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(54) **ULTRASOUND IMAGING BEAM-FORMER APPARATUS AND METHOD**

ULTRASCHALL-STRAHLENFORMER-GERÄT UND VERFAHREN

DISPOSITIF ET PROCEDE DE FORMATION DE FAISCEAU D'IMAGERIE ULTRASONORE

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Description

BACKGROUND

Field of the Invention :

[0001] The present invention relates to ultrasonic diagnostic imaging systems and methods. More specifically, the preferred embodiments relate to a device and method for ultrasound imaging beam-forming that may be incorporated in a substantially integrated hand-held ultrasonic diagnostic imaging instrument.

Introduction :

[0002] Medical imaging is a field dominated by high cost systems that may be so complex as to require specialized technicians for operation and the services of experienced medical doctors and nurses for image interpretation. Medical ultrasound, which is considered a low cost modality, utilizes imaging systems costing as much as $250K. These systems may be operated by technicians with two years of training or specialized physicians. This high-tech, high-cost approach works very well for critical diagnostic procedures. However it makes ultrasound impractical for many of the routine tasks for which it would be clinically useful.

[0003] A number of companies have attempted to develop low cost, easy to use systems for more routine use. The most notable effort is that by Sonosite. Their system produces very high quality images at a system cost of approximately $20,000. While far less expensive than high-end systems, these systems are still very sophisticated and require a well-trained operator. Furthermore, at this price few new applications may be opened.

[0004] Many ultrasonic imaging systems utilize an array transducer that is connected to beamformer circuitry through a cable, and a display that is usually connected directly to or integrated with the beam-former. This approach is attractive because it allows the beamformer electronics to be as large as is needed to produce an economical system. In addition, the display may be of a very high quality.

[0005] Some conventional system architectures have been improved upon through reductions in beam-former size. One of the most notable efforts has been undertaken by Advanced Technologies Laboratories and then continued by a spin-off company, Sonosite. U. S. Patent No. 6,135, 961 to Pflugrath et al., entitled "Ultrasonic Signal Processor for a Hand Held Ultrasonic Diagnostic Instrument," describes some of the signal processing employed to produce a highly portable ultrasonic imaging system. The Pflugrath'961 patent makes reference to an earlier patent, U. S. Patent No. 5,817, 024 to Ogle et al., entitled, "Hand Held Ultrasonic Diagnostic instrument with Digital Beamformer". In U. S. Patent No. 6,203, 498 to Bunce et al., entitled "Ultrasonic Imaging Device with Integral Display," however, the transducer, beamformer, and display may be all integrated to produce a very small and convenient imaging system.

[0006] Other references of peripheral interest are US 6,669, 641 to Poland, et al., entitled "Method of and system for ultrasound imaging," which describes an ultrasonic apparatus and method in which a volumetric region of the body is imaged by biplane images. One biplane image has a fixed planar orientation to the transducer, and the plane of the other biplane image can be varied in relation to the fixed reference image.

[0007] US 6,491, 634 to Leavitt, et al., entitled "Sub-beam-forming apparatus and method for a portable ultrasound imaging, "describes a sub-beam-forming method and apparatus that is applied to a portable, one-dimensional ultrasonic imaging system. The sub-beam-forming circuitry may be included in the probes assembly housing the ultrasonic transducer, thus minimizing the number of signals that are communicated between the probe assembly and the portable processor included in the imaging system.

[0008] US 6,380, 766 to Savord, entitled "Integrated circuitry for use with transducer elements in an imaging system, "describes integrated circuitry for use with an ultrasound transducer of an ultrasound imaging system.

[0009] US 6,013, 032 to Savord, entitled "Beam-forming methods and apparatus for three-dimensional ultrasound imaging using two-dimensional transducer array," describes an ultrasound imaging system including a two-dimensional array of ultrasound transducer elements that define multiple sub-arrays, a transmitter for transmitting ultrasound energy into a region of interest with transmit elements of the array, a sub-array processor and a phase shift network associated with each of the sub-arrays, a primary beam-former and an image generating circuit.

[0010] US 6,126, 602 to Savord, et al., entitled "Phased array acoustic systems with intra-group processors," describes an ultrasound imaging apparatus and method that uses a transducer array with a very large number of transducer elements or a transducer array with many more transducer elements than beam-former channels.

[0011] US 5,997, 479 to Savord, et al., entitled "Phased array acoustic systems with intra-group processors," describes an ultrasound imaging apparatus and method that uses a transducer array with a very large number of transducer elements or a transducer array with many more transducer elements than beam-former channels.

[0012] US 6,582, 372 to Poland, entitled "Ultrasound system for the production of 3-D images," describes an ultrasound system that utilizes a probe in conjunction with little or no specialized 3-D software/hardware to produce images having depth cues.

[0013] US 6,179, 780 to Hossack, et al., entitled "Method and apparatus for medical diagnostic ultrasound real-time 3-D transmitting and imaging," describes a medical diagnostic ultrasound real-time 3-D transmitting and imaging system that generates multiple transmit beam sets using a 2-D transducer array.

[0014] US 6,641, 534 to Smith, et al., entitled "Methods and devices for ultrasound scanning by moving sub-apertures of cylindrical ultrasound transducer arrays in two dimensions," describes methods of scanning using a two dimensional (2-D) ultrasound transducer array.

[0015] US 4,949, 310 to Smith, et al., entitled "Maltese cross processor: a high speed compound acoustic imaging system," describes an electronic signal processing device which forms a compound image for any pulse-echo ultrasound imaging system using a two-dimensional array transducer.

[0016] US 6,276, 211 to Smith, entitled "Methods and systems for selective processing of transmit ultrasound beams to display views of selected slices of a volume," describes the selection of a configuration of slices of a volume, such as B slices, I slices, and/or C slices.

[0017] Commercial ultrasound systems have been limited to one-dimensional (1- D) or linear transducer arrays until fairly recently. A typical number of transducers in such an array may be 128. Providing separate multiplex and receive circuitry is manageable with this many transducers, albeit with significant use of expensive high-voltage switches. Newer arrays, however, may be likely to be two-dimensional (2-D) or square arrays. The number of transducers in a two-dimensional array may range up to 128 X 128 or 16,384, and is often in the thousands. Maintaining separate receive, transmit, and multiplex partitioning for the transducers in such an array creates a tremendous burden in terms of cost, space, and complexity. The power consumption and heat dissipation of thousands high-voltage multiplexers is enough to discourage the use of two-dimensional arrays in portable ultrasound imaging systems.

[0018] Current beam-forming strategies can be broadly classified into the two approaches depicted in Fig. 5. One approach is to use digital time delays to focus the data, as illustrated in 5 (a). Geometric delays are calculated and applied to the digitized data on each channel. In such beam-formers, the data needs to either be sampled at a very high sampling rate or interpolated. Implementation of time delays requires sufficient memory to hold a few hundred samples per channel to implement an adequate delay envelope, constraining system complexity.

[0019] In the second approach, systems combine time delays with complex phase rotation, as depicted in 5 (b). Coarse focusing is implemented by delaying the digitized data on each channel. Fine focusing is accomplished by phase rotation of data that has undergone complex demodulation at the center frequency. Such systems require circuitry to perform complex demodulation on every channel. Time delay beam-forming requires significant fast memory to implement a reasonable delay envelope.

A conventional ultrasound beamforming apparatus according to the preamble of claim 1 which combines time delays with complex phase rotation is disclosed in European Patent Application EP 0 713 681 A1.

[0020] Conventional approaches to generating I/Q data may also include analog/digital baseband demodulation, or use a Hilbert transform. Using a demodulation based approach to generate I/Q data may necessitate significant extra circuitry on each channel, while use of the Hilbert transform may require a significant amount of memory to hold the raw RF data.

[0021] Existing ultrasound systems with thousands of separate transmit and receive switches may be too expensive for many applications. While a variety of systems and methods may be known, there remains a need for improved systems and methods.

BRIEF DESCRIPTION OF THE DRAWINGS

[0022] The preferred embodiments of the present invention are shown by a way of example, and not limitation, in the accompanying figures, in which:

FIG. 1 is a schematic diagram of an ultrasound imaging beam-forming apparatus according to a first embodiment of the invention;
FIG. 2 is a schematic diagram of a protection circuit for use with an embodiment of the invention;
FIG. 3 is a schematic diagram of a protection circuit for use with an embodiment of the invention;
FIG. 4 is a schematic diagram of an ultrasound imaging beam-forming apparatus according to a second embodiment of the invention;
FIG. 5 is a schematic diagram of a conventional ultrasound imaging beam-forming apparatus;
FIG. 6 are graphs of signals for use with an embodiment of the invention; and
FIG. 7 is a schematic diagram of a signal receiver for use with an embodiment of the invention.

SUMMARY OF THE INVENTION

[0023] The present invention ultrasound imaging beam-former may be incorporated in an ultrasonic imaging system convenient enough to be a common component of nearly every medical examination and procedure. The present invention ultrasound imaging beam-former provides the potential to have a broad and significant impact in healthcare. The instant document identifies various clinical applications of the present invention ultrasound imaging beam-forming apparatus, but should not be limited thereto, and other applications will become attained as clinicians gain access to the system and method.

[0024] The preferred embodiments of the present invention may improve significantly upon existing methods and/or apparatuses. In particular, the present invention comprises an ultrasound imaging beam-former that may be used in a hand held ultrasonic instrument such as one provided in a portable unit which performs B-mode or C-Mode imaging and/or collects three dimensional (3-D) image data.

An ultrasound beamforming apparatus according to the invention is defined in claim 1. A corresponding method of beamforming for ultrasound imaging according to the present invention is defined in claim 18.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

[0025] The device and method for ultrasound imaging beam-forming may be utilized with various products and services as discussed below, but is not limited thereto. Technicians may attempt to insert needles into a vein based on the surface visibility of the vein coupled with their knowledge of anatomy. While this approach works quite well in thin, healthy individuals, it can prove extremely difficult in patients who may be ill or obese. It may be desirable to have a relatively small, inexpensive, and portable ultrasound imaging system for guiding the insertion of intravenous (IV) devices like needles and catheters into veins, or for drawing blood.

[0026] Sleep apnea (obstruction of the air passage in the of the throat) may affect more than eighteen million Americans. Obstructive sleep apnea may be among the most common variants of sleep apnea. Obstructive sleep apnea may represent a significant risk to the patient. It is difficult and expensive to diagnose obstructive sleep apnea. Typical diagnostic methods require an overnight hospital stay in an instrumented laboratory. Many at-risk patients refuse this inconvenient testing regime and thus go undiagnosed. It may be desirable to have a relatively small, inexpensive, and portable ultrasound imaging system to aid in the diagnosis of obstructive sleep apnea in a minimally obtrusive manner.

[0027] Manual palpation is an exceedingly common diagnostic procedure. Clinicians use their sense of touch to feel for subcutaneous lumps or even to estimate the size of lymph nodes or other masses. While palpation undoubtedly yields valuable qualitative information, numerous studies have shown it to have extremely poor sensitivity and that quantitative size estimates may be completely unreliable. It is desirable to have a relatively small, inexpensive, and portable ultrasound imaging system to aid in observing subcutaneous tissues.

[0028] It may be desirable to place an image display at a transducer. It may be desirable to have a relatively small, inexpensive, and portable ultrasound imaging system to aid in placing the image display at the transducer.

[0029] Ultrasound may be used to search for internal defects in metallic or ceramic parts in a broad variety of industrial applications. Current systems may be cost effective, but areun-wieldy and acquire limited data, making it difficult to ensure that a thorough search has been performed. It is desirable to have a relatively small, inexpensive, and portable ultrasound imaging system to aid in non-destructive evaluation.

[0030] Furthermore, new users may expect ultrasound images to produce representations parallel to the skin's surface, i.e. C-Scan images. It is desirable for a low cost, system to be capable of producing C-Scan images. It is further desirable to display data in the intuitive C-scan format to allow clinicians with little or no training in reviewing ultrasound images to make use of the device.

[0031] Ultrasound imaging devices may be too expensive for some applications. It may be desirable for an ultrasound imaging device to rely primarily or exclusively on receive side beam-forming to reduce or eliminate transmit-side circuitry, enabling the beam-former to be implemented using large scale integration or as software, and enabling system to be produced at a lower cost.

[0032] It may further be desirable for an ultrasound imaging device to rely primarily or exclusively on phase rotation for focusing, enabling the beam-former to be implemented using large scale integration or as software, and enabling system to be produced at a lower cost.

[0033] Ultrasound imaging devices may be insufficiently portable for some applications. It may be desirable for an ultrasonic imaging device to be of a small size to make it easy to carry the device in a pocket or on a belt attachment. This may make the device as convenient as a stethoscope and will thus open new applications. It may be desirable for an ultrasound imaging device to rely primarily or exclusively on receive side beam-forming to reduce or eliminate transmit-side circuitry, enabling the beam-former to be implemented using large scale integration or as software, and enabling the system to be made portable. It may further be desirable for an ultrasound imaging device to rely primarily or exclusively on phase rotation for focusing to reduce or eliminate transmit-side circuitry, enabling the beam-former to be implemented using large scale integration or as software, and enabling the system to be made portable.

[0034] Since it would be desirable for a beam-former to be simple, small, and low cost, it would be further desirable for the size and speed requirements of digital memory in such a beam-former to be minimized. It would be further desirable for focusing to be performed solely by phase rotation of I/Q data, thus eliminating the need for some circuitry, and allowing some of the remaining circuitry to be implemented as an integrated circuit. This may also allow the use of slower memory and reduce the computational complexity of the beam-former.

[0035] It would be further desirable for I/Q data to be generated by sampling an RF signal directly. In one embodiment, an analytic signal (I/Q data) is generated by sampling the received RF signal directly, in a manner analogous to the Hilbert transform. In one embodiment, focusing is implemented via phase rotation of this I/Q data.

[0036] In Fig. 1 is shown an ultrasound imaging beam-former apparatus 100 according to a first embodiment of the invention. Ultrasound imaging beam-former apparatus 100 includes a signal generator 102 for producing an outgoing signal 104 having an outgoing amplitude 106 at an outgoing time 108, as shown in Fig. 6A. In several embodiments, outgoing signal 104 may be an electrical signal, an electro-magnetic signal, or an optical signal.

[0037] If outgoing signal 104 is an optical signal, cross-talk between the circuits of ultrasound imaging beam-former apparatus 100 may be reduced or eliminated, since optical signals do not, in general, interfere with one another. This may allow ultrasound imaging beam-former apparatus 100 to be made smaller than an equivalent electronic device by increasing the density of the circuits. In one case, outgoing signal 104 may be processed as an optical signal and converted to an electrical signal to drive a transducer. An integrated circuit comprising ultrasound imaging beam-forming apparatus 100 may be implemented out of gallium-arsenide (GaAs) so that the both the optical circuits and the electrical circuits can be implemented on the same device. In another case, a transducer utilizing sono-luminescence to convert light directly into sound may be used, dispensing entirely with any need for an electrical-optical interface.

[0038] In several embodiments, signal generator 102 may be a storage device, such as a read-only memory (ROM), an oscillator such as a crystal oscillator, a resonant circuit such as a resistor-inductor-capacitor (RLC) or tank circuit, a resonant cavity such as a ruby laser or a laser diode or a tapped delay line.

[0039] In the event that signal generator 102 is a storage device, outgoing signal 104 may have been stored previously, to be read out when needed. In this embodiment, several versions of outgoing signal 104 may be stored for use with various objects 170 to be imaged. Ultrasound imaging beam-forming apparatus 100 may thus be set to produce a signal appropriate for a particular object 170 to be imaged by choosing one of the stored versions of outgoing signal 104.

[0040] In the event that signal generator 102 is an oscillator, outgoing signal 104 may be a sinusoid of varying frequencies. In this case, outgoing signal 104 may be generated at an arbitrarily high clock speed and still be forced through filters of arbitrarily small bandwidth. This may be advantageous, for example, if a wide band signal is inconvenient. A resonant circuit or a resonant cavity may work in a similar manner. Furthermore, an oscillator may be used to produce a range of frequencies, from which a frequency that generates an optimum response may be selected.

[0041] In the event that signal generator 102 is tapped delay line, outgoing signal 104 could be generated in a manner similar to a spreading code in a code division multiple access (CDMA) format cell phone system. In this case outgoing signal 104 would not need to be a pure sinusoid, but may be a code with a fixed repetition length, such as a Walsh or a Gold code. This may, for example, allow an autocorrelation length of outgoing signal 104 to be adjusted to enhance or suppress coded excitation of an incoming signal.

[0042] If signal generator 102 is a tapped delay line it may be followed by an equalizer to bias or pre-emphasize a range of frequencies in outgoing signal 104. In one embodiment, the equalizer may be an adaptive equalizer that operates on an incoming signal analogous to the sound reflected by the imaged object 170. In this case, the incoming signal could be measured and the result applied to the adaptive equalizer to compensate for frequency attenuation of the sound by amplifying one or more frequencies of the incoming signal or outgoing signal 104 as necessary. This may be useful if, for example, object 170 attenuates or absorbs sound to the point that no return signal is available for imaging. The adaptive equalizer can be placed in parallel with signal generator 102 and in series with the incoming signal.

[0043] An equalizer can be placed in series with signal generator 102. In this case the equalizer can emphasize a particular frequency or frequencies in outgoing signal 104. The equalizer may, for example, place a bias or pre-emphasis toward lower frequencies on outgoing signal 104. This embodiment may be appropriate if, for example, object 170 to be imaged is expected to have features that attenuate lower frequencies significantly more than higher frequencies to the extent that imaging may be difficult. The converse may be true as well, in that the equalizer may have a bias or pre-emphasis toward higher frequencies.

[0044] Outgoing signal 104 may be amplified. Signal generator 102 may include a generator amplifier 158 for amplifying outgoing signal 104. Generator amplifier 158 may pre-emphasize certain frequencies of outgoing signal 104 to suit the attenuation characteristics of object 170 to be imaged as well. Signal generator 102 may also include an oscillator to produce an appropriate modulation frequency, such as a radio frequency (RF) signal, with which to modulate outgoing signal 104.

[0045] A transducer 110 converts outgoing signal 104 to outgoing ultrasound 112. Transducer 110 may be a piezoelectric element, a voice coil, a crystal oscillator or a Hall effect transducer 110. Reversals of outgoing signal 104 produce vibration of a surface of transducer 110 at substantially the frequency of outgoing signal 104. Reversals of outgoing signal 104 may also produce vibrations of a surface of transducer 110 at frequencies that are significantly higher or lower than the frequency of outgoing signal 104, such as harmonics of outgoing signal 104. This vibration may, in turn, produce successive compressions and rarefactions of an atmosphere surrounding the surface of transducer 110, also at substantially the frequency of outgoing signal 104. If the frequency of outgoing signal 104 is substantially higher than a frequency at which sound may be heard, the successive compressions and rarefactions of the atmosphere may be termed ultrasound.

[0046] Transducer 110 may include a plurality of transducers 110. A plurality of transducers 110 may be arranged in an array 166. The array 166 may be a linear array, a phased array, a curvilinear array, an unequally sampled 2-D array, a 1.5-D array, an equally sampled 2D array, a sparse 2D array, or a fully sampled 2D array.

[0047] If outgoing ultrasound 112 is reflected by object 170, some of outgoing ultrasound 112 may return to ultrasound imaging system 100 as reflected ultrasound 182. Reflected ultrasound 182 can be converted to an incoming signal 114 having a period 116, as shown in Fig. 6B. Incoming signal 114 may be an electro-magnetic signal, an electrical signal or an optical signal. Incoming signal 114 may be amplified, pre-amplified, or stored.

[0048] Outgoing ultrasound 112 may be delayed or attenuated partially by object 170. A first portion 174 of outgoing ultrasound 112, for example, may be reflected immediately upon encountering a nearer surface 178 of object 170 while a second portion 176 of outgoing ultrasound 112 is not reflected until it encounters a further surface 180 of object 170. A round trip of second portion 176 will thus be longer than a round trip of first portion 174, resulting in a delay of second portion 176 relative to first portion 174, as well as delays of both first and second portions 174,176 relative to outgoing ultrasound 112. Furthermore, second portion 176 may be damped or attenuated by a material of object 170. The delays may be measured for disparate points of object 170, producing an image 168 of object 170.

[0049] Apparatus 100 includes a signal receiver 118 for processing incoming signal 114. Signal receiver 118 may be implemented as a digital signal processor 164. Signal receiver 118 may also be implemented as an integrated circuit.

[0050] Ultrasonic transducers associated with ultrasound imaging systems may be driven from a single terminal with the second terminal grounded. A transducer may be used to transmit ultrasound signals as well as receive reflected ultrasound. A signal received at a transducer may typically be several orders of magnitude smaller than the signal that was transmitted due to, inter alia, signal attenuation by the target tissue. Some of the signal may be lost due to transducer inefficiencies as well. It may be thus necessary to couple the transducer to a high-voltage transmit signal while the ultrasound is being transmitted, and then to a sensitive low-noise pre-amplifier while the reflected ultrasound is being received.

[0051] A switch that couples the transducer to the transmit and receive signals must be capable of withstanding high peak transmit voltages (typically 50-200 volts) while isolating the pre-amplifier input from those voltage levels, since they would otherwise destroy the pre-amplifier. If a receiver for the signals from the transducers is implemented as a high-density, low-voltage integrated circuit (IC), the switches themselves may need to be implemented off-chip in a separate package from materials and devices that can withstand the high voltage transmit pulses.

[0052] Ultrasound imaging system 100 includes a protection circuit 172 to allow both transmit and receive operations, as shown in Fig. 2. A piezoelectric transducer array 202, shown on the left, acts as an interface to a signal processor by converting electrical signals to acoustic pulses and vice versa. Images may be formed by transmitting a series of acoustic pulses from the transducer array 202 and displaying signals representative of the magnitude of the echoes received from these pulses. A beam-former 214 applies delays to the electrical signals to steer and focus the acoustic pulses and echoes.

[0053] Image formation begins when a state of a transmit/receive switch (TX/RX switch) 204 is altered to connect the transducer elements 202 to individual transmit circuits. Next, transmit generators 206 output time varying waveforms with delay and amplitude variations selected to produce a desired acoustic beam. Voltages of up to 200 Volts may be applied to the transducer elements 202. Once transmission is complete, the state of the TX/RX switch 204 is altered again to connect the transducer elements 202 to individual receive circuitry associated with each element.

[0054] Signals representative of incoming echoes are amplified by pre-amplifiers 208 and time gain control (TGC) 210 circuits to compensate for signal losses due to diffraction and attenuation. Note that the transducer array 202 shown in Fig. 2 has one common electrode 212, and the non-common electrodes may be multiplexed between high-voltage transmit and low-voltage receive signals. This conventional TX/RX switch 204 is the source of considerable expense and bulk in typical ultrasound systems.

[0055] In Fig. 3 is shown an alternative ultrasound imaging beam-forming apparatus 300 with a protection circuit for use with an embodiment of the invention. Ultrasound imaging beam-forming apparatus 300 includes a signal generator 302 for producing an outgoing signal 304.

[0056] Ultrasound imaging beam-forming apparatus 300 also includes a transducer 306 for converting outgoing signal 304 to outgoing ultrasound 308 at a frequency of outgoing signal 304. The transducer 306 has a transmit side 314 forming an interface with outgoing signal 304.

[0057] Transmit side 314 is connected operably to a transmit switch 318. Transmit switch 318 may be an electronic switch, an optical switch, a micro-mechanical switch, a transistor, a field-effect transistor (FET), a bi-polar transistor, a metal-oxide-semiconductor (MOS) transistor, a complementary metal-oxide-semiconductor (CMOS) transistor, or a metal-oxide-semiconductor field-effect transistor (MOSFET). Transmit switch 318 may be connected switchably to signal generator 302 and a ground 320.

[0058] The transducer 306 converts at least a portion of reflected ultrasound 310 to an incoming signal 312. Incoming signal 312 may be an electro-magnetic signal, an electrical signal, or an optical signal. Transducer 306 may have a receive side 316 forming an interface with incoming signal 312.

[0059] The receive side 316 may be connected operably to a receive switch 322. In several embodiments, receive switch 322 may be an electronic switch, an optical switch, a micro-mechanical switch, a transistor, a field-effect transistor, a bi-polar transistor, a MOS transistor, a CMOS transistor, or a MOSFET transistor. Receive switch 322 may be connected switchably to a signal receiver 324 and ground 320.

[0060] Transmit switch 318 connects transmit side 314 to signal generator 302 for a first predetermined period of time while signal generator 302 generates outgoing signal 304. In this embodiment, receive switch 322 connects receive side 316 to signal receiver 324 for a second predetermined period of time while signal receiver 324 receives incoming signal 312. Transmit switch 318 connects transmit side 314 to ground 320 during substantially second predetermined period of time while signal receiver 324 receives incoming signal 312, and receive switch 322 connects receive side 316 to ground 320 during substantially first predetermined period of time while signal generator 302 generates outgoing signal 304. Transmit side 314 and receive side 316 may be on separate transducers 306.

[0061] Signal receiver 118 includes a receiver amplifier 160 for amplifying incoming signal 114. Signal receiver 118 includes a receiver pre-amplifier 162 for amplifying incoming signal 114. Signal receiver 118 may include a band-pass filter for filtering incoming signal 114.

[0062] In the preferred embodiment, signal receiver 118 includes an in-phase sample- and-hold 120 connected receivably to transducer 110 for sampling incoming signal 114 at an incoming time 122 and outputting an in-phase amplitude 124 of incoming signal 114 at substantially incoming time 122. The signal receiver 118 includes an in-phase analog-to-digital converter 126 connected receivably to in-phase sample-and-hold 120 for assigning an in-phase digital value 128 to in-phase amplitude 124 and outputting in-phase digital value 128.

[0063] The signal receiver 118 further includes a quadrature sample- and-hold 130 connected receivably to transducer 110 for sampling incoming signal 114 at substantially one-quarter of period 116 after incoming time 122, quadrature sample-and-hold 130 outputting a quadrature amplitude 132 of incoming signal 114 at substantially one-quarter of period 116 after incoming time 122. One-quarter of period 116 is merely exemplary. Incoming signal 114 may be sampled at any appropriate interval or fraction of period 116. The signal receiver 118 includes a quadrature analog-to-digital converter 134 connected receivably to quadrature sample-and-hold 130 for assigning a quadrature digital value 136 to quadrature amplitude 132 and outputting quadrature digital value 136.

[0064] The signal receiver 118 may include a magnitude calculator 138 connected receivably to in-phase analog-to-digital converter 126 and quadrature analog-to-digital converter 134 for receiving incoming time 122, in-phase digital value 128, and quadrature digital value 136 and outputting a magnitude 140. The signal receiver 118 may further include a phase calculator 142 connected receivably to in-phase analog-to-digital converter 126 and quadrature analog-to-digital converter 134 for receiving incoming time 122, in-phase digital value 128, and quadrature digital value 136 and outputting a phase 144.

[0065] The incoming signal 114 may be band-pass filtered by the band-pass filter and diverted to in-phase sample-and-hold 120 and quadrature sample-and-hold 130. An in-phase clock signal 184 driving in-phase sample-and-hold 120 may be of the same frequency as a quadrature clock signal 186 driving quadrature sample-and-hold 130. Quadrature clock signal 186 may, however, be offset by a quarter of period 116 with respect to in-phase clock signal 184 at an assumed center frequency of incoming signal 114. An output of in-phase sample-and-hold 120 may be digitized by in-phase analog-to-digital converter 126 while an output of quadrature sample-and-hold 130 is digitized in quadrature analog-to-digital converter 134, forming I and Q channel data.

[0066] Reflected ultrasound 182 may be considered to be real part of an amplitude and phase modulated complex exponential signal, or analytic signal. The modulating signal may be expressed mathematically as $A(t)e^{j\phi(t)}$ with instantaneous amplitude A (t) and phase $\phi(t)$. This is superimposed on a carrier signal $e^{-j\omega 0t}$, where $\omega_0 = 2\pi f_0$ and f_0 is the frequency of the signal. Therefore the analytic signal S(t) can be written as,

$$\begin{aligned} S(t) &= A(t)e^{-j(\omega_0 t-\phi(t))} \\ &= A(t)\cos(\omega_0 t-\phi(t)) - jA(t)\sin(\omega_0 t-\phi(t)) \end{aligned} \tag{1}$$

[0067] Only the real part of S(t), which is equivalent to reflected ultrasound 182, is able to be acquired experimentally.

$$I(t) = \mathrm{Re}\{S(t)\} = A(t)\cos(\omega_0 t - \phi(t)) \tag{2}$$

[0068] The output of in-phase analog-to-digital converter 126 is the signal in equation 2 after sampling, or

$$\hat{I}(nT) = InT) = A(nT)\cos(\omega_0 nT - \phi(nT)), n = 0,1,2,3... \tag{3}$$

where T is the sample interval. However, we also require the imaginary component of S (t), shown below in equation 4, to perform beam-forming.

$$Q(t) = \mathrm{Im}\{S(t)\} = -A(t)\sin(\omega_0 t - \phi(t)) \tag{4}$$

[0069] Quadrature clock signal 186 has a time lag of a quarter period at the assumed center frequency relative to the in-phase clock signal 184, as shown schematically in Fig. 7. Therefore the relative time lag is $\frac{1}{4f_0}$, or $\frac{\pi}{2\omega_0}$.

[0070] The output of quadrature sample-and-hold 130 is,

$$\begin{aligned} \hat{Q}(nT) &= I\left\{nT + \frac{\pi}{2\omega_0}\right\} \\ &= A\left(nT + \frac{\pi}{2\omega_0}\right)\cos\left(\omega_0\left(nT + \frac{\pi}{2\omega_0}\right) - \phi\left(nT + \frac{\pi}{2\omega_0}\right)\right) \end{aligned} \tag{5}$$

[0071] We assume that the modulating signal $A(t)e^{j\phi(t)}$ varies slowly with time and approximate,

$$A\left(nT + \frac{\pi}{2\omega_0}\right) \approx A(nT) \tag{6}$$

and

$$\phi\left(nT + \frac{\pi}{2\omega_0}\right) \approx \phi(nT) \tag{7}$$

[0072] Equation 5 can now be rewritten as follows.

$$\begin{aligned}\hat{Q}(nT) &\approx A(nT)\cos\left(\omega_0\left(nT+\frac{\pi}{2\omega_0}\right)-\phi(nT)\right), n=0,1,2,3\ldots \\ &\approx -A(nT)\sin(\omega_0 nT-\phi(nT)) \\ &\approx Q(nT)\end{aligned} \tag{8}$$

[0073] We therefore approximate the imaginary component of S(t), or Q(t) in equation 4, by estimating it to be the output of quadrature sample-and-hold 130.

[0074] Geometric time delays can be calculated and converted to phase delays at the assumed center frequency. Complex weights that implement apodization and focus with the calculated phase delays can be applied to the I/Q data. The signal receiver 118 includes an apodizer 146 for applying a difference 148 between outgoing amplitude 106 and magnitude 140 and applying a first illumination 150-1 to an image points 154 in substantial proportion to difference 148. The signal receiver 118 further includes a phase rotator 152 for applying a second illumination 150-2 to image point 154 in substantial proportion to phase 144.

[0075] In a second embodiment of the invention, shown in Fig. 4, apparatus 100 includes a second transducer 110-2 for converting outgoing signal 104 to second outgoing ultrasound 112-2. Some of second outgoing ultrasound 112-2 may return to second transducer 110-2 if it is reflected by object 170 as well. Second transducer 110-2 converts at least a portion of outgoing ultrasound 112 and second outgoing ultrasound 112-2 to a second incoming signal 114-2 having a second period 116-2, as shown in Fig. 6C.

[0076] Signal receiver 118 includes a second in-phase sample-and-hold 120-2 connected receivably to second transducer 110-2 for sampling second incoming signal 114-2 at incoming time 122 and outputting a second in-phase amplitude 124-2 of second incoming signal 114-2 at substantially incoming time 122. Signal receiver 118 further includes a second in-phase analog-to-digital converter 126-2 connected receivably to second in-phase sample-and-hold 120-2 for assigning a second in-phase digital value 128-2 to second in-phase amplitude 124-2 and outputting second in-phase digital value 128-2.

[0077] Signal receiver 118 includes a second quadrature sample-and-hold 130-2 connected receivably to second transducer 110-2 for sampling second incoming signal 114-2 at substantially one-quarter of second period 116-2 after incoming time 122, second quadrature sample-and-hold 130-2 outputting a second quadrature amplitude 132-2 of second incoming signal 114-2 at substantially one-quarter of second period 116-2 after incoming time 122. Signal receiver 118 further includes a second quadrature analog-to-digital converter 134-2 connected receivably to second quadrature sample-and-hold 130-2 for assigning a second quadrature digital value 136-2 to second quadrature amplitude 132-2 and outputting second quadrature digital value 136-2.

[0078] Signal receiver 118 may further include a second magnitude calculator 138-2 connected receivably to second in-phase analog-to-digital converter 126-2 and second quadrature analog-to-digital converter 134-2 for receiving incoming time 122, second in-phase digital value 128-2, and second quadrature digital value 136-2 and outputting a second magnitude 140-2. Signal receiver 118 includes a second phase calculator 142-2 connected receivably to second in-phase analog-to-digital converter 126-2 and second quadrature analog-to-digital converter 134-2 for receiving incoming time 122, second in-phase digital value 128-2, and second quadrature digital value 136-2 and outputting a second phase 144-2.

[0079] Signal receiver 118 includes a second apodizer 146-2 and a second phase rotator 152-2. Signal receiver 118 further includes a summer 156.

[0080] A method of beam-forming for ultrasound imaging includes the steps of generating an outgoing signal 104 having an outgoing amplitude 106 at an outgoing time 108, transducing outgoing signal 104 to outgoing ultrasound 112, receiving at least a portion of reflected outgoing ultrasound 112, transducing reflected ultrasound to an incoming signal 114 having a period 116, sampling incoming signal 114 at an incoming time 122 to produce an in-phase amplitude 124 of incoming signal 114, assigning an in-phase digital value 128 to in-phase amplitude 124 sampling incoming signal 114 at substantially one-quarter of period 116 after incoming time 122 to produce a quadrature amplitude 132 of incoming signal 114, assigning a quadrature digital value 136 to quadrature amplitude 132, calculating a magnitude 140 at incoming time 122 based on in-phase digital value 128 and quadrature digital value 136, calculating a phase 144 at incoming time 122 based on in-phase digital value 128 and quadrature digital value 136, measuring a difference 148 between outgoing amplitude 106 and magnitude 140, applying a first illumination 150-1 to an image point 154 in substantial proportion to difference 148, and applying a second illumination 150-2 to image point 154 in substantial proportion to phase 144.

[0081] The method of beam-forming for ultrasound imaging may further include the steps of transducing outgoing signal 104 to second outgoing ultrasound 112-2, receiving at least a portion of reflected outgoing ultrasound 112 and second outgoing ultrasound 112-2, transducing reflected outgoing ultrasound 112 and second outgoing ultrasound 112-2 to a second incoming signal 114-2 having a second period 116-2, sampling second incoming signal 114-2 at incoming time 122 to produce a second in-phase amplitude 124-2 of second incoming signal 114-2, assigning a second in-phase digital value 128-2 to second in-phase amplitude 124-2, sampling second incoming signal 114-2 at substantially one-quarter of second period 116-2 after incoming time 122 to produce a second quadrature amplitude 122-2 of second incoming signal 114-2, assigning a second quadrature digital value 136-2 to second quadrature amplitude 122-2, calculating a second magnitude 140-2 at incoming time 122 based on second in-phase digital value 128-2 and second quadrature digital value 136-2, calculating a second phase 144-2 at incoming time 122 based on second in-phase digital value 128-2 and second quadrature digital value 136-2, measuring a second difference 148-2 between outgoing amplitude 106 and second magnitude 140-2, summing difference 148, second difference 148-2, phase 144, and second phase 144-2, applying a third illumination 150-3 to image point 154 in substantial proportion to second difference 148-2, and applying a fourth illumination 150-4 to image point 154 in substantial proportion to second phase 144-2.

[0082] The method of beam-forming may be repeated to produce a plurality of image points 154 forming an image 168. In several embodiments, image 168 may be viewed, used to guide insertion of a needle, used to guide insertion of a catheter, used to guide insertion of an endoscope, used to estimate blood flow, or used to estimate tissue motion. In one embodiment, plurality of image points 154 may be focused. In one embodiment, focusing may be repeated on reflected outgoing ultrasound 112 at plurality of image points 154.

[0083] Plurality of image points 154 may be along a line at a range of interest. A line may also be formed at a plurality of ranges to form a planar image. In one embodiment, the planar image may be a B-mode image. In one embodiment, plurality of image points 154 may lie within a plane at a range of interest. In one embodiment, plurality of image points 154 may form a C-scan. In one embodiment, the plane may be formed at multiple ranges. In one embodiment, several planes may form a complex 1-D image.

[0084] An envelope of magnitude 140 may be displayed. In one embodiment, phase 144 may be used to compensate for a path difference 148 between various transducers and object 170. In one embodiment, a main lobe resolution and a side lobe level may be balanced based on magnitude 140. In one embodiment, a sum squared error between a desired system response and a true system response may be minimized.

[0085] One skilled in the art would appreciate that a variety of tissue information may be obtained through judicious pulse transmission and signal processing of received echoes with the current invention. Such information could be displayed in conjunction with or instead of the aforementioned echo information.

[0086] One such type of information is referred to as color flow Doppler as described in U. S. Patent No. 4,573, 477 to Namekawa et al., entitled "Ultrasonic Diagnostic Apparatus." Another useful type of information is harmonic image data as described in U. S. Patent No. 6,251, 074 to Averkiou et al., entitled "Ultrasonic Tissue Harmonic Imaging" and U. S. Patent no. 5,632, 277 to Chapman et al., entitled "Ultrasound Imaging System Employing Phase Inversion Subtraction to Enhance the Image". Yet another type of information that may be obtained and displayed is known as Power Doppler as described in U. S. Patent No. 5,471, 990 to Thirsk, entitled "Ultrasonic Doppler Power Measurement and Display System".

[0087] Angular scatter information might also be acquired using a method described in a copending U. S. Patent Application No. 10/030, 958, entitled "Angular Scatter Imaging System Using Translating Apertures Algorithm and Method Thereof," filed June 3, 2002.

Speckle is a common feature of ultrasound images. While it is fundamental to the imaging process, many users find its appearance confusing and it has been shown to limit target detectability. A variety of so called compounding techniques have been described which could be valuable for reducing the appearance of speckle in ultrasound transducer drive images. These techniques include spatial compounding and frequency compounding, both of which are well described in the literature.

[0088] One skilled in the art would appreciate that the common practice of frequency compounding could be readily applied to the current invention. By transmitting a plurality of pulses at different frequencies and forming separate detected images using the pulses one may obtain multiple unique speckle patterns from the same target. These patterns may then be averaged to reduce the overall appearance of speckle.

[0089] The well known techniques of spatial compounding may also be applied to the current invention. The most conventional form of spatial compounding, which we call two-way or transmit-receive spatial compounding, entails the acquisition of multiple images with the active transmit and receive apertures shifted spatially between image acquisitions. This shifting operation causes the speckle patterns obtained to differ from one image to the next, enabling image averaging to reduce the speckle pattern.

[0090] In another technique, which we term one-way or receive-only spatial compounding, the transmit aperture is held constant between image acquisitions while the receive aperture is shifted between image acquisitions. As with two-way spatial compounding, this technique reduces the appearance of speckle in the final image.

[0091] In many ultrasound applications the received echoes from tissue have very small amplitude, resulting in an image with poor signal to noise ratio. This problem may be addressed through the use of a technique known as coded excitation. In this method the transmitted pulse is long in time and designed so that it has a very short auto-correlation length. In this manner the pulse is transmitted and received signals are correlated with the transmitted pulse to yield a resultant signal with good signal to noise ratio, but high axial resolution (short correlation length). This method could be readily applied in the present invention ultrasound transducer drive device and method to improve the effective signal to noise ratio. The coded excitation technique is described in U. S. Patent No. 5,014, 712 to O'Donnell, entitled "Coded Excitation for Transmission Dynamic Focusing of Vibratory Energy Beam".

[0092] An aspect in fabricating a system like the present invention ultrasound imaging beam-forming apparatus is in construction of the transducer array. Both cost and complexity could be reduced by incorporating a transducer implemented using photolithographic techniques, i. e. the transducer is formed using micro electro mechanical systems (MEMS). One particularly attractive approach has been described in U. S. Patent No. 6,262, 946 to Khuri-Yakub et al., entitled "Capacitive Micromachined Ultrasonic Transducer Arrays with Reduced Cross-Coupling".

Claims

1. An ultrasound beamformer apparatus, comprising:

a signal generator (102) for producing an outgoing signal (104) having an outgoing amplitude (106) at an outgoing time (108);
a transducer (110, 110-2) for converting said outgoing signal (104) to outgoing ultrasound (112, 112-2);
a plurality of transducers (110, 110-2) for converting at least a portion of said outgoing ultrasound that is reflected to incoming signals (114, 114-2), said incoming signals (114, 114-2) having a period (116);
a plurality of signal receivers (118) and analog-to-digital converters (126, 126-2; 134, 134-2) for converting each of said incoming signals (114, 114-2) to a pair, or time series of pairs of digital in-phase (128, 128-2) and digital quadrature values (136, 136-2);
an apodizer (146, 146-2) and a phase rotator (152, 152-2) for combining said digital in-phase (128, 128-2) and digital quadrature values (136, 136-2) to yield a focused in-phase / quadrature sample,
characterized in that each signal receiver (118) comprises:

an in-phase sample-and-hold (120, 120-2) connected receivably to said transducer (110, 110-2) for sampling said incoming signal (114, 114-2) at an incoming time (122) and outputting an in-phase amplitude (124, 124-2) of said incoming signal (114, 114-2) at substantially said incoming time;
a quadrature sample-and-hold (130, 130-2) connected receivably to said transducer (110, 110-2) for sampling said incoming signal (114, 114-2) at substantially one-quarter of said period after said incoming time, said quadrature sample-and-hold (130, 130-2) outputting a quadrature amplitude (132, 132-2) of said incoming signal (114, 114-2) at substantially one-quarter of said period after said incoming time;
wherein said analog-to-digital converters (126, 126-2; 134, 134-2) are connected receivably to said in-phase sample-and-hold (120, 120-2) and said quadrature sample-and-hold (130, 130-2), respectively.

2. The ultrasound beamformer apparatus of claim 1, wherein:

complex demodulated echo data obtained from a single range from each transducer array element (202) are sampled;
complex echo signals are multiplied by complex weightings; and the results are summed to focus at a specific point at the range of interest.

3. The ultrasound beamformer apparatus of claim 2, wherein:

the complex demodulation is performed using an analog demodulation circuit on each element.

4. The ultrasound beamformer apparatus according to any of the preceding claims, wherein:

the in-phase sample-and-hold circuit (120, 120-2) and the quadrature sample-and-hold circuit (130, 130-2) are configured to sample the incoming signal at two times separated in time by approximately 1/4 of the period (116) of the incoming signal.

5. The ultrasound beamformer apparatus according to any of the preceding claims, wherein:

the ultrasound beamformer apparatus is configured such that combining is repeated on the same set of echo data at a plurality of points to form complex image data at that range.

6. The ultrasound beamformer apparatus of claim 5, wherein:

the plurality of points are along a line at the range of interest.

7. The ultrasound beamformer apparatus of claim 5, wherein:

the plurality of points lie within a plane at the range of interest and thereby form a complex c-scan.

8. The ultrasound beamformer apparatus of claim 6, wherein:

the process of forming image lines is repeated at numerous ranges to form a planar ultrasound image, possibly being a b-mode image.

9. The ultrasound beamformer apparatus of claim 7, wherein:

the process of forming image planes is repeated at multiple ranges to form a complex 3D image.

10. The ultrasound beamformer apparatus according to any of the preceding claims, wherein:

the envelope of the magnitude is taken for display to the user.

11. The ultrasound beamformer apparatus according to any of the preceding claims, wherein:

the phase rotator (152, 152-2) is adapted for focusing to compensate for path length differences between different transducer array elements and a focal point.

12. The ultrasound beamformer apparatus according to any of the preceding claims, wherein:

the apodizer (146, 146-2) is adapted to maintain a reasonable balance between main-lobe resolution and side-lobe levels in the system response.

13. The ultrasound beamformer apparatus according to any of the preceding claims, wherein:

the transducer (110, 110-2) is part of a transducer array (202) employed for imaging, said transducer array (202) consisting of a plurality of array elements, said transducer elements placed in a linear configuration selected from the group consisting of: a linear array, a phased array, and a curvilinear array.

14. The ultrasound beamformer apparatus according to any of the preceding claims, wherein:

the transducer (110, 110-2) is part of a transducer array (202) employed for imaging, said transducer array (202) consisting of one of the group consisting of:

a plurality of elements arranged in an unequally sampled 2D configuration or a 1.5-D array;
a plurality of elements that are placed in an equally sampled 2D configuration;
a fraction of the elements of the array are utilized such that the resulting array is a sparse 2D array;
all elements of the array are utilized such that the resulting array is a fully sampled 2D array;
connections are made to individual array elements such that individual elements may be used for either transmission or reception, but not both, thereby eliminating the need for receive protection circuitry; and
only a fraction of elements are used to form any given image point.

15. The ultrasound beamformer apparatus according to any of the preceding claims, wherein:

the ultrasound beamformer apparatus is configured such that combining is repeated for different fractions of the aperture thereby obtaining multiple redundant views of the same target location.

16. The ultrasound beamformer apparatus of claim 15, wherein:

the multiple looks are averaged after taking their magnitudes so as to reduce the appearance of speckle in the resulting image.

17. The ultrasound beamformer apparatus of any of the preceding claims, wherein:

the sampling operation is performed at multiple ranges for a single transmit event, thereby increasing the image formation rate.

18. A method of beamforming for ultrasound imaging, comprising:

generating an outgoing signal (104) having an outgoing amplitude (106) at an outgoing time (108);
transducing said outgoing signal (104) to outgoing ultrasound (112, 112-2);
receiving at least a portion of reflected outgoing ultrasound,
transducing said reflected ultrasound to incoming signals (114, 114-2) having a period (116);

A/D converting each of said incoming signals (114, 114-2) to a pair, or time series of pairs of digital in-phase (128, 128-2) and digital quadrature values (136, 136-2); and combining said digital in-phase (128, 128-2) and digital quadrature values (136, 136-2) to yield a focused in-phase/quadrature sample;

characterized by comprising the further steps of:

sampling said incoming signal (114, 1114-2) at an incoming time (122) to produce an in-phase amplitude (124, 124-2) of said incoming signal (114, 114-2) at substantially said incoming time;
sampling said incoming signal (114, 114-2) at substantially one-quarter of said period after said incoming time to produce a quadrature amplitude (132, 132-2) of said incoming signal (114, 114-2) at substantially one-quarter of said period after said incoming time;
wherein said in-phase amplitude (124, 124-2) and said quadrature amplitude (132, 132-2) are A/D converted to generate said pair, or time series of pairs of digital in-phase (128, 128-2) and digital quadrature values (136, 136-2).

19. The method of beamforming for ultrasound imaging of claim 18, further comprising amplifying said outgoing signal (104).

20. The method of beamforming for ultrasound imaging of claim 18, further comprising an operation selected from the group consisting of:

amplifying said incoming signal (114, 114-2);
pre-amplifying said incoming signal (114, 114-2); and
storing said incoming signal (114, 114-2).

21. The method of beamforming for ultrasound imaging of claim 18, further comprising repeating the method of beamforming to produce a plurality of image points forming an image.

22. The method of beamforming for ultrasound imaging of claim 21, further comprising focusing said plurality of image points.

23. The method of beamforming for ultrasound imaging of claim 22, wherein said focusing is repeated on said reflected outgoing ultrasound at said plurality of image points.

24. The method of beamforming for ultrasound imaging of claim 21, wherein the plurality of image points are along a line at a range of interest.

25. The method of beamforming for ultrasound imaging of claim 24, wherein the line is formed at a plurality of ranges to form a planar image.

26. The method of beamforming for ultrasound imaging of claim 25, wherein the planar image is a B-mode image.

27. The method of beamforming for ultrasound imaging of claim 21, wherein the plurality of image points lie within a plane at a range of interest.

28. The method of beamforming for ultrasound imaging of claim 27, wherein the plurality of image points form a C-scan.

29. The method of beamforming for ultrasound imaging of claim 27, wherein the plane is formed at multiple ranges.

30. The method of beamforming for ultrasound imaging of claim 27, wherein the planes form a complex 3D image.

Patentansprüche

1. Ultraschall-Strahlenbündelvorrichtung mit:

einem Signalgenerator (102) zum Erzeugen eines abgehenden Signals (104) mit einer Ausgangsamplitude (106) zu einem Ausgangszeitpunkt (108);
einem Wandler (110, 110-2) zum Umwandeln des abgehenden Signals (104) in abgehenden Ultraschall (112, 112-2);
einer Mehrzahl von Wandlern (110, 110-2) zum Umwandeln zumindest eines Teils des reflektierten abgehenden Ultraschalls in ankommende Signale (114, 114-2), wobei die ankommenden Signale (114, 114-2) eine Periode (116) aufweisen;
einer Mehrzahl von Signalempfängern (118) und Analog-Digital-Wandlern (126, 126-2; 134, 134-2) zum Umwandeln jedes der ankommenden Signale (114, 114-2) in ein Paar oder in eine Zeitreihe von Paaren aus digitalen Inphase-Werten (128, 128-2) und digitalen Quadraturwerten (136, 136-2) und
einem Apodisierer (146, 146-2) und einem Phasendreher (152, 152-2) zum Kombinieren der digitalen Inphase-Werte (128, 128-2) und der digitalen Quadraturwerte (136, 136-2), um einen gebündelten Inphase-/Quadratur-Abtastwert zu erhalten,

dadurch gekennzeichnet, dass jeder Signalempfänger (118) Folgendes aufweist:

eine Inphase-Abtast- und -Halteschaltung (120, 120-2), die empfangsfähig mit dem Wandler (110, 110-2) verbunden ist, zum Abtasten des ankommenden Signals (114, 114-2) zu einem Eingangszeitpunkt (122) und zum Ausgeben einer Inphase-Amplitude (124, 124-2) des ankommenden Signals (114, 114-2) zu im Wesentlichen dem Eingangszeitpunkt und
eine Quadratur-Abtast- und -Halteschaltung (130, 130-2), die empfangsfähig mit dem Wandler (110, 110-2) verbunden ist, zum Abtasten des ankommenden Signals (114, 114-2) zu im Wesentlichen einem Viertel der Periode nach dem Eingangszeitpunkt, wobei die Quadratur-Abtast- und -Halteschaltung (130, 130-2) eine Quadratur-Amplitude (132, 132-2) des ankommenden Signals (114, 114-2) zu im Wesentlichen einem Viertel der Periode nach dem Eingangszeitpunkt ausgibt,

wobei die Analog-Digital-Wandler (126, 126-2; 134, 134-2) empfangsfähig mit der Inphase-Abtast- und -Halteschaltung (120, 120-2) bzw. der Quadratur-Abtast- und -Halteschaltung (130, 130-2) verbunden sind.

2. Ultraschall-Strahlenbündelvorrichtung nach Anspruch 1, **dadurch gekennzeichnet, dass**
komplex demodulierte Echo-Daten abgetastet werden, die aus einem einzelnen Bereich von jedem Wandler-Array-Element (202) erhalten werden,
komplexe Echo-Signale mit komplexen Gewichtungen multipliziert werden und
die Ergebnisse summiert werden, um eine Bündelung an einem bestimmten Punkt in dem interessierenden Bereich zu erhalten.

3. Ultraschall-Strahlenbündelvorrichtung nach Anspruch 2, **dadurch gekennzeichnet, dass** die komplexe Demodulation unter Verwendung einer analogen Demodulationsschaltung an jedem Element durchgeführt wird.

4. Ultraschall-Strahlenbündelvorrichtung nach einem der vorhergehenden Ansprüche, **dadurch gekennzeichnet, dass** die Inphase-Abtast- und -Halteschaltung (120, 120-2) und die Quadratur-Abtast- und -Halteschaltung (130, 130-2) so konfiguriert sind, dass sie das ankommende Signal zu zwei Zeitpunkten abtasten, die zeitlich um etwa ¼ der Periode (116) des ankommenden Signals voneinander getrennt sind.

5. Ultraschall-Strahlenbündelvorrichtung nach einem der vorhergehenden Ansprüche, **dadurch gekennzeichnet, dass** die Ultraschall-Strahlenbündelvorrichtung so konfiguriert ist, dass das Kombinieren an ein und demselben Satz von Echo-Daten an einer Vielzahl von Punkten wiederholt wird, um komplexe Bilddaten in diesem Bereich zu erzeugen.

6. Ultraschall-Strahlenbündelvorrichtung nach Anspruch 5, **dadurch gekennzeichnet, dass** die Vielzahl von Punkten entlang einer Linie in dem interessierenden Bereich liegt.

7. Ultraschall-Strahlenbündelvorrichtung nach Anspruch 5, **dadurch gekennzeichnet, dass** die Vielzahl von Punkten in einer Ebene in dem interessierenden Bereich liegt und dadurch eine komplexe c-Abtastung entsteht.

8. Ultraschall-Strahlenbündelvorrichtung nach Anspruch 6, **dadurch gekennzeichnet, dass** des Prozess des Erzeugens von Bildlinien in zahlreichen Bereichen wiederholt wird, um ein ebenes Ultraschallbild zu erzeugen, das möglicherweise ein b-Modus-Bild ist.

9. Ultraschall-Strahlenbündelvorrichtung nach Anspruch 7, **dadurch gekennzeichnet, dass** der Prozess des Erzeugens von Bild-Ebenen in mehreren Bereichen wiederholt wird, um ein komplexes 3D-Bild zu erzeugen.

10. Ultraschall-Strahlenbündelvorrichtung nach einem der vorhergehenden Ansprüche, **dadurch gekennzeichnet, dass** die Enveloppe der Größe zum Anzeigen für den Nutzer verwendet wird.

11. Ultraschall-Strahlenbündelvorrichtung nach einem der vorhergehenden Ansprüche, **dadurch gekennzeichnet, dass** der Phasendreher (152, 152-2) zum Bündeln eingerichtet ist, um Weglängen-Unterschiede zwischen verschiedenen Wandler-Array-Elementen und einem Brennpunkt auszugleichen.

12. Ultraschall-Strahlenbündelvorrichtung nach einem der vorhergehenden Ansprüche, **dadurch gekennzeichnet, dass** der Apodisierer (146, 146-2) so eingerichtet ist, dass er ein sinnvolles Gleichgewicht zwischen einer Hauptlappen-Auflösung und Nebenlappen-Niveaus bei der Reaktion des Systems aufrechterhält.

13. Ultraschall-Strahlenbündelvorrichtung nach einem der vorhergehenden Ansprüche, **dadurch gekennzeichnet, dass** der Wandler (110, 110-2) Teil eines Wandler-Arrays (202) ist, das für die Bildgebung verwendet wird, wobei das Wandler-Array (202) aus einer Vielzahl von Array-Elementen besteht, wobei die Wandler-Elemente in einer linearen Konfiguration angeordnet sind, die aus der Gruppe lineares Array, phasengesteuertes Array und krummliniges Array gewählt ist.

14. Ultraschall-Strahlenbündelvorrichtung nach einem der vorhergehenden Ansprüche, **dadurch gekennzeichnet, dass** der Wandler (110, 110-2) Teil eines Wandler-Arrays (202) ist, das für die Bildgebung verwendet wird, wobei das Wandler-Array (202) in einer Komponente aus der folgenden Gruppe besteht:

    eine Vielzahl von Elementen, die in einer ungleichmäßig abgetasteten 2D-Konfiguration oder einem 1,5-D-Array angeordnet sind;
    eine Vielzahl von Elementen, die in einer gleichmäßig abgetasteten 2D-Konfiguration angeordnet sind;
    ein Teil der Elemente des Arrays wird so genutzt, dass das resultierende Array ein dünn besetztes 2D-Array ist;
    ein Teil der Elemente des Arrays wird so genutzt, dass das resultierende Array ein vollständig abgetastetes 2D-Array ist;
    Verbindungen zu einzelnen Array-Elementen werden so hergestellt, dass einzelne Elemente entweder zum Senden oder zum Empfangen, aber nicht für beides, verwendet werden können, wodurch die Notwendigkeit einer Empfangsschutzschaltung eliminiert wird;
    und
    nur ein Teil der Elemente wird zum Erzeugen eines gegebenen Bildpunkts verwendet.

15. Ultraschall-Strahlenbündelvorrichtung nach einem der vorhergehenden Ansprüche, **dadurch gekennzeichnet, dass** die Ultraschall-Strahlenbündelvorrichtung so konfiguriert ist, dass das Kombinieren für verschiedene Teile der Apertur wiederholt wird, wodurch mehrere redundante Ansichten ein und desselben Zielorts erhalten werden.

16. Ultraschall-Strahlenbündelvorrichtung nach Anspruch 15, **dadurch gekennzeichnet, dass** aus den mehreren Ansichten nach dem Bestimmen ihrer Größen der Mittelwert ermittelt wird, um das Auftreten der Fleckigkeit in dem resultierenden Bild zu verringern.

17. Ultraschall-Strahlenbündelvorrichtung nach einem der vorhergehenden Ansprüche, **dadurch gekennzeichnet, dass** das Abtasten für einen einzigen Sendevorgang in mehreren Bereichen durchgeführt wird, wodurch die Bilderzeugungsgeschwindigkeit erhöht wird.

18. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung mit den folgenden Schritten:

Erzeugen eines abgehenden Signals (104) mit einer Ausgangsamplitude (106) zu einem Ausgangszeitpunkt (108);
Umwandeln des abgehenden Signals (104) in abgehenden Ultraschall (112, 112-2);
Empfangen zumindest eines Teils des reflektierten abgehenden Ultraschalls;
Umwandeln des reflektierten Ultraschalls in ankommende Signale (114, 114-2), die eine Periode (116) haben;
Durchführen einer Analog-Digital-Umwandlung jedes der ankommenden Signale (114, 114-2) in ein Paar oder in eine Zeitreihe von Paaren aus digitalen Inphase-Werten (128, 128-2) und digitalen Quadraturwerten (136, 136-2) und
Kombinieren der digitalen Inphase-Werte (128, 128-2) und der digitalen Quadraturwerte (136, 136-2), um einen gebündelten Inphase-/Quadratur-Abtastwert zu erhalten,

dadurch gekennzeichnet, dass das Verfahren die folgenden weiteren Schritte aufweist:

Abtasten des ankommenden Signals (114, 114-2) zu einem Eingangszeitpunkt (122), um eine Inphase-Amplitude (124, 124-2) des ankommenden Signals (114, 114-2) zu im Wesentlichen dem Eingangszeitpunkt zu erzeugen; und
Abtasten des ankommenden Signals (114, 114-2) zu im Wesentlichen einem Viertel der Periode nach dem Eingangszeitpunkt, um eine Quadratur-Amplitude (132, 132-2) des ankommenden Signals (114, 114-2) zu im Wesentlichen einem Viertel der Periode nach dem Eingangszeitpunkt zu erzeugen,

wobei die Inphase-Amplitude (124, 124-2) und die Quadratur-Amplitude (132, 132-2) einer Analog-Digital-Umwandlung unterzogen werden, um das Paar oder die Zeitreihe von Paaren aus digitalen Inphase-Werten (128, 128-2) und digitalen Quadraturwerten (136, 136-2) zu erzeugen.

19. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung nach Anspruch 18, das weiterhin das Verstärken des abgehenden Signals (104) aufweist.

20. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung nach Anspruch 18, das weiterhin eine Operation aufweist, die aus der folgenden Gruppe gewählt ist:

Verstärken des ankommenden Signals (114, 114-2);
Vorverstärken des ankommenden Signals (114, 114-2) und
Speichern des ankommenden Signals (114, 114-2).

21. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung nach Anspruch 18, das weiterhin das Wiederholen des Verfahrens zum Bündeln von Strahlen aufweist, um eine Vielzahl von Bildpunkten zu erzeugen, die ein Bild bilden.

22. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung nach Anspruch 21, das weiterhin das Bündeln der Vielzahl von Bildpunkten aufweist.

23. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung nach Anspruch 22, **dadurch gekennzeichnet, dass** das Bündeln an dem abgehenden Ultraschall wiederholt wird, der an der Vielzahl von Bildpunkten reflektiert wird.

24. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung nach Anspruch 21, **dadurch gekennzeichnet, dass** sich die Vielzahl von Bildpunkten entlang einer Linie in dem interessierenden Bereich befindet.

25. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung nach Anspruch 24, **dadurch gekennzeichnet, dass** die Linie in einer Vielzahl von Bereichen erzeugt wird, um ein ebenes Bild zu erzeugen.

26. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung nach Anspruch 25, **dadurch gekennzeichnet,**

dass das ebene Bild ein B-Modus-Bild ist.

27. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung nach Anspruch 21, **dadurch gekennzeichnet, dass** die Vielzahl von Bildpunkten in einer Ebene in einem interessierenden Bereich liegt.

28. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung nach Anspruch 27, **dadurch gekennzeichnet, dass** durch die Vielzahl von Bildpunkten eine c-Abtastung entsteht.

29. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung nach Anspruch 27, **dadurch gekennzeichnet, dass** die Ebene in mehreren Bereichen erzeugt wird.

30. Verfahren zum Bündeln von Strahlen für die Ultraschall-Bildgebung nach Anspruch 27, **dadurch gekennzeichnet, dass** die Ebenen ein komplexes 3D-Bild erzeugen.

Revendications

1. Appareil formeur de faisceau d'ultrasons, comprenant :

   un générateur de signal (102) pour produire un signal de sortie (104) ayant une amplitude de sortie (106) à un temps de sortie (108) ;
   un transducteur (110, 110-2) pour convertir ledit signal de sortie (104) en ultrasons de sortie (112, 112-2) ;
   une pluralité de transducteurs (110, 110-2) pour convertir au moins une partie desdits ultrasons de sortie qui sont réfléchis en signaux d'entrée (114, 114-2), lesdits signaux d'entrée (114, 114-2) ayant une période (116) ;
   une pluralité de récepteurs de signal (118) et des convertisseurs analogique-numérique (126, 126-2 ; 134, 134-2) pour convertir chacun desdits signaux d'entrée (114, 114-2) en une paire, ou une série temporelle de paires de valeurs en phase numériques (128, 128-2) et de quadrature numériques (136, 136-2) ;
   un apodiseur (146, 146-2) et un dispositif de rotation de phase (152, 152-2) pour combiner lesdites valeurs en phase numériques (128, 128-2) et de quadrature numériques (136, 136-2) pour obtenir un échantillon en phase / quadrature focalisé,
   caractérisé en ce que chaque récepteur de signal (118) comprend :

      un échantillonneur bloqueur en phase (120, 120-2) raccordé pour réception audit transducteur (110, 110-2) pour échantillonner ledit signal d'entrée (114, 114-2) à un temps d'entrée (122) et transmettre une amplitude en phase (124, 124-2) dudit signal d'entrée (114, 114-2) sensiblement audit temps d'entrée ;
      un échantillonneur bloqueur à quadrature (130, 130-2) raccordé pour réception audit transducteur (110, 110-2) pour échantillonner ledit signal d'entrée (114, 114-2) à sensiblement un quart de ladite période après ledit temps d'entrée, ledit échantillonneur bloqueur à quadrature (130, 130-2) transmettant une amplitude de quadrature (132, 132-2) dudit signal d'entrée (114, 114-2) à sensiblement un quart de ladite période après ledit temps d'entrée ; dans lequel lesdits convertisseurs analogique-numérique (126, 126-2 ; 134, 134-2) sont raccordés pour réception audit échantillonneur bloqueur en phase (120, 120-2) et audit échantillonneur bloqueur à quadrature (130, 130-2), respectivement.

2. Appareil formeur de faisceau d'ultrasons de la revendication 1, dans lequel :

   des données d'écho démodulées complexes obtenues à partir d'une plage unique de chaque élément de réseau de transducteurs (202) sont échantillonnées ;
   des signaux d'écho complexes sont multipliés par des pondérations complexes ; et
   les résultats sont additionnés pour se focaliser à un point spécifique dans la plage d'intérêt.

3. Appareil formeur de faisceau d'ultrasons de la revendication 2, dans lequel :

   la démodulation complexe est effectuée en utilisant un circuit de démodulation analogique sur chaque élément.

4. Appareil formeur de faisceau d'ultrasons selon l'une quelconque des revendications précédentes, dans lequel :

   le circuit échantillonneur bloqueur en phase (120, 120-2) et le circuit échantillonneur bloqueur à quadrature (130, 130-2) sont configurés pour échantillonner le signal d'entrée à deux temps séparés dans le temps par approximativement 1/4 de la période (116) du signal d'entrée.

5. Appareil formeur de faisceau d'ultrasons selon l'une quelconque des revendications précédentes, dans lequel :

l'appareil formeur de faisceau d'ultrasons est configuré de sorte que la combinaison soit répétée sur le même ensemble de données d'écho à une pluralité de points pour former des données d'image complexes à cette distance.

6. Appareil formeur de faisceau d'ultrasons de la revendication 5, dans lequel : la pluralité de points sont le long d'une ligne à la distance d'intérêt.

7. Appareil formeur de faisceau d'ultrasons de la revendication 5, dans lequel :

la pluralité de points sont situés dans un plan à la distance d'intérêt et forment ainsi un scan C complexe.

8. Appareil formeur de faisceau d'ultrasons de la revendication 6, dans lequel :

le processus de formation de lignes d'image est répété à de nombreuses distances pour former une image échographique plane, étant éventuellement une image en mode B.

9. Appareil formeur de faisceau d'ultrasons de la revendication 7, dans lequel :

le processus de formation de plans d'image est répété à des distances multiples pour former une image 3D complexe.

10. Appareil formeur de faisceau d'ultrasons selon l'une quelconque des revendications précédentes, dans lequel :

l'enveloppe de l'amplitude est utilisée pour affichage à l'utilisateur.

11. Appareil formeur de faisceau d'ultrasons selon l'une quelconque des revendications précédentes, dans lequel :

le dispositif de rotation de phase (152, 152-2) est adapté pour focalisation pour compenser les différences de longueur de chemin entre différents éléments de réseau de transducteurs et un point focal.

12. Appareil formeur de faisceau d'ultrasons selon l'une quelconque des revendications précédentes, dans lequel :

l'apodiseur (146, 146-2) est adapté pour maintenir un équilibre raisonnable entre la résolution de lobe principal et les niveaux de lobe latéral dans la réponse du système.

13. Appareil formeur de faisceau d'ultrasons selon l'une quelconque des revendications précédentes, dans lequel :

le transducteur (110, 110-2) fait partie d'un réseau de transducteurs (202) utilisé pour imagerie, ledit réseau de transducteurs (202) étant constitué d'une pluralité d'éléments de réseau, lesdits éléments transducteurs étant placés dans une configuration linéaire choisie dans le groupe constitué de : un réseau linéaire, un réseau à commande de phase, et un réseau curviligne.

14. Appareil formeur de faisceau d'ultrasons selon l'une quelconque des revendications précédentes, dans lequel :

le transducteur (110, 110-2) fait partie d'un réseau de transducteurs (202) utilisé pour imagerie, ledit réseau de transducteurs (202) étant constitué du groupe constitué de :

une pluralité d'éléments agencés dans une configuration 2D inégalement échantillonnée ou un réseau 1,5-D ;
une pluralité d'éléments qui sont placés dans une configuration 2D également échantillonnée ; une fraction des éléments du réseau sont utilisés de sorte que le réseau résultant soit un réseau 2D à faible densité ;
tous les éléments du réseau sont utilisés de sorte que le réseau résultant est un réseau 2D totalement échantillonné ;
des connexions sont établies avec des éléments de réseau individuels de sorte que les éléments individuels

puissent être utilisés pour transmission ou réception, mais pas les deux, de manière à éliminer le besoin d'un circuit de réception ; et
seule une fraction des éléments sont utilisés pour former un point d'image donné.

15. Appareil formeur de faisceau d'ultrasons selon l'une quelconque des revendications précédentes, dans lequel :

l'appareil formeur de faisceau d'ultrasons est configuré de sorte que la combinaison soit répétée pour différentes fractions de l'ouverture de manière à obtenir des vues redondantes multiples du même emplacement cible.

16. Appareil formeur de faisceau d'ultrasons de la revendication 15, dans lequel :

les vues multiples sont moyennées après avoir acquis leurs amplitudes de manière à réduire l'aspect de granularité dans l'image résultante.

17. Appareil formeur de faisceau d'ultrasons de l'une quelconque des revendications précédentes, dans lequel :

l'opération d'échantillonnage est effectuée à des distances multiples pour un événement de transmission unique, de manière à augmenter le taux de formation d'image.

18. Procédé de formage de faisceau pour imagerie échographique, comprenant :

la génération d'un signal de sortie (104) ayant une amplitude de sortie (106) à un temps de sortie (108) ;
la transduction dudit signal de sortie (104) en ultrasons de sortie (112, 112-2) ;
la réception d'au moins une partie d'ultrasons de sortie réfléchis,
la transduction desdits ultrasons réfléchis en signaux d'entrée (114, 114-2) ayant une période (116) ;
la conversion A/N de chacun desdits signaux d'entrée (114, 114-2) en paire, ou série temporelle de paires de valeurs en phase numériques (128, 128-2) et de quadrature numériques (136, 136-2) ; et la combinaison desdites valeurs en phase numériques (128, 128-2) et de quadrature numériques (136, 136-2) pour obtenir un échantillon en phase/de quadrature focalisé ;
caractérisé en ce qu'il comprend les étapes supplémentaires de :

échantillonnage dudit signal d'entrée (114, 114-2) à un temps d'entrée (122) pour produire une amplitude en phase (124, 124-2) dudit signal d'entrée (114, 114-2) sensiblement audit temps d'entrée ;
échantillonnage dudit signal d'entrée (114, 114-2) à sensiblement un quart de ladite période après ledit temps d'entrée pour produire une amplitude de quadrature (132, 132-2) dudit signal d'entrée (114, 114-2) à sensiblement un quart de ladite période après ledit temps d'entrée ;
dans lequel ladite amplitude en phase (124, 124-2) et ladite amplitude de quadrature (132, 132-2) sont soumises à conversion A/N pour générer ladite paire, ou la série temporelle de paires de valeurs de quadrature en phase numérique (128, 128-2) et valeurs de quadrature numériques (136, 136-2).

19. Procédé de formage de faisceau pour imagerie échographique de la revendication 18, comprenant en outre l'amplification dudit signal de sortie (104).

20. Procédé de formage de faisceau pour imagerie échographique de la revendication 18, comprenant en outre une opération choisie dans le groupe constitué de :

l'amplification dudit signal d'entrée (114, 114-2) ;
la préamplification dudit signal d'entrée (114, 114-2) ; et
le stockage dudit signal d'entrée (114, 114-2).

21. Procédé de formage de faisceau pour imagerie échographique de la revendication 18, comprenant en outre la répétition du procédé de formage de faisceau pour produire une pluralité de points d'image formant une image.

22. Procédé de formage de faisceau pour imagerie échographique de la revendication 21, comprenant en outre la focalisation de ladite pluralité de points d'image.

23. Procédé de formage de faisceau pour imagerie échographique de la revendication 22, dans lequel ladite focalisation est répétée sur lesdits ultrasons de sortie réfléchis au niveau de ladite pluralité de points d'image.

24. Procédé de formage de faisceau pour imagerie échographique de la revendication 21, dans lequel la pluralité de points d'image sont le long d'une ligne à une distance d'intérêt.

25. Procédé de formage de faisceau pour imagerie échographique de la revendication 24, dans lequel la ligne est formée à une pluralité de distances pour former une image plane.

26. Procédé de formage de faisceau pour imagerie échographique de la revendication 25, dans lequel l'image plane est une image en mode B.

27. Procédé de formage de faisceau pour imagerie échographique de la revendication 21, dans lequel la pluralité de points d'image sont situés dans un plan à une plage d'intérêt.

28. Procédé de formage de faisceau pour imagerie échographique de la revendication 27, dans lequel la pluralité de points d'image forment un scan C.

29. Procédé de formage de faisceau pour imagerie échographique de la revendication 27, dans lequel le plan est formé à des distances multiples.

30. Procédé de formage de faisceau pour imagerie échographique de la revendication 27, dans lequel les .plans forment une image 3D complexe.

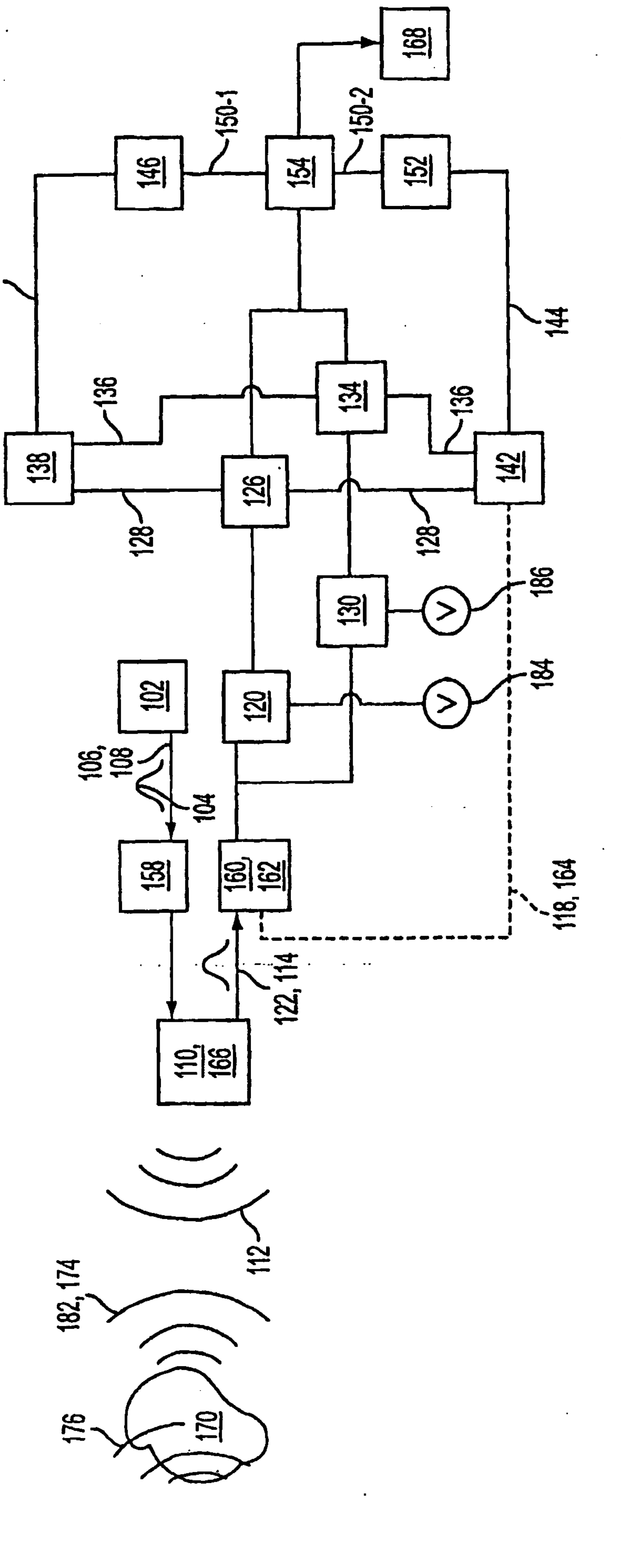

102
106, 108
104
158
110, 166
160, 162
122, 114
118, 164
112
182, 174
176
170
120
126
130
134
138
142
146
152
154
168
128
136
140
144
150-1
150-2
184
186
V

FIG. 1

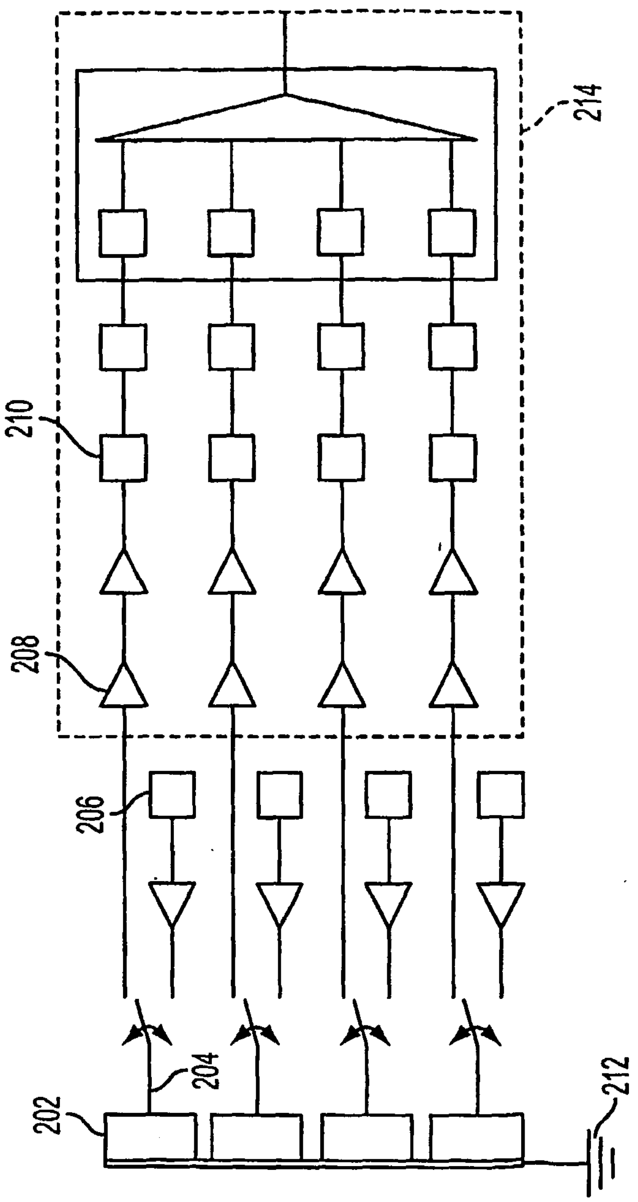


FIG. 2

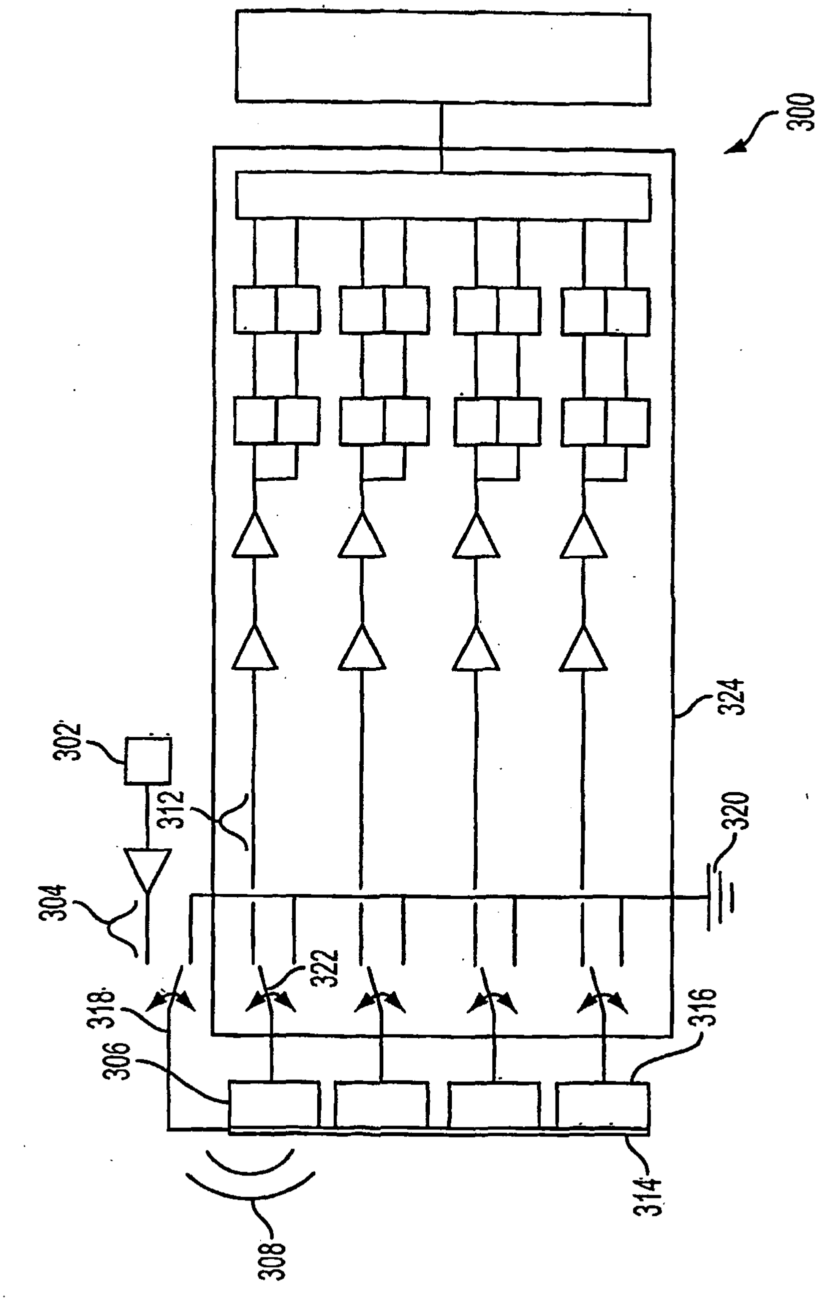
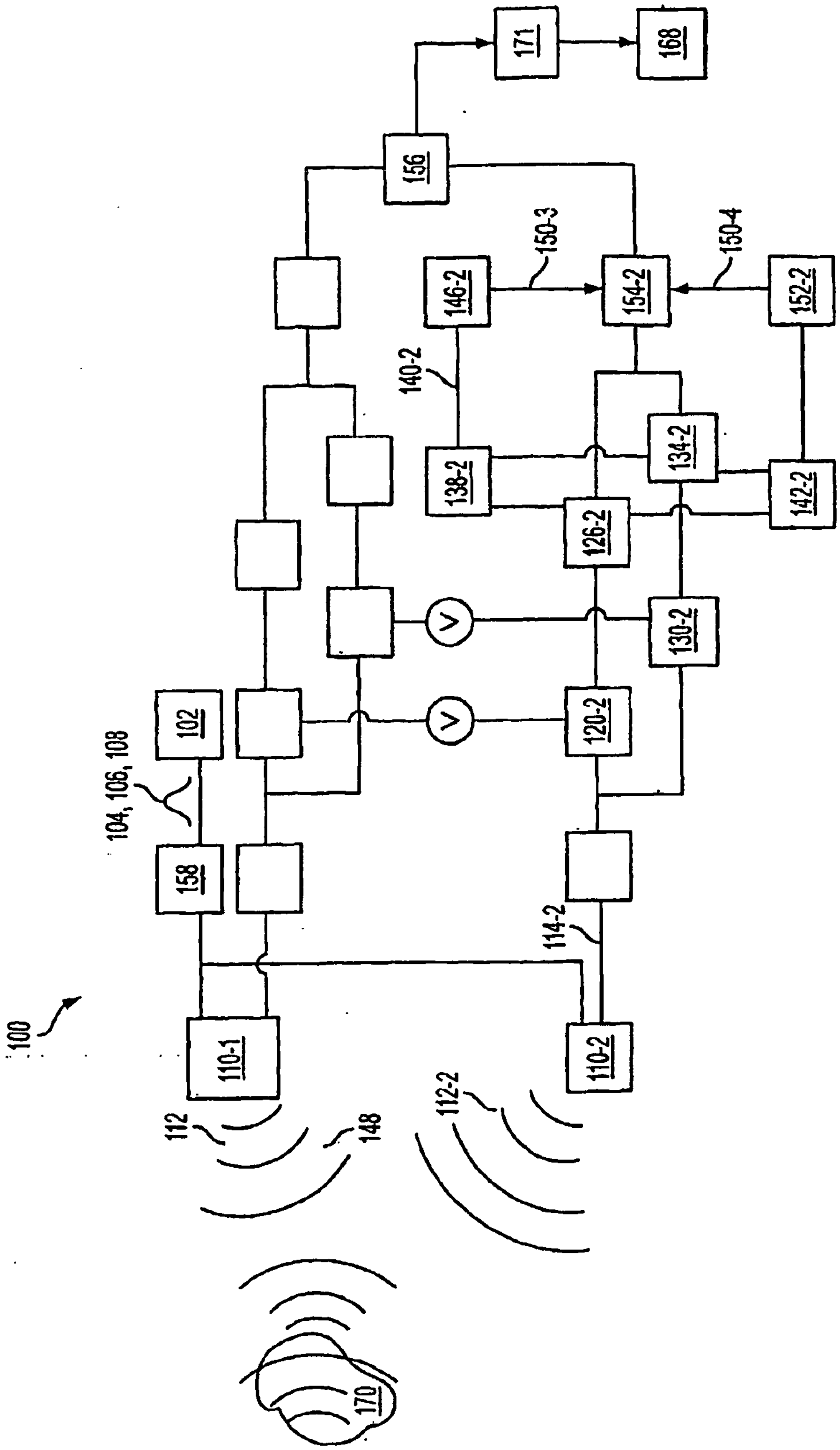
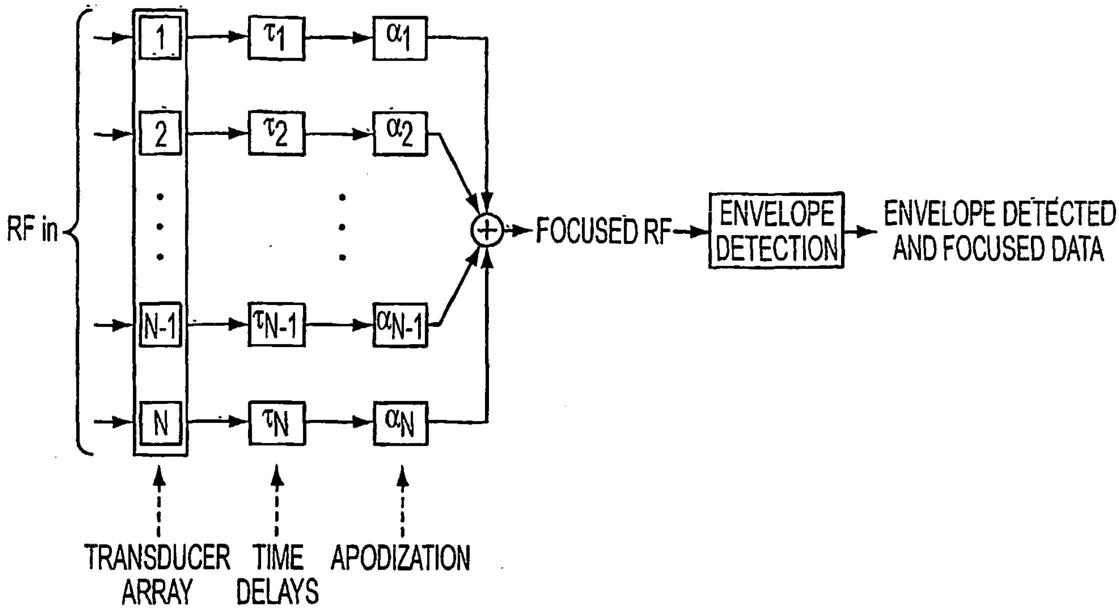


FIG. 3

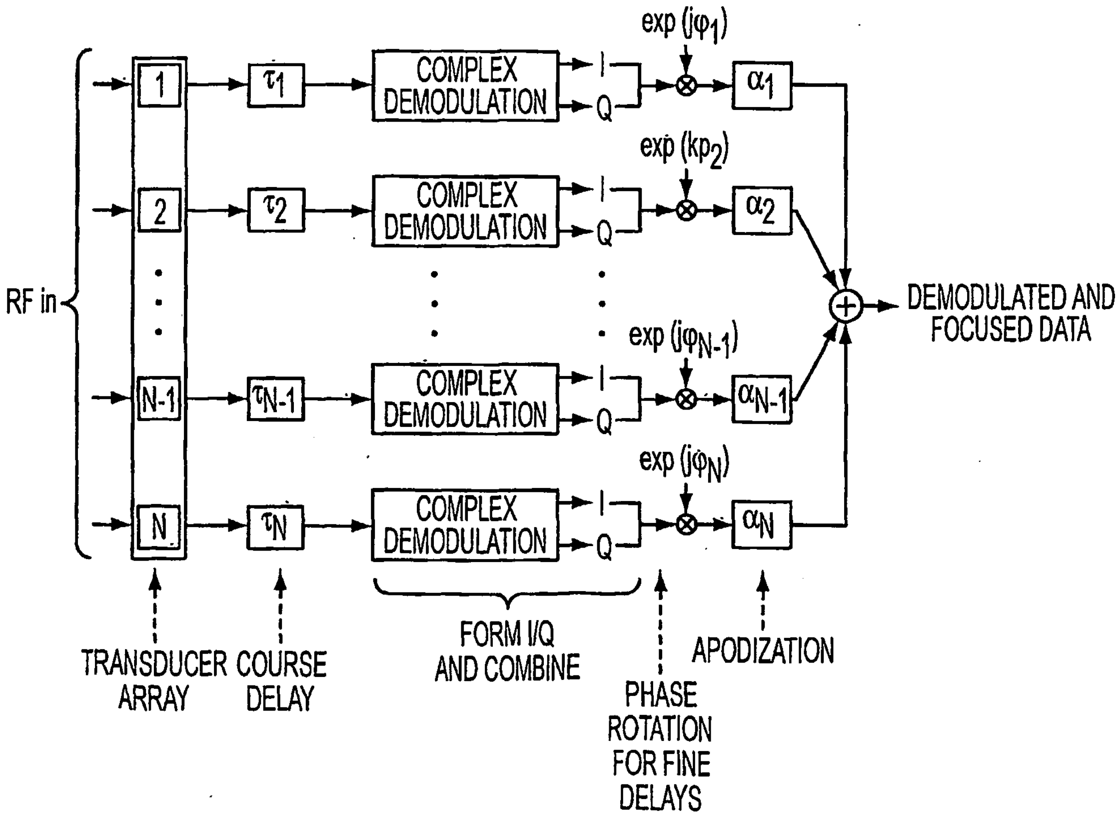


100
110-1
110-2
112
148
112-2
114-2
158
102
104, 106, 108
120-2
130-2
126-2
134-2
138-2
142-2
140-2
146-2
154-2
152-2
150-3
150-4
156
171
168
170
V
V

FIG. 4



CONVENTIONAL ART
FIG. 5A



CONVENTIONAL ART
FIG. 5B

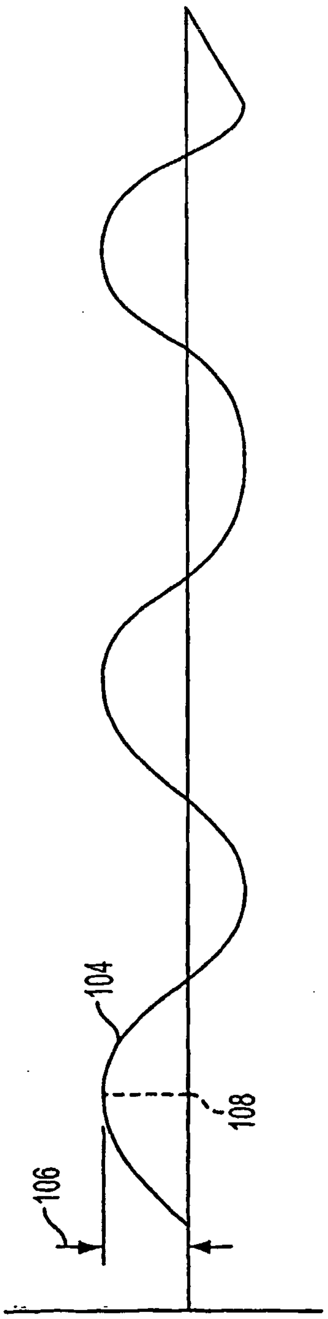


FIG. 6A

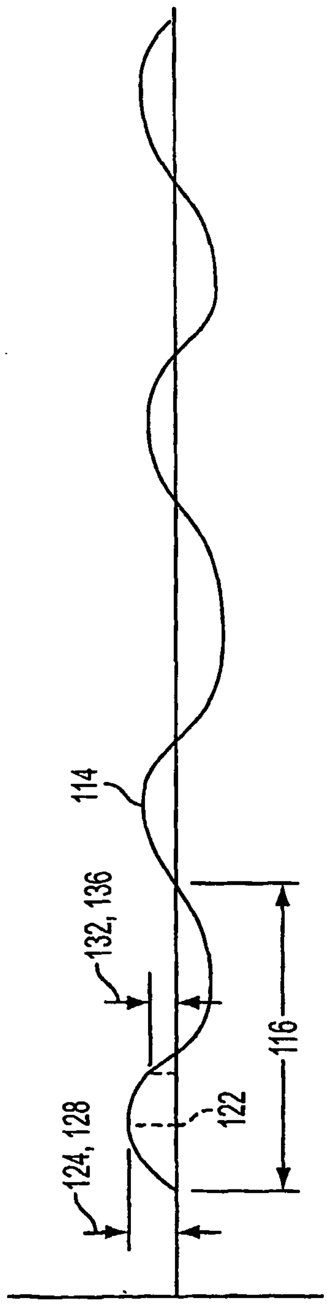


FIG. 6B

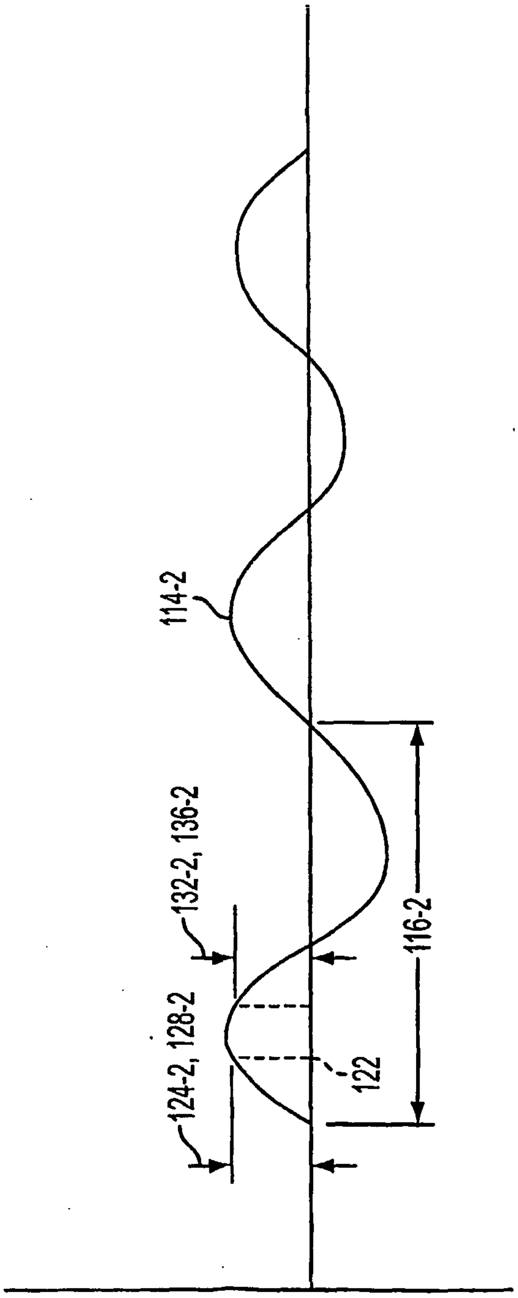


FIG. 6C

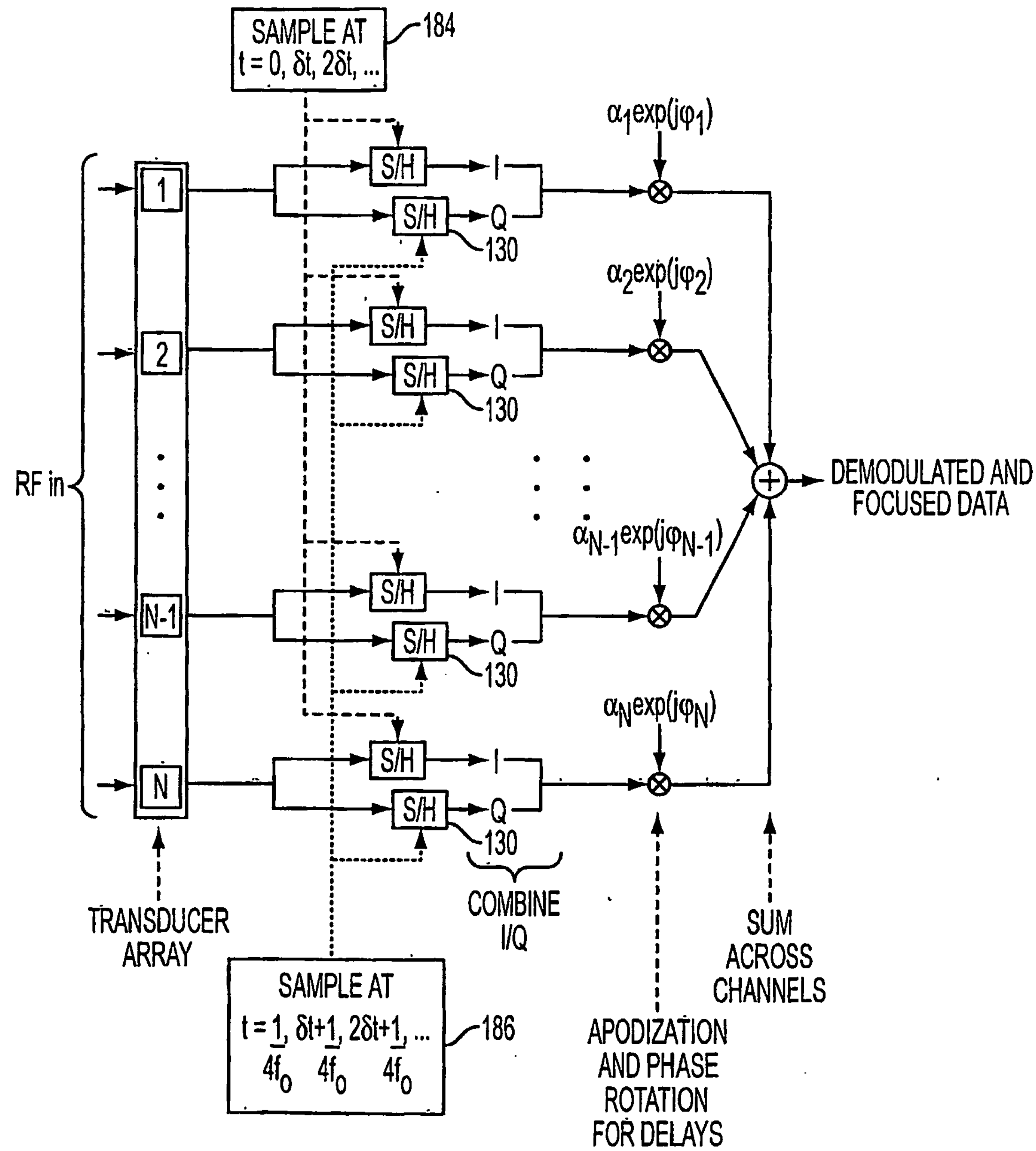

SAMPLE AT
t = 0, δt, 2δt, ...
184
α1exp(jφ1)
α2exp(jφ2)
αN-1exp(jφN-1)
αNexp(jφN)
1
2
N-1
N
S/H
I
Q
130
RF in
DEMODULATED AND
FOCUSED DATA
TRANSDUCER
ARRAY
COMBINE
I/Q
SUM
ACROSS
CHANNELS
APODIZATION
AND PHASE
ROTATION
FOR DELAYS
SAMPLE AT
t = 1/4f0, δt+1/4f0, 2δt+1/4f0, ...
186

FIG. 7

REFERENCES CITED IN THE DESCRIPTION

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patsnap

专利名称(译)	超声成像光束成形装置和方法		
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申请(专利权)人(译)	VIRGINIA专利大学基金会		
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其他公开文献	EP1622514A2 EP1622514B1		
外部链接	Espacenet		

摘要(译)

在一些说明性实施例中，来自超声成像波束成形器装置中的换能器的输入信号被应用于同相采样和保持以及正交采样和保持。正交采样保持可以在同相采样保持之后的四分之一周期内计时。采样保持器的输出应用于同相和正交模数转换器。幅度计算器接收同相和正交数字值，并输出幅度。相位计算器接收同相和正交数字值，并输出相位。变迹器应用输出信号的幅度和幅度之间的差值，并且将第一照度应用于与差值基本成比例的图像点，并且相位旋转器将第二照度应用于与相位成大比例的图像点。

$$\hat{Q}(nT) = I\left\{nT + \frac{\pi}{2\omega_0}\right\}$$

$$= A\left(nT + \frac{\pi}{2\omega_0}\right)\cos\left(\omega_0\left(nT + \frac{\pi}{2\omega_0}\right) - \phi\left(nT + \frac{\pi}{2\omega_0}\right)\right)$$