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(54) METHOD AND APPARATUS TO PRODUCE ULTRASONIC IMAGES USING MULTIPLE APERTURES

VERFAHREN UND VORRICHTUNG ZUR HERSTELLUNG VON ULTRASCHALLBILDERN MITHILFE MEHRERER ÖFFNUNGEN

PROCÉDÉ ET APPAREIL DE PRODUCTION D'IMAGES ULTRASONORES AU MOYEN D'UNE PLURALITÉ D'ORIFICES

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Description

BACKGROUND OF THE INVENTION

Technical Field

[0001] The present invention relates generally to imaging techniques used in medicine, and more particularly to medical ultrasound, and still more particularly to an apparatus for producing ultrasonic images using multiple apertures.

Background Art

[0002] In conventional ultrasonic imaging, a focused beam of ultrasound energy is transmitted into body tissues to be examined and the returned echoes are detected and plotted to form an image. In echocardiography the beam is usually stepped in increments of angle from a center probe position, and the echoes are plotted along lines representing the paths of the transmitted beams. In abdominal ultrasonography the beam is usually stepped laterally, generating parallel beam paths, and the returned echoes are plotted along parallel lines representing these paths. The following description will relate to the angular scanning technique for echocardiography (commonly referred to as a sector scan). However, the same concept with modifications can be implemented in abdominal scanners.

[0003] The basic principles of conventional ultrasonic imaging are well described in the first chapter of Echocardiography, by Harvey Feigenbaum (Lippincott Williams & Wilkins, 5th ed., Philadelphia, 1993). These will not be repeated here except as necessary to illustrate the differences between the conventional techniques and the present invention.

[0004] It is well known that the average velocity v of ultrasound in human tissue is about 1500 m/sec, the range in soft tissue being 1440 to 1670 m/sec (see for example P. N. T. Wells, Biomedical Ultrasonics, Academic Press, London, New York, San Francisco, 1977). Therefore, the depth of an impedance discontinuity generating an echo can be estimated as the round-trip time for the echo multiplied by v/2, and the amplitude is plotted at that depth along a line representing the path of the beam. After this has been done for all echoes along all beam paths, an image is formed, such as the image **10** shown in Figure 1, in which a circle has been imaged. The gaps between the scan lines are typically filled in by interpolation. One of the earliest interpolation algorithms applied to echocardiography was described in U.S. Patent 4,271,842, to Specht, et al.

[0005] In order to insonify the body tissues, a beam formed either by a phased array or a shaped transducer is scanned over the tissues to be examined. Traditionally, the same transducer or array is used to detect the returning echoes. This design configuration lies at the heart of one of the most significant limitations in the use of ultrasonic imaging for medical purposes; namely, poor lateral resolution. Theoretically the lateral resolution could be improved by increasing the aperture of the ultrasonic probe, but the practical problems involved with aperture size increase have kept apertures small and lateral resolution large. Unquestionably, ultrasonic imaging has been very useful even with this limitation, but it could be more effective with better resolution.

[0006] In the practice of cardiology the limitation on single aperture size is dictated by the space between the ribs (the intercostal spaces). For scanners intended for abdominal and other use, the limitation on aperture size is not so obvious, but it is a serious limitation nevertheless. The problem is that it is difficult to keep the elements of a large aperture array in phase because the speed of ultrasound transmission varies with the type of tissue between the probe and the area of interest. According to the book by Wells (cited above), the speed varies up to plus or minus 10% within the soft tissues. When the aperture is kept small, the intervening tissue is, to a first order of approximation, all the same and any variation is ignored. When the size of the aperture is increased to improve the lateral resolution, the additional elements of a phased array may be out of phase and may actually degrade the image rather than improving it. The instant disclosure teaches methods to maintain all of the information from an extended phased array "in phase" and thus to achieve sought-after improved lateral resolution.

[0007] In the case of cardiology, it has long been thought that extending the phased array into a second or third intercostal space would improve the lateral resolution, but this idea has met with two problems. First, elements over the ribs have to be eliminated, leaving a sparsely filled array. New theory is necessary to steer the beam emanating from such an array. Second, the tissue speed variation described above, but not adequately addressed until this time, needs to be compensated. The same solution taught in this disclosure is equally applicable for multi-aperture cardiac scanning, or for extended sparsely populated apertures for scans on other parts of the body.

[0008] US 2006/0122506 discloses an ultrasound imaging system comprising ultrasound transmitters and receivers.

Disclosure of Invention

[0009] According to the present invention there is provided an ultrasound imaging system according to claim 1 and a method of reconstructing an image according to claim 5.

[0010] The present invention solves both the problem of using more than one intercostal space and the problem of accommodating unknown phase delays from using elements spread over a large sparse aperture. The solution involves separating the insonifying probe from the imaging elements.

[0011] A single omni-directional receive element (such as a receive transducer) can gather all of the information necessary to reproduce a two-dimensional section of the body. Each time a pulse of ultrasound energy is transmitted along a particular path, the signal received by the omnidirectional probe can be recorded into a line of memory. [The terms "omni-directional probe," "omni probe" and/or "omni," are used synonymously herein to mean an omnidirectional probe.] When this is done for all of the lines in a sector scan, the memory can be used to reconstruct the image. This can be accomplished in the same time as data is being collected for the next frame.

[0012] There are numerous advantages to this approach, and these comprise the objects and advantages of the present invention. They include, among others:

[0013] The dominance of specular reflections so prominent in reconstructing images by returns to the main probe is greatly attenuated.

[0014] More than one omnidirectional probe can be used. Each one can be used to reconstruct an entire sector image but with different point spread functions. These can be combined to produce an image with a sharper point spread function.

[0015] Compensations can be made for different delays in different paths through the tissue.

[0016] Many more scan lines can be reconstructed than the number of pulses generated by the main probe. This overcomes the traditional limit of the number of scan lines by the speed of ultrasound in tissue, tissue depth of interest, and the time allowed between frames, which is typically 1/30th second.

[0017] Artificial scan lines can be considered as overlapping, and each pixel on an output image can be imaged from information from more than one omni line of data. Therefore the output pixel can be averaged from multiple data, thus improving the signal-to-noise ratio.

[0018] Omnidirectional probes can be placed in multiple intercostal spaces, the suprasternal notch, the substernal window, multiple apertures along the abdomen and other parts of the body, and even on the end of a catheter. An advantage in using omnidirectional probes for catheter placement is that no steering is required of the probe.

[0019] Probes can be placed either on the image plane, off of it, or any combination. When placed away from the image plane, omni probe information can be used to narrow the thickness of the sector scanned.

[0020] There has thus been broadly outlined the more important features of the invention in order that the detailed description that follows may be better understood, and in order that the present contribution to the art may be better appreciated. Additional objects, advantages and novel features of the invention will be set forth in part in the description as follows, and in part will become apparent to those skilled in the art upon examination of the following. Furthermore, such objects, advantages and features may be learned by practice of the invention, or may be realized and attained by means of the instrumentalities and combinations particularly pointed out in the appended claims.

[0021] Still other objects and advantages of the present invention will become readily apparent to those skilled in this art from the following detailed description, which shows and describes only the preferred embodiments of the invention, simply by way of illustration of the best mode now contemplated of carrying out the invention. As will be realized, the invention is capable of modification in various obvious respects without departing from the invention. Accordingly, the drawings and description of the preferred embodiment are to be regarded as illustrative in nature, and not as restrictive.

Brief Description of the Drawings

[0022]

Fig 1 is a diagrammatic view of a simulation showing a circular object imaged by a conventional sector scanner.

Fig 2 is a schematic diagram of the axes representing the relative positions of the insonifying and omni-directional probes.

Fig 3 is a graph showing the orientation of the point spread function of an imaging system using a single omnidirectional probe as a function of the position of the omnidirectional probe.

Fig 4 is graph showing the orientation of the point spread function of an imaging system using a single omnidirectional probe as a function of the position of the omnidirectional probe when it is not in the scan plane.

Fig 5 is a diagrammatic view of a simulation showing the same circular object of Fig 1 imaged by data received by a single omni element located at x_0 = 40 mm, y_0 = 0 mm, z_0 = 0 mm.

Fig 6 is a schematic diagram illustrating the relative positions of probes showing, in addition, points A-A', which have equal round trip distances and times.

Fig 7 is a schematic diagram showing a possible fixture for positioning an omni-directional probe relative to the main (insonifying) probe.

Fig 8 is a schematic diagram showing a non-instrumented linkage for two probes.

Fig 9 is a schematic diagram showing variables for computation of x and y positions from received echoes.

Fig. 10a is a phantom image taken with a standard Acuson 128 XP-10 with a 3.5 MHz transducer and harmonic processing.

Fig 10b is the same phantom image as that shown in Fig. 10A and taken with the same XP-10, wherein the center 64 elements were obscured but external processing employed to show improved lateral resolution. The progressions of anechoic areas on the phantom are 8mm diameter, 6mm, 4mm, 3mm, and 2mm.

Fig 11 is an image of the same phantom produced by the same transducer as the images in Figs. 10a and 10b, with the center obscured, but with substantial image processing over multiple scans. Note that even though the total aperture is only 19mm that the 2mm diameter anechoic areas are now visible. Lateral resolution could be greatly improved if the two parts of the transducer were physically separated and the phased delays reprogrammed for the resulting geometry.

Figs. 12a is a schematic perspective view showing an adjustable, extendable hand held two-aperture probe especially adapted for use in cardiology US imaging. This view shows the probe in a partially extended configuration.

Fig 12b is a side view in elevation thereof showing the probe in a collapsed configuration.

Fig 12c shows the probe extended so as to place the heads at a maximum separation distance permitted under the probe design, and poised for pushing the separated probe apertures into a collapsed configuration.

Fig 12d is a side view in elevation again showing the probe in a collapsed configuration, with adjustment means shown (i.e., as scroll wheel).

Fig 12e is a detailed perspective view showing surface features at the gripping portion of the probe.

Fig 13 is a 3D image highlighting the anechoic tubes of the ATS Model 539 phantom.

Best Mode for Carrying Out the Invention

[0023] A key element of the present invention is that returned echoes in ultrasonography can be detected by a separate relatively non-directional receive transducer located away from the insonifying probe (transmit transducer), and the non-directional receive transducer can be placed in a different intercostal space from the insonifying probe. It can also be located in the suprasternal, subcostal, or apical positions. This probe will be called an omni-directional probe because it can be designed to be sensitive to a wide field of view.

[0024] If the echoes detected at the omni probe are stored separately for every pulse from the insonifying transducer, it is surprising to note that the entire two-dimensional image can be formed from the information received by the one omni. Additional copies of the image can be formed by additional omni-directional probes collecting data from the same set of insonifying pulses.

[0025] A large amount of straightforward computation is required to plot the amplitude of echoes received from the omni. Refering now to Figure 2, in which there is shown the position of the omni-directional probe **100** relative to the position of the insonifying (main) probe **120** and the insonifying beam **130.** The position of the omni-directional probe relative to the beam is indicated by x_0, y_0 and z_0 **140,** where x_0 and y_0 are in the scan plane **150** scanned by the insonifying beam and z_0 is distance perpendicular to that plane. Instead of simply plotting the depth along the scan line d = t v/2 (where t is the round-trip time, it is now computed as that point d = sqrt(x^2 + y^2) for which t v = d + sqrt((x-x_0)^2+(y-y_0)^2 +z_0^2).

[0026] This procedure will produce a sector scan image similar to that using the conventional technique except that the point spread function will be rotated. Figure 3 shows the orientation **200** of the psf as a function of the position of the omni probe. A single point at x = 0, y = 70mm, z = 0 is the point being imaged, and the groups of dots **210** each indicate the orientation of the psf if the omni-probe were placed at the location of the center of each group. The insonifying probe (main probe) is located at the center group **220** on the bottom of the figure. In this simulation the horizontal (x axis) **230** shown goes from -40 mm to +40 mm. The vertical (y axis) **240** goes from 0 to 80 mm depth.

[0027] Figure 4 shows the orientation **300** of the psf as a function of the position of the omni probe if it is not in the scan plane. In this simulation the horizontal (x axis) **310** shown again goes from -40 mm to + 40 mm, but the vertical axis **320** is the z axis (distance away from the scan plane) and goes from 0 to 80 mm away from the scan plane.

[0028] Figure 5 shows a plot of the same circular object **20** as in Figure 1, plotted, however, from data received by a single omni element. Other complete two dimensional reconstructions may be formed using data from additional omni elements, if desired.

[0029] Specular reflection is reduced using the omni probe compared to using the main probe for both insonification and detection. This is because all parts of a surface normal to the main beam are insonified with the same phase. When the phase is such that maximum echo is returned, all the echoes add to produce a specular echo. When the signals are normalized to accommodate the dynamic response of a particular display device, non-specular echoes will tend to drop out.

[0030] In contrast to this, and referring now to Figure 6, there is also shown the relative positions of probes, but there is also shown points A-A' **400,** which have equal round trip distances and times. Points A-A' are on a surface normal to the bisector line $a_1 = a_2$ **410**, which also have equal round trip distances and times, are not insonified with the same phase, and do not all reflect equally. This attenuation of the specular reflection is particularly important when visualizing circular structures which frequently have surfaces normal to the main beam.

[0031] An algorithm to plot this data on a rectangular grid is: (a) for each point on the x,y grid, convert x,y to depth and angle; (b) then find closest angle k scanned by the insonifying beam; (c) if it is sufficiently close, then convert x,y to distance to the omni; (d) compute time t=(distance to insonifying beam + distance to omni)/v; and (e) plot amplitude recorded by the omni for the k scan at x,y.

[0032] However, more information is now available and should be used. It is possible to use the same technique to plot additional scan lines which were not explicitly insonified by the insonifying probe. Because of the inherently wide beam width of medical ultrasonic probes, much tissue between intentional scan lines is also insonified and returns echoes. Making use of this information is particularly important when capturing cardiac motion because the number of pulses that can be generated is strictly limited by the speed of ultrasound in tissue and the scan repetition rate desired.

[0033] It is expected that the reconstructed image would get better as the angle between the main beam and the omni gets larger. However it is not necessary to focus a narrow beam on every element of tissue to be imaged as is true if the data is not stored and then processed before display. The lateral resolution can be reconstructed using a Wiener Filter to be much better than the beam width if the noise spectrum is low enough. In one simulation of 2 circles of diameter 2.2 mm and 4.0 mm, both imaged well enough that the center was clear even though the beam width was 4.4 mm tapered from 1 to 0 by a cosine function. The Wiener Filter is described in the next section.

[0034] There are four main sources of noise in ultrasonic imaging: (1) blur due to array size not wide enough; (2) shot noise; (3) reverberation from big interfaces; and (4) speckle.

[0035] Multiple probes give independent measures of shot noise, but using closely spaced elements in the main probe (if it is a phased array) will not give independent noise for the other three sources. Adding one or more omni probes will change the look angle, which will thereby change the speckle pattern and the reverberation pattern. These can be averaged out to lower the noise power spectrum. The Wiener filter can then be employed to cancel the blur.

[0036] Another way to eliminate speckle is to obtain a good sample of it for estimates of the noise spectrum to then be used in the Wiener filter.

[0037] It is hoped that de-blurring and de-noising by these techniques using only an external omni probe or probes will make it possible to visualize the coronary arteries to assess degree of opening in the lumen or patency of bypass grafts without resort to invasive catheterization techniques.

[0038] When combining more than one image such as from the main probe and one or more omni probes for the purpose of averaging out the various sources of noise, it is necessary to compensate for the variation is ultrasound velocity through different paths. Experiments have shown that small unaccounted errors in path velocity will displace the reconstructed image in both horizontal and vertical positions. Cross correlation techniques should be used to find the displacement with one image taken as reference before addition or other combination of images.

[0039] Two possibilities exist with regard to the Wiener filter. In one, a Wiener filter can be used separately on each image and then combine them. Alternatively, one can first combine the images (yielding a more complex point spread function) and then employ Wiener filtering.

[0040] In order to perform the indicated computations, it is necessary to either measure or estimate the position (x_0, y_0, z_0) of the omni relative to the main probe. Exact knowledge of these positions is not required because, as we have seen, the displacement of the image from the omni probe is also affected by the variation of the velocity of ultrasound in different types of tissue. For this reason it is necessary to use cross correlation or some other matching criterion to make a final correction of the position of the omni-generated image before combining with the reference image or images.

[0041] Determining the Position of the Omni Probe(s): There are many ways to determine the position of the omni probe. Referring now to Figure 7, one way is to provide apparatus 500 for pivotally and/or swivelingly mounting the omni probe **510** on a fixture **520** attached to the insonifying probe **530.** The fixture preferably includes articulated joints **540** with sensors (not shown) to measure angles and distances **550** of the links. Figure 7 illustrates a simplified version of such a fixture, wherein fixed hinges allow movement of only x_0 and y_0.

[0042] Another method is to have no mechanical connection between the omni probe and the main probe (except wires for signals and power). Instead, the omni probe can transmit a signal using radio frequencies, light, or a different frequency ultrasound to triangulation receivers mounted on the main probe or a separate platform.

[0043] A third method again has no mechanical connection between the omni probe and main probe. For this method the omni probe (or probes) can be attached to the patient with tape, and the ultrasonographer can manipulate the main probe to find the best image without regard to positioning the omni probe(s). As indicated in Figure 4, a two dimensional image can be formed separately from the echoes received from the omni probe and from the main probe. By adjusting four variables (x_0, y_0, z_0, and D, the average difference in time for ultrasound to go through different tissue types instead of traveling though idealized tissue of constant ultrasound velocity), the images can be made to coincide. The four variables can be adjusted iteratively to maximize cross-correlation or another measure of similarity. Standard multi-dimensional search techniques that could be used include gradient ascent, second order (Newton's method), simulated annealing, and genetic algorithms. A fifth variable, the angle of the psf which is necessary for deconvolution, is implied from the first four variables. Misregistration of the images can be caused by inaccurate estimation of any of the four variables, but good registration can be achieved by simply adjusting D which will tend to compensate for errors in estimates of the others.

[0044] When the application requires the highest resolution compatible with capturing motion at a high frame rate, the four variables can be estimated over several frames of information. When the ultrasonographer has selected a good view angle, the frames can be combined at high rate holding x_0, y_0, z_0, and D constant.

[0045] When the application requires the highest possible resolution, data can be captured (perhaps with EKG gating to capture separate images at systole and diastole) and the multi-dimensional search to optimize matching can be done more accurately although not in real time. Two advantages of this approach is that different values of D can be found for systole and diastole, and that different psf's can be used for deconvolution at different depths in the image.

[0046] Determining the Position of the Omni Probe(s) Using Correlation of the Scan Line Data Rather than Complete 2D Sectors: A fourth method for determining the position of the omni probe(s) entails replacing the omni probe or probes with a "semi-omni probe" or probes. The reason for this is to increase the signal to noise ratio by restricting the sensitive region of the receive transducer to a plane rather than a hemisphere. Because of this restriction it is necessary to have a mechanical linkage to ensure that both the transmit and receive transducers are focused on the same plane.

[0047] Two probes could be placed in any two acoustic windows. One would likely be in the normal parasternal window which typically gives the clearest view of the whole heart. A second window available in most patients is the apical view. Another window usually available is the subcostal. The two probes could even be placed on either side of the sternum or in parasternal windows in two intercostal spaces.

[0048] One probe could be the standard phased array cardiac probe. The second (and third, etc.) would be used as receive only. Theoretically it could be omnidirectional, but that would necessarily provide lower signals and therefore low signal to noise ratios (S/N). A better alternative is to use a probe which is ground to be sensitive to a plane of scan but omnidirectional within that plane. A single piece of PZT would work well, but to minimize the amount of new design required it is also possible to use a second probe head similar to the main probe and then disable all elements except a small group combined to act as a single element. The design goal is to use as many elements as possible to maximize signal to noise ratio while using few enough to minimize angle sensitivity.

[0049] In this embodiment **600** (see Figure 8), the two probes **610, 620,** may be linked together with an articulating mechanical linkage, which ensures that the plane of scan of each probe includes the other, but the distance between them is unconstrained. A slave servomechanism is also possible, but the mechanical linkage will be described here.

[0050] The procedure is to aim the main probe **620** at the target **630** (e.g., heart) and position the secondary probe **610** at a second window with maximum received signal strength. One possibility is that the main probe be positioned for a long axis view with the secondary probe over the apex of the heart. Some slight deviation of the long axis view may be necessary in order to maintain the secondary probe in its most sensitive spot.

[0051] The secondary probe would now be held on the patient with an elastic strap which allows a rocking motion. The disadvantage of having two probes in fixed positions on the body is that the plane of scan must include these two points. The only degree of freedom is the angle at which the scan plane enters the patient's body. For a conventional 2D examination this is a severe limitation, but if the goal is to gather three dimensional information, this is not a limitation. The 3D information is obtained by rocking the main probe back and forth through a sufficient angle so that the entire heart is insonified. The secondary probe also rocks back and forth by virtue of the mechanical linage between the probes. The instantaneous angle of rocking must be monitored-perhaps by reference to a gyroscope mounted with the main probe. The rocking could be actuated by the hand motion of the ultrasound technician, or it could be motorized for a more-uniform angle rate. In an alternative preferred embodiment, the main probe and an array of omni probes are placed in adjacent intercostal spaces using a mechanism as shown in Figure 12.

[0052] The need for and one important use of the 3D information is covered in co-pending United States Utility Patent Application Serial 11/532,013, also by the present inventor.

[0053] Yet another variation on this theme is to have the secondary transducer mechanically linked to the primary so that each plane of scan contains the other transducer (as above), but allow rotation of the main probe about its own axis. In this case the secondary probe would be allowed to move on the patient's body (properly prepared with ultrasound gel). It would have many elements, and an attached computer would scan them all to find those elements which have the strongest return signal.

[0054] Estimating Relative Probe Positions from Reflected Signals: For image reconstruction it is essential to know the position of the secondary probe (x, y) relative to the main probe. This has to be evaluated separately for each frame of data because of the motion of the patient, technician, and/or motorized angle actuator. Since the linkage will prevent any difference in position (z) perpendicular to the scan plane, only x and y need be assessed.

[0055] Note that any tilt of the main probe will change the reference axes so that x and particularly y will change too.

[0056] When a pulse is transmitted from the main probe it insonifies a sequence of tissues in the path of the beam. The returns from the tissues will be received by both the main probe and the secondary, digitized and stored in the computer. Echoes from relatively proximate tissues will be different for the two probes, but echoes from mid- to far range will be similar. It is possible to use cross correlation to find similar small patches in the two stored returns. They will be similar except for the time delay relative to the launching of the pulse from the main probe. The time delay will be related to the offsets x and y. Values for x and y cannot be determined from one set of time delays, but can be determined by solving a set of simultaneous equations from two detected similar returns. These could be different patches of the same pulse return or from returns from differently directed main pulses.

[0057] Referring now to Figure 9, if the main probe **700** transmits at angle 90° - α **710** relative to its centerline and an identifiable packet of returns occurs at time t_{1m} at the main probe and at time t_{1s} at the secondary (omni) probe **720,** then:

tissue packet at (x_1, y_1) is received at time t_{1m}, and distance $m_1 = \text{sqrt}(x_1^2 + y_1^2)$
tissue packet at (x_2, y_2) is received at time t_{2m}, and distance $m_2 = \text{sqrt}(x_2^2 + y_2^2)$
t_{1m} corresponds to time of two trips of distance m_1
t_{1m} s = 2 m_1, where s = speed of ultrasound in same units as m = approx. 1.54 x 10^6 mm/sec

$$t_{1s}\,s = m_1 + \text{sqrt}((x_1 - x)^2 + (y_1 - y)^2).$$

[0058] Similarly, t_{2m} S = 2 m_2

$$t_{2s}\,s = m_2 + \text{sqrt}((x_2 - x)^2 + (y_2 - y)^2)$$
$$(t_{1s}\,s - 0.5\,t_{1m}\,s)^2 = (x_1 - x)^2 + (y_1 - y)^2 \qquad (1)$$
$$t_{2s}\,s - 0.5\,t_{2m}\,s)^2 = (x_2 - x)^2 + (y_2 - y)^2$$

[0059] Since x_1, y_1, x_2, y_2 and the times are known, one can solve the last two simultaneous equations for x and y.

[0060] Similarly, if a z offset between the two probes is allowed, x, y, and z can be calculated by solving three simultaneous equations.

[0061] Many more measurements from packet pairs are available. One could make a measurement on several or every scan line (angle) as measured from the main probe. Then we would have many equations in 2 unknowns which can be used to make more-accurate estimations of the 2 unknowns. Since these are nonlinear equations, a search technique can be utilized. One way to accomplish this is to compute error squared over a grid of (x, y) points using the equation:

$$E^2 = \sum_{i=1}^{N} (\sqrt{(x_i - x)^2 + (y_i - y)^2} - t_{is} + 0.5 t_{im} s)^2 \qquad (2)$$

[0062] The minimum E^2 will indicate the minimum squared error estimate of (x, y). The search should be conducted over the expected range of x and y to save time and to avoid spurious ambiguous minima.

[0063] When the z component of the relative position is not constrained to be zero, the comparable error squared equation is:

$$E^2 = \sum_{i=1}^{N} (\sqrt{(x_i - x)^2 + (y_i - y)^2 + (z_i - z)^2} - t_{is} + 0.5 t_{im} s)^2 \qquad (3)$$

[0064] The minimum E^2 will indicate the minimum squared error estimate of (x, y, z).

[0065] If the speed of sound on the return path to the secondary (omni) transducer is different from s due to different types of tissues being traversed, the values of x and y (and z if used) will be different from the geometric values. However, use of these values in the image reconstruction algorithm will automatically compensate for the different speeds.

[0066] The Wiener Filter: The Wiener filter itself is not new, but since it is important for the de-blurring step it will be described briefly here in the context of the present invention.

[0067] The Wiener filter is the mean squared error optimal stationary linear filter for images degraded by additive noise and blurring. Wiener filters are usually applied in the frequency domain. Given a degraded image i(n.m.), one takes the discrete Fourier Transform (DFT) or the Fast Fourier Transform (FFT) to obtain I(u,v). The true image spectrum is estimated by taking the product of I(u,v) with the Wiener filter G(u,v):

$$\hat{S} = G(u,v)I(u,v)$$

[0068] The inverse DFT or FFT is then used to obtain the image estimate s(n.M.) from its spectrum. The Wiener filter is defined in terms of the following spectra:

(a) H(u,v) - Fourier transform of the point spread function (psf);
(b) P_s(u,v) - Power spectrum of the signal process, obtained by taking the Fourier transform of the signal autocorrelation;
(c) P_n(u,v) - Power spectrum of the noise process, obtained by taking the Fourier transform of the noise autocorrelation;

[0069] The Wiener filter is:

$$G(u,v) = \frac{H^*(u,v)P_s(u,v)}{\left|H(u,v)\right|^2 P_s(u,v) + P_n(u,v)}$$

[0070] The ratio P_s / P_n can be interpreted as signal-to-noise ratio. At frequencies with high signal to noise ratio, the Wiener filter becomes H^{-1}(u,v), the inverse filter for the psf. At frequencies for which the signal to noise ratio is low, the Wiener filter tends to 0 and blocks them out.

[0071] $P_s(u,v) + P_n(u,v) = |I(u, v)|^2$. The right hand function is easy to compute from the Fourier transform of the observed data. P_n(u,v) is often assumed to be constant over (u,v). It is then subtracted from the total to yield P_s(u,v).

[0072] The psf can be measured by observing a wire phantom in a tank using the ultrasound instrument. The Fourier transform of the psf can then be stored for later use in the Wiener filter when examining patients.

[0073] Because the psf is not constant as a function of range, the Wiener filter will have to be applied separately for several range zones and the resulting images will have to be pieced together to form one image for display. A useful compromise might be to optimize the Wiener filter just for the range of the object of interest such as a coronary artery or valve. It will be necessary to store separate Wiener filters for each omni-directional probe and for the main probe when it is used as a receive transducer.

[0074] An alternative to the Wiener Filter is the least mean square (LMS) adaptive filter described in co-pending U.S. Patent Application Serial Number 11/532,013. LMS Filtering is used in the spatial domain rather than the frequency domain, and can be applied to the radial scan line data, the lateral data at each depth, or both together.

[0075] Another alternative for the deblurring step is to use unsharp masking. Because aperture blur is much more pronounced in the lateral dimension (perpendicular to the insonifying beam) than in the radial dimension, it is necessary to perform unsharp masking in only one dimension. When using a sector scanner, this masking should be performed before scan conversion. When using a linear phased array, the unsharp masking should be performed on each data set of constant range. Unsharp masking consists of intentionally blurring an image, subtracting the result from the original image, multiplying the difference by an arbitrary factor, and adding this to the original image. In one dimension this is the same as blurring a line of data, subtracting it from the original line, and adding a multiple of the difference to the original line.

[0076] Multiple Active Transducers-Two Alternative Approaches: It is possible to use more than one active transducer placed at multiple acoustic windows in order to achieve the same goals of increased lateral resolution and noise suppression. A practical method of providing multiple omni probes is to use a second phased array head in a second acoustic window and then treating each element or group of elements of the second phased array as a separate omni. With this configuration of probes it would be possible to switch the functions of the two probe heads on alternate scans thereby generating images with different speckle patterns which can be averaged out.

[0077] Multiple phased array heads can also be used together so that both are active on the same scan. When two (or more) phased array transducers are placed in the same scan plane, they can be programmed with delays such that they act as a single array with a gap in the array of transducer elements. The advantages of having a gap in the array include a) achieving the lateral resolution of a wide aperture without the expense of filling in acoustic elements through the gap, and b) the gap in the probe or between probes can be fitted over ribs or the sternum. The first advantage applies equally to applications other than cardiac. The disadvantages of multiple active probes is that both the transmit and receive delays have to be recomputed for each new gap dimension and/or angular orientation of one probe relative to the others.

[0078] An active probe with a gap has been demonstrated to produce lateral resolution almost as good as the probe without the gap. Referring now to Figure 10a, there is illustrated the image **800** of an ATS Laboratories Model 539 phantom was imaged using an Acuson 128 XP-10 ultrasonic scanner with a 4V2c probe. In Figure 10b, the same probe was used to image the phantom with its center 64 elements totally obscured by aluminum foil and electrical tape. As can be seen, the lateral resolution in the image **900** is nearly as good as the original although the image quality is degraded by speckle and other noise.

[0079] Figure 11 shows an image **1000** of the same phantom produced by the same transducer with the center obscured, but with substantial image processing over multiple scans.

[0080] Figures 12a-e are various views showing an adjustable, extendable hand held two-aperture probe **1100** adapted for use in cardiology US imaging. This apparatus embodies the inventive concept of separating the insonifying probe **1110** (a transmit transducer) from the imaging elements **1120** (receiver transducer). This comfortable device includes adjustment means **1130,** such as a scroll wheel, which selectively drives the elements either closer or further apart along either a medial telescoping portion **1140** or a medial insertable sleeve, and thereby provides a range of separation at predetermined distances. The gripping portion **1150** provides easy access to the scroll wheel and places the user's hand in the functional position to minimize overuse injury. Figure 12a shows the probe in a partially extended configuration. Figure 12b shows the probe in a collapsed configuration. Figure 12c shows the probe extended so as to place the heads at a maximum separation distance permitted under the probe design. Figure 12d shows the probe in a collapsed configuration, with adjustment means shown. And Figure 12e is a detailed perspective view showing surface features at the gripping portion of the probe.

[0081] Figure 13 shows a three-dimensional display **1200** of the anechoic tubes of the Model 539 phantom. This 3D display was formed from 13 parallel slices produced with the same transducer with the center obscured. When the total aperture is increased it will be possible to display smaller anechoic tubes such as the coronary arteries. The processing involved for this display is a combination of the techniques of the instant patent and those of co-pending U.S. Patent Application Serial 11/532,013.

Claims

1. An ultrasound imaging system, comprising:

a transmit transducer array (620) configured for insonifying a selected object and/or area of living tissue from a first acoustic window with a plurality of directional ultrasound pulses;

at least one receive transducer (610) configured for detecting and recording ultrasound echo returns from a second acoustic window, wherein said transducer is relatively non-directional in a plane of scan; and

a computer in electronic communication with said transmit transducer array and said receive transducer(s), said computer having hardware and software configured for digitizing and storing each ultrasound echo in a line of computer memory, determining the location of said transmit transducer in relation to said receive transducer(s), reconstructing scan lines as lines on a reconstructed image based on the determination of the location of the receive transducer(s) relative to that of the transmit transducer and the average speed of ultrasound in the tissue object, and image processing to fill in gaps between scan lines and optionally correct distortions caused by the point spread function of the system,

characterized in that the transmit transducer array (620) and the at least one receive transducer (610) are linked together with an articulating mechanical linkage configured to ensure that the plane of scan of the transmit transducer array and the at least one receive transducer includes the other while the distance between the transmit transducer array and the at least one receive transducer is unconstrained.

2. The system of claim 1, further including 3D volume filling means to fill in a 3D volume when said apparatus is used for imaging 2D slices through living tissue mounted on a platform so as to successively scan slices in parallel planes and record the relative positions of each slice, said 3D volume filling means comprising a computer in electronic communication with said transducer arrays, said computer including application software which is configured to reconstruct each two dimensional image, assemble the 2D images into a 3D volume in computer memory using interpolation algorithms, align and perform 3D segmentation on the volume, and display the 3D volume as a projection on a plane, or a 2D slice of arbitrary orientation.

3. The system of claim 1, further including 3D volume filling means to fill in a 3D volume when said apparatus is used for imaging 2D slices through living tissue mounted on a platform so as to scan slices in successive angles and record the relative positions and angles of each slice, said 3D volume filling means comprising a computer in electronic communication with said transducer arrays, said computer including application software which is configured to reconstruct each two dimensional image, assemble the 2D images into a 3D volume in computer memory using interpolation algorithms, align and perform 3D segmentation on the volume, and display the 3D volume as a projection on a plane, or a 2D slice of arbitrary orientation,

4. The system of claim 1, wherein the transmit transducer array and said at least one receive transducer are each pivotally connected to an articulating arm so as to be mechanically linked and to move in the same plane.

5. A method of reconstructing an image, comprising the steps of:

(a) linking a transmit transducer (620) and a receive transducer (610) together with an articulating mechanical linkage that ensures that the plane of scan of the transmit transducer and the receive transducer includes the other while the distance between the transmit transducer and the receive transducer is unconstrained;
(b) insonifying an object one scan line at a time with a directional ultrasound pulse generated from a transmit transducer (620) positioned in a first acoustic window;
(c) recording echoes received by a receive transducer (610) that is relatively non-directional and positioned in a second acoustic window;
(d) storing the digitized echoes each in a line of computer memory;
(e) reconstructing scan lines as lines on the reconstructed image based on knowledge of the location of the receive transducer relative to that of the transmit transducer and the average speed of ultrasound in the object; and
(f) image processing to fill in gaps between scan lines and optionally correcting distortions caused by the point spread function of the system,
wherein the reconstructed scan lines are at arbitrary angles or locations and/or the number of such scan lines exceeds the number of actual transmitted pulses.

6. The method according to claim 5, wherein the transmit transducer (620) is a phased array type sector scanner which sends out pulses in successive angles from a common origin.

7. The method according to claim 5, wherein the transmit transducer (620) is a linear phased array which sends out pulses on parallel paths.

8. The method according to claim 5, wherein a separate image is reconstructed for each receive transducer (610) and these images are superimposed by finding the displacement which maximizes correlation to one chosen as the reference image and then averaging the images.

9. The method according to claim 5, wherein a separate image is reconstructed for each receive transducer (610), each is deconvolved based on its point spread function and an estimate of the noise spectra, and then the images are superimposed by finding the displacement which maximizes correlation to one chosen as the reference image, and then averaging the images.

10. The method according to claim 5, further including the step of determining the position of the receive transducer(s) (610) relative to the transmit transducer (620) by using an instrumented arm connecting the receive transducer(s) to the probe containing the transmit transducer.

11. The method according to claim 5, further including the step of determining the position of the receive transducer(s) (610) relative to the transmit transducer (620) by triangulation using IR, laser, or radio transponders attached to the receive and transmit ultrasound transducers.

12. The method according to claim 5, wherein the image processing step includes Wiener filtering based on estimation of the point spread function of the system measured on a phantom as a function of distance and angle subtended by transmit transducer - point in object - receive transducer, wherein the Wiener filter can be either continuously varied across the image for optimal filtering, or optimized only for the region of the image requiring greatest detail.

13. The method according to claim 5, wherein said image processing step (step (e)) involves using least mean square (LMS) adaptation based on estimation of the point spread function of the system measured on a phantom as a function of distance and angle subtended by transmit transducer - point in object - receive transducer, and wherein the LMS filter can be either continuously varied across the image for optimal filtering, or optimized only for the region of the image requiring greatest detail.

14. The method according to claim 5, wherein said image processing step (step (e)) involves using an image sharpening technique selected from either ID or 2D unsharp masking.

15. The method of claim 5, wherein the recording step (b) comprises recording the echo received by two or more relatively non-directional receive transducers (610); the reconstructing step (d) comprises reconstructing scan lines as lines on the reconstructed image based on knowledge of the location of the receive transducers relative to that of the transmit transducer (620) and the average speed of ultrasound in the object; and the image processing step comprises combining overlapping scan lines.

16. The method of claim 5, wherein the transmit transducer array and said at least one receive transducer are each pivotally connected to an articulating arm so as to be mechanically linked and to move in the same plane.

Patentansprüche

1. Ultraschallbilderzeugungssystem, Folgendes umfassend:

eine Sendewandleranordnung (620), die konfiguriert ist, ein ausgewähltes Objekt und/oder einen Bereich von lebendem Gewebe aus einem ersten akustischen Fenster mit mehreren gerichteten Ultraschallimpulsen zu beschallen;
mindestens einen Empfangswandler (610), der zum Erfassen und Aufzeichnen von Ultraschallechorücksendungen aus einem zweiten akustischen Fenster konfiguriert ist, wobei der Wandler in einer Abtastebene vergleichsweise ungerichtet ist; und
einen Computer in elektronischer Kommunikation mit der Sendewandleranordnung und dem/den Empfangswandler(n), wobei der Computer Hardware und Software aufweist, die konfiguriert sind zum Digitalisieren und Speichern jedes Ultraschallechos in einer Zeile eines Computerspeichers, zum Bestimmen des Standorts des Sendewandlers in Bezug auf den/die Empfangswandler, zum Rekonstruieren von Abtastzeilen als Zeilen auf einem rekonstruierten Bild basierend auf der Bestimmung des Standorts des/der Empfangswandler(s) relativ zu dem des Sendewandlers und der Durchschnittsgeschwindigkeit des Ultraschalls in dem Gewebeobjekt und zum Verarbeiten des Bildes zum Ausfüllen von Lücken zwischen Abtastzeilen und optional zum Korrigieren von Verzerrungen, die durch die Punktverteilungsfunktion des Systems verursacht werden,
dadurch gekennzeichnet, dass:
die Sendewandleranordnung (620) und der mindestens eine Empfangswandler (610) über eine mechanische Gelenkverbindung miteinander verbunden sind, die dazu konfiguriert ist, sicherzustellen, dass die Abtastebene der Sendewandleranordnung und des mindestens einen Empfangswandlers das jeweils andere Element einschließt, während der Abstand zwischen der Sendewandleranordnung und dem mindestens einen Empfangswandler nicht eingeschränkt ist.

2. System nach Anspruch 1, ferner mit 3D-Volumen-Füllmitteln zum Füllen eines 3D-Volumens, wenn die Vorrichtung zum Abbilden von 2D-Schnitten durch auf einer Plattform angeordnetes lebendes Gewebe verwendet wird, um Schnitte in parallelen Ebenen nacheinander abzutasten und die relativen Positionen jedes Schnitts aufzuzeichnen, wobei die 3D-Volumenfülleinrichtung einen Computer in elektronischer Kommunikation mit den Wandleranordnungen umfasst, wobei der Computer eine Anwendungssoftware umfasst, die konfiguriert ist, jedes zweidimensionale Bild zu rekonstruieren, die 2D-Bilder unter Verwendung von Interpolationsalgorithmen zu einem 3D-Volumen im Computerspeicher zusammenzusetzen, das Volumen auszurichten und eine 3D-Segmentierung des Volumens durchzuführen und das 3D-Volumen als Projektion auf eine Ebene oder als 2D-Schnitt mit beliebiger Ausrichtung anzuzeigen.

3. System nach Anspruch 1, ferner mit 3D-Volumen-Füllmitteln zum Füllen eines 3D-Volumens, wenn die Vorrichtung zum Abbilden von 2D-Schnitten durch auf einer Plattform angeordnetes lebendes Gewebe verwendet wird, um Schnitte in nebeneinanderliegenden Winkeln abzutasten und die relativen Positionen und Winkel jedes Schnitts aufzuzeichnen, wobei die 3D-Volumenfülleinrichtung einen Computer in elektronischer Kommunikation mit den Wandleranordnungen umfasst, wobei der Computer eine Anwendungssoftware umfasst, die konfiguriert ist, jedes zweidimensionale Bild zu rekonstruieren, die 2D-Bilder unter Verwendung von Interpolationsalgorithmen zu einem 3D-Volumen im Computerspeicher zusammenzusetzen, das Volumen auszurichten und eine 3D-Segmentierung des Volumens durchzuführen und das 3D-Volumen als Projektion auf eine Ebene oder als 2D-Schnitt mit beliebiger Ausrichtung anzuzeigen.

4. System nach Anspruch 1, wobei die Sendewandleranordnung und der mindestens eine Empfangswandler jeweils schwenkbar mit einem Gelenkarm verbunden sind, um mechanisch verbunden zu sein und sich in der gleichen Ebene zu bewegen.

5. Verfahren zum Rekonstruieren eines Bildes, die folgenden Schritte umfassend:

(a) Verbinden eines Sendewandlers (620) und eines Empfangswandlers (610) über eine mechanische Gelenkverbindung miteinander, die sicherstellt, dass die Abtastebene des Sendewandlers und des Empfangswandlers das jeweils andere Element einschließt, während der Abstand zwischen dem Sendewandler und dem Empfangswandler nicht eingeschränkt ist;
(b) Beschallen eines Objekts mit jeweils einer Abtastzeile durch einen von einem Sendewandler (620) erzeugten gerichteten Ultraschallimpuls, der in einem ersten akustischen Fenster angeordnet ist;
(c) Aufzeichnen von durch einen Empfangswandler (610) empfangenen Echos, der relativ ungerichtet und in einem zweiten akustischen Fenster angeordnet ist;
(d) Speichern der digitalisierten Echos jeweils in einer Zeile eines Computerspeichers;
(e) Rekonstruieren von Abtastzeilen als Zeilen auf dem rekonstruierten Bild basierend auf der Kenntnis des Standorts des Empfangswandlers relativ zu dem des Sendewandlers und der durchschnittlichen Geschwindigkeit von Ultraschall in dem Objekt; und
(f) Bildverarbeiten zum Ausfüllen von Lücken zwischen den Abtastzeilen und optional zum Korrigieren von Verzerrungen, die durch die Punktverteilungsfunktion des Systems verursacht werden,
wobei sich die rekonstruierten Abtastzeilen in beliebigen Winkeln oder Positionen befinden und/oder die Anzahl dieser Abtastzeilen die Anzahl der tatsächlich übertragenen Impulse überschreitet.

6. Verfahren nach Anspruch 5, wobei der Sendewandler (620) ein Sektorabtaster vom Phased-Array-Typ ist, der Impulse in nebeneinanderliegenden Winkeln von einem gemeinsamen Ursprung aussendet.

7. Verfahren nach Anspruch 5, wobei der Sendewandler (620) ein lineares Phased-Array ist, das Impulse auf parallelen Pfaden aussendet.

8. Verfahren nach Anspruch 5, wobei für jeden Empfangswandler (610) ein separates Bild rekonstruiert wird und diese Bilder überlagert werden, indem die Verschiebung ermittelt wird, welche die Korrelation zu einem als das Referenzbild ausgewählten maximiert, gefolgt vom Mitteln der Bilder.

9. Verfahren nach Anspruch 5, wobei für jeden Empfangswandler (610) ein separates Bild rekonstruiert wird, jedes davon dann auf der Grundlage seiner Punktverteilungsfunktion und einer Abschätzung der Rauschspektren entfaltet wird, und dann die Bilder überlagert werden, indem die Verschiebung ermittelt wird, welche die Korrelation zu einem als das Referenzbild ausgewählten maximiert, und dann die Bilder gemittelt werden.

10. Verfahren nach Anspruch 5, ferner enthaltend den Schritt zum Bestimmen des Standorts des/der Empfangswandler(s) (610) relativ zu dem Sendewandler (620) unter Einsatz eines instrumentierten Arms, der den/die Empfangswandler mit der Sonde verbindet, die den Sendewandler enthält.

11. Verfahren nach Anspruch 5, ferner enthaltend den Schritt zum Bestimmen des Standorts des/der Empfangswandler(s) (610) relativ zu dem Sendewandler (620) durch Triangulation unter Verwendung von IR, Laser oder Radiotranspondern, die an den Empfangs- und Sendeultraschallwandlern befestigt sind.

12. Verfahren nach Anspruch 5, wobei der Bildverarbeitungsschritt Wiener-Filterung auf der Grundlage einer Schätzung der Punktverteilungsfunktion des Systems einschließt, gemessen an einem Phantom als eine Funktion von Distanz und Winkel zwischen Sendewandler, Punkt im Objekt und Empfangswandler, wobei die Wiener-Filterung für eine optimale Filterung entweder kontinuierlich über das Bild variiert werden oder nur für den Bereich des Bildes, der die genauesten Details erfordert, optimiert werden kann.

13. Verfahren nach Anspruch 5, wobei der Bildverarbeitungsschritt (Schritt (e)) mindestens eine LMS(Least Mean Square)-Anpassung auf der Grundlage einer Schätzung der Punktverteilungsfunktion des Systems einschließt, gemessen an einem Phantom als eine Funktion von Distanz und Winkel, begrenzt durch Sendewandler, Punkt im Objekt und Empfangswandler, und wobei die LMS-Filterung für eine optimale Filterung entweder kontinuierlich über das Bild variiert werden oder nur für den Bereich des Bildes, der die genauesten Details erfordert, optimiert werden kann.

14. Verfahren nach Anspruch 5, wobei der Bildverarbeitungsschritt (Schritt (e)) das Einsetzen einer Bildschärfungstechnik beinhaltet, ausgewählt aus ID- oder 2D-Unschärfemaskierung.

15. Verfahren nach Anspruch 5, wobei der Aufzeichnungsschritt (b) das Aufzeichnen des von zwei oder mehr vergleichsweise ungerichteten Empfangswandlern (610) empfangenen Echos umfasst; der Rekonstruktionsschritt (d) das Rekonstruieren von Abtastzeilen als Zeilen auf dem rekonstruierten Bild basierend auf der Kenntnis des Standorts der Empfangswandler relativ zu dem des Sendewandlers (620) und der durchschnittlichen Geschwindigkeit von Ultraschall in dem Objekt umfasst; und der Bildverarbeitungsschritt das Kombinieren überlappender Abtastzeilen umfasst.

16. Verfahren nach Anspruch 5, wobei die Sendewandleranordnung und der mindestens eine Empfangswandler jeweils schwenkbar mit einem Gelenkarm verbunden sind, um mechanisch verbunden zu sein und sich in der gleichen Ebene zu bewegen.

Revendications

1. Système d'imagerie par ultrasons, comprenant :

un réseau de transducteurs d'émission (620) configuré pour insonifier un objet et/ou une zone de tissu vivant choisi(e) à partir d'une première fenêtre acoustique, au moyen d'une pluralité d'impulsions ultrasonores directionnelles ;
au moins un transducteur de réception (610) configuré pour détecter et enregistrer les retours d'écho ultrasonore provenant d'une seconde fenêtre acoustique, ledit transducteur étant relativement non directionnel dans un plan de balayage ; et
un ordinateur en communication électronique avec ledit réseau de transducteurs d'émission et ledit ou lesdits transducteurs de réception, ledit ordinateur comportant un matériel et un logiciel configurés pour numériser et stocker chaque écho ultrasonore dans une ligne de mémoire d'ordinateur, déterminer l'emplacement dudit transducteur d'émission par rapport audit ou auxdits transducteurs de réception, reconstruire des lignes de balayage sous forme de lignes sur une image reconstruite sur la base de la détermination de l'emplacement du ou des transducteurs de réception par rapport à celui du transducteur d'émission et de la vitesse moyenne des ultrasons dans l'objet de tissu, et pour traiter des images pour combler les espaces entre les lignes de balayage et éventuellement corriger les distorsions causées par la fonction d'étalement de point du système,
caractérisé en ce que :
le réseau de transducteurs d'émission (620) et l'au moins un transducteur de réception (610) sont reliés ensemble au moyen d'une liaison mécanique articulée configurée pour garantir que le plan de balayage du réseau de transducteurs d'émission et l'au moins un transducteur de réception comprend l'autre, tandis que la distance entre le réseau de transducteurs d'émission et l'au moins un transducteur de réception n'est pas limitée.

2. Système selon la revendication 1, comprenant en outre des moyens de remplissage de volume 3D permettant de remplir un volume 3D lorsque ledit appareil est utilisé pour l'imagerie de coupes 2D à travers des tissus vivants montés sur une plate-forme de manière à balayer successivement des coupes dans des plans parallèles et à enregistrer les positions relatives de chaque coupe, lesdits moyens de remplissage de volume 3D comprenant un ordinateur en communication électronique avec lesdits réseaux de transducteurs, ledit ordinateur comprenant un logiciel d'application qui est configuré pour reconstruire chaque image bidimensionnelle, assembler les images 2D en un volume 3D dans la mémoire de l'ordinateur en utilisant des algorithmes d'interpolation, aligner et effectuer une segmentation 3D sur le volume, et pour afficher le volume 3D comme une projection sur un plan ou une coupe 2D d'orientation arbitraire.

3. Système selon la revendication 1, comprenant en outre des moyens de remplissage de volume 3D permettant de remplir un volume 3D lorsque ledit appareil est utilisé pour l'imagerie de coupes 2D à travers des tissus vivants montés sur une plate-forme de manière à balayer des coupes selon des angles successifs et à enregistrer les positions et angles relatifs de chaque coupe, lesdits moyens de remplissage de volume 3D comprenant un ordinateur en communication électronique avec lesdits réseaux de transducteurs, ledit ordinateur comprenant un logiciel d'application qui est configuré pour reconstruire chaque image bidimensionnelle, assembler les images 2D en un volume 3D dans la mémoire de l'ordinateur en utilisant des algorithmes d'interpolation, aligner et effectuer une segmentation 3D sur le volume, et pour afficher le volume 3D comme une projection sur un plan ou une coupe 2D d'orientation arbitraire.

4. Système selon la revendication 1, dans lequel le réseau de transducteurs d'émission et ledit au moins un transducteur de réception sont chacun reliés de manière pivotante à un bras articulé de manière à être liés mécaniquement et à se déplacer dans le même plan.

5. Procédé de reconstruction d'une image, comprenant les étapes consistant à :

a) relier un transducteur d'émission (620) et un transducteur de réception (610) au moyen d'une liaison mécanique articulée qui garantit que le plan de balayage du transducteur d'émission et du transducteur de réception comprend l'autre, tandis que la distance entre le transducteur d'émission et le transducteur de réception n'est pas limitée ;
b) insonifier un objet une ligne de balayage à la fois au moyen d'une impulsion ultrasonore directionnelle générée à partir d'un transducteur d'émission (620) positionné dans une première fenêtre acoustique ;
c) enregistrer des échos reçus par un transducteur de réception (610) qui est relativement non directionnel et positionné dans une seconde fenêtre acoustique ;
d) stocker les échos numérisés chacun dans une ligne de mémoire d'ordinateur ;
e) reconstruire des lignes de balayage sous forme de lignes sur l'image reconstruite sur la base de la connaissance de l'emplacement du transducteur de réception par rapport à celui du transducteur d'émission et de la vitesse moyenne des ultrasons dans l'objet ; et à
f) traiter des images afin de combler les espaces entre les lignes de balayage et éventuellement corriger les distorsions causées par la fonction d'étalement de point du système,
les lignes de balayage reconstruites se trouvant à des angles ou à des emplacements arbitraires et/ou le nombre de telles lignes de balayage dépassant le nombre d'impulsions réelles transmises.

6. Procédé selon la revendication 5, dans lequel le transducteur d'émission (620) est un scanner sectoriel de type réseau phasé qui envoie des impulsions selon des angles successifs à partir d'une origine commune.

7. Procédé selon la revendication 5, dans lequel le transducteur d'émission (620) est un réseau en phase linéaire qui envoie des impulsions sur des trajets parallèles.

8. Procédé selon la revendication 5, dans lequel une image séparée est reconstruite pour chaque transducteur de réception (610), et ces images sont superposées en trouvant le déplacement qui maximise la corrélation avec celle choisie comme image de référence puis en faisant la moyenne des images.

9. Procédé selon la revendication 5, dans lequel une image séparée est reconstruite pour chaque transducteur de réception (610), chacune étant déconvoluée sur la base de sa fonction d'étalement de point et d'une estimation des spectres de bruit, puis les images sont superposées en trouvant le déplacement qui maximise la corrélation avec celle choisie comme image de référence, puis en faisant la moyenne des images.

10. Procédé selon la revendication 5, comprenant en outre l'étape consistant à déterminer la position du ou des transducteurs de réception (610) par rapport au transducteur d'émission (620) en utilisant un bras instrumenté reliant le ou les transducteurs de réception à la sonde contenant le transducteur d'émission.

11. Procédé selon la revendication 5, comprenant en outre l'étape consistant à déterminer la position du ou des transducteurs de réception (610) par rapport au transducteur d'émission (620) par triangulation en utilisant des transpondeurs IR, laser ou radio fixés aux transducteurs ultrasonores de réception et d'émission.

12. Procédé selon la revendication 5, dans lequel l'étape de traitement d'image consiste à effectuer un filtrage de Wiener sur la base d'une estimation de la fonction d'étalement de point du système mesurée sur un fantôme en fonction de la distance et de l'angle sous-tendus par le transducteur d'émission (point de l'objet), le transducteur de réception, le filtre de Wiener pouvant être modifié de façon continue à travers l'image pour un filtrage optimal, ou optimisé uniquement pour la région de l'image nécessitant le plus de détails.

13. Procédé selon la revendication 5, dans lequel ladite étape de traitement d'image (étape e)) consiste à utiliser l'adaptation des moindres carrés moyens (LMS) basée sur l'estimation de la fonction d'étalement de point du système mesurée sur un fantôme en fonction de la distance et de l'angle sous-tendus par le transducteur d'émission (point de l'objet), le transducteur de réception, et le filtre LMS pouvant être modifié de façon continue à travers l'image pour un filtrage optimal, ou optimisé uniquement pour la région de l'image nécessitant le plus de détails.

14. Procédé selon la revendication 5, dans lequel ladite étape de traitement d'image (étape e)) consiste à utiliser une technique d'affinage d'image choisie parmi un masquage flou ID ou 2D.

15. Procédé selon la revendication 5, dans lequel l'étape d'enregistrement b) consiste à enregistrer l'écho reçu par deux ou plusieurs transducteurs de réception relativement non directionnels (610) ; l'étape de reconstruction d) consiste à reconstruire des lignes de balayage sous forme de lignes sur l'image reconstruite sur la base de la connaissance de l'emplacement des transducteurs de réception par rapport à celui du transducteur d'émission (620) et de la vitesse moyenne des ultrasons dans l'objet ; et l'étape de traitement d'image consiste à combiner des lignes de balayage se chevauchant.

16. Procédé selon la revendication 5, dans lequel le réseau de transducteurs d'émission et ledit au moins un transducteur de réception sont chacun reliés de manière pivotante à un bras articulé de manière à être liés mécaniquement et à se déplacer dans le même plan.

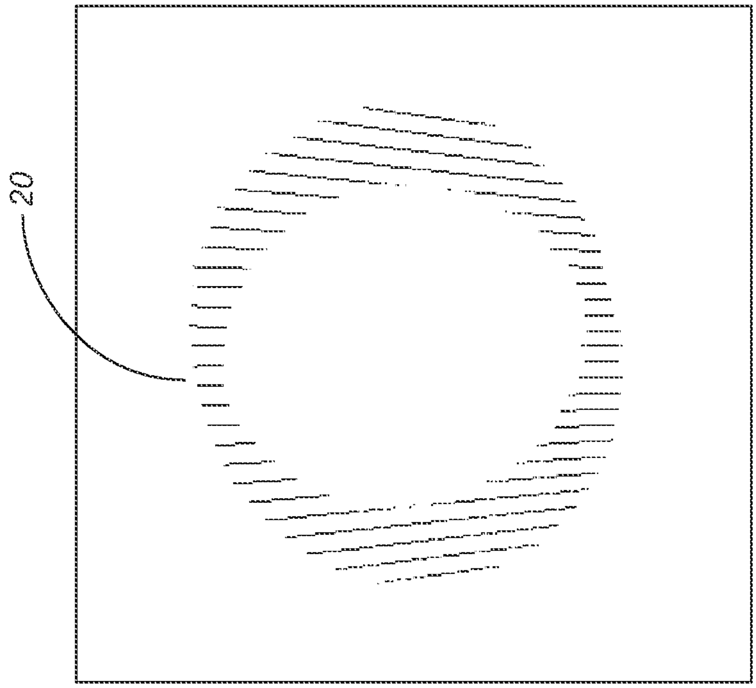


FIG. 5

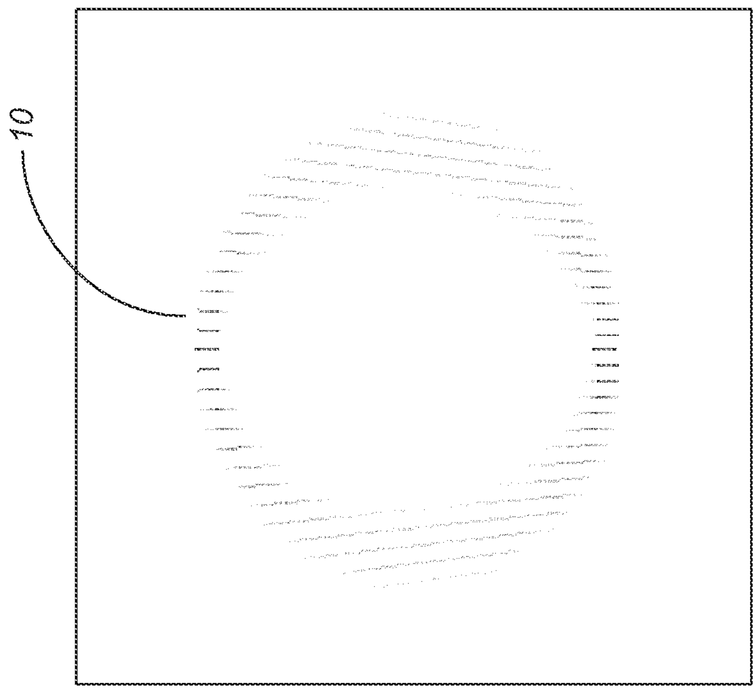


FIG. 1

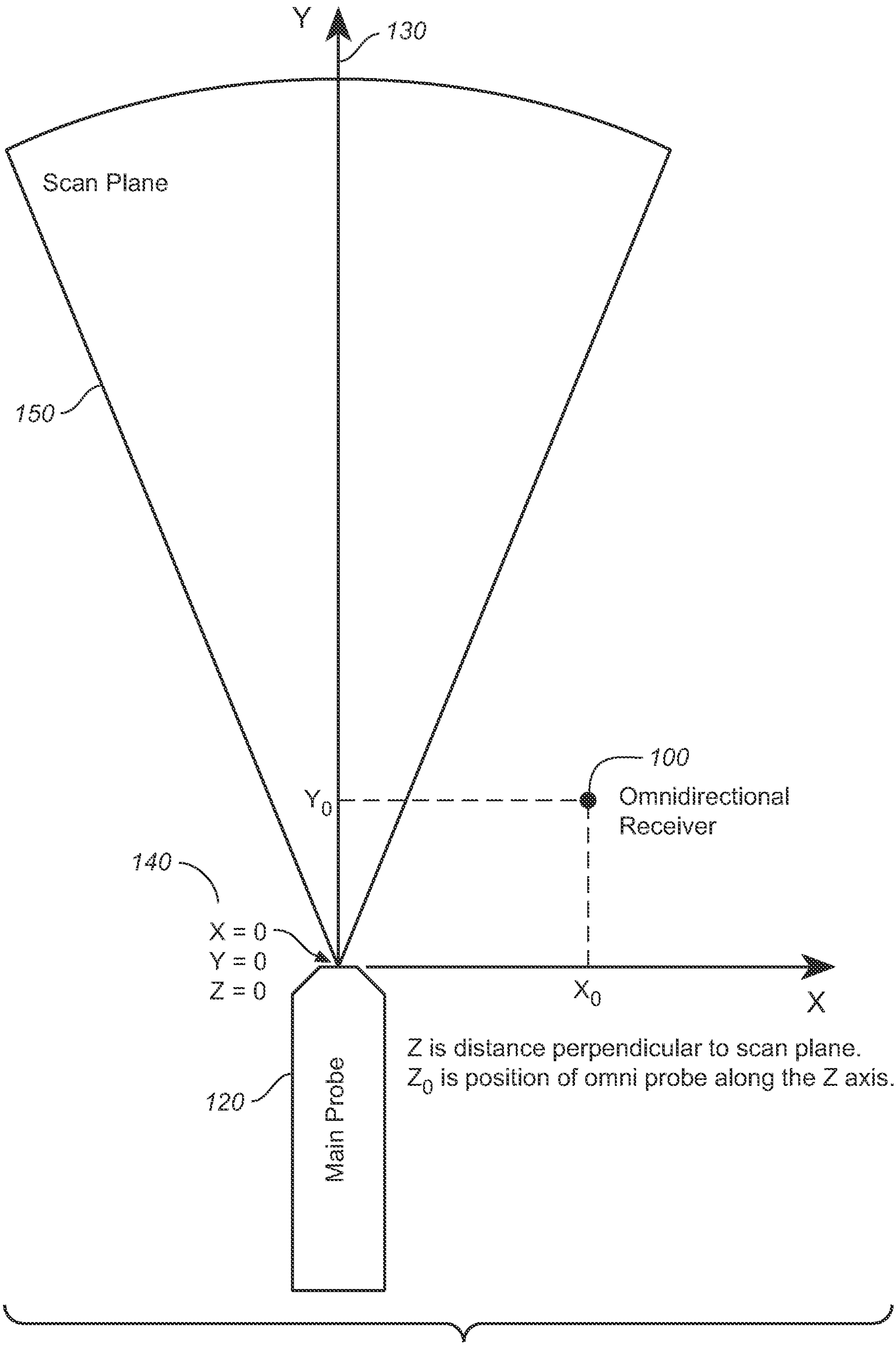

Y
130
Scan Plane
150
100
Y_0
Omnidirectional
Receiver
140
X = 0
Y = 0
Z = 0
X_0
X
Z is distance perpendicular to scan plane.
Z_0 is position of omni probe along the Z axis.
120
Main Probe

FIG. 2

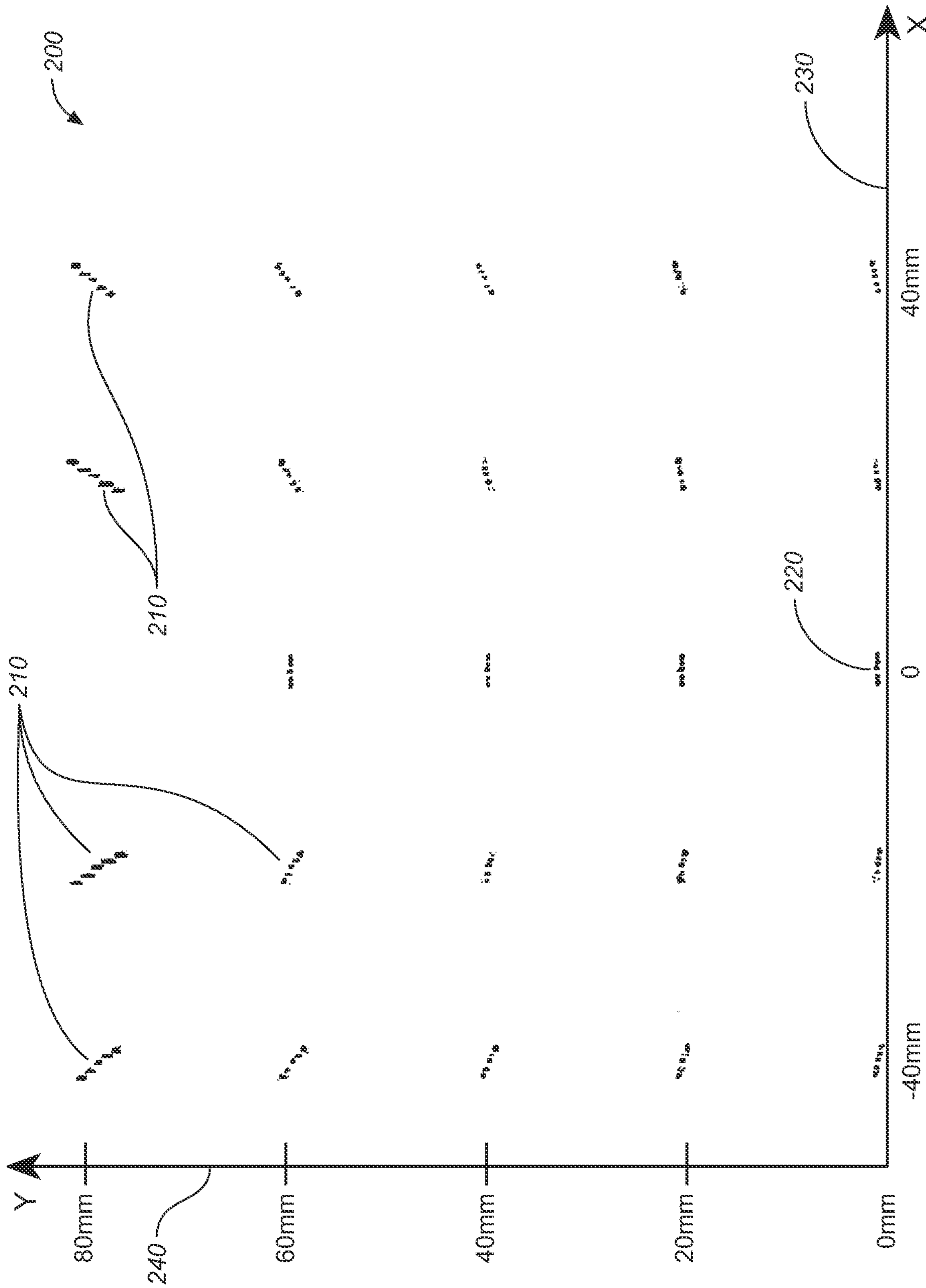
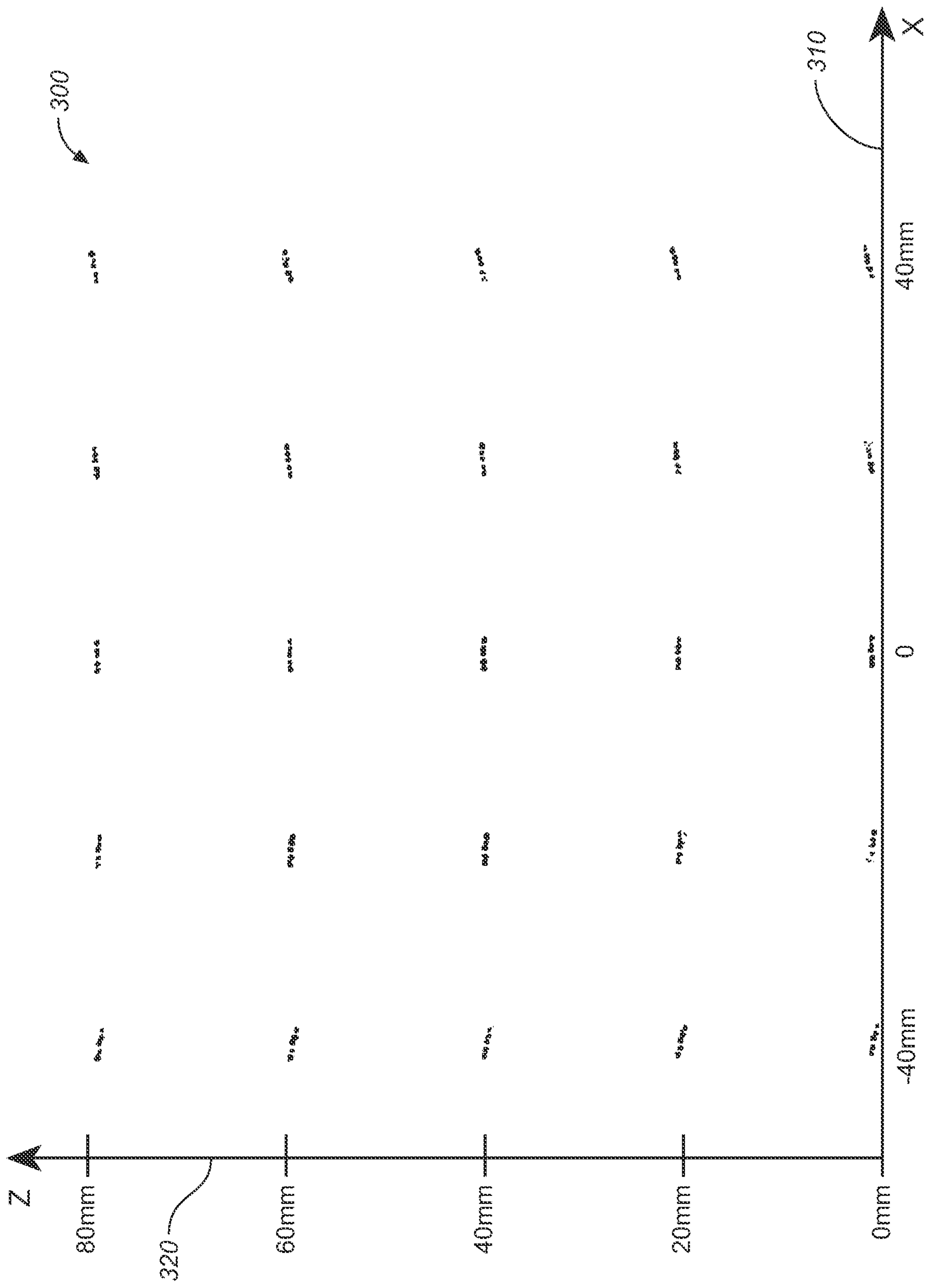


FIG. 3

Orientation of the point spread function of an imaging system using a single omnidirectional probe as a function of the postion of the omnidirectional probe



Orientation of the point spread function of an imaging system using a single omnidirectional probe as a function of the postion of the omnidirectional probe when it is not in the scan plane y=0

FIG. 4

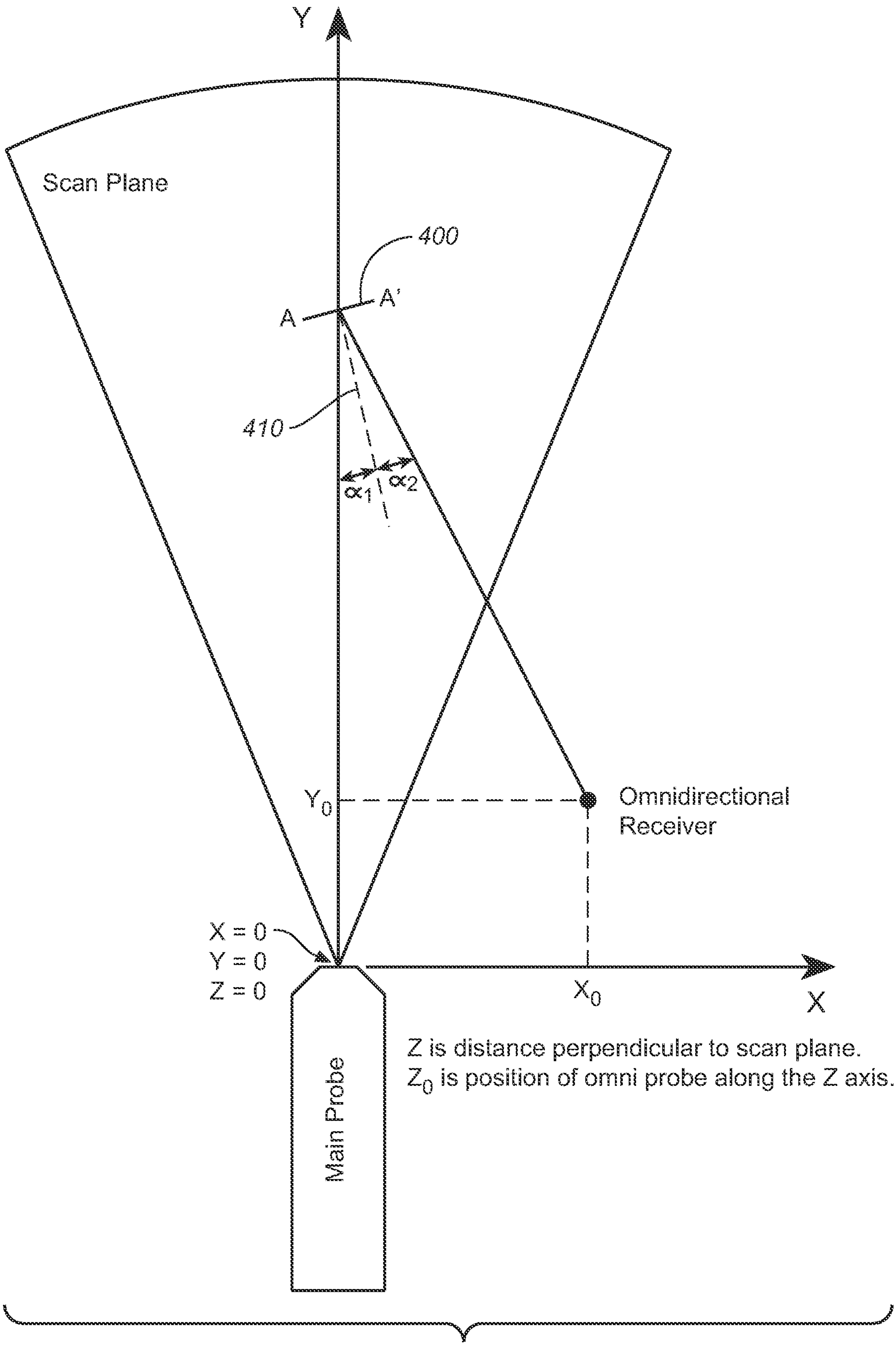

Y
Scan Plane
400
A
A'
410
α1
α2
Y0
Omnidirectional Receiver
X = 0
Y = 0
Z = 0
X0
X
Main Probe
Z is distance perpendicular to scan plane.
Z0 is position of omni probe along the Z axis.

FIG. 6

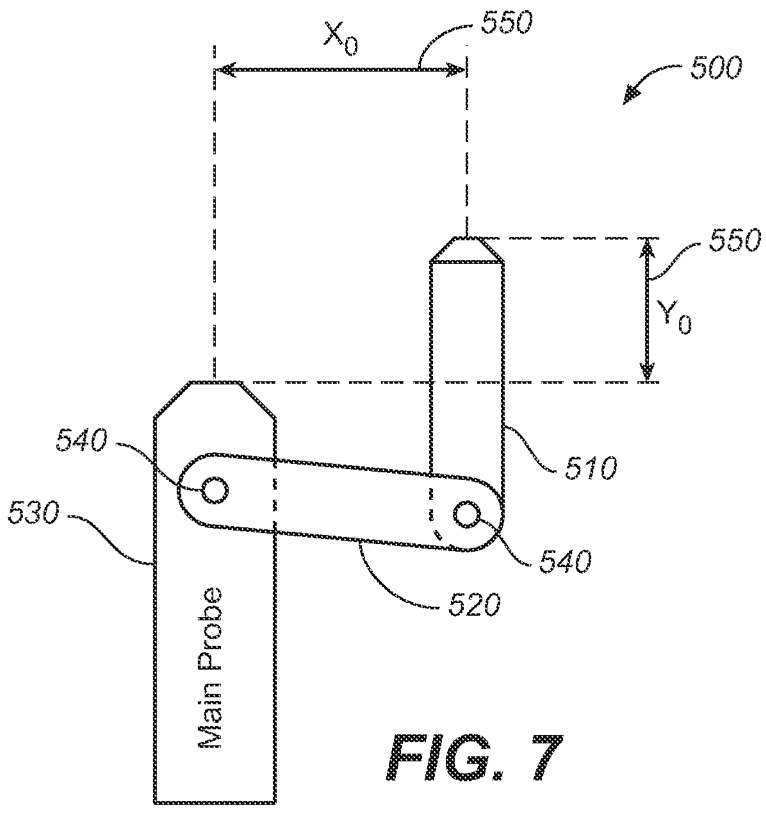


FIG. 7

FIG. 8

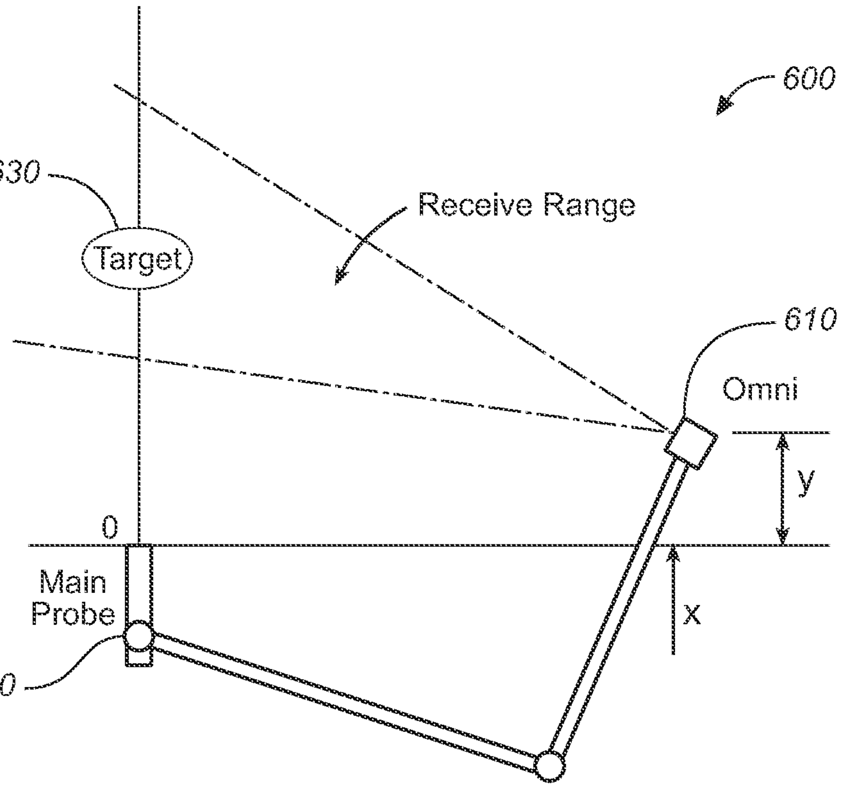
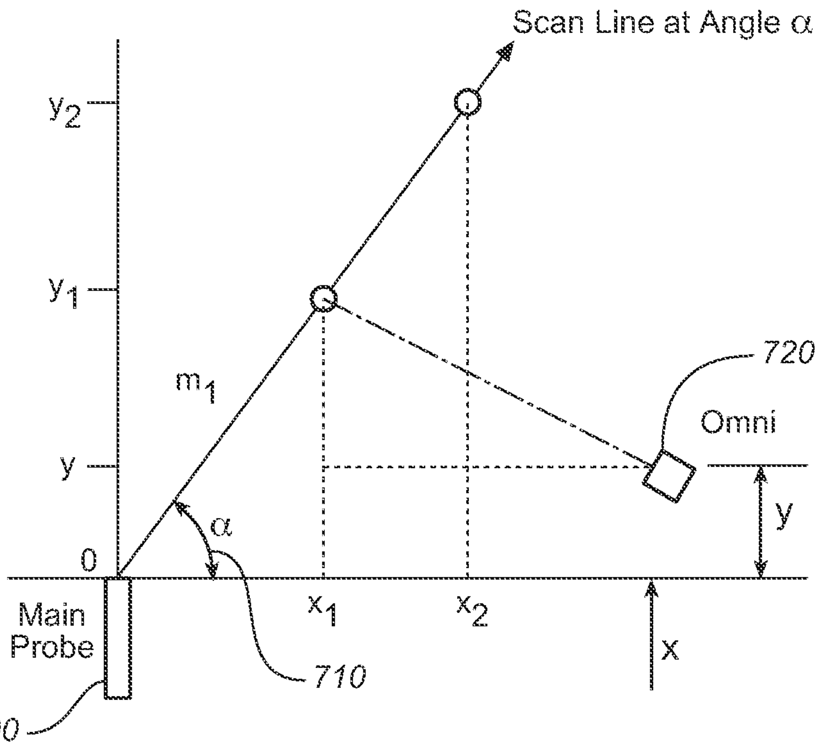

600
630
Target
Receive Range
610
Omni
y
0
Main
Probe
x
620

FIG. 9


Scan Line at Angle α
y2
y1
m1
720
Omni
y
α
0
y
x1
x2
Main
Probe
x
710
700

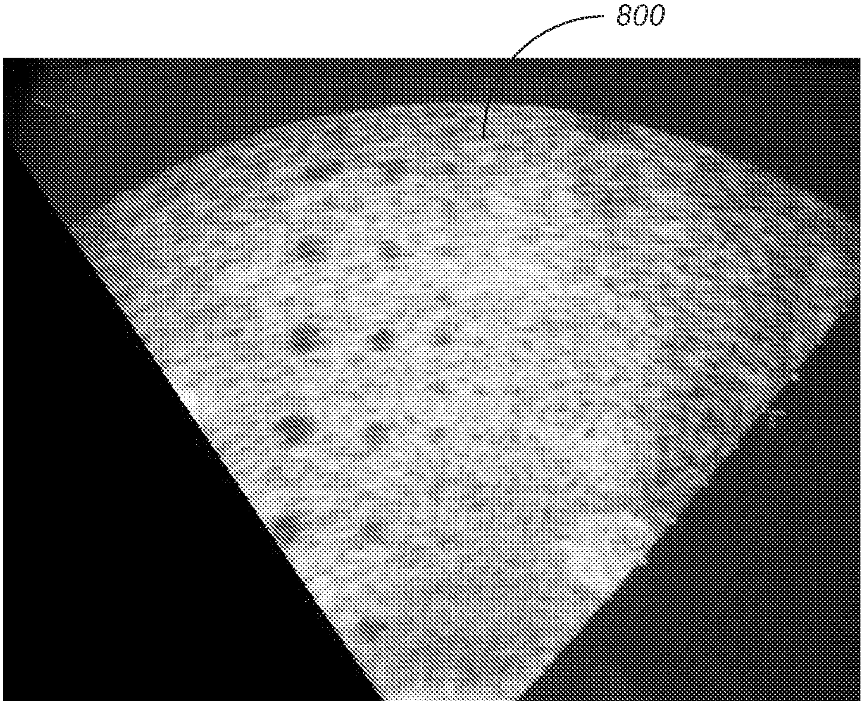


FIG. 10A

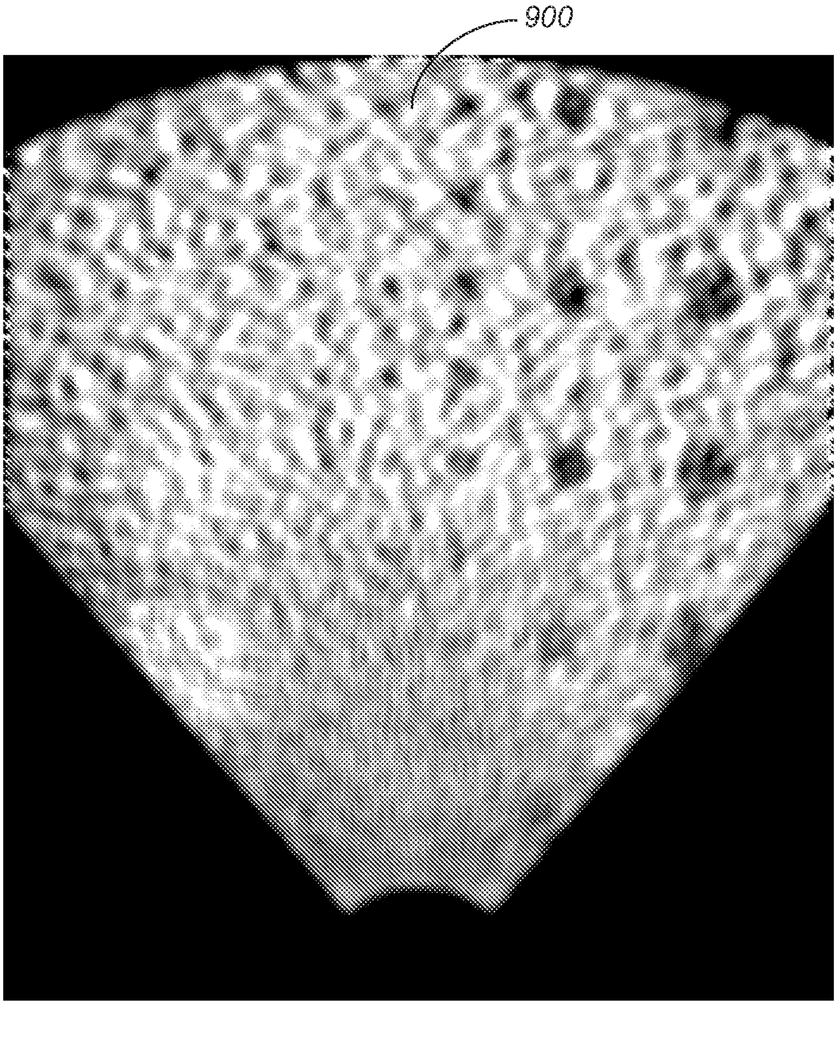


FIG. 10B

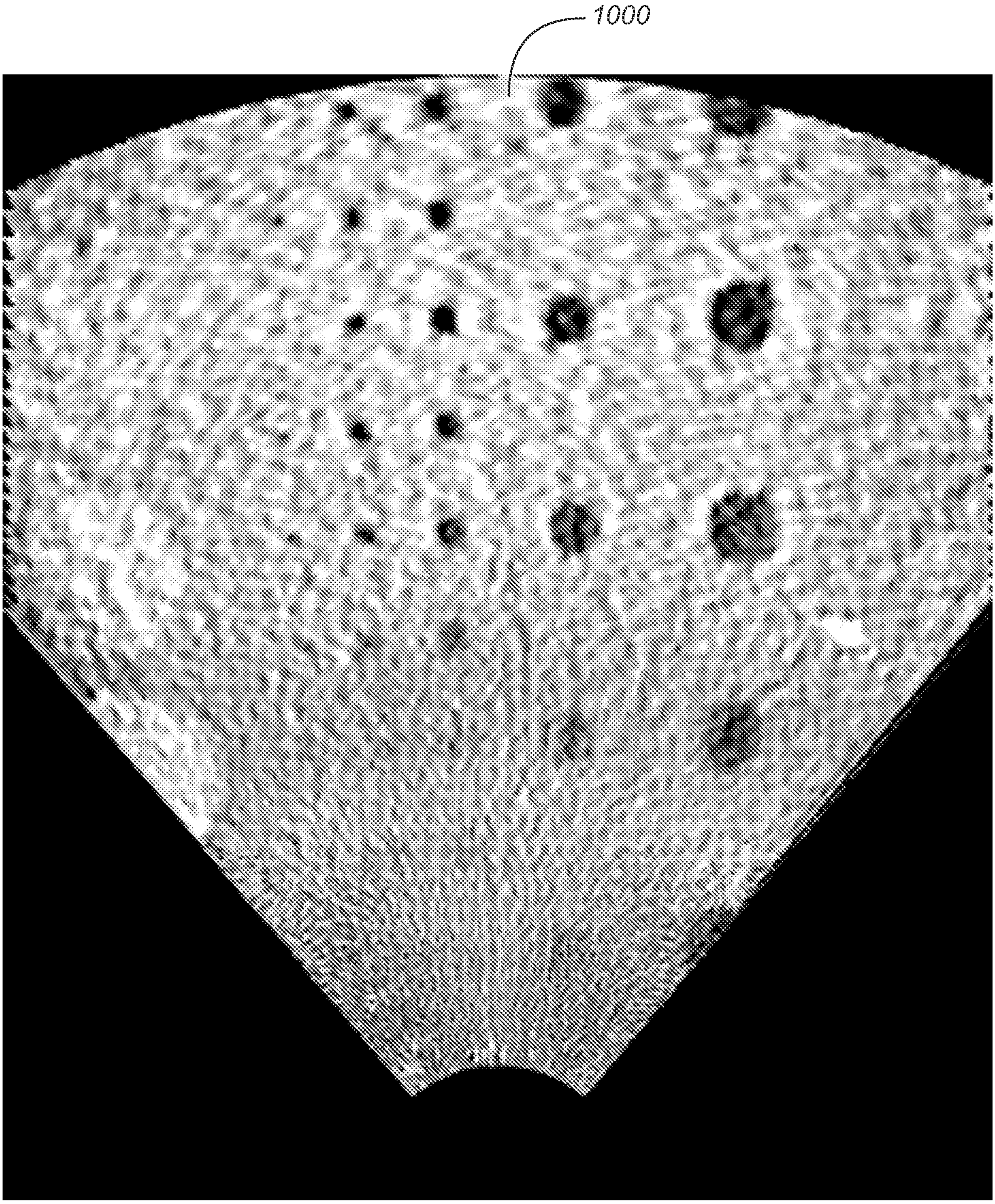


FIG. 11

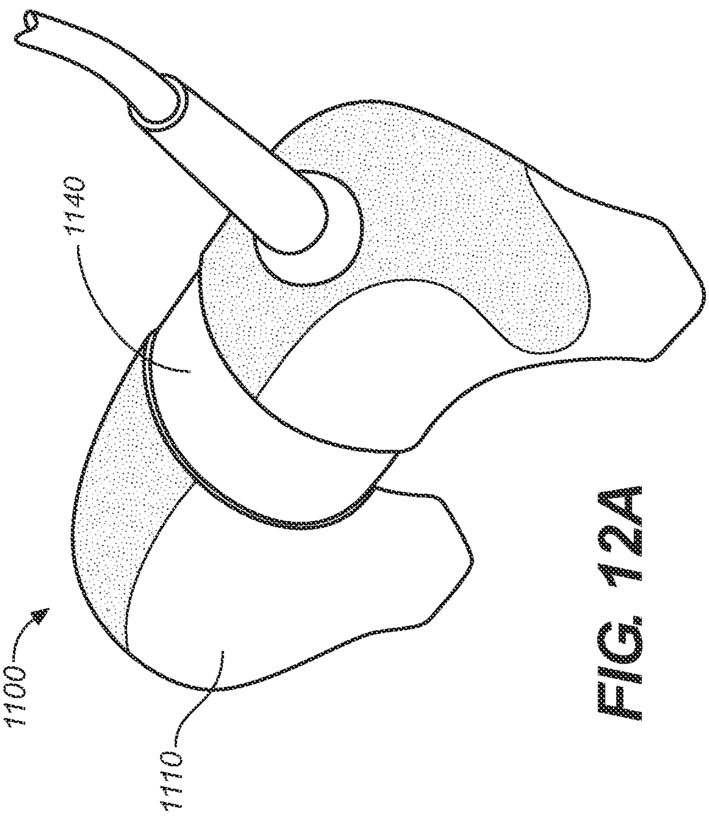


FIG. 12A

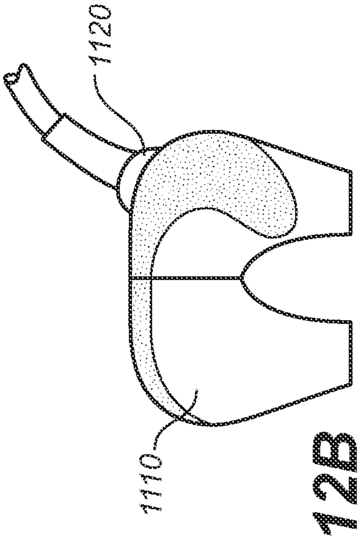


FIG. 12B

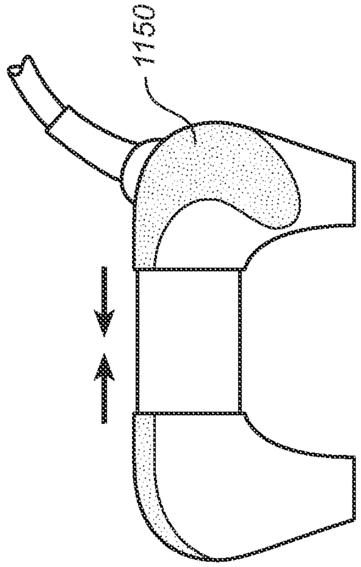


FIG. 12C

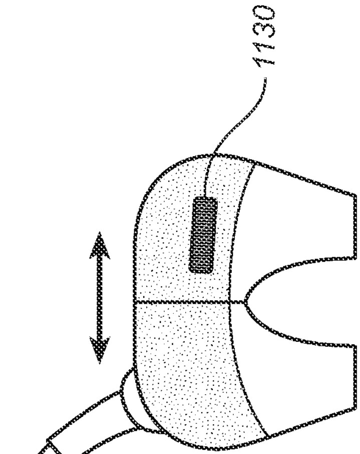


FIG. 12D

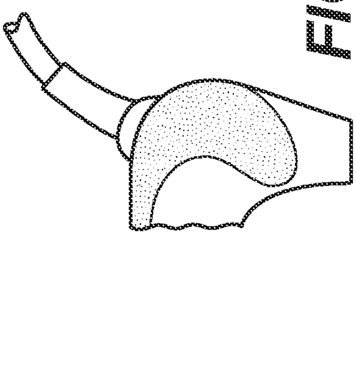


FIG. 12E

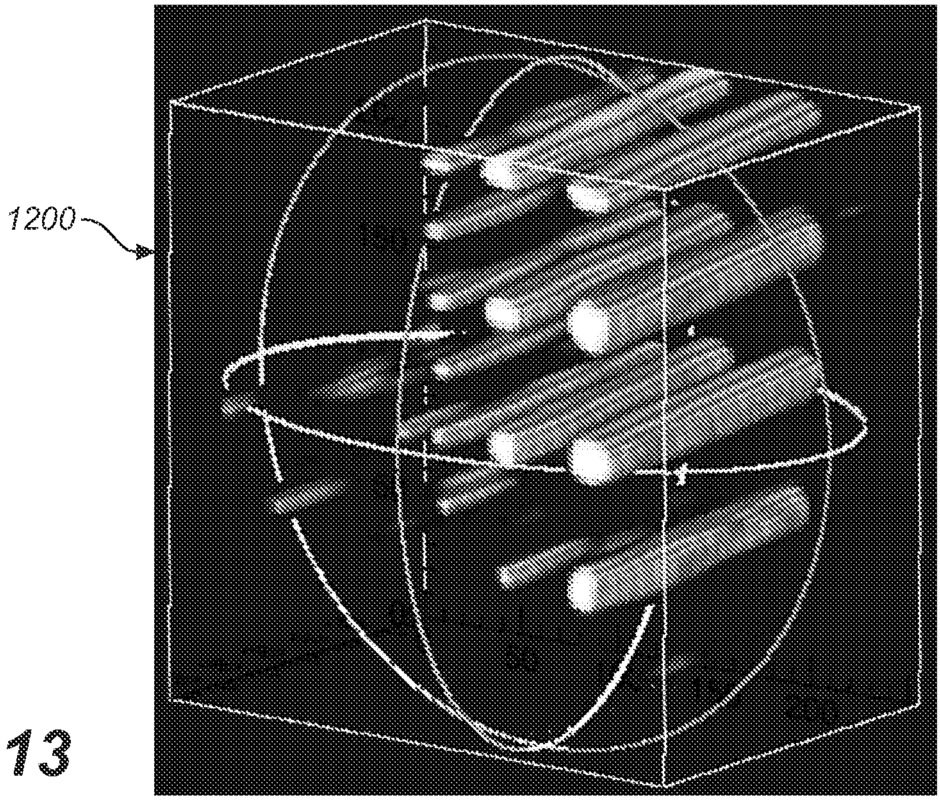


FIG. 13

REFERENCES CITED IN THE DESCRIPTION

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- US 20060122506 A **[0008]**
- US 11532013 B **[0052]**
- US 532013 **[0074] [0081]**

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- **HARVEY FEIGENBAUM.** Echocardiography. Lippincott Williams & Wilkins, 1993 **[0003]**
- **P. N. T. WELLS.** Biomedical Ultrasonics. Academic Press, 1977 **[0004]**

专利名称(译)	具有多个开口的超声图像的产生方法和装置		
公开(公告)号	[EP2088932B1](EP2088932B1)	公开(公告)日	2020-04-08
申请号	EP2007840422	申请日	2007-07-16
[标]申请(专利权)人(译)	茂伊成像股份有限公司		
申请(专利权)人(译)	MAUI IMAGING , INC.		
当前申请(专利权)人(译)	MAUI IMAGING , INC.		
[标]发明人	SPECHT DONALD F		
发明人	SPECHT, DONALD F.		
IPC分类号	A61B8/00		
CPC分类号	A61B8/42 A61B8/4209 A61B8/4218 A61B8/4281 A61B8/4455 A61B8/4483 A61B8/483 A61B8/543 G01S7/52046 G01S7/5205 G01S15/8977 A61B5/725 A61B8/08 A61B8/085 A61B8/145 A61B8/4245 A61B8/4494 A61B8/5207 A61B8/5253		
代理机构(译)	POTTER CLARKSON		
优先权	60/862951 2006-10-25 US 60/940261 2007-05-25 US		
其他公开文献	EP2088932A2 EP2088932A4		
外部链接	Espacenet		

摘要(译)

描述了超声扫描仪和全向接收换能器的组合，用于从由单个全向换能器接收的回波中产生二维图像。 可以从其他换能器接收到的回波中构建具有不同噪声成分的二维图像。 这些可以组合以产生具有更好信噪比和横向分辨率的图像。 还公开了一种基于信息内容来补偿通过介入组织的不同路径的不同延迟的方法。 即使使用从声波探头移开的单个全向接收器，镜面反射也会衰减。 所公开的技术在医学成像中具有广泛的应用，但是理想地适合于使用两个或更多个肋间隙的多孔心脏成像。 由于横向分辨率主要由端部元件定义的孔确定，因此不必用等距元素填充整个孔。 实际上，可以留出间隙以容纳横跨患者的肋骨，或者只是减少大孔径阵列的成本。 使用这些方法的多个切片可以组合以形成三维图像。

$$t_{is} = m_i + \mathrm{sqrt}\left((x_i - x)^2 + (y_i - y)^2\right)$$