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(54) **ULTRASOUND TRANSDUCER DEVICE AND METHOD OF OPERATION**

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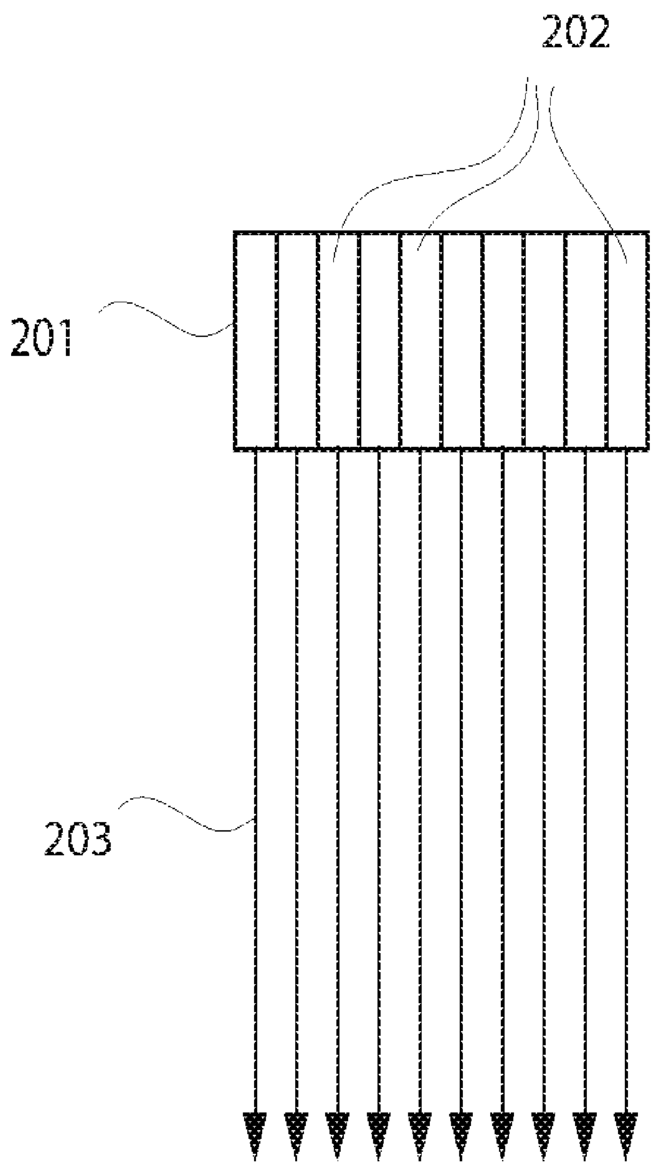
(57) **ABSTRACT**

A method and device for ultrasound imaging including an array of transducers wherein the distance between adjacent transducers is greater than the minimum separation of adjacent scanlines required to produce an ultrasound image of a selected resolution. The transducer array is adapted to be mechanically moved so that the array may be swept to generate scanlines.



Prior Art

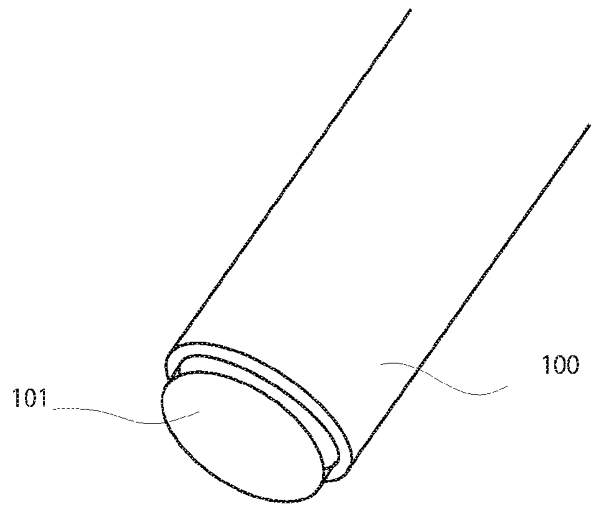


Figure 1a

Prior Art

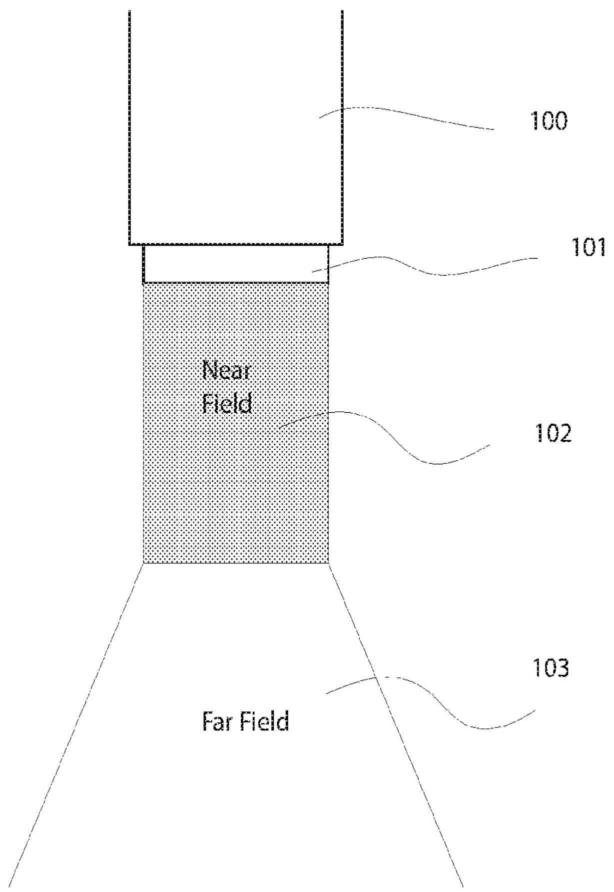


Figure 1b

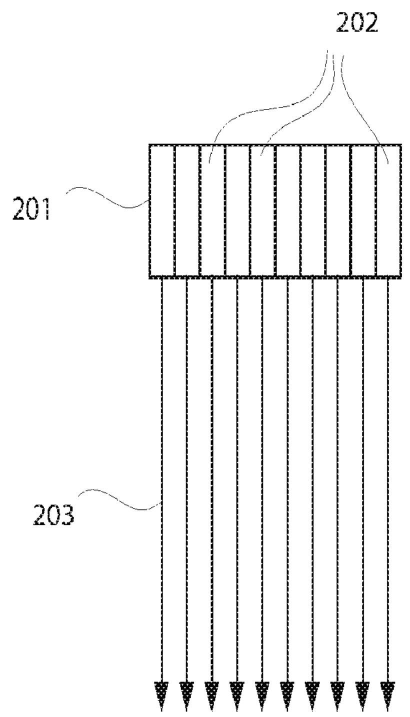


Figure 2a

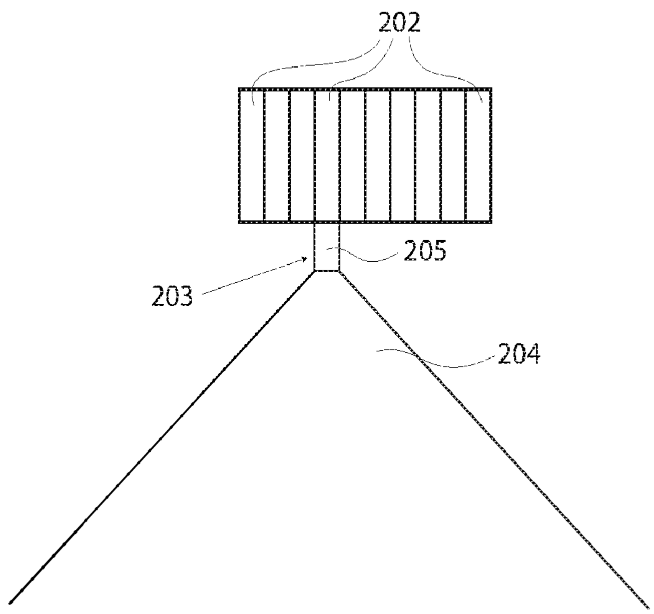


Figure 2b

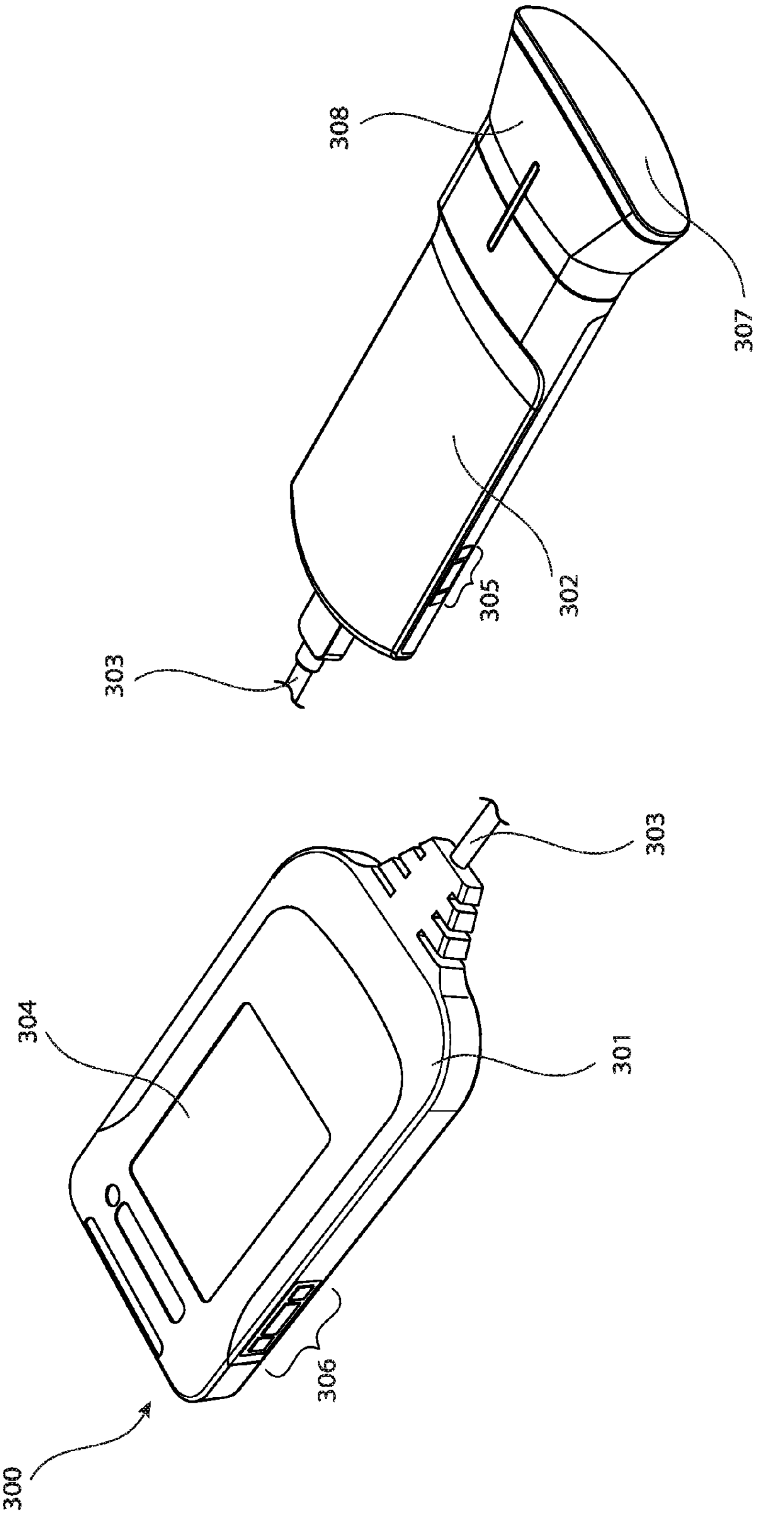


Figure 3

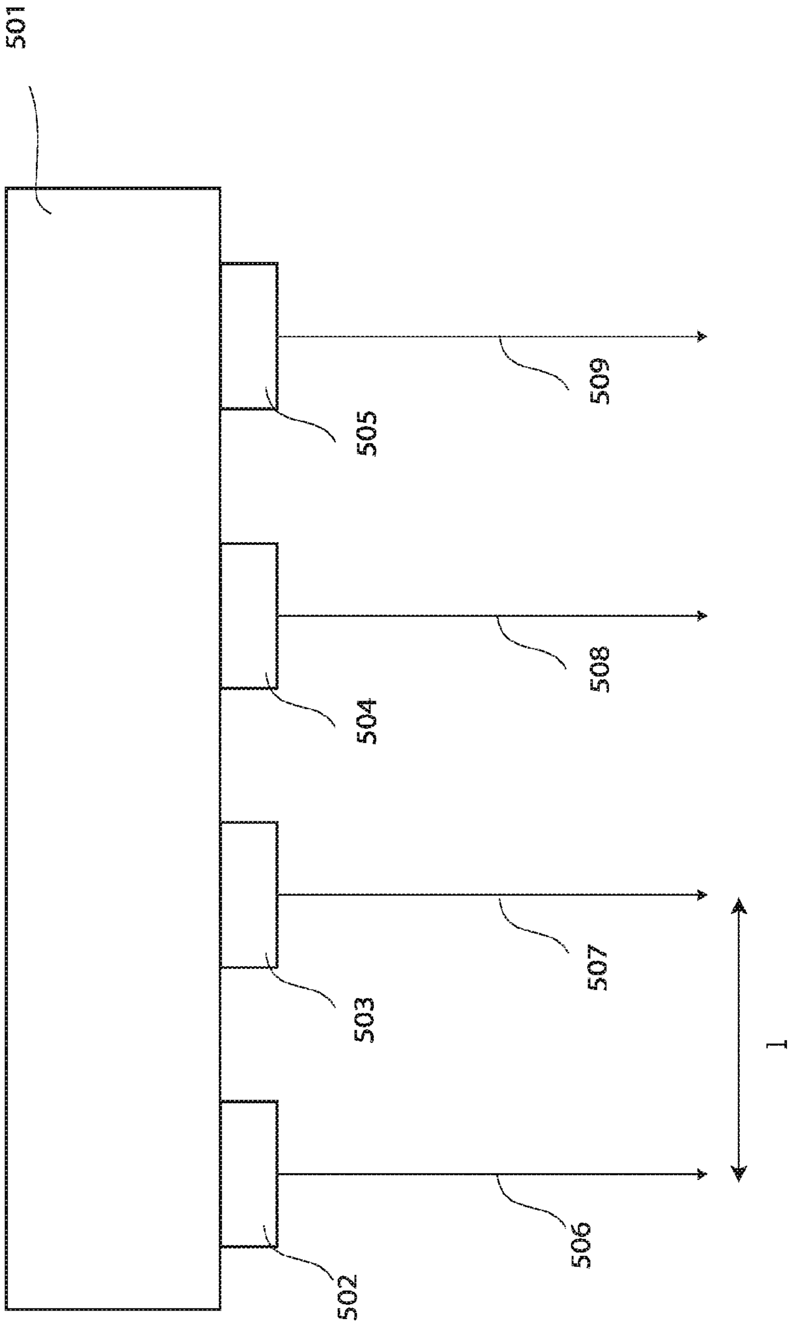


Figure 4

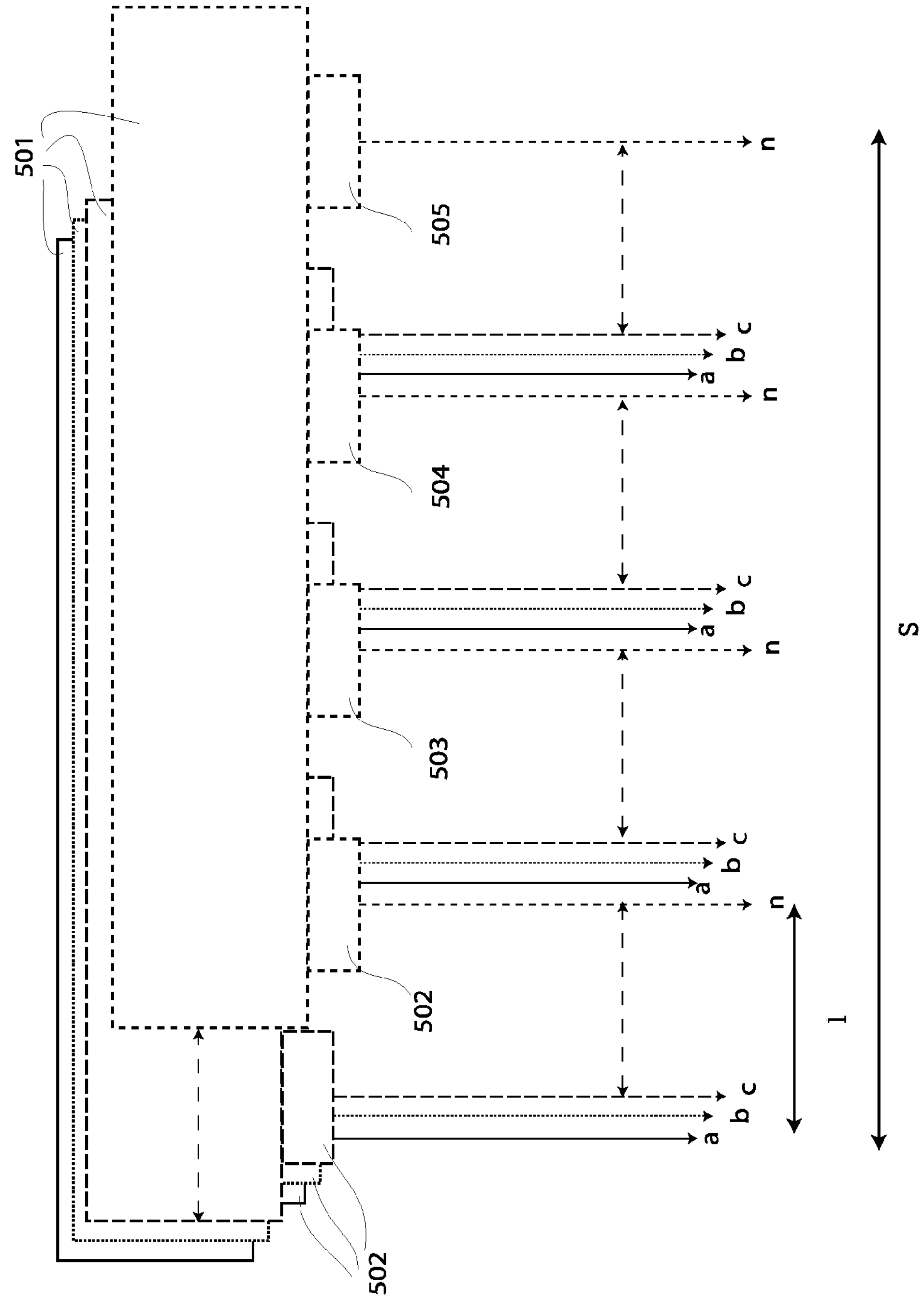


Figure 5

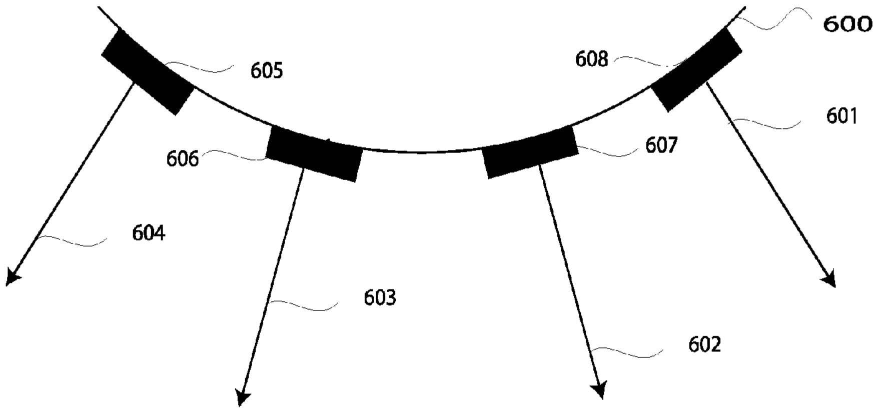


Figure 6a

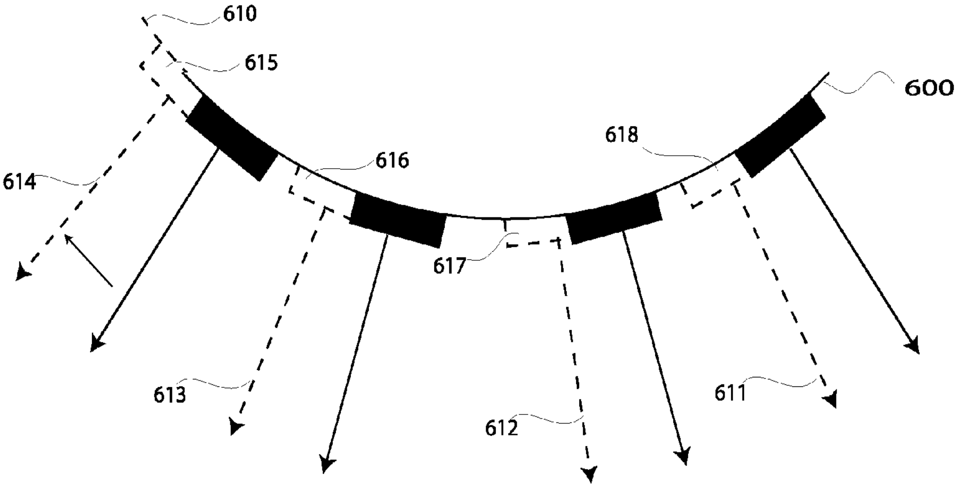


Figure 6b

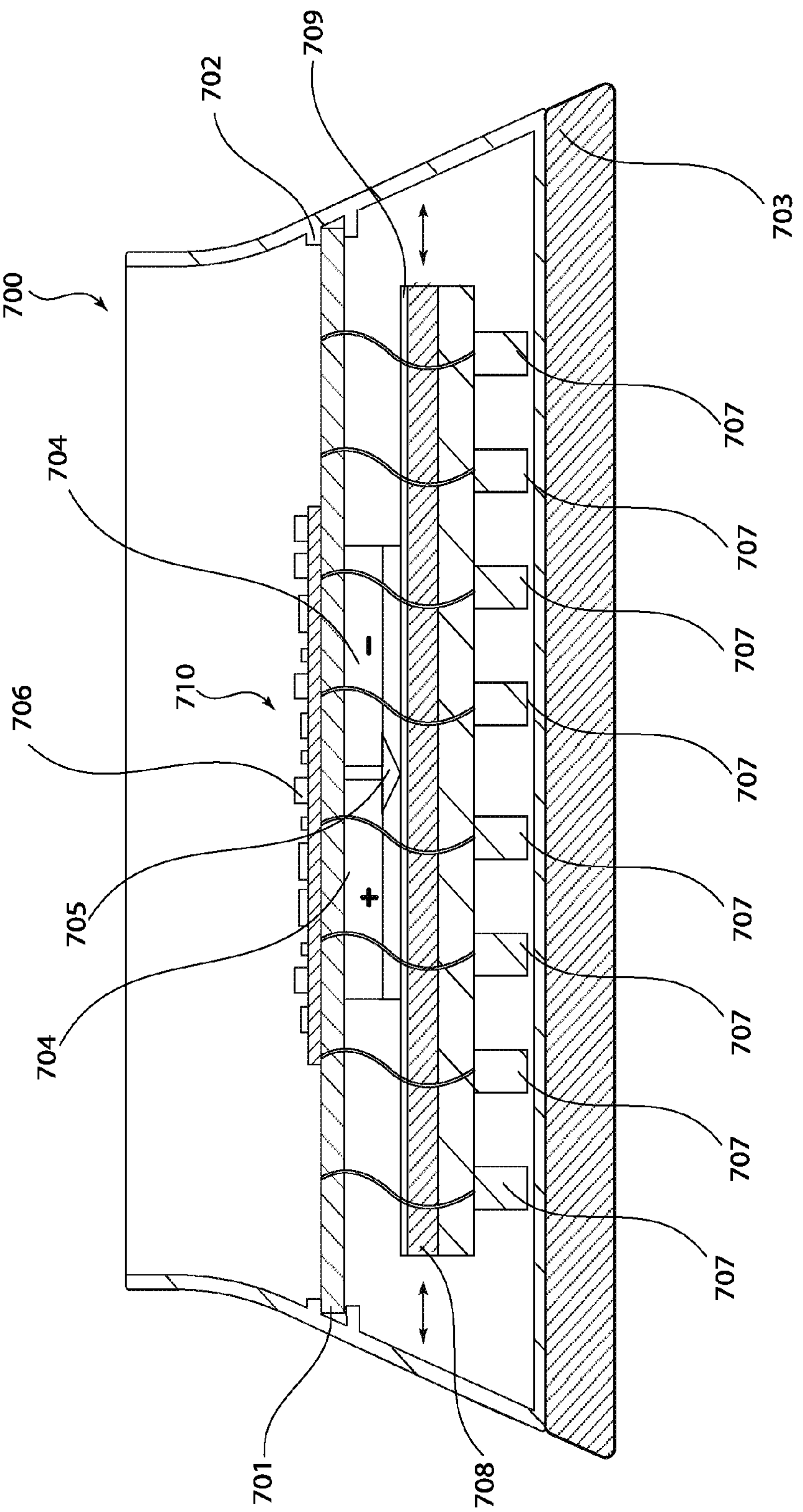

700
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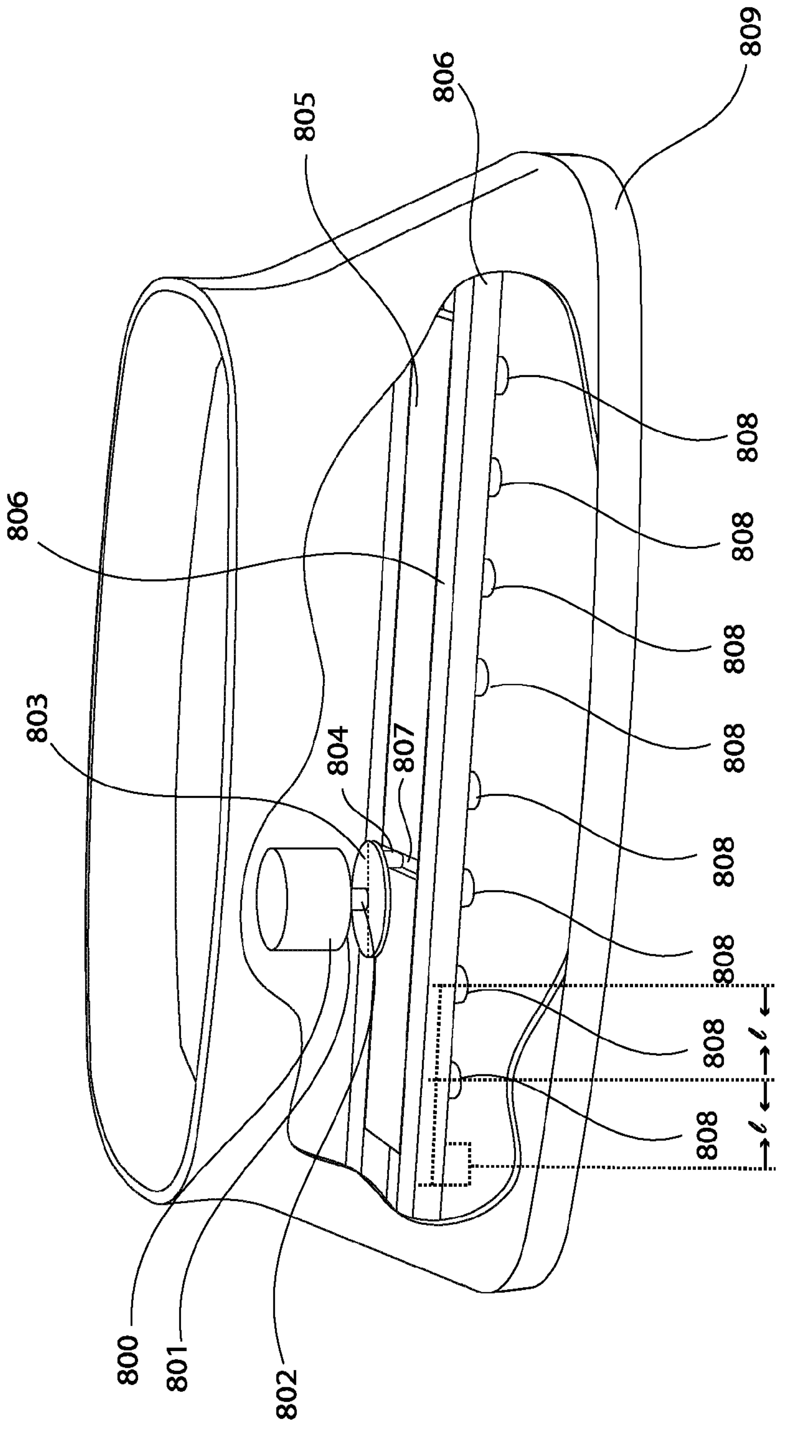
Figure 7


800
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ℓ

Figure 8

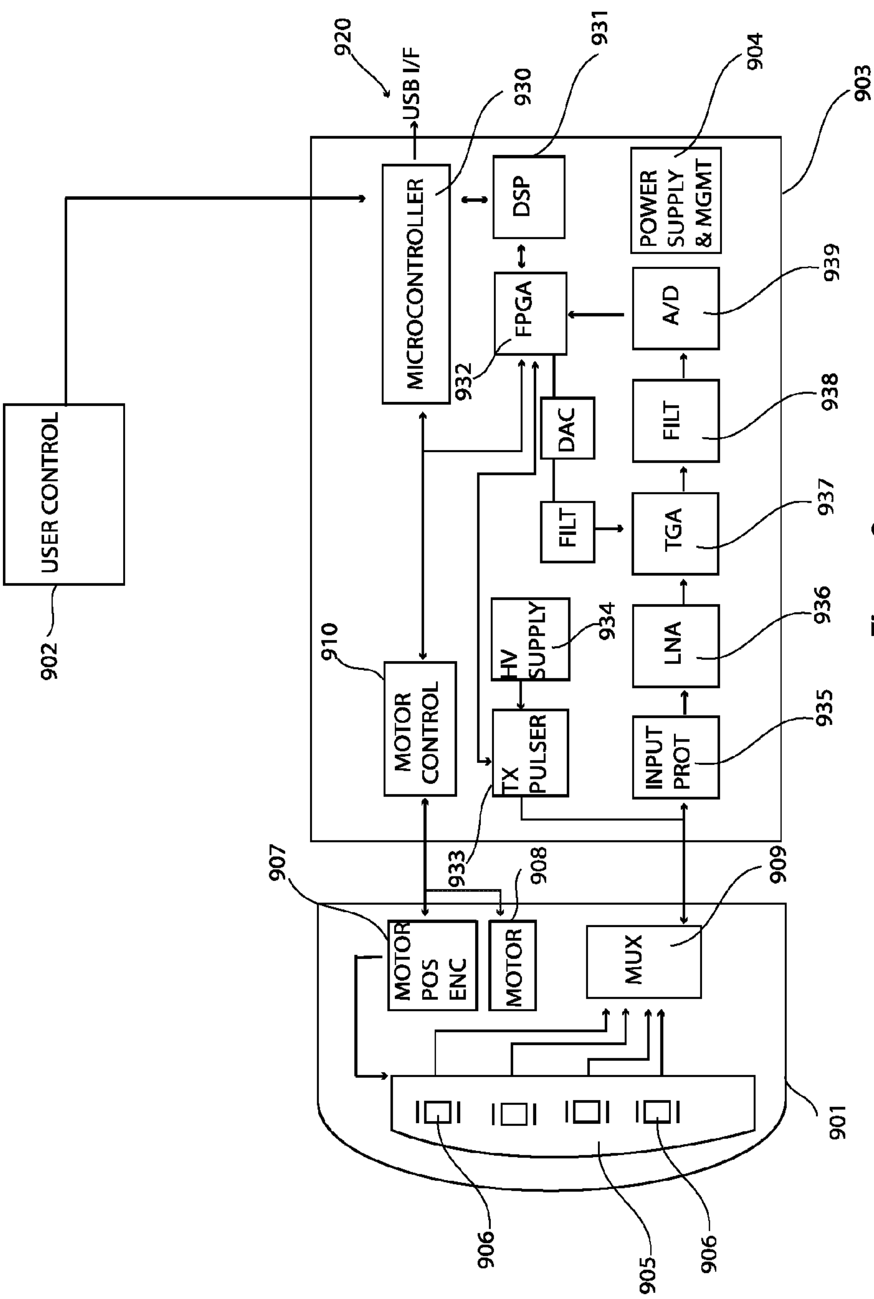
USER CONTROL
902
920
USB I/F
930
931
904
903
MICROCONTROLLER
DSP
POWER SUPPLY & MGMT
FPGA
932
A/D
939
DAC
FILT
938
FILT
TGA
937
HV SUPPLY
934
LNA
936
910
MOTOR CONTROL
TX PULSER
INPUT PROT
935
907
933
908
909
MOTOR POS ENC
MOTOR
MUX
901
906
905
906

Figure 9

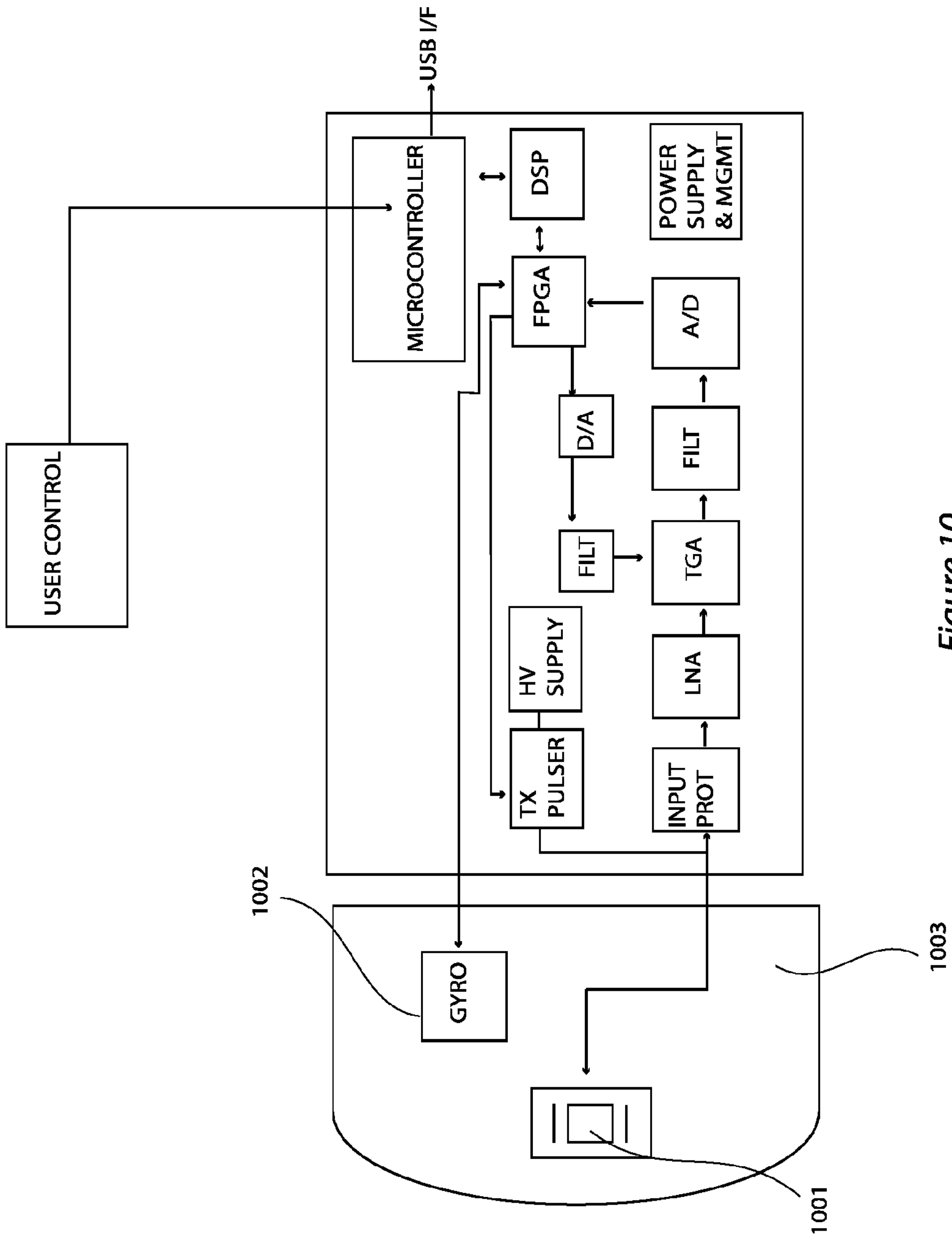
USER CONTROL
USB I/F
MICROCONTROLLER
DSP
POWER SUPPLY & MGMT
FPGA
D/A
A/D
FILT
FILT
TGA
LNA
HV SUPPLY
TX PULSER
INPUT PROT
GYRO
1001
1002
1003

Figure 10

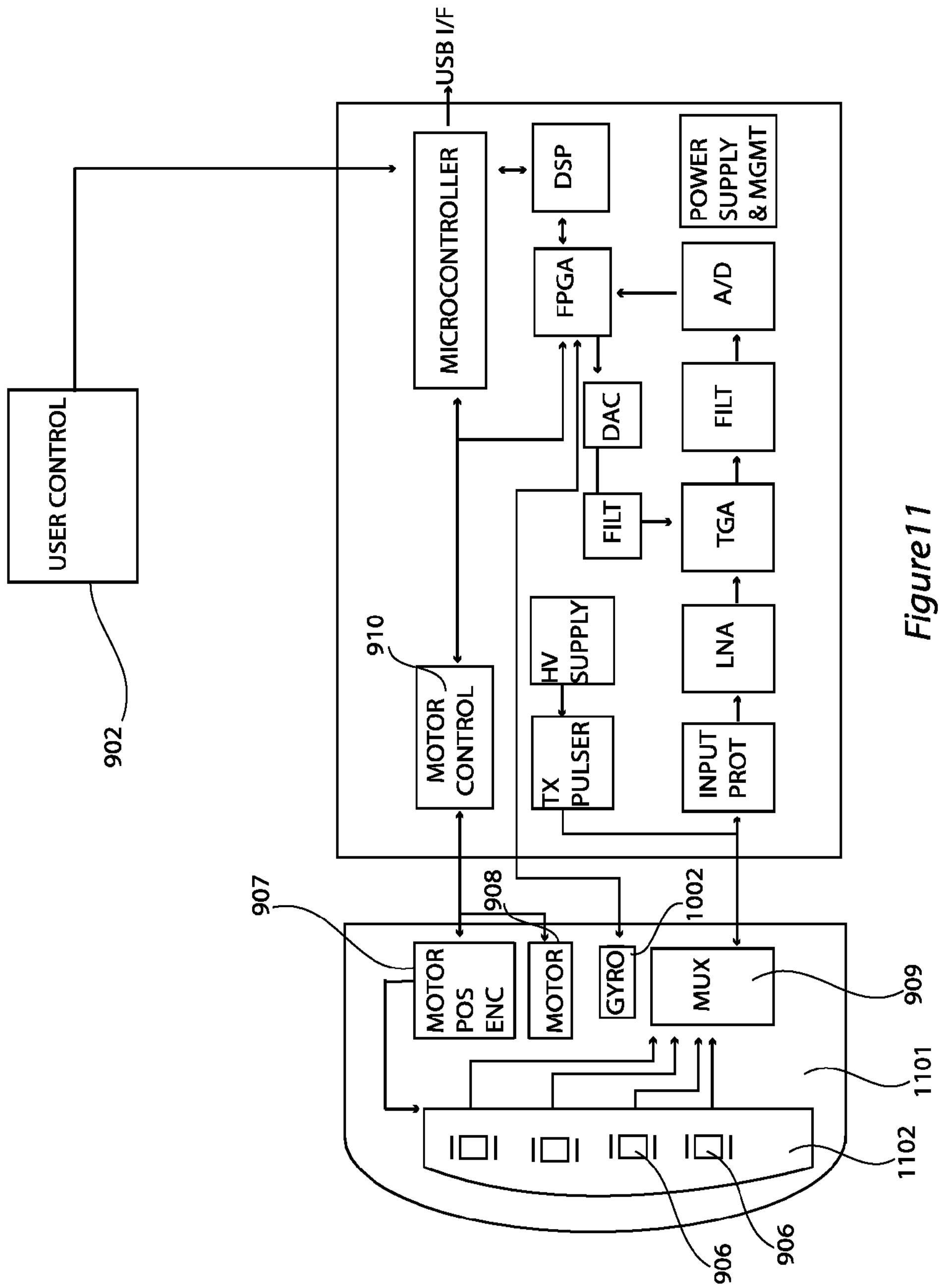
USB I/F
MICROCONTROLLER
DSP
POWER SUPPLY & MGMT
FPGA
A/D
DAC
FILT
FILT
TGA
LNA
INPUT PROT
HV SUPPLY
TX PULSER
MOTOR CONTROL
910
USER CONTROL
902
907
908
1002
MOTOR POS ENC
MOTOR
GYRO
MUX
909
1101
1102
906
906

Figure11

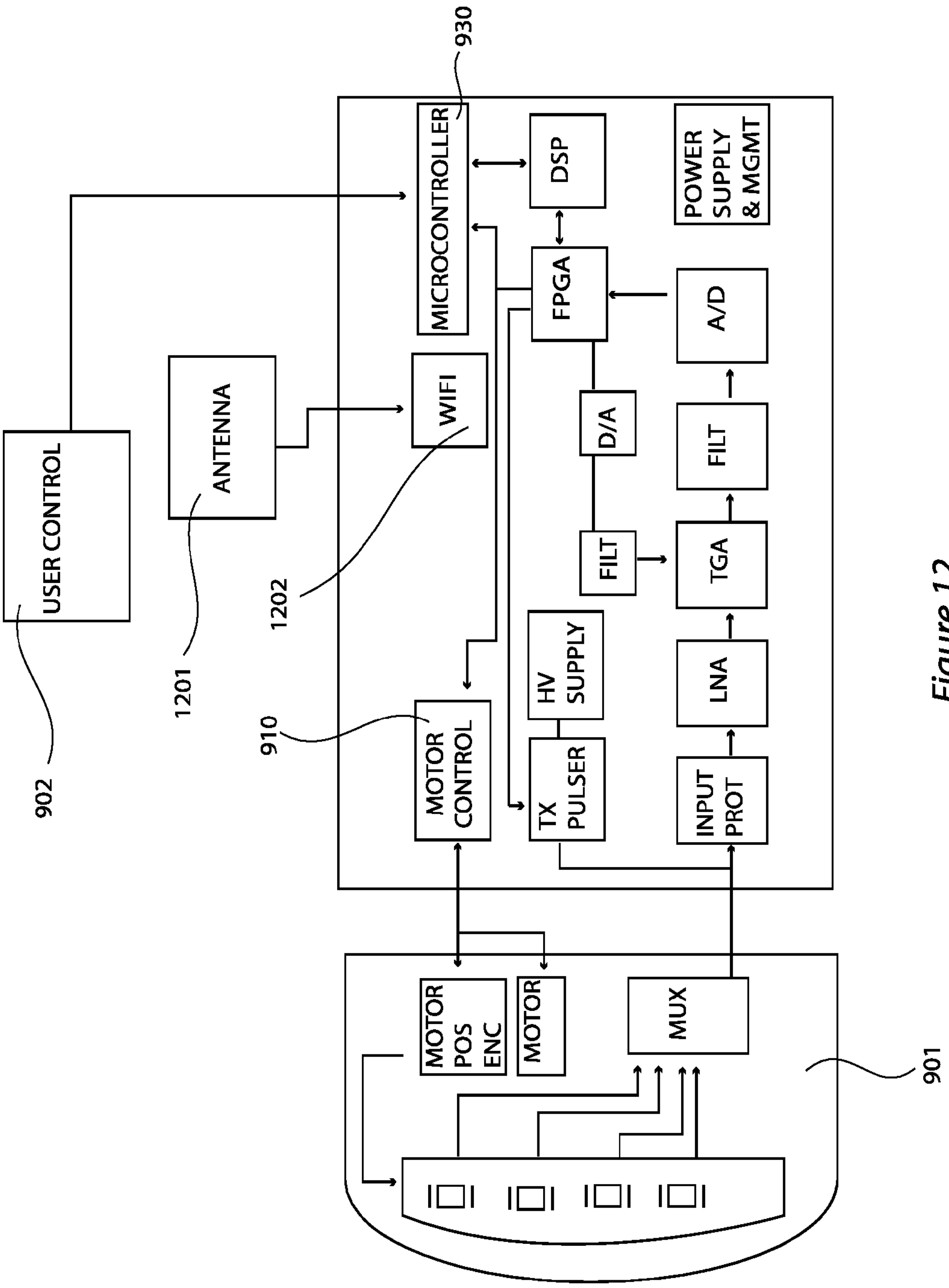
930
MICROCONTROLLER
DSP
POWER SUPPLY & MGMT
FPGA
A/D
WIFI
D/A
FILT
ANTENNA
USER CONTROL
FILT
TGA
HV SUPPLY
LNA
1202
1201
910
MOTOR CONTROL
TX PULSER
INPUT PROT
902
MOTOR POS ENC
MOTOR
MUX
901

Figure 12

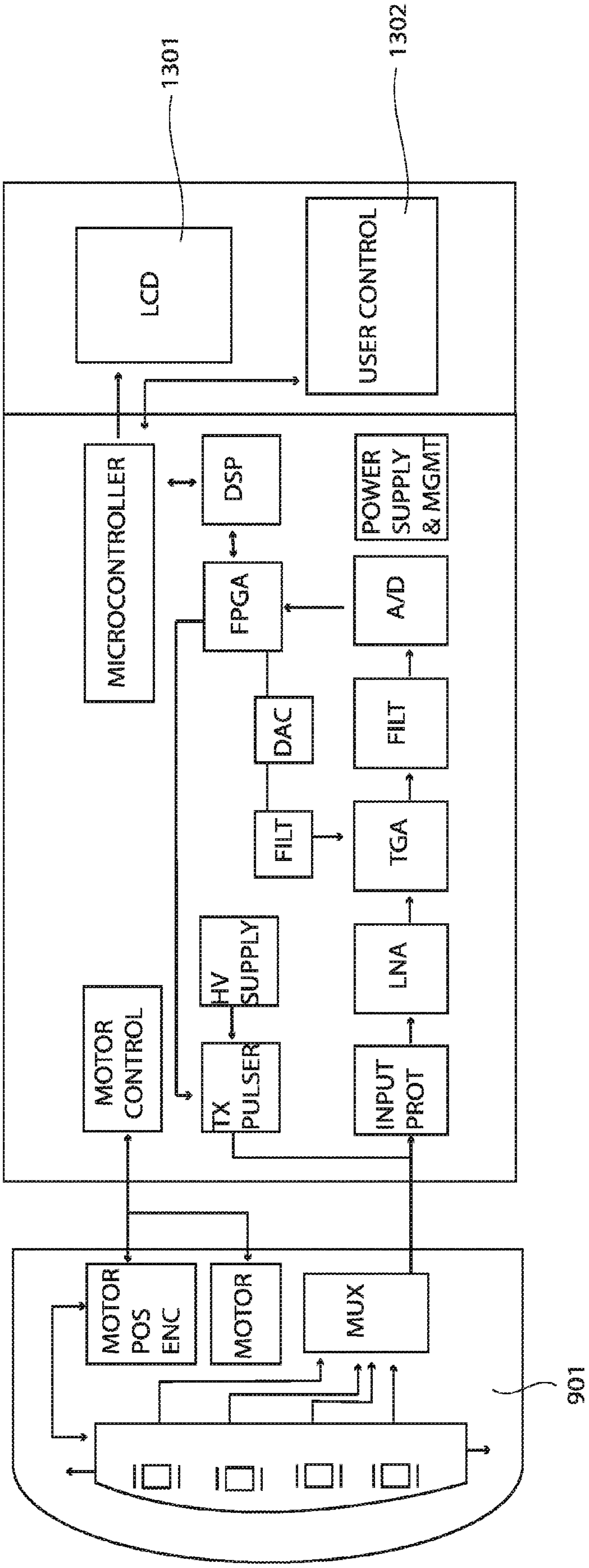
1301
1302
LCD
USER CONTROL
MICROCONTROLLER
DSP
POWER SUPPLY & MGMT
FPGA
A/D
DAC
FILT
FILT
TGA
HV SUPPLY
LNA
MOTOR CONTROL
TX PULSER
INPUT PROT
MOTOR POS ENC
MOTOR
MUX
901

Figure 13

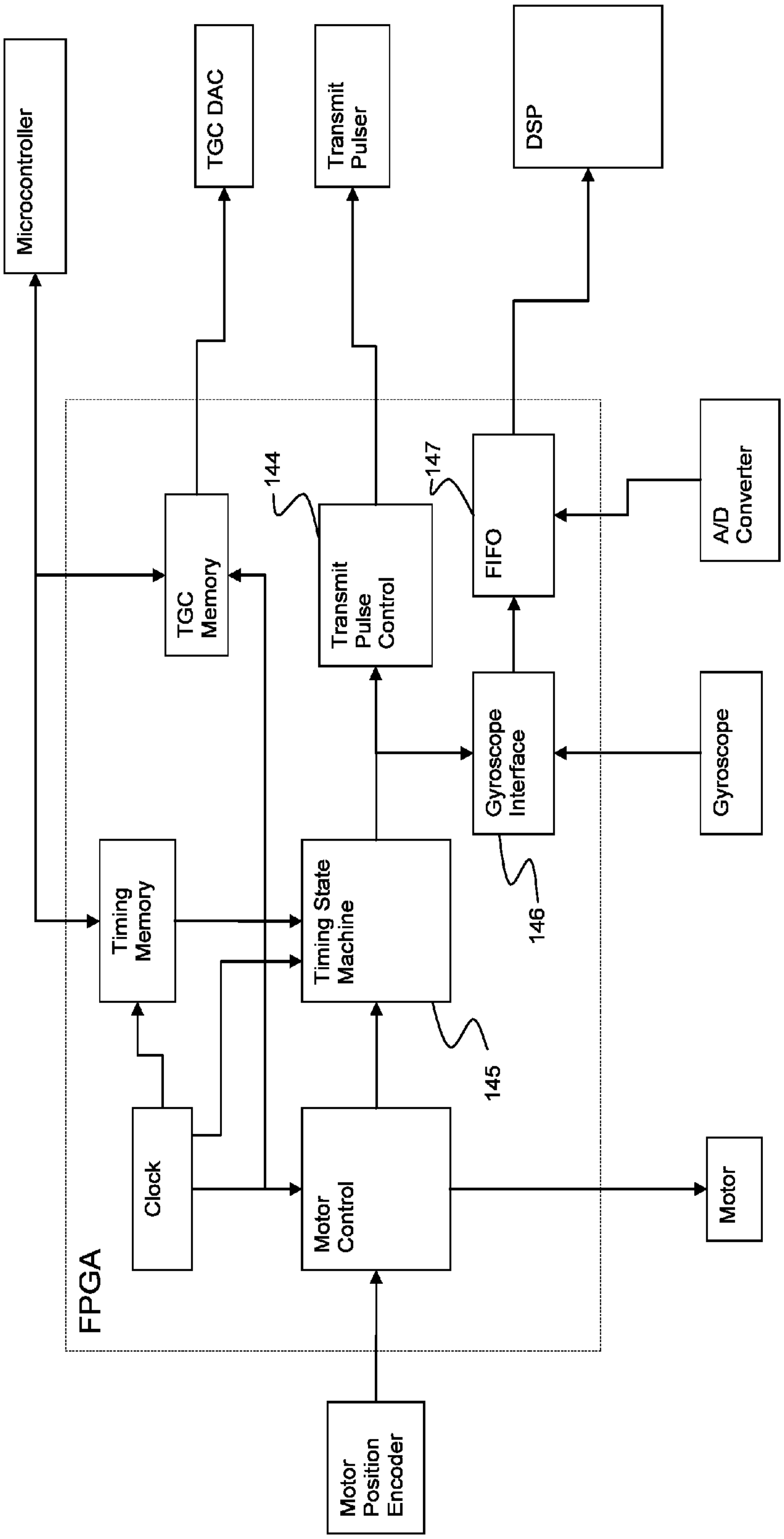
Microcontroller
TGC DAC
Transmit Pulser
DSP
FPGA
TGC Memory
Transmit Pulse Control
FIFO
A/D Converter
144
147
Timing Memory
Timing State Machine
Gyroscope Interface
Gyroscope
146
145
Clock
Motor Control
Motor
Motor Position Encoder

Figure 14

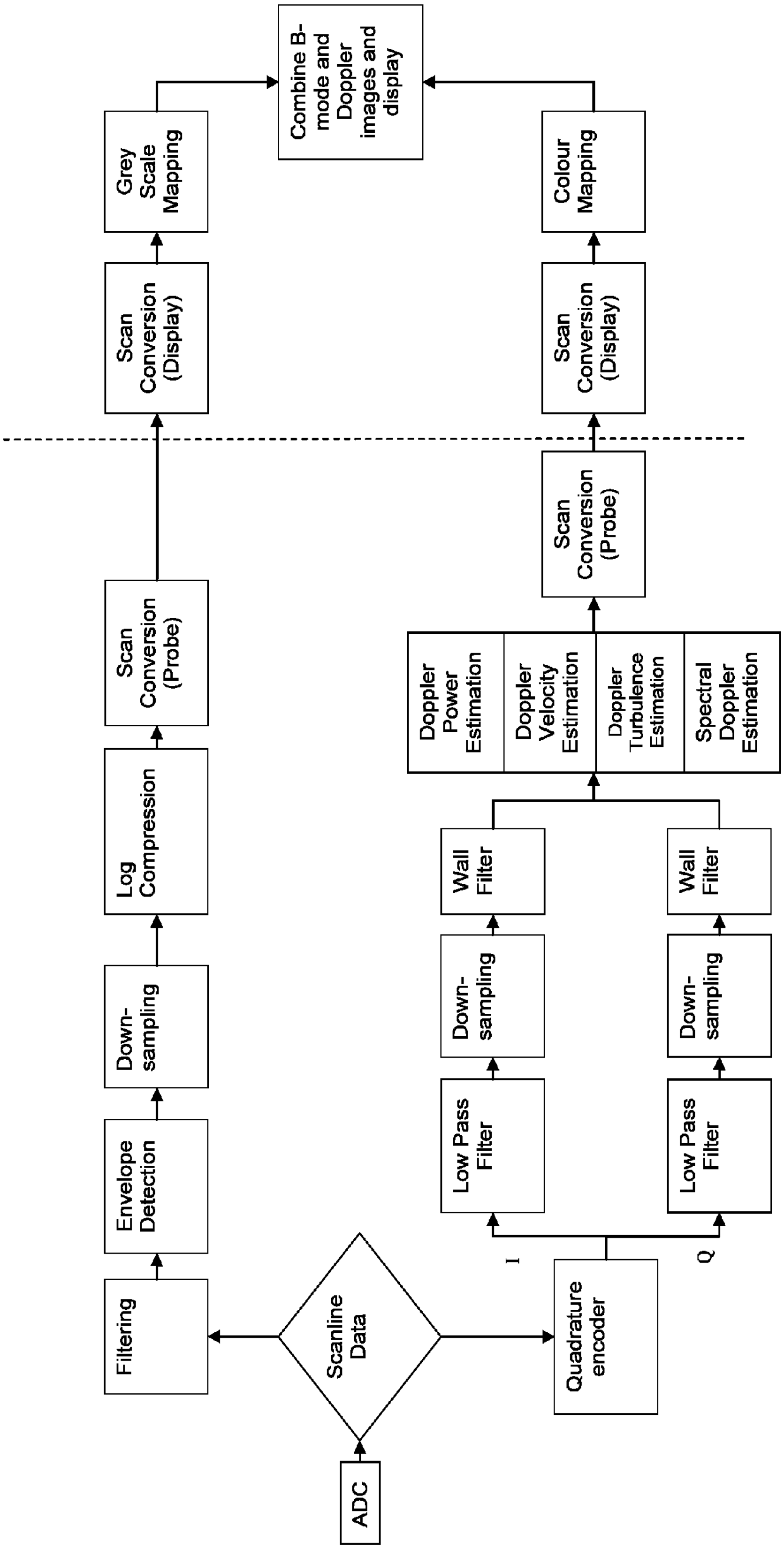
ADC
Scanline Data
Filtering
Envelope Detection
Down-sampling
Log Compression
Scan Conversion (Probe)
Scan Conversion (Display)
Grey Scale Mapping
Combine B-mode and Doppler images and display
Quadrature encoder
I
Q
Low Pass Filter
Down-sampling
Wall Filter
Low Pass Filter
Down-sampling
Wall Filter
Doppler Power Estimation
Doppler Velocity Estimation
Doppler Turbulence Estimation
Spectral Doppler Estimation
Scan Conversion (Probe)
Scan Conversion (Display)
Colour Mapping

Figure 15

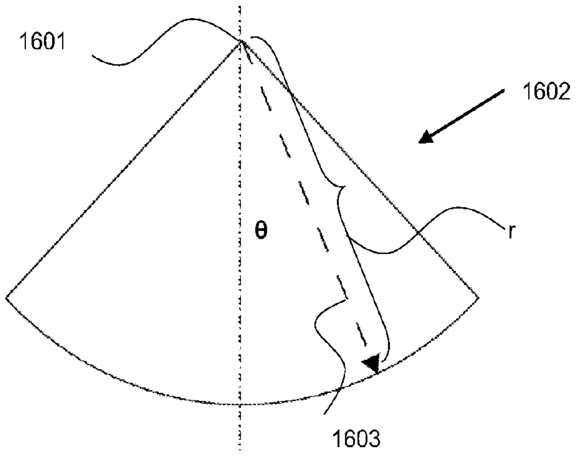

Figure 16a

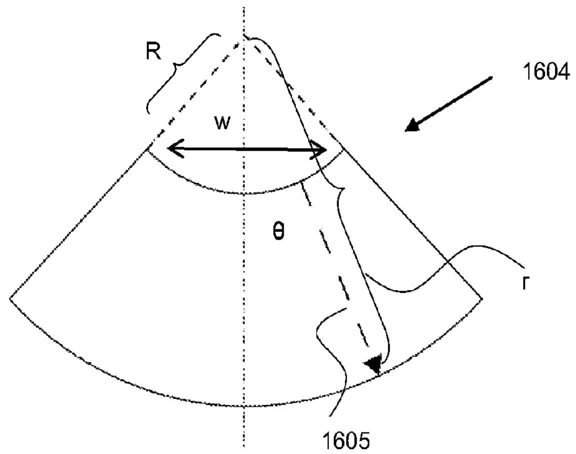

Figure 16b

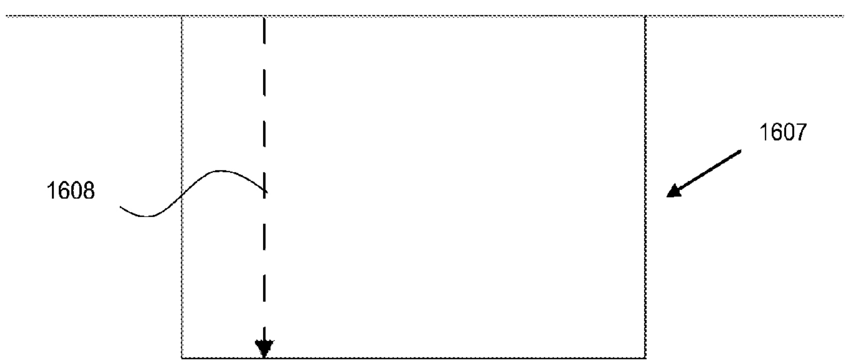

Figure 16c

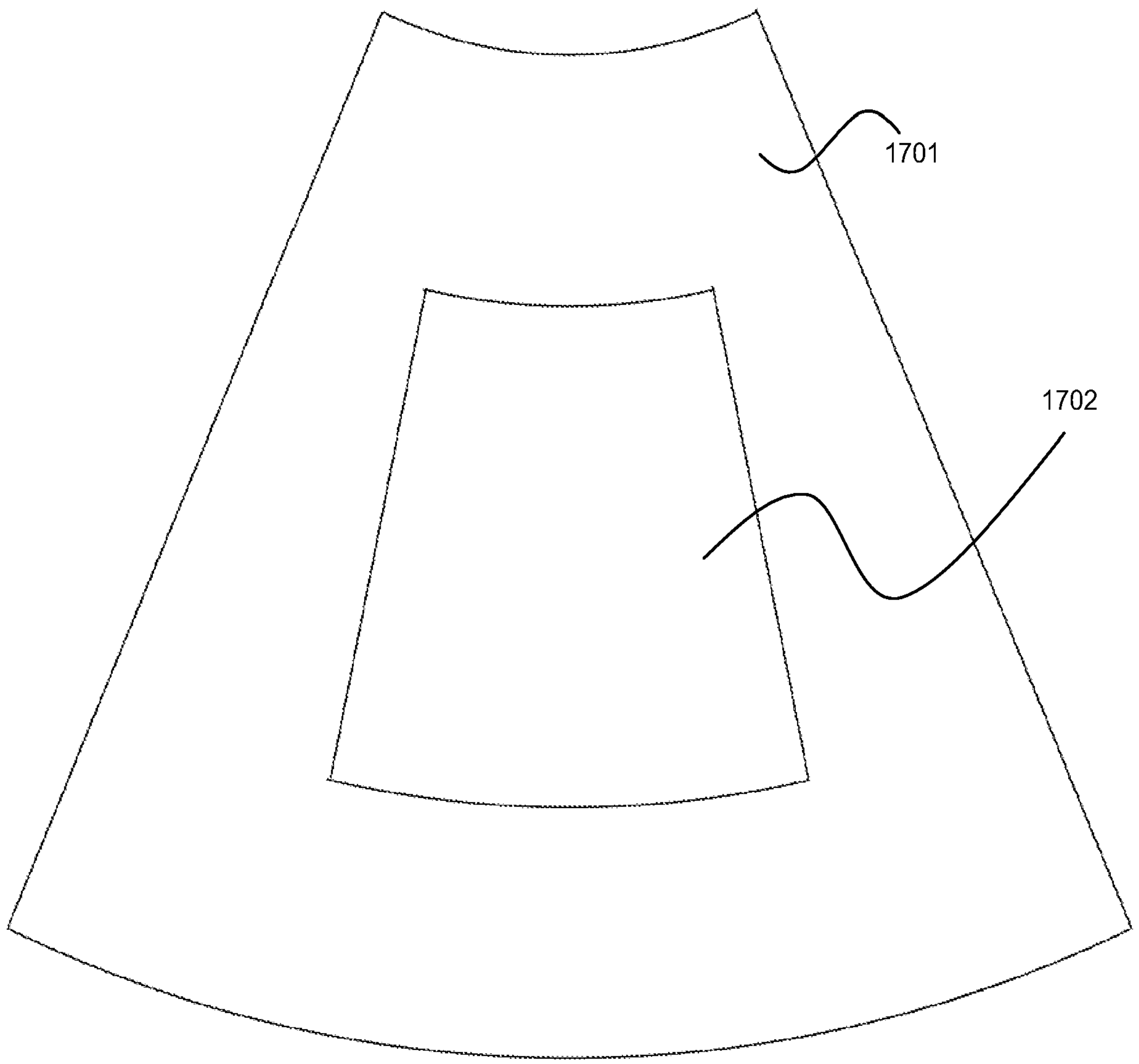

Figure 17

ULTRASOUND TRANSDUCER DEVICE AND METHOD OF OPERATION

FIELD OF THE INVENTION

[0001] The invention generally relates to a real time medical ultrasound imaging system, and more specifically to a real tie medical ultrasound imaging system employing a sparse array transducer having a motor drive.

BACKGROUND OF THE INVENTION

[0002] Ultrasound is a non invasive technique for generating image scans of interior body organs. There are a number of types of real time ultrasound systems supporting a wide variety of ultrasound transducers. These systems can be divided into electronic systems such as phased array, curvilinear array, and linear array transducers which employ fully electronic techniques for beam forming and for directing the ultrasonic beam; and mechanical systems where a transducer or a transducer array is moved mechanically to direct the beam.

[0003] Mechanical scanners are traditionally the simplest and least expensive types of real time imaging systems. These systems utilize one or more piezoelectric crystals which transmit the sensing ultrasound signal, and receive the echoes returned from the body being imaged. To be effective, the ultrasound signal has significant directionality and may be described as an ultrasound beam. An electromagnetic motor is employed to move the crystal in a repetitive manner in order for the beam to cover an area to be imaged. The motor may be of any type, depending upon the movement characteristics required. Devices using stepping motors, DC motors and linear motors are known.

[0004] Mechanical ultrasound scanners generally employ one of the following techniques for moving the beam and generating an image.

[0005] The first technique is the rotating wheel transducer where one or more crystals are rotated through 360° such that a beam emitted from the crystal would sweep out a circle. A sector of that circle constitutes the area to be imaged. The ultrasound signal is only transmitted and received while that sector is being swept out.

[0006] The second type of mechanical ultrasound scanner employs an oscillating transducer where a single crystal is moved back and forth by an electromagnetic motor such that the ultrasound beam emitted by the transducer sweeps out the region of interest.

[0007] Electromagnetic motors were invented more than a hundred years ago. These motors still dominate the industry, but can be considered to be at the limits of technological development: it is unlikely that incremental improvements in such motors will result in performance improvements in mechanically driven ultrasound scanners. Conventional electromagnetic motors are difficult to produce at the very small sizes which are desirable for use in a portable ultrasound probe.

[0008] Electromagnetic motors are very hard to control with precision and have low position resolution. They are also characterized by high speed and low torque and have a slow response.

[0009] The limitations in electromagnetic motors cause a number of problems with the current single crystal mechanical transducers.

[0010] Such transducers are prone to registration artifacts, which are spurious apparent reflections which occur because the motor is free running, and hence image frames produced from consecutive sweeps may not exactly align. This may also result in image jitter which can make the images annoying or fatiguing to view.

[0011] The motor is noisy and imparts a vibration to the ultrasound probe which is in contact with the patient. This may be disconcerting or uncomfortable for the patient.

[0012] In addition any form of Doppler imaging is not possible due to wobble and positional error in the motor.

[0013] Electronic systems which do not use motors, including phased array, curvilinear, and linear transducers overcome many of the problems with mechanical systems, including image registration, ability to perform Doppler imaging, vibration and noise. However, electronic systems have other shortcomings. They are more expensive to manufacture, have relatively high power consumption, and are relatively larger because, to achieve satisfactory performance, they include a large number of transducer elements and associated electronic channels.

SUMMARY OF THE INVENTION

[0014] A transducer array with few elements, set relatively far apart gives good image width for less cost, but sacrifices the number of scanlines in the image. Moving the transducer allows the relatively wide gap between the transducer elements to be "filled in" with scanlines.

[0015] The invention utilizes a method of producing an ultrasound image using a sparse array of ultrasound transducers wherein scanlines are captured from the transducers, the array is then moved an incremental distance, and further scanlines are captured.

[0016] These steps are repeated while the array is moved in a reciprocating or other repetitive fashion.

[0017] The captured scanlines are displayed in a spatial relationship proportional to the spatial relationship in which the scanlines were captured to display a real time ultrasound image.

[0018] The invention also involves an ultrasound imaging device including an array of transducers wherein the distance between adjacent transducers is greater than the minimum separation of adjacent scanlines required to produce an ultrasound image of a selected resolution.

[0019] In preference, the array is adapted to be mechanically moved in a reciprocating motion a selected linear distance along a linear path, the linear separation between the transducers of the transducer array along the linear path being of approximately the same magnitude as the linear distance, the probe unit being operated to acquire a plurality of scanlines during each reciprocation.

BRIEF DESCRIPTION OF THE DRAWINGS

[0020] FIG. **1***a* shows a transducer of the prior art.

[0021] FIG. **1***b* shows a diagrammatic representation of an ultrasound field of a transducer of the prior art of the type illustrated in FIG. **1***a*.

[0022] FIG. **2***a* shows a diagrammatic representation of a linear segmental array transducer of the prior art.

[0023] FIG. **2***b* shows a diagrammatic representation of an ultrasound field of a transducer of the prior art of the type illustrated in FIG. **2***a*.

[0024] FIG. **3** shows an ultrasound scanning device incorporating the invention.

[0025] FIG. **4** shows a diagrammatic representation of an ultrasound transducer array according to the invention.

[0026] FIG. **5** shows a diagrammatic representation of the motion of the ultrasound transducer slider of FIG. **4**.

[0027] FIG. **6***a* shows a diagrammatic representation of a curvilinear transducer array of the invention.

[0028] FIG. **6***b* shows a diagrammatic representation of the motion of the ultrasound transducer slider of FIG. **6***a*.

[0029] FIG. **7** is a cross-section view of a preferred version of the scan head.

[0030] FIG. **8** shows a cross-section of an ultrasound probe unit scan head incorporating a rotary motor.

[0031] FIG. **9** shows a system block diagram of an ultrasound probe unit scan head having a curvilinear-shaped transducer and including a USB interface connection to a host display.

[0032] FIG. **10** shows a system block diagram of an ultrasound probe unit scan head having a single channel ultrasound element and including a USB interface connection to a host display.

[0033] FIG. **11** shows a system block diagram of an ultrasound probe unit scan head having a curvilinear-shaped transducer and including a gyroscope and a USB interface connection to a host display.

[0034] FIG. **12** shows a system block diagram of an alternative version of the ultrasound probe unit scan head including a wireless interface to a host display.

[0035] FIG. **13** shows a system block diagram of an ultrasound probe unit scan head having an integrated display.

[0036] FIG. **14** is a block diagram of an FPGA timing and control unit of the system of FIG. **1**.

[0037] FIG. **15** is a block diagram of the DSP software of the system of FIG. **1**.

[0038] FIGS. **16***a*, **16***b*, and **16***c* illustrate the scan conversion requirements for different transducer shapes which may be incorporated in the invention.

[0039] FIG. **17** illustrates a Doppler processing window.

DETAILED DESCRIPTION OF THE INVENTION

[0040] FIG. **1***a* shows a diagrammatic representation of a simple, single crystal ultrasound transducer as used in early ultrasound scanners. This has a probe unit body **100**, which supports a circular ultrasound crystal **101**. This crystal is relatively large. FIG. **1***b* shows the unit of FIG. **1***a* in cross-section, with a representation of the ultrasound field produced by the transducer. This arrangement has several desirable features. The ultrasound beam produced by the crystal has a relatively long, even near field **102**, before the beam diverges in the far field **103**. The lateral resolution is best in the non-divergent field, so the long near field is advantageous. The far field also diverges relatively slowly, improving the lateral resolution in this area when compared to transducers with a more rapidly diverging far field.

[0041] However, such an arrangement produces only one scanline. In order to scan a region it is necessary to move the transducer over an area to be scanned, obtaining scanlines at different positions, and to keep very accurate track of the movement so that the scanlines can be displayed in correct spatial relation to form an ultrasound image. This was achieved using large articulated arms, with exact position control. These were effective, but severely restricted the ways and places in which ultrasound could be deployed.

[0042] In order to avoid the need for such arms, it would be possible to use an array of crystals to form a transducer. This would allow for an area to be scanned without the cumbersome articulated arm, or the need to keep track of the position probe, since the positional relationship between the crystals, and hence the scanlines, would be fixed. However, the large crystals mean that relatively few scanlines would be gathered from a scan region, significantly reducing the utility of the image.

[0043] In practice, the problem was solved by using a segmental linear array, as shown diagrammatically in FIG. **2**. The segmental linear array consists of a transducer made from a single, rectangular piece of piezoelectric material **201**. This piezoelectric material is divided into a number of individual transducer elements **202**. Each element is separately electrically connected to the ultrasound driving electronics, and may be individually excited to produce a scanline **203**.

[0044] Each individual element is relatively narrow, allowing more scanlines to be collected from a given region.

[0045] However, the smaller transducer element has a less desirable beam shape. As shown in FIG. **2***b*, the beam **203** has a shorter near field **205**, and a more divergent far field **204**. This leads to reduced lateral resolution, which is undesirable.

[0046] The prior art includes solutions involving firing the elements in groups to improve beam shape, and to firing the elements in very specific patterns to form and electronically steer a beam across the area to be scanned. In prior arrangements, the signal of each transducer element is combined with a number of adjacent elements, usually 32, to produce one scanline.

[0047] All of these solutions lead to compromises of the ideal beam shape provided by the circular crystal transducer of the original devices. Sophisticated control and signal processing techniques are used to counter these problems, but these introduce further cost and complexity. These solutions also introduce very great complexity and cost into the physical design of the transducers, with the need to provide large numbers of transducer elements each requiring precise, matched electrical connections.

[0048] FIG. **3** shows a view of an ultrasound scan device **300**. This includes a display unit **301** and a probe unit **302** connected by communications cable **303**. The display unit **301** has a display **304**. The display screen **304** may be a touch screen allowing a user to control the functionality of the display unit **301** and the probe unit **302**. In the illustrated version, control members **306** are provided on the display unit **301** in the form of push buttons and a scrollwheel. Control members **305** are provided on the probe unit **302**. Either of both of these control members **305** and **306** may be absent.

[0049] The probe unit **302** includes or is connected to a probe unit scan head **308**. The scan head **308** includes a transducer which may be, without limitation, an array transducer having multiple transducer elements, a single element transducer or an array of separate, individual transducers.

[0050] The scan head **308** includes a front panel **307** which contacts the patient during scanning. This front panel **307** is acoustically transparent. It is desirable to provide the best possible acoustic coupling between the transducer elements and the body to be imaged.

[0051] In use, the probe unit **302** is held against the body of a patient adjacent to the internal part of the body which is to be imaged, with the front panel **307** in contact with the patient's skin. Electronics in the probe unit **302** stimulate the emission of ultrasound energy from the transducer. This beam is reflected back to the transducer as echoes by the features to be imaged. The transducer receives these echoes which are amplified and converted to digital scanline data.

[0052] A motor moves the transducer such that the ultrasound beam or beams sweep out an area to be imaged. Electronics for control of the motor are provided in the probe unit. The motor may be a linear or a rotary motor, most preferably a linear or rotary ultrasonic motor, though it may be another type of motor (e.g., an electromagnetic motor).

[0053] The term "ultrasonic motor" is used throughout this specification. Other terms may be used for devices having the same principle of operation but varying in size, configuration and/or application. These terms include, without limitation, piezomotor, piezoelectric actuator, piezoactuator, and ultrasonic actuator. The term "ultrasonic motor" as used in this specification covers all of these and any other possible terminology which may be used to describe ultrasonically driven moving devices which may be used to perform the invention.

[0054] FIG. **4** shows a diagrammatic representation of an ultrasound transducer of the invention. There is a transducer slider **501** which supports ultrasound transducers **502-505**. Each transducer **502-505** is able to transmit ultrasound energy and to receive the resultant echoes from a body to be imaged to gather one scanline **506-509**. There may be as few as one transducer crystals, with the maximum number being limited only by practical considerations of cost, size and complexity. The transducers **502-505** form a sparse array, defined as an array of transducers wherein the distance between adjacent transducers or transducer elements is greater than the minimum separation of adjacent scanlines required to produce an ultrasound image of a desirable resolution.

[0055] In use the slider **501** is driven in a reciprocating fashion such that the beams from the transducers **502-505** cover an area to be imaged. The slider **501** may be driven in a reciprocating motion by any convenient means.

[0056] The scanlines produced by adjacent ultrasound transducers **502-505** are separated by a distance **1**.

[0057] The slider **501** is moved, continuously or in steps, the distance **1**. The transducers **502-505** transmit and receive ultrasound energy while moving (or at each step) in order to receive a scanline **506-509**, which is a series of echo intensity values returned from features at various depths along a line running into the body to be imaged.

[0058] FIG. **5** shows the effect of moving the transducer slider **501**. The series of dotted line diagrams of the transducer slider **501** represent the movement of the ultrasound slider **501** through the distance **1**. Each of the transducers **502-505** is driven to transmit and receive a series of scanlines a, b, c . . . , n. The combination of these n scanlines for all of the transducers produces a scanset which covers the distance S. The distance S is the width of the B mode ultrasound image which may be produced by the probe unit.

[0059] All of the scanlines captured during a single traverse of the transducer of the distance **1** form a scanset. The scanlines in each scanset are processed and displayed on the display screen **304** in the same spatial relationship as they were acquired by the transducer. This produces an ultrasound image of the area to be imaged. Each scanset corresponds to a single frame ultrasound image. The distance moved by the transducer slider between successive firings of the same transducer, for example between the acquisition of scanlines a and b and between the acquisition of scanlines b and c is sufficiently small to provide the desired image resolution.

[0060] Post-processing may be applied to enhance the image in a variety of ways.

[0061] The slider **501** is moved in a reciprocating fashion. A new scanset is obtained on each pass over the area to be imaged. Scansets are displayed sequentially as they are received to form a real time display.

[0062] In a prior art linear array transducer ultrasound probe, as shown in FIG. **2**, the lateral distance between scanlines is proportional to the linear distance between successive transducer elements in the array.

[0063] To provide a higher resolution image without sacrificing image coverage, or to provide a wider image coverage without sacrificing resolution, requires more scanlines and hence more transducer elements. Providing additional elements is expensive. The larger transducer is itself more expensive, but the additional wiring and electronics required to access and control the additional elements (usually called channels) is also expensive.

[0064] An array with few elements, set relatively far apart gives good image width for less cost, but sacrifices the number of scanlines in the image. Moving the transducer as described allows the relatively wide gap between the transducer elements to be "filled in" with scanlines, thereby providing similar performance and image size to systems with more channels, at greatly reduced cost and complexity. The scanline density is not restricted by the physical size of the transducer elements or the spacing or number of elements.

[0065] Conventional linear array systems have transducers with a large number of elements, each cut in a rectangular shape. This generates an acoustic beamshape with a non-symmetrical pattern. It is desirable to use circular crystals, where the beamshape is circular and symmetrical. Since high end systems may have 1024 or more elements, this is not practical for conventional linear array systems.

[0066] Using a sparse array of individual transducers makes the use of circular transducers in a linear array practical.

[0067] In order to be displayed as part of a scan image, in the spatial relationship in which the scanlines were received, each scanline must include, or be associated with, data indicating the relative location of the transducer element which received the scanline data at the time it was received. This can be done by any means known in the art. A method based on knowledge of the rotational or linear position of the motor may be used, or the position of the slider **501** may be directly monitored by a device such as a linear encoder. This data may be transmitted to the display unit **301** as part of the scanline data; as a separate data stream; or the information may be inferred and calculated by a processor in the display unit.

[0068] The image captured by the linear sparse array of FIG. **5** has a maximum width of S, being the width of the transducer slider **501** plus the distance traveled by the slider **501**. This can be increased by increasing the width of the slider **501**, but this increases cost. Further, a larger slider requires a larger probe unit **302**, which is undesirable from an ease of use perspective.

[0069] FIG. **6** shows a diagrammatic representation of the invention applied to a curvilinear scan head form factor.

[0070] There is a transducer slider **600** to which are attached one or more transducers, with the illustrated version having four transducers **605-608**. The transducer slider **600** is shaped as the arc of a circle. Each transducer **605-608** is able to transmit and receive ultrasound energy in order to produce a single ultrasound scanline **601-604**. The transducer slider

600 is incorporated in and forms part of an ultrasound probe unit 302 as illustrated in FIG. 3.

[0071] In use, the transducer slider 600 is rotated about an axis through the center of the circle of which the transducer 605-608 forms an arc. This may be done by use of a pivot arm with a pivot point at the center. Alternatively the transducer slider 600 may be guided by arc-shaped guides. Other means to constrain the slider 600 to move along the path of the arc of the circle may be employed.

[0072] FIG. 6*b* shows the rotated transducer slider 610 with transducers 615-618 producing scanlines 611-614, superimposed on the original position of the slider 600.

[0073] The placement of the transducers 605-608 on the outside of a circle means that the overall area covered by the ultrasound beams is a sector of a circle. This allows broader coverage than the width of the transducer probe unit 302. The greater the curvature of the slider 600, the greater is the amount by which the width of the scanned area can exceed the width of the probe unit 302.

[0074] The received scanlines are processed for display as an ultrasound image having a form factor similar to the well known sector of a circle ultrasound image. The transducer slider 600 is reciprocated to achieve real time coverage of the area being imaged. Substantially all of the scanlines captured during one reciprocation are displayed to form an image frame. Successive display of the frames as the scanlines are received gives a real time display. Real time frame rate on the order of 20 frames per second are achieved.

[0075] FIG. 7 shows of an version of the invention wherein the slider of the ultrasonic motor is separate from the transducer array. There is a scan head 700 which includes an ultrasonic motor 710, including excitation electrodes 704, vibratory driver 705 and slider 708 held by a frame and guide rails. The slider 708 includes a friction layer 709 which facilitates drive friction between the vibratory driver 705 and the slider 708. The motor 710 is attached to a circuit board (PCB) 701, which is positioned and supported within the scan head 700 by locating guides 702. The PCB 701 includes (or has attached to it) ultrasonic motor control circuitry 706. This control circuitry 706 supplies signals to the excitation electrodes 704 in order to drive the motor 710.

[0076] A transducer array 707 is attached to the slider 708. The transducers making up the array 707 may be of any suitable type. The transducers 707 are attached to a backing element, which is mechanically connected to the slider 708 of the motor 710. This arrangement simplifies the assembly and manufacture of the system, as the main PCB 701 is mechanically connected to the motor 710 and ultrasound transducers 707, and the motor 710 can operate independently of the enclosure.

[0077] The transducers 707 are electrically connected to circuitry on the PCB 701 which excites the transducers 707 to produce ultrasonic output and receives and processes the electrical signals returned from the transducers 707.

[0078] In the illustrated version, a non-liquid friction reduction means is provided to allow the transducer array 707 to slide easily over the acoustic window 703 which, in use, is in contact with the patient. This may be, without limitation, a solid lubricant, or the material of the acoustic window 703 may inherently provide low friction.

[0079] An acoustically matched lubricant may be provided on the interior side of the front panel 703 of the scan head 700 filling the gap between the transducers 707 and the panel.

[0080] In a further version, a strip of solid material having a low coefficient of friction and acoustic matching properties is provided on the interior side of the front panel 703 of the scan head 700 filling the gap between the transducers 707 and the front panel.

[0081] In other versions, a flexible, liquid-proof, membrane may be provided between the transducer array 707 and the motor 710, forming a closed compartment with the acoustic window 703. This enclosed compartment encompasses the transducer array 707. With the motor 710 thus protected from liquid, the transducer array 707 may be immersed in a liquid, which provides acoustic coupling between the transducers of the transducer array 707 and the acoustic window 703. In this case the transducer array 707 need not touch the acoustic window 703 in use. Any gap between the transducers 707 and the acoustic window 703 will not cause significant reverberation in the received ultrasound signal, since the gap will be filled with the acoustic matching liquid.

[0082] In the illustrated version, the transducer array 707 is an array of eight piezo-electric transducers. In other versions, other numbers and other types of transducer elements may be employed. In particular versions, capacitive micro-machined ultrasonic transducers (CMUTS) or other MEMS based transducers may be used.

[0083] Referring to FIG. 8, the slider 805 may be driven by a rotary motor, which in a preferred version is an ultrasonic motor.

[0084] The range of the reciprocating movement, being the linear movement of the slider 805, is approximately equal to d, the diameter of the drive disk 803. The number of transducer elements 808 is N. The linear distance between successive transducer elements 808 is 1.

[0085] In order to achieve full coverage of the area to be scanned with the same density of scanlines, it is necessary that 1 should be less than or equal to d. Preferably 1 should be approximately equal to d, and further:

$$d \sim 1 = S/N$$

[0086] The number of transducer elements 808 used can vary depending on a number of factors including coverage area, motor speed, disk size and motor response time. In the illustrated version a 7.5 MHz scan head with eight 3 mm circular crystals 808 mounted on the slider 805 combined with a 6 mm width disk 803 provides a scan head with 48 mm (8×6 mm) coverage.

[0087] The rotational motion of the ultrasonic motor 800 is translated into reciprocating motion. There is a drive disk 803 which has an off-center pin 804 protruding from one face. The disk 803 is attached to a shaft 802 driven by the rotary motor 800. The slider 805 to which the transducer element array 808 is mounted includes a slot 807. The off-center pin 804 of the disk 803 is located into the slot 807. Rotation of the disk 803 causes the pin 804 to impart a reciprocating motion to the slider 805. The slider 805 is restrained from motion other than reciprocation in two dimensions by rails 806 on each side.

[0088] The part of the scan head containing the transducer array 808, may be filled with a liquid medium to eliminate any air interfaces between the moving transducer elements 808 and the protective front panel 809 of the scan head. The transducer array compartment is separated from the motor 800 by a seal 801 around the motor rotor 802. This prevents the liquid medium from entering and potentially damaging the motor.

[0089] In further versions the transducers may be implemented such that the array **808** is moved in a curvilinear pattern. This is illustrated diagrammatically in FIG. **6**.

[0090] In such versions, the slider is shaped as a circle segment, while the guide rails, if used, and the front panel are shaped as arcs of a circle. Either a rotary or linear motor may be used to drive the slider. For a linear motor, the slider and guides may form a linear motor, or a separate linear motor may be provided. Where a rotary motor is used, the slider may have a slot which accepts the offset pin drive as for the illustrated version using a rotary motor. The slot will be trapezoidal in cross section to accommodate the fixed orientation of the drive pin. Where the radius of curvature of the slider segment is sufficiently small, a rotary motor may move the slider by directly driving a rotor arm.

[0091] A rotary motor shaft may be used to drive a fully annular slider. This slider will have transducers on the outer face. These transducers may be grouped to form a curvilinear array in the form of an arc of a circle, or they may be evenly distributed around the circumference of the annular slider.

[0092] In a preferred version, the motor moves the transducer array slider across the width of the area to be scanned in a reciprocating fashion. Each transducer element is fired in turn. Each firing results in a single scanline of data being collected. The delay between firing each transducer element should be at least sufficient to allow for the return of echo from that depth within the body being imaged where it is desired for an image to be obtained. The firing process repeats. Generally, each transducer element will be fired many times in the time it takes for a single reciprocation of the slider. Preferably the firing is timed such that on each successive reciprocation, the firing of each transducer element is in substantially the same places with respect to the probe unit body.

[0093] Preferably, the delay between successive firings of a transducer element is related to the width of the beam from the transducer elements such that the entire length of the area being scanned is covered by a focused beam.

[0094] In an alternative version, the system operates by firing each transducer element individually with the transducer slider at rest to acquire a set of scanlines. The motor then moves the slider a step, with the slider coming to a halt before the transducer elements are again fired acquiring a second set of scanlines. This continues until the distance **1** is covered. The process repeats.

[0095] The size of the step is determined to provide the required scanline density over the area to be imaged. It may be varied in use in order to optimize characteristics of the scan system such as frame rate or motor speed.

[0096] All of the acquired scanlines for a single full traverse of the slider are then combined to form one frame of a real-time ultrasound display. In order to be displayed as part of a scan image, each scanline must include, or be associated with, data indicating the relative location of the transducer element which received the scanline data at the time it was received. This can be done by any means known in the art, for example, the position of the slider may be directly monitored by a device such as a linear encoder.

[0097] The number of steps used is determined by the scanline density required and the magnitude of the transducer element separation **1**.

[0098] The number of transducer elements used can vary depending on a number of factors including coverage area, motor speed, scanline density desired and motor response time.

[0099] All of the scanlines acquired in one full transverse movement of the transducer constitute a single frame for display. This means that the firing rate of the transducer as a whole is determined as the product of the number of elements, the frame rate and the number of scanlines required to cover the distance between the elements at the desired density.

[0100] Other element firing regimes are possible. Where this can be done without undue interference between the scanlines, more than one element may be fired at a time. Without limitation, a means of achieving this is for the transmit pulses to be coded, for example with Barker, Golay, or Gold codes, such that the correlation between pulses from separate transducers is extremely low.

[0101] In a further version, the slider length is less than the image width S, by an amount greater than the separation of the transducer elements **1**. It may be much less. The scan head and rails cover the full image width. The slider is driven in incremental steps to cover the required image width. In this version, the elements of the transducer array may be arranged at such a separation that adjacent elements give the desired scanline density. In this case, the slider may move an amount equal to its full length between each full firing of the array.

[0102] In the most general case, the transducer may consist of a single element. This is not the currently preferred version since one of the limitations of ultrasonic motors is that they have low linear and rotational speeds. This poses a constraint on the speed the ultrasonic crystal can scan the region of interest and hence limits the system's frame rate. It is therefore preferred to use an array incorporating more than one crystal, which decreases the commuting distance of the crystals and reduces the motor speed requirements, but further improvements in motor design may make this alternative practical.

[0103] Pulsed Wave Doppler imaging is achieved by electronically stimulating a transducer element to produce a pulse of ultrasound and then to listen for echoes before another pulse is generated. This is done rapidly at a single location, and the variable frequency shift in the returned echoes is used to detect movement in the feature being imaged, for example blood flow. These modes are not possible with motor driven transducer systems of the prior art, since they cannot be held stationary nor rapidly and accurately repositioned.

[0104] This can be achieved by the device of the current invention because ultrasonic motors have excellent response time and can be positioned extremely accurately, with errors in the nanometer range. This allows the transducer to be completely stationary at a known position when sending and receiving ultrasound pulses.

[0105] Color Doppler combines real time imaging with Doppler imaging. Real time images are displayed for the full image area at the same time as Doppler information is displayed for one or more scanlines. This is not possible for any prior art system with a moving transducer, since Doppler requires that the transmitting element be stationary in a known position and real time scanning requires that the transducer be moved.

[0106] In an version providing color Doppler functionality, the transducer array is controlled to move to produce a real time moving image on a display for a user. The user is able to select an area of the displayed image for acquisition of color

Doppler information. A selected element or elements of the transducer, while located above the selected area, fires the required number of pulses and receives the necessary echoes to acquire the movement information, being the color Doppler frequency shift data. The transducer array is stationary whilst the Doppler information is acquired. The real time scan operation then continues, with the transducer being moved by the ultrasonic motor. Real time image scanlines are acquired whilst the transducer array is moving. When transducer has moved a selected distance, the transducer stops and the selected element again fires multiple pulses, acquiring Doppler information. This continues until a full real time image frame has been acquired, along with Doppler information for a subset of that frame. The process then repeats to acquire a real time image display.

[0107] The color Doppler and real time information are combined into a single display, as is known in the art.

[0108] The fast response and excellent positioning accuracy of ultrasonic motors allows the transducer to come to complete stop, fire the required number of ultrasound pulses, listen to the echoes, and then move to the next line while simultaneously performing real time imaging. Generally the real time frame rate will be reduced while color Doppler is in use.

[0109] Versions using ultrasonic motors may also use a curvilinear array where the linear slider is replaced with a curvilinear slider with N transducer elements mounted on it or machined as part of the slider. The rails and the front panel would also be manufactured in a curvilinear construction. Ultrasonic motors are readily manufactured in different shapes and sizes. In versions where the slider itself is part of the motor, no special arrangement or shaping of the slider is necessary to allow it be driven by an external motor.

[0110] In alternative versions, a linear ultrasonic motor separate from the curvilinear slider may be used to move a curvilinear slider through an arc of a circle.

[0111] The slider may be driven through a full circle in a continuous revolution. In this case, the transducer would be active only when it was over the body to be scanned. This is especially relevant for micro curvilinear shaped scan heads, which have a relatively tight radius of curvature so as to enable imaging between ribs. Two or more transducers may be provided, each having a different operating frequency, allowing simultaneous or separate scanning at different frequencies. Doppler imaging would be possible as described for other versions.

[0112] One of the drawbacks of ultrasonic motors is the limited lifetime due to the frictional drive. A spring-like mechanism is required to maintain the force between the stator and the slider/rotor. Contact areas are required between the stator and the slider to provide the frictional drive, and these contact surfaces wear. This wear increases the separation between the contact surfaces, reducing the effectiveness of the drive. This problem can be mitigated by using a piezoelectric actuator as the spring-like mechanism to provide the force between the stator and slider/rotor. The force imposed by the actuator is proportional to a DC voltage across the actuator. This DC voltage in effect pushes the piezoelectric driver closer into the frictional coating and hence the coupling force between the stator and the slider is maintained throughout the life time of the motor. The magnitude of the DC voltage can be controlled so that it is increased with the motor's operating life to compensate for the layer wear.

[0113] Ultrasonic motor efficiency is insensitive to size and hence they are superior in the mini-motor area. This allows for the construction of devices of small size and low weight.

[0114] Ultrasonic motors do not produce the electromagnetic interference which is inherent in the operation of electromagnetic motors. This is an advantage for ultrasound systems where the electrical signals being received are inherently of low power and susceptible to interference. This has a further advantage for hand held ultrasound systems in reducing the need for shielding between the electronic components and the motor, which reduces size, weight and cost.

[0115] In order to produce ultrasound energy for scanning, the ultrasound transducer elements are stimulated by a high voltage power supply. This high power supply would generally supply power at voltage magnitudes on the order of tens or hundreds of volts. The ultrasonic elements which form the ultrasonic motors in those versions which employ ultrasonic motors are also driven by high voltage power supplies which may have similar output characteristics. In further versions, the ultrasound transducer elements and the ultrasonic motor are driven from a common power supply or power supplies. This results in lower costs.

[0116] FIG. **9** shows the architecture of an exemplary version of the invention. The block diagram shows a probe unit, including a scan head **901** and a probe unit body **903**. There may also be a host unit (not illustrated) providing a display and at least part of a user interface, connected to the system using a USB interface **920**.

[0117] Small size and weight are desirable for the probe unit to facilitate hand held use. In an version, the size of the probe unit is about 15 cm×8 cm×3 cm. In a further version, the weight of the probe unit is 500 g or less. In a preferred version the size of the probe unit is about 11 cm×6 cm×2 cm and the weight is about 200 g.

[0118] The scan head **901** includes transducers **906**, arranged as a transducer array on a slider element **905**. The slider element **905** is driven by ultrasonic motor **908**. The position of the slider **905** is monitored by position encoder **907**.

[0119] The transducers **906** are connected to multiplexer **909**. Electrical signals to excite the transducers to produce ultrasound output, and the electrical signals from the transducers in response to received echoes, are passed through the multiplexer **909**. The multiplexer may be provided within the probe body **903**, or omitted, at the expense of providing more connections between the scan head **901** and the probe unit body **903**.

[0120] User control of the ultrasound scanning system is via a user interface which may be provided wholly on the host unit, wholly on the probe unit, or split or duplicated between the probe unit and the host unit. In the illustrated version there is a User Control Panel **902** incorporated in the probe unit, the Panel **902** including a freeze button to start and stop scanning, a set of buttons providing increment, decrement and select functionality, and a "back" button. These buttons are used for basic control of the system, including without limitation, some or all of start and stop scan, depth adjustments, gain adjustments, operating mode selections, and preset selections as well as other basic settings. The user control panel **902** sends data identifying the state of the buttons to a microcontroller **930**, which in this version is part of a combined microcontroller/DSP device such as the OMAP3525 Applications Processor from Texas Instruments. The microcontroller **930** monitors the state of the control panel **902** and provides appropriate commands to control the appropriate electronic circuitry to allow the user to control the ultrasound system. Where this requires a graphical user interface, this is displayed on the host unit.

[0121] The microcontroller **930** contains a set of parameters for controlling the operation of the device during scanning. At initial power up, a default set of parameters are created, which can then be modified by the user before or during scanning. The parameters include without limitation, the operating frequency, the active scanning mode, the gain curve, the scanning depth, the Doppler gate depth and angle if required, color Doppler ranges if required, power Doppler ranges if required, and M-mode pulse repetition rate. The scanning modes available may include, without limitation any or all of B-mode, M-mode, and modes available using Pulse Wave Doppler, including color Doppler, power Doppler and spectral Doppler, and Duplex Doppler.

[0122] The microcontroller **930** passes the relevant parameters to the Digital Signal Processor (DSP) **931** and Field Programmable Gate Array (FPGA) **932** when scanning is activated or the parameters change.

[0123] When a user commences a scan, either by pressing a button on the control panel or by activating a control on the host, the microcontroller **930** sends a command to the DSP **931** and Field Programmable Gate Array (FPGA) **932** to activate scanning, along with any updated configuration parameters. The DSP **931** is configured to receive and process ultrasound data according to the parameters, which may include parameters concerning Doppler processing.

[0124] The microcontroller **930** and the FPGA **932** together provide the functionality to control the scan head ultrasonic motor. The ultrasonic motor position encoder **907** produces a value proportional to the position of the ultrasonic motor **908** at any point in time. This position is saved with a timestamp in a register in the FPGA **932**, and the microcontroller **930** reads this information to calculate a velocity. The velocity is compared with the desired velocity. The motor control unit **910** is instructed to alter the voltage and frequency of the drive signal applied to the ultrasonic motor **908** in order to achieve the desired velocity.

[0125] FIG. **14** illustrates the FPGA functionality. The FPGA **932** receives and decodes the output of the ultrasonic motor position encoder **907**, and provides the information to the microcontroller **930** in order for it to recalculate the ultrasonic motor **908** drive signals. The FPGA includes a timing state machine **145** which uses the decoded position information to determine the appropriate time to generate the next ultrasound pulse sequence to be output from the transducers **906**, which is achieved by driving the transmit pulse controller **144**.

[0126] Transmit pulse controller **144** generates control signals corresponding to the type of scan line required to be generated. For a B-mode scan line, a single pulse is generated at the imaging frequency at the voltages provided by the High Voltage power supply (HV Supply) **934**. These voltages are typically up to +−100V. For Doppler, a sequence of several pulses, typically eight pulses, at the Doppler imaging frequency is generated. These Doppler pulses are typically of longer duration than B-mode pulses. The multiplexer is preconfigured before the Tx Pulser **933** fires to steer the transmit pulse to the appropriate circular transducer crystal. The transducer crystal **906** generates an ultrasonic waveform in response to the electrical pulse, which is transmitted into a body to be imaged.

[0127] The ultrasonic waveform is reflected by changes in acoustic impedance, producing echoes which are received back at the transducer crystal **906** at lower pressure. The crystal converts this lower pressure waveform into an electrical signal, which is directed through the multiplexer to the input protection circuitry **935**. The input protection circuitry protects the sensitive low noise amplifier (LNA) **936** from the high voltage transmit pulse while letting through a low voltage receive signal. Several input protection circuits are known in the prior art.

[0128] The low noise amplifier (LNA) **936** provides amplification of the small receive signal, while adding little noise to the output signal. The LNA **936** is typically single ended, and generates a differential output voltage for feed into a time gain amplifier (TGA) **937**. The TGA **937** provides further amplification, with the amplification provided being dependent on time. Ultrasound signals attenuate as they propagate through tissue, and the TGA **937** compensates for this attenuation by increasing the gain dependent on the time from the beginning of the received pulse, which is proportional to the depth of the echo reflection.

[0129] The output of the time gain amplifier **937** is filtered to remove as much noise as possible and to prevent aliasing. Typically a bandpass filter **938** is used. The output of the filter **938** is input to the analog to digital converter **939**. A sampling frequency of at least 4× the imaging frequency is preferred. The analog to digital output is input to a first in first out memory (FIFO) **147** in the FPGA **932**. The FPGA adds some header information to each complete scan line, including the type of scan line (for example, B-mode or Doppler), the motor positional encoder input if required, and a time count. This information may be used at a later stage in processing dependent on the configuration parameters.

[0130] The DSP **931** reads the scan lines out of the FPGA FIFO memory **147**, and performs appropriate processing on the data depending on the configuration parameters.

[0131] FIG. **10** is a configuration of the system which does not have real-time ultrasound capability, but uses a lower cost single channel system where the ultrasound beam is stationary with respect to the system. A single transducer **1001** is provided, which is in a fixed relationship to the probe unit body. There is also provided a gyroscope **1002**. Images created by moving the probe unit in an arc with the point of contact of the scan head **1003** with the patient being substantially constant while the transducer **1001** is pulsed. This will generate a sequence of scan lines which together make up a sector image. The gyroscope **1002** provides information as to the relative position of each scan line, enabling the scan lines to be assembled into an image.

[0132] FIG. **11** illustrates a further version of the invention, where the scan head **1101** contains a motor **908** and a gyroscope **1103**. The gyroscope **1103** is oriented so it provides angular measurements perpendicular to the image plane of the B-mode image which is produced from the transducer array **1102**. By providing the gyroscope **1103**, the system is able to construct 3D images from a sequence of 2D image scans, and determine volumes. In use, a series of B-mode images are made as for the system of FIG. **9**. The probe unit is moved in an arc perpendicular to the image plane of the B-mode images. The operation is otherwise similar to the system as described in FIG. **9**.

[0133] Referring to FIG. **14**, The FPGA timing state machine **145** interfaces to the gyroscope in versions where a gyroscope is provided. The gyroscope angular measurement is combined with the ultrasound scan line information in a first in first out (FIFO) memory **147**, and read by the digital signal processor (DSP). Where a single transducer is used, the gyroscope information is used to assemble the scan lines to form a sector image. Where a transducer array is used, the gyroscope information is used to assemble B-mode sector images into a representation of a volume which may also be used for any applications which require 3D volume calculations.

[0134] FIG. **12** illustrates an alternative version of the invention, whereby the interface to the host display and control system is via a wireless interface. There is provided a wireless interface module **1202** controlled by the microcontroller **930**. An antenna **1201** is provided for signal transmission. This interface provides communication to and from the host unit. No USB interface is required for communication with the host unit, although one may be provided for access to third party hardware. In a presently preferred version the wireless protocol used is WiFi Direct.

[0135] FIG. **13** illustrates an alternative version of the invention, wherein there is no host unit. There is provided a display **1301** which is integral with the probe unit. There is also provided an enhanced user control module **1302**. This includes the functionality of the control panel of previously described versions, but further includes functionality allowing for full control of the system in the absence of a host unit. The control module **1302** may include a graphical user interface displayed on the display **1301**. The display **1301** may be a touchscreen, able to provide input to the user interface.

[0136] The operation of the DSP in different versions will vary. Where a host control and display system is connected, some processing and control functions and the display function may be performed by the host unit. Where no host is used, all functions are performed by the probe unit, increasing the load on the DSP. Limitations on the division of functionality between a host and the probe unit come from the processing power of the host and the bandwidth available for transmission between the host and the probe unit. In general, the implementation and subdivision of algorithms is designed to minimize the host processing requirements and the bandwidth required for transmission.

[0137] FIG. **15** illustrates the algorithms for processing combined B-mode/Doppler, with a division of the tasks between the ultrasound probe unit and the host display system.

[0138] The first step in the B-Mode processing chain is to filter the digitized incoming rf scan line data using an FIR bandpass filter. The filter is adjusted depending on the transducer connected to the system and the imaging frequency. For example, for a 3 MHz imaging frequency a bandpass filter of 1 to 5 MHz could be used. For an 8 MHz imaging frequency a bandpass filter of 4.5 to 11.5 MHz could be used. Following filtering, the envelope of the rf scan line data is generated. The preferred method is to use a Hilbert transform to generate the in-phase (I) and quadrature (Q) components of the rf scan line. The final envelope is generated by summing the squares of the I and Q components, and taking the square root of the result.

$$\text{Envelope}=\sqrt{I^2+Q^2}$$

[0139] There are a number of algorithms for generating Hilbert Transforms. The preferred version is using an FIR approximation.

[0140] As part of the envelope generation, the scan line can be downsampled or decimated. Downsampling by a factor of 2-4 is possible, depending on the scan conversion algorithm used. In the preferred version, the scan line is downsampled by a factor of 4 with scan conversion using bilinear interpolation. An alternative is to downsample by a factor of 2 and use a less computationally intensive interpolation algorithm, such as one which computes pixel intensity from the average of sample points inside the pixel area, and interpolates for other pixels between adjacent pixels.

[0141] Following downsampling, the scan line is compressed from the analog to digital word size into an 8 bit word size to map the signal into the desired grey scale levels for display.

[0142] The downsampled scan lines are then converted to a rectangular image display through scan conversion. The scan conversion is performed in 2 stages. The first stage is converting the image into a compressed rectangular array with high resolution, preferably with pixel resolution of less than half the axial resolution of the ultrasound pulse. A common scan conversion algorithm is a 2×2 bilinear interpolation, mapping the scan line points from a polar co-ordinate system to a rectangular co-ordinate system. Referring to FIG. **16***a* there is shown an idealized scan from a phased array transducer. The scan area **1602** can be seen as a sector of a circle, with origin **1601**. In practice, a phased array transducer is not a point source, but it is small compared to the depth of scan. The location of each scanline **1603** can be characterized in polar co-ordinates by a length r and an angle θ. Referring to FIG. **16***b*, there is shown a sector scan **1604** for a curvilinear array with a radius of curvature R and a width W. The scan is a truncated sector of a circle. Each scanline **1605** can be characterized by polar co-ordinates of a length R+r and an angle θ.

[0143] FIG. **16***c* shows an idealized scan **1607** for a linear array, with scanline **1608**. The characterizing coordinates are linear, but in general will not correspond to the linear coordinate system desired for display.

[0144] In each of the cases illustrated in FIG. **16**, scan conversion is required to convert the acquired scanline data into pixel data for display of the image to a user.

[0145] For sector shaped images, the compression algorithm reduces the image size by ignoring image area which is not containing actual image data. Several methods are possible to perform this, with one being a simple format where each pixel row contains a header with the starting pixel and number of pixels with valid data. This technique yields a compression ratio of close to 2. Another lossless technique which would provide reasonable results is run-length encoding. LZW encoding, or Huffman based encoding such as png could be used.

[0146] The high resolution compressed rectangular array is handled differently depending on the system configuration. The array is stored in a local memory, with typically up to 100 compressed frames being stored locally. At the same time the current frame is transmitted from the ultrasound probe to a host display system. The host display system completes the processing by decompressing the image into a high resolution buffer, interpolating from the high resolution buffer to an interim display image buffer, applying any grey scale adjustment, combining with any Doppler image information if required, and finally writing to the display buffer.

[0147] FIG. **15** also illustrates the steps required for processing and displaying Doppler information overlayed on the B-mode image. When generating Doppler information, the system generates a sequence of Doppler pulses with a consistent phase and processes the received set of scan lines. The raw input rf scanlines are quadrature encoded, with the inphase (I) and quadrature (Q) components extracted by multiplying the echo signal by the cosine and sine of the transducer excitation frequency. The I and Q outputs from the quadrature encoder are low pass filtered and decimated or downsampled. A downsampling ratio of four produces satisfactory results, but other factors may be employed. The output of the downsampler is saved into a buffer until a complete set of scan lines is received. Typically a set will be eight scan lines, although the size of the set can be adjusted. The data set is filtered using a Wall filter. The function of the wall filter is to reduce any contribution from large, slow moving features such as an abdominal wall.

[0148] The wall filter in the illustrated version is an FIR type high pass filter. An alternative is to use a state-space formulation IIR filter, which reduces the transient response length of the IIR filter. The state-space initialization provides satisfactory attenuation using a step initialization scheme.

[0149] In the preferred version the filtered set of scan lines is processed using autocorrelation techniques to generate velocity, power, and turbulence information. The Doppler power of the set of scan lines is calculated as follows:

$$P_d = \sum_{k=0}^{n-2} s_{d,k} \cdot s_{d,k}^*$$

[0150] The correlation between adjacent scan line points is calculated as follows:

$$c_d = \sum_{k=0}^{n-2} s_{d,k+1} \cdot s_{d,k}^*$$

[0151] The velocity estimation is calculated as follows:

$$v_d = \tan^{-1}\left(\frac{\mathrm{Im}(c_d)}{\mathrm{Re}(c_d)}\right)$$

[0152] Turbulence estimation is calculated as follows:

$$t_d = 1 - \left(\frac{|c_d|}{P_d}\right)$$

where

[0153] s is the complex representation of a scan line (s=I+j Q)

[0154] d is the sample at a depth in a scan line.

[0155] k is the scan line number in a set of scan lines.

[0156] When the system is configured for power Doppler, only the power estimation is required. The correlation between adjacent scan lines, velocity, and turbulence are not required. When the system is configured for color Doppler, the system calculates velocity, and optionally power and turbulence. The output of the flow and turbulence estimation is scan converted to a rectangular array of size 200×200×8 for each flow estimator. The rectangular arrays are compressed using a simple lossless compression algorithm. FIG. **17** provides an illustration of a typical Doppler window **1702**, where the Doppler processing area is limited with respect to the full B-mode image **1701** size. This improves the compression of the 200×200×8 arrays, as a large proportion of the array is empty. The compressed arrays are transmitted to the host, with a further interpolation step to the required Doppler image window, and color map conversion. Positive velocities are encoded in red shades, while negative velocities are encoded in blue shades. Turbulence is encoded in green. When power Doppler is selected, only red shades are used to represent the total power of the Doppler signals.

[0157] An alternative version of the invention is to provide a system which plots power Doppler with different color schemes used depending on whether the direction of flow is positive or negative. The power Doppler is calculated as above, scan converted, and transmitted to the host. In parallel, an efficient method is used to determine the direction of the flow. The full correlation between adjacent scan line points can be used, with the sign of the real and imaginary outputs used for each point to determine a direction. Alternatively, for imaging where sensitivity is not an issue, the correlation to decode direction can be shortened, and use a smaller set of scan lines. Power is optimized by providing a system which has feedback, where a small set of scan lines is used, and if the direction result is unstable (changing rapidly) the number of scan lines is increased to maximum. Then when the result is stable, the system runs a test correlation in parallel on a smaller set of scan lines and if that result is stable moves to using the smaller set. The system therefore optimizes the power transmitted into the body, and minimizes power consumption.

[0158] The system supports Cine playback or video recording. For Cine playback, the ultrasound probe stored compressed 500×500×8 B-mode frames into local DSP memory. 100 frames provides around five seconds of automatic Cine recording. When the user instructs the host user interface to scroll back through recorded frames, the DSP transmits the compressed frames up to the host and scan conversion is completed as per normal operation. For video recording, when the user instructs the host to record a segment to video, the DSP activates operations to complete any scan conversion and Doppler combining, and runs a video encoder which records the real-time images to a video format saving directly to memory, preferably at 640×480 pixel resolution. The video file can then be played back by a video decoder streaming data to a frame buffer which is transmitted to the host for display.

[0159] The system can be configured to operate in M-mode or duplex mode where a B-mode image and M-mode image are shown on the same screen.

[0160] In M-mode or Duplex mode, the DSP reads the M-mode scan lines and interpolates each scan line to a 500×1×8 buffer. The M-mode buffer is transmitted to the host which performs grey scale adjustment and renders the scan line to the display.

[0161] For duplex mode, two sorts of M-mode are possible. Full duplex is where an M-mode scan line is placed on a realtime B-mode image, and the M-mode pulse repetition rate is the same rate as the B-mode image update rate, typically 20 frames per second.

[0162] For quasi duplex mode, a high M-mode frame rate is required, such as for looking at rapidly moving heart valves. In this mode, an M-mode line is placed on a B-mode image, and when the select button is pressed the B-mode image is frozen and the M-mode graph is rendered to the screen.

[0163] A quasi duplex gated Doppler mode is also provided. A Doppler line is placed by the user on a B-mode image, and a gate area moved along the line to select the area for Doppler analysis. The gate angle can also be adjusted. When the select button is activated, the B-mode image is frozen, and a spectral Doppler graph—which displays time, frequency and magnitude of the Doppler shift—is plotted. The Doppler signal is processed in a similar way to color Doppler, except the scanlines are repeatedly pulsed only along the selected line (rather than moving across the image). In addition, the user selects a particular depth of interest, and the received echoes are 'range gated' such that only Doppler information from said depth of interest is analysed. This provides a large number of scanlines from the region of interest; frequency analysis is carried out on large groups of these scanlines (say 256) at a time, with the preferred method being applying the Fast Fourier Transform (FFT) on the complex representation of a group of scanlines. Each FFT performed provides a frequency spectrum of the Doppler signal shift, and these spectra can be displayed together to generate a time-frequency plot of the flow characteristics at the depth of interest which is often called 'spectral Doppler'. Spectral Doppler enables visualisation of how the velocity and power of blood flow changes with time for the selected region of interest.

[0164] The drawings and foregoing description relate to exemplary versions of the invention, and it should be recognized other versions of the invention are possible. The scope of this patent is not to be limited to the details described herein, but extends to all versions of the invention which are literally encompassed by the following claims, and to equivalents of such versions of the invention.

The claimed invention is:

1. A method of producing an ultrasound image, the method including the steps of:
- a. acquiring an ultrasound scanline from the transducer array, wherein this acquiring step includes:
  - (1) firing at least one of the transducers into a body to be imaged, and
  - (2) receiving ultrasound echoes returned from the body,
- b. moving the transducer array an incremental distance,
- c. acquiring further ultrasound scanlines from the transducer array,
- d. repeating the foregoing steps b. and c. until the transducer array is moved a selected distance, the selected distance being substantially greater than the incremental distance,
- e. displaying the acquired scanlines in a spatial relationship proportional to the spatial relationship in which the scanlines were captured, and
- f. repeating the foregoing steps b. through e. to generate a real time ultrasound image.

wherein the distance between adjacent transducers in the transducer array is greater than the minimum distance between adjacently-acquired scanlines needed to produce an ultrasound image having a desired resolution.

2. The method of claim **1** wherein the transducer array repetitively moves along a fixed path.

3. The method of claim **2** wherein the total distance traveled by each of the transducers along the fixed path is at least substantially equal to the distance between adjacent transducers within the transducer array.

4. The method of claim **2** wherein:
- a. the fixed path is circular,
- b. each transducer within the array only acquires scanlines over a defined sector of the circle.

5. The method of claim **1** wherein:
- a. after moving the transducer array an incremental distance, the transducer array is held stationary, whereby the transducer array is moved in a series of discrete steps; and
- b. the ultrasound scanlines are acquired when the transducer array is stationary.

6. The method of claim **1** wherein:
- a. the movement of the transducer array, and
- b. the acquisition of ultrasound scanlines,

are performed substantially continuously.

7. The method of claim **1** wherein:
- a. the transducer array is within a probe, and
- b. the motion of the transducer array occurs within the probe.

8. A method of producing an ultrasound image, the method including the steps of:
- a. moving an array of one or more ultrasound transducers in a repetitive motion along a path with respect to a body to be imaged,
- b. generating an ultrasound image, the image including scanlines acquired by the transducers at consecutive locations along the path of motion, with each transducer contributing multiple scanlines to the image.

9. The method of claim **8** wherein:
- a. during the repetitive motion of the transducer array, the transducer array is periodically held stationary, whereby the transducer array is moved in a series of discrete steps; and
- b. the ultrasound scanlines are acquired by the transducers when the transducer array is stationary.

10. An ultrasound imaging device configured to produce an ultrasound image having a selected resolution, the device including a transducer array wherein the distance between adjacent transducers is greater than the minimum distance between adjacent scanlines required to produce an ultrasound image having the selected resolution.

11. The device of claim **10** wherein the transducer array is supported by a moving element configured to repetitively move the transducer array along a fixed path.

12. The device of claim **11** wherein:
- a. the transducer array is situated within a probe, and
- b. the repetitive motion of the transducer array occurs within the probe.

13. The device of claim **10** wherein:
- a. the transducer array is supported by a translating element, the translating element being configured to move in a reciprocating motion by a selected translational distance along a path,
- b. the linear distance between the transducers of the transducer array along the linear path being at least substantially the same as the selected translational distance,
- c. the transducer array is configured to acquire multiple scanlines during each reciprocation.

14. The device of claim **13** wherein the translating element is driven by an ultrasonic motor.

15. The device of claim **13** wherein the translating element is driven by a linear motor.

16. The device of claim **13** wherein the translating element is driven by a rotary motor.

17. The device of claim **13** wherein:

a. the translating element has a curvilinear shape, and

b. the linear path is a curvilinear path.

18. The device of claim **10** wherein:

a. the transducer array is supported by a rotating element, with the transducers being arranged along an arc on the rotating element; and

b. the rotating element is configured to rotate such that the scanline acquired by each transducer sweeps out a sector of a circle.

19. The device of claim **18** including a controller controlling the stimulation of the transducers to acquire scanlines therefrom, the stimulation being controlled such that the transducers are stimulated only when positioned above a sector of a circle which extends into a body to be imaged.

20. The device of claim **18** wherein the rotating element is an annulus.

* * * * *

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摘要(译)

一种用于超声成像的方法和设备，包括换能器阵列，其中相邻换能器之间的距离大于产生所选分辨率的超声图像所需的相邻扫描线的最小间隔。换能器阵列适于机械移动，以便可以扫描阵列以产生扫描线。