
4,803,990 A 2/1989 Bonnefous et al.
4,917,097 A * 4/1990 Proudian et al. 600/463
5,119,342 A 6/1992 Harrison, Jr. et al.
5,135,000 A 8/1992 Akselrod et al.

FOREIGN PATENT DOCUMENTS

JP 09-131344 5/1997

OTHER PUBLICATIONS

W. Wilkening, et al., "Phase-Coded Pulse Sequence for Non-Linear
Imaging," IEEE Ultrasonics Symposium 2000.

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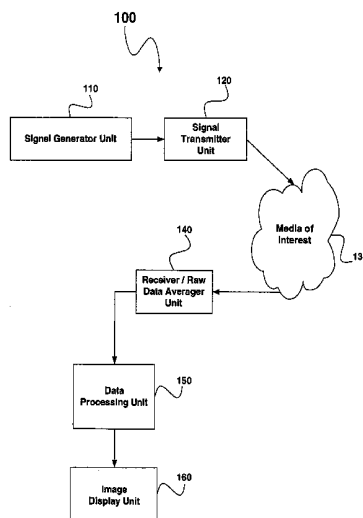
(57) **ABSTRACT**

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

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

providing for constructive interference of the k^{th} order har-
monic 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 an 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.

19 Claims, 6 Drawing Sheets



US 7,699,781 B2

Page 2

U.S. PATENT DOCUMENTS

5,295,485 A	3/1994	Shinomura et al.	6,055,861 A	5/2000	Banta, Jr. et al.
5,302,372 A	4/1994	Lin et al.	6,056,693 A	5/2000	Haider
5,349,524 A *	9/1994	Daft et al. 600/441	6,089,096 A	7/2000	Alexandru
5,410,516 A	4/1995	Uhlendorf et al.	6,108,572 A	8/2000	Panda et al.
5,453,575 A	9/1995	O'Donnell et al.	6,113,545 A	9/2000	Chiao et al.
5,456,257 A	10/1995	Johnson et al.	6,117,082 A	9/2000	Bradley et al.
5,483,963 A	1/1996	Butler et al.	6,120,448 A *	9/2000	Bradley et al. 600/443
5,505,203 A	4/1996	Deitrich et al.	6,132,377 A	10/2000	Bolorforosh et al.
5,632,277 A *	5/1997	Chapman et al. 600/443	6,135,961 A	10/2000	Pflugrath et al.
5,667,373 A	9/1997	Wright et al.	6,139,498 A	10/2000	Katsman et al.
5,706,819 A	1/1998	Hwang et al.	6,139,501 A	10/2000	Roundhill et al.
5,722,412 A	3/1998	Pflugrath et al.	6,168,565 B1	1/2001	Napolitano
5,740,806 A	4/1998	Miller	6,174,286 B1	1/2001	Ramamurthy et al.
5,793,701 A	8/1998	Wright et al.	6,193,662 B1	2/2001	Hwang
5,817,024 A	10/1998	Ogle et al.	6,193,663 B1	2/2001	Napolitano et al.
5,839,442 A	11/1998	Chiang et al.	6,203,498 B1	3/2001	Bunce et al.
5,893,363 A	4/1999	Little et al.	6,206,833 B1	3/2001	Christopher
5,897,501 A	4/1999	Wildes et al.	6,210,334 B1	4/2001	Phillips
5,904,652 A	5/1999	Gilbert et al.	6,213,951 B1	4/2001	Krishnan et al.
5,905,692 A	5/1999	Dolazza et al.	6,221,018 B1	4/2001	Ramamurthy et al.
5,925,967 A	7/1999	Toda	6,238,346 B1	5/2001	Mason
5,970,025 A	10/1999	Cole et al.	6,251,073 B1	6/2001	Imran et al.
5,973,438 A	10/1999	Toda	6,506,158 B2	1/2003	Kawagishi et al.
6,023,977 A	2/2000	Langdon et al.	6,524,252 B1	2/2003	Yu et al.

* cited by examiner

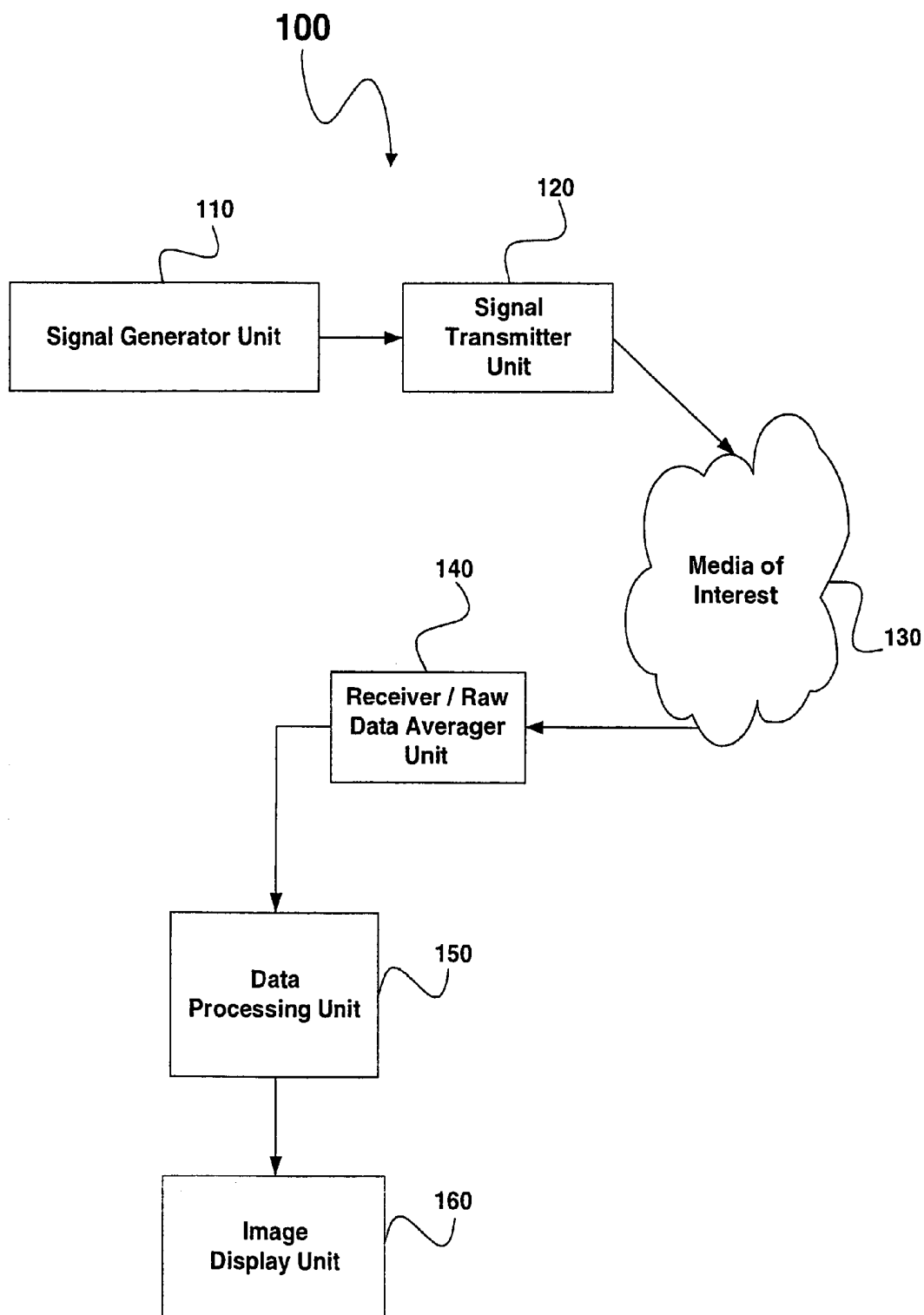


FIG. 1

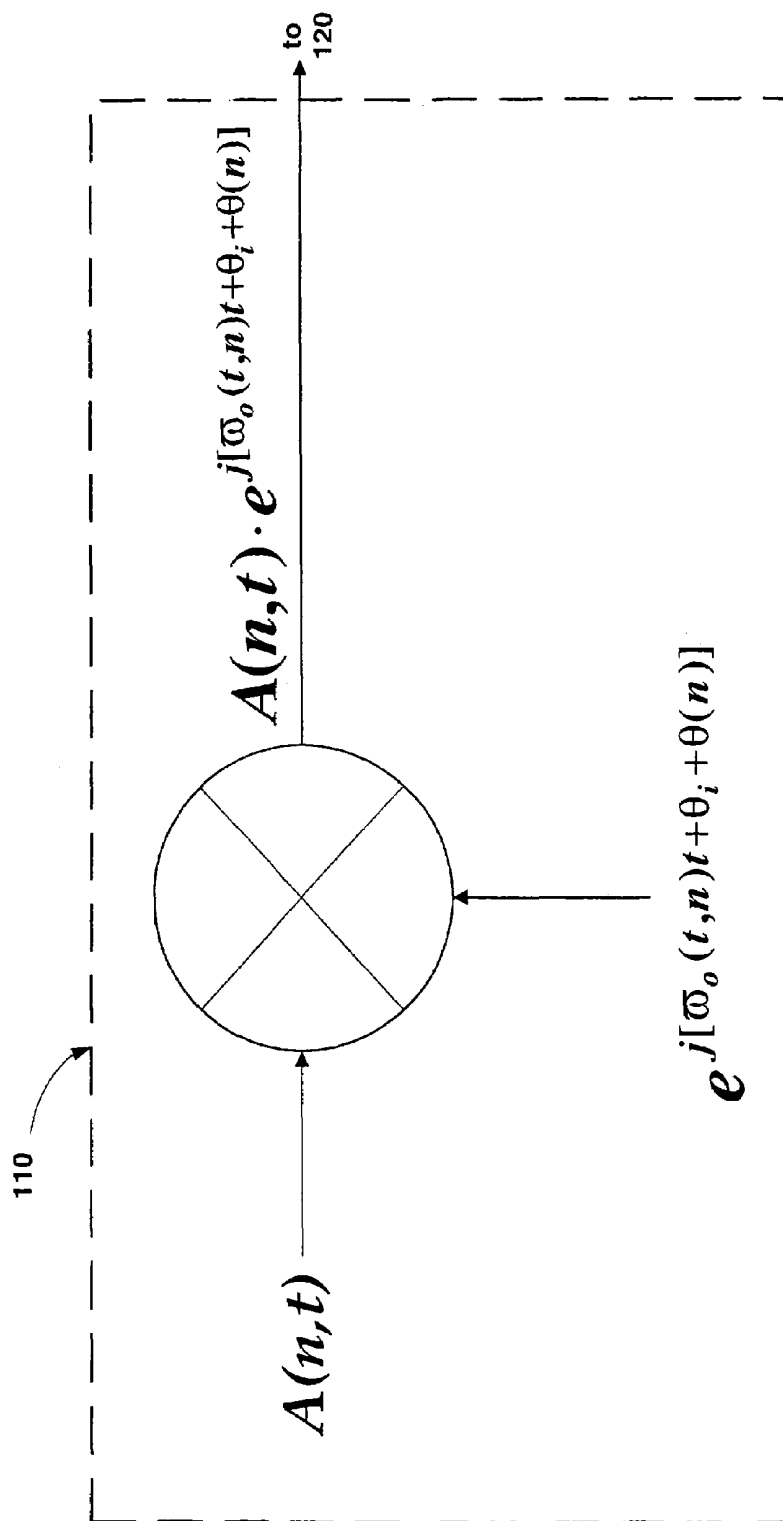
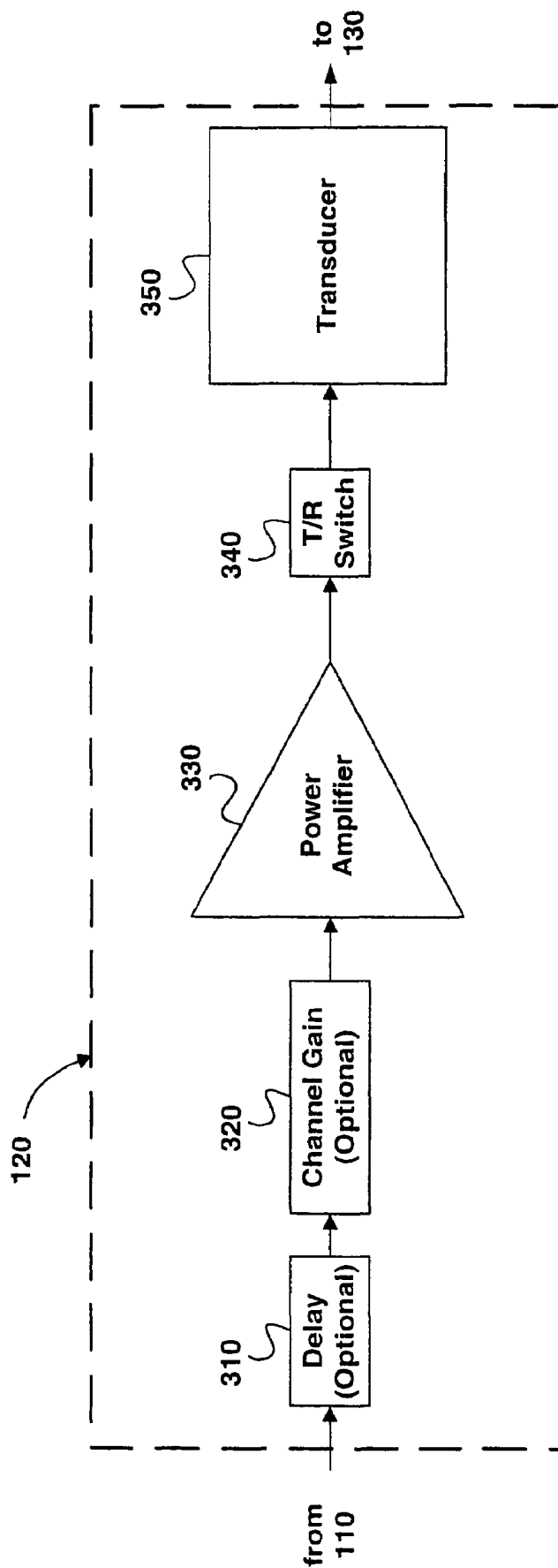


FIG. 2

**FIG. 3**

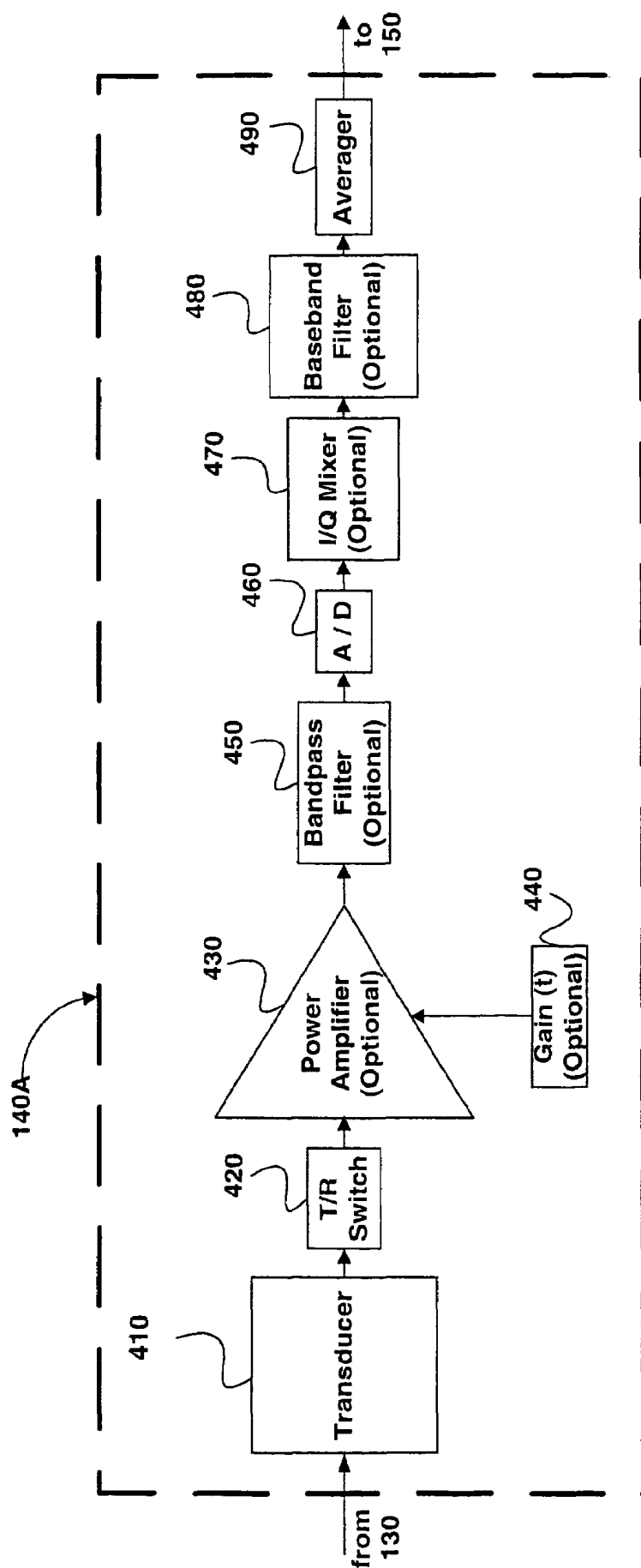
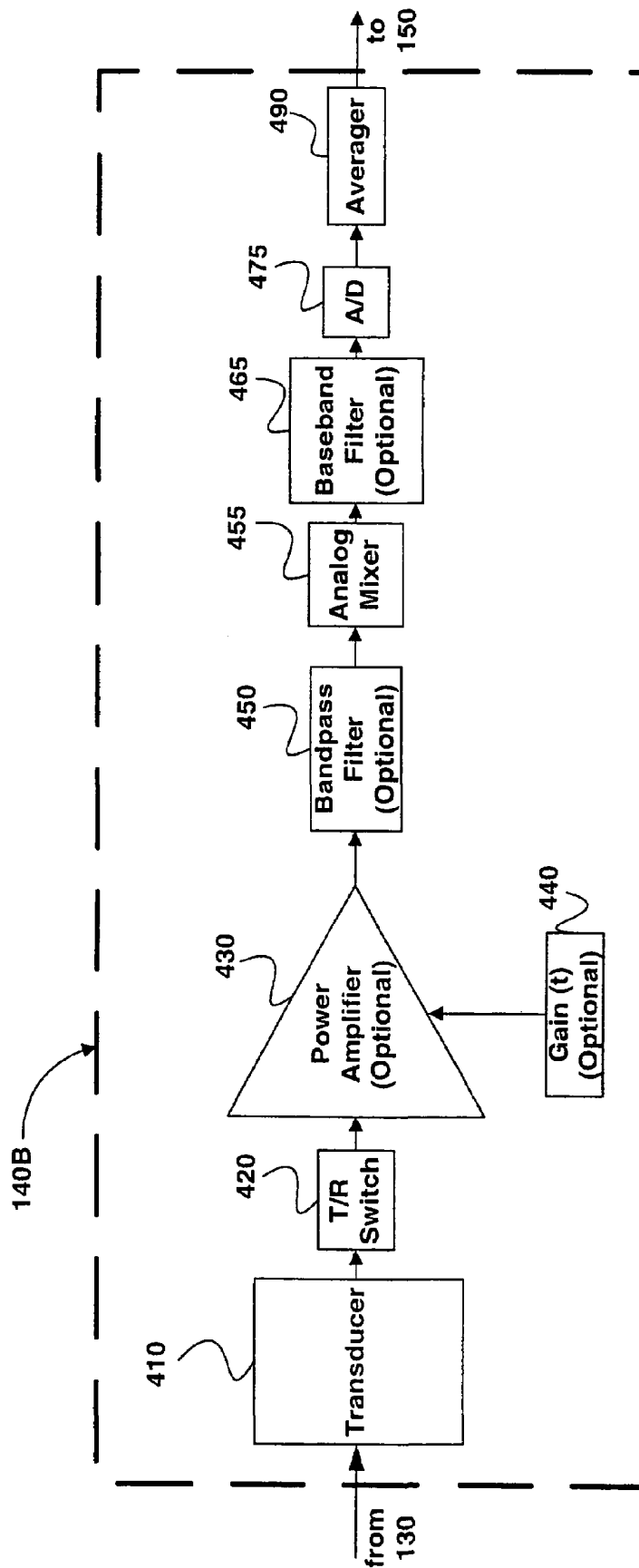


FIG. 4A

**FIG. 4B**

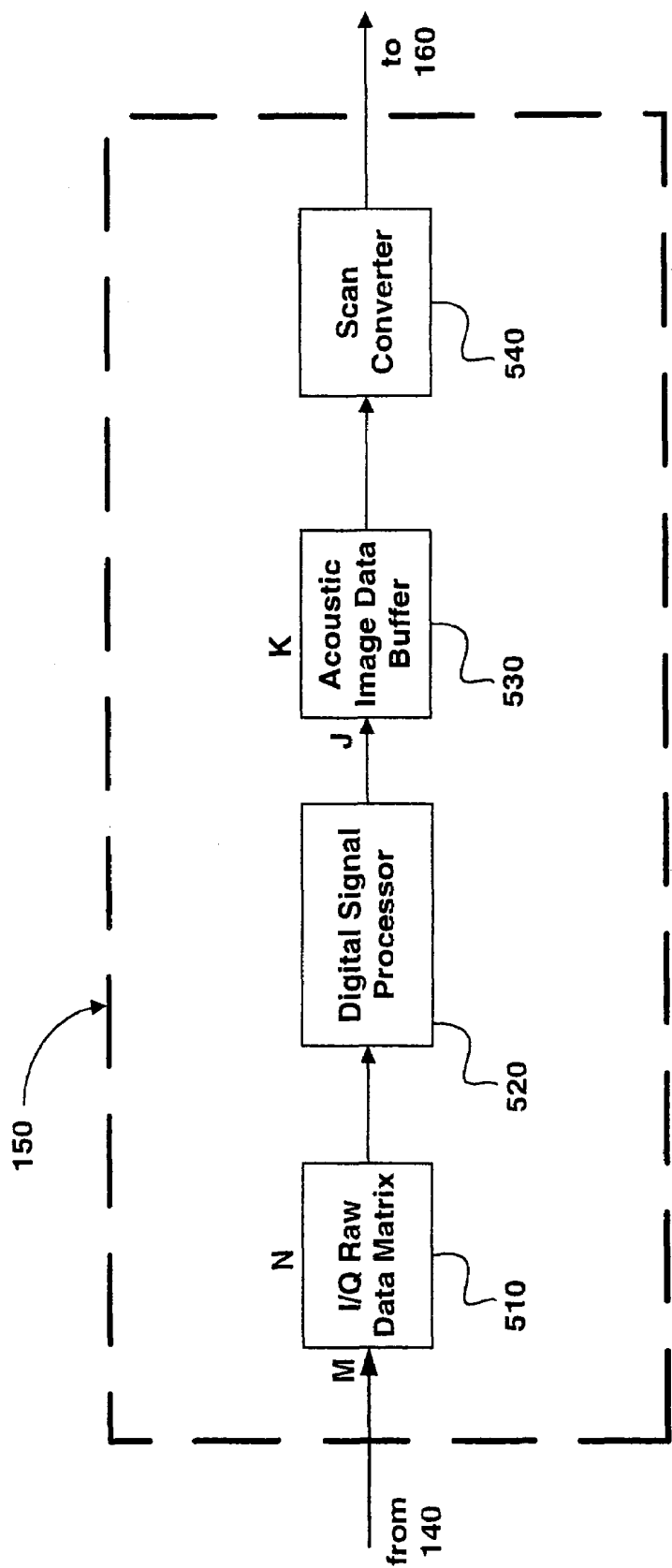


FIG. 5

SYSTEM FOR PHASE INVERSION ULTRASONIC IMAGING

CROSS-REFERENCE TO RELATED APPLICATIONS

This application is a continuation—and claims the priority benefit—of commonly owned U.S. patent application Ser. No. 09/872,541, entitled “Method for Phase Inversion Ultrasonic Imaging,” filed May 31, 2001. The subject matter of this application is incorporated herein by reference.

This application is related to commonly owned U.S. patent application Ser. No. 10/226,843, entitled “Method for Phase Inversion Ultrasonic Imaging,” filed Aug. 21, 2002. U.S. patent application Ser. No. 10/226,843 is a divisional of U.S. patent application Ser. No. 09/872,541. The subject matter of this application is incorporated herein by reference.

BACKGROUND OF THE INVENTION

1. Field of the Invention

The present invention relates to ultrasound imaging, and more particularly, to a system 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 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

$$\frac{360^\circ}{k},$$

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

3

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

$$\frac{360^\circ}{k} = \frac{360^\circ}{2}$$

or 180 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 k^{th} 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 $n \times m$ 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 (B^2 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,

4

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 $e^{j[\omega_0(t, n)t + \theta_i + \theta(n)]}$ giving the final waveform $A(n, t)e^{j[\omega_0(t, n)t + \theta_i + \theta(n)]}$. 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 $i=1 \dots 2$). The phase varies for different pulse sequences within a given pulse set and is indicated by the θ_i 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

$$\frac{360^\circ}{k} = \frac{360^\circ}{3}$$

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 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 $M \times N$ 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 $\times 2$ (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

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

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 $J \times K$ 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,

$$x = r \cos(\phi) \text{ and } y = r \sin(\phi).$$

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 system for ultrasonic imaging, comprising:

a signal generator unit for generating at least two out-of-phase pulses, wherein the signal generator unit convolves at least two out-of-phase sine waves with an envelope function to produce the at least two out-of-phase pulses, and wherein the at least two out-of-phase sine waves are modulated in a way to produce a chirped Gaussian pulse width modulated waveform;

a signal transmitter unit coupled to the signal generator unit for transmitting the at least two out-of-phase pulses into media of interest;

a receiver and raw data averager unit for receiving the at least two out-of-phase pulses modified by the media of interest, the receiver and raw data averager unit provid-

7

ing a point-by-point arithmetic average of the received at least two out-of-phase pulses modified by the media of interest; and

a data processing unit coupled to the receiver and raw data averager unit, the data processing unit constructing an area of acoustic image based on the point-by-point arithmetic average provided by the receiver and raw data averager unit.

2. The system of claim 1, wherein the signal generator unit is a digital waveform generator.

3. The system of claim 1, wherein the envelope function is a Gaussian waveform.

4. The system of claim 1, wherein the envelope function is a chirped waveform.

5. The system of claim 1 wherein the signal transmitter unit comprises a power amplifier, a transmit/receive switch, and a transducer.

6. The system of claim 5, further comprising a digital delay circuit configured to delay the transmission of the at least two out-of-phase pulses into media of interest.

7. The system of claim 5, further comprising an analog delay circuit configured to delay the transmission of the at least two out-of-phase pulses into media of interest.

8. The system of claim 5, further comprising a channel gain circuit to drive the power amplifier.

9. The system of claim 1, wherein the at least two out-of-phase pulses are alternately transmitted by the signal transmitter unit to produce a pulse set.

10. The system of claim 1, wherein the receiver and raw data averager unit comprises a transducer, a transmit/receive switch, an analog-to-digital converter, and an averager.

8

11. The system of claim 10, wherein the receiver and raw data averager unit further comprises a power amplifier, a bandpass filter, and a baseband filter.

12. The system of claim 10, wherein the receiver and raw data averager unit further comprises an in-phase and quadrature mixer configured to produce a single side-band signal.

13. The system of claim 1, wherein the signal generator unit and the receiver and raw data averager unit share a transducer.

14. The system of claim 1, wherein the data processing unit comprises an in-phase and quadrature mixer, a digital signal processor, an acoustic image data buffer, and a scan converter.

15. The system of claim 1, wherein the data processing unit comprises an in-phase and quadrature mixer, an application specific integrated circuit, an acoustic image data buffer, and a scan converter.

16. The system of claim 1, further comprising an image display unit coupled to the data processing unit.

17. The system of claim 16, wherein the image display unit is a computer monitor, the computer monitor configured to display the area of acoustic image.

18. The system of claim 16, wherein the image display unit is a flat-panel display, the flat-panel display configured to display the area of acoustic image.

19. The system of claim 16, wherein the image display unit is a liquid-crystal display, the liquid-crystal display configured to display the area of acoustic image.

* * * * *

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[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		
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外部链接	Espacenet USPTO		

摘要(译)

一种用于超声成像的系统，其利用在幅度，频率，相位和/或脉冲宽度上不同的多组发射脉冲。一个实施例具有k个发射信号之间的相位差 $360k\text{degrees}$ 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。这些脉冲组被发送到感兴趣的介质中，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。

