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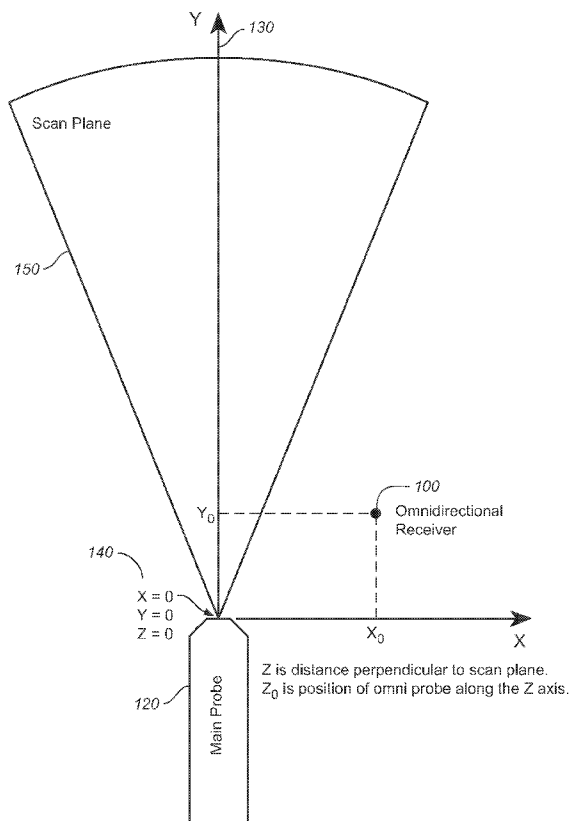
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(57) Abstract: A combination of an ultrasonic scanner and an omnidirectional receive transducer for producing a two-dimensional image from the echoes received by the single omnidirectional transducer is described. Two-dimensional images with different noise components can be constructed from the echoes received by additional transducers. These can be combined to produce images with better signal to noise ratios and lateral resolution. Also disclosed is a method based on information content to compensate for the different delays for different paths through intervening tissue is described. Specular reflections are attenuated by using even a single omnidirectional receiver displaced from the insonifying probe. The disclosed techniques have broad application in medical imaging but are ideally suited to multi-aperture cardiac imaging using two or more intercostal spaces. An alternative method employs multiple transducers to increase the lateral resolution and reduce noise and includes using two or more active phased array transducers in the same scan plane. In this method the phase delays are computed such that the transducers operate as one larger phased array with a gap (or gaps) in the middle. Lateral resolution is determined primarily by the aperture defined by the end elements, wherein a gap necessitated by spanning the patient's ribs can be accommodated. Multiple slices using these methods can be combined to form three-dimensional images.

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METHOD AND APPARATUS TO PRODUCE ULTRASONIC IMAGES USING MULTIPLE APERTURES

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BACKGROUND OF THE INVENTION

Technical Field

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[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

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[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.

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[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).

5 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
10 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
15 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
20 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
25 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.

Disclosure of Invention

[0008] 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. The separation can be a physical separation or simply a separation in concept wherein some of the elements of the array can be shared for the two functions.

[0009] 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.

[0010] There are numerous advantages to this approach, and these comprise the objects and advantages of the present invention. They include, among others:

[0011] The dominance of specular reflections so prominent in reconstructing images by returns to the main probe is greatly attenuated.

5 [0012] 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.

[0013] Compensations can be made for different delays in different paths through the tissue.

10 [0014] 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.

15 [0015] 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.

[0016] 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
20 catheter placement is that no steering is required of the probe.

[0017] 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.

25 [0018] 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.

5 [0019] 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
10 regarded as illustrative in nature, and not as restrictive.

Brief Description of the Drawings

[0020] Fig 1 is a diagrammatic view of a simulation showing a circular object imaged by a conventional sector scanner.

15 [0021] Fig 2 is a schematic diagram of the axes representing the relative positions of the insonifying and omni-directional probes.

[0022] 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.

20 [0023] 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.

[0024] 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.
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[0025] 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.

[0026] Fig 7 is a schematic diagram showing a possible fixture for positioning an omni-

directional probe relative to the main (insonifying) probe.

[0027] Fig 8 is a schematic diagram showing a non-instrumented linkage for two probes.

[0028] Fig 9 is a schematic diagram showing variables for computation of x and y positions from received echoes.

5 **[0029]** Fig. 10a is a phantom image taken with a standard Acuson 128 XP-10 with a 3.5 MHz transducer and harmonic processing.

[0030] 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.

10 **[0031]** 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.

[0032] 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.

20 **[0033]** Fig 12b is a side view in elevation thereof showing the probe in a collapsed configuration.

[0034] 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.

25 **[0035]** 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).

[0036] Fig 12e is a detailed perspective view showing surface features at the gripping portion of the probe.

[0037] Fig 13 is a 3D image highlighting the anechoic tubes of the ATS Model 539 phantom.

Best Mode for Carrying Out the Invention

5 [0038] 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
10 probe because it can be designed to be sensitive to a wide field of view.

[0039] 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
15 insonifying pulses.

[0040] A large amount of straightforward computation is required to plot the amplitude of echoes received from the omni. Referring 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
20 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 $t v = d + \sqrt{(x-x_0)^2 + (y-y_0)^2 + z_0^2}$).

25 [0041] 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 $x = 0$, $y = 70\text{mm}$, $z = 0$ 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.

5 **[0042]** 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.

[0043] Figure 5 shows a plot of the same circular object **20** as in Figure 1, plotted,
10 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.

[0044] 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
15 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.

[0045] 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**,
20 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.

25 **[0046]** 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 $t=(\text{distance to insonifying beam} + \text{distance to omni})/v$; and (e) plot amplitude recorded by the

omni for the k scan at x,y.

[0047] 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.

[0048] 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.

[0049] 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.

[0050] 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.

[0051] 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.

[0052] 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.

[0053] 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 in 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.

[0054] 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.

[0055] 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.

[0056] 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 .

[0057] 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.

[0058] 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.

[0059] 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.

[0060] 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.

[0061] 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.

[0062] 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.

[0063] 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.

[0064] 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.

[0065] 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.

5 **[0066]** 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 linkage 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.

10 **[0067]** 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, and which application is incorporated in its entirety by reference herein.

15 **[0068]** 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.

20 **[0069]** 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.

[0070] Note that any tilt of the main probe will change the reference axes so that x and particularly y will change too.

5 **[0071]** 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.

10 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.

15 **[0072]** Referring now to Figure 9, if the main probe **700** transmits at angle $90^\circ - \alpha$ **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:

[0073] tissue packet at (x_1, y_1) is received at time t_{1m} , and distance $m_1 = \text{sqrt}(x_1^2 + y_1^2)$

[0074] tissue packet at (x_2, y_2) is received at time t_{2m} , and distance $m_2 = \text{sqrt}(x_2^2 + y_2^2)$

20 **[0075]** t_{1m} corresponds to time of two trips of distance m_1

[0076] $t_{1m} s = 2 m_1$, where s = speed of ultrasound in same units as m = approx. 1.54×10^6 mm/sec

[0077] $t_{1s} s = m_1 + \text{sqrt}((x_1 - x)^2 + (y_1 - y)^2)$.

[0078] Similarly, $t_{2m} s = 2 m_2$

25 **[0079]** $t_{2s} s = m_2 + \text{sqrt}((x_2 - x)^2 + (y_2 - y)^2)$

[0080] $(t_{1s} s - 0.5 t_{1m} s)^2 = (x_1 - x)^2 + (y_1 - y)^2$ (1)

[0081] $(t_{2s} s - 0.5 t_{2m} s)^2 = (x_2 - x)^2 + (y_2 - y)^2$

[0082] 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 .

[0083] Similarly, if a z offset between the two probes is allowed, $x, y,$ and z can be calculated by solving three simultaneous equations.

5 [0084] 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)
 10 points using the equation:

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

[0085] 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
 15 spurious ambiguous minima.

[0086] When the z component of the relative position is not constrained to be zero, the comparable error squared equation is:

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

20 [0087] The minimum E^2 will indicate the minimum squared error estimate of (x, y, z) .

[0088] 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.

[0089] 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.

[0090] 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 $G(u,v)$:

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

[0091] 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:

[0092] (a) $H(u,v)$ – Fourier transform of the point spread function (psf);

[0093] (b) $P_s(u,v)$ – Power spectrum of the signal process, obtained by taking the Fourier transform of the signal autocorrelation;

[0094] (c) $P_n(u,v)$ – Power spectrum of the noise process, obtained by taking the Fourier transform of the noise autocorrelation;

[0095] The Wiener filter is:

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

[0096] 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.

[0097] $P_s(u,v) + P_n(u,v) = |I(u, v)|^2$. 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 $P_s(u,v)$.

[0098] 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.

5 [0099] 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.

10 [0100] 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.

15 [0101] 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
20 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.

25 [0102] 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.

[0103] 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.

[0104] 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.

[0105] 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.

[0106] 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.

[0107] 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.

[0108] Having fully described several embodiments of the present invention, many other equivalents and alternative embodiments will be apparent to those skilled in the art. These and other equivalents and alternatives are intended to be included within the scope of the present invention.

CLAIMS

What is claimed as invention is:

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

- 5 (a) insonifying the object one scan line at a time with a directional ultrasound pulse generated from a transmit transducer;
- (b) recording the echo received by a relatively non-directional receive transducer;
- (c) storing the digitized echoes each in a line of computer memory;
- (d) reconstructing scan lines as lines on the reconstructed image based on knowledge
10 of the location of the receive transducer relative to that of the transmit transducer and the average speed of ultrasound in the object; and
- (e) image processing to fill in gaps between scan lines and optionally correcting distortions caused by the point spread function of the system.

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

- (a) insonifying the object one scan line at a time with a directional ultrasound pulse generated from a transmit transducer;
- (b) recording the echo received by two or more relatively non-directional receive transducers;
- 20 (c) storing the digitized echoes each in a line of computer memory;
- (d) 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 and the average speed of ultrasound in the object; and
- (e) image processing to fill in the gaps between scan lines, to combine overlapping
25 scan lines and optionally to correct distortions caused by the point spread function of the system.

3. The method of either of claims 1 and 2, wherein the receive transducer is separated

from the transmit transducer.

4. The method of either one of claims 1 and 2, wherein the receive transducer is an element of the transmit transducer.

5

5. The method according to any one of claims 1 through 4, inclusive, wherein the reconstructed scan lines can be at arbitrary angles or locations and the number of such scan lines can exceed the number of actual transmitted pulses.

10

6. The method according to any one of claims 1 through 4, inclusive, wherein the insonifying transducer is a phased array type sector scanner which sends out pulses in successive angles from a common origin.

15

7. The method according to any one of claims 1 through 4, inclusive, wherein the insonifying transducer is a linear phased array which sends out pulses on parallel paths.

20

8. The method according to any one of claims 2 through 4, inclusive, wherein a separate image is reconstructed for each receive transducer and these images are superimposed by finding the displacement which maximizes correlation to one chosen as the reference images and then averaging the images.

25

9. The method according to any one of claims 2 through 4, inclusive, wherein a separate image is reconstructed for each receive transducer, 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 any one of claims 1 through 4, inclusive, further including the step of determining the position of the receive transducer(s) relative to the transmit transducer by using an instrumented arm connecting the receive transducer(s) to the probe containing the transmit transducer.

5

11. The method according to any one of claims 1 through 4, inclusive, further including the step of determining the position of the receive transducer(s) by using a rigid construction of the receive transducer in the body of the main probe containing the transmit transducer.

10

12. The method according to any one of claims 1 through 4, inclusive, further including the step of determining the position of the receive transducer(s) relative to the transmit transducer by triangulation using IR, laser, or radio transponders attached to the receive and transmit ultrasound transducers.

15

13. The method according to any one of claims 1 through 4, 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.

20

14. The method according to any one of claims 1 through 4, inclusive, 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

25

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.

5 15. The method according to any one of claims 1 through 4, inclusive, wherein said image processing step (step (e)) involves using an image sharpening technique.

 16. The method of claim 15, wherein said image sharpening technique is either 1D or 2D unsharp masking.

10

 17. A method of determining the position of a receive transducer relative to the insonifying transducer, comprising the steps of:

 (a) using a main probe having a transmit transducer to transmit an ultrasound pulse at one or more angles;

15

 (b) recording the echo returns on both the main probe and on a receive transducer;

 (c) using correlation techniques to recognize distinctive short periods of echoes

 (d) noting the time of return for each patch through the different paths;

 (e) using sets of data from two or more patches; and

 (f) solving the simultaneous equations for the relative position (x,y).

20

 18. The method of claim 17, wherein step (e) involves using sets of data from three or more patches, and wherein step (f) involves solving the simultaneous equations for the relative position (x,y,z) when z is not constrained to be zero.

25

 19. The method of claim 17, wherein step (e) involves using more sets of data from patches than there are unknowns, and further including the step of finding the minimum of an error function to find the relative position (x,y) or (x,y,z).

20. A method of reconstructing an image using two or more active phased array ultrasound transducer sections separated by a gap or gaps, said method comprising the steps of:

5 (a) insonifying the object one scan line at a time with a directional ultrasound pulse formed by phasing elements from a plurality of transducer sections;

(b) recording the echoes received by elements in multiple transducer sections;

10 (c) reconstructing scan lines as lines on a reconstructed image based on knowledge of the position of each transducer element relative to a reference and the instantaneous position along the scan line from which echoes are coming at a particular time relative to the time of the transmit pulse, and

(d) image processing to fill in gaps between scan lines and optionally to correct distortions caused by the point spread function of the system.

21. An ultrasound imaging system, comprising:

15 a transmit transducer array for insonifying selected object and/or area of living tissue with a directional ultrasound pulse;

at least one receive transducer for detecting and recording ultrasound echo returns, wherein said transducer is relatively non-directional in the plane of scan;

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

22. The apparatus of claim 21, wherein at least one of said receive transducers is physically separated from said transmit transducer array.

23. The apparatus of claim 21, wherein at least one of said receive transducers is co-located with said transmit transducer array and comprises an element in said transmit transducer array.

24. The apparatus of claim 21, wherein said transmit transducer is a phased array type sector scanner which sends out pulses in successive angles from a common origin.

25. The apparatus of claim 21, wherein said transmit transducer is a linear phased array which sends out pulses on parallel paths.

26. The apparatus of claim 21, wherein said software includes means to reconstruct a separate image for each of said receive transducers.

27. The apparatus of claim 26, wherein said software further includes means to superimpose images for each of said receives transducers by finding the displacement which maximizes correlation to one chosen as the reference images and then averaging the images.

28. The apparatus of claim 26, wherein said software includes means to deconvolve the separate images for each of said receive transducers based on its point spread function and an estimate of the noise spectra, and then superimpose the images by finding the displacement which maximizes correlation to one chosen as the reference image, and then averaging the images.

29. The apparatus of claim 21, further including a transponder attached to each of said transmit and receiver transducers for determining the position of said receive transducer(s) relative to said transmit transducer.

5 30. The apparatus of claim 29, wherein said transponder is selected from the group consisting of IR, laser, and radio transponder.

31. The apparatus of claim 21, wherein said image processing software includes a Wiener filter.

10

32. The apparatus of claim 21, wherein said image processing software includes means for least mean square filtering.

15 33. The apparatus of claim 21, wherein said image processing software includes image sharpening means.

34. The apparatus of claim 33, wherein said image sharpening means is selected from the group consisting of 1D and 2D unsharp masking.

20 35. The apparatus of claim 21, wherein during use each of said transmit transducer and said receive transducer have a position (x,y) , and wherein said software includes correlation techniques to recognize distinctive short periods of echoes, note the time of return for each patch through the different paths, use sets of data from two or more patches, and solve simultaneous equations for the relative position (x,y) .

25

36. The apparatus of claim 21, wherein said transmit transducer array and said receiver transducer are each pivotally connected to an articulating arm so as to be mechanically linked and to move in the same plane.

37. The apparatus of claim 36, wherein said receive transducers are sensitive in the same plane as the plane of scan of the transmit transducer array.

5 38. The apparatus of claim 36, further including position sensors disposed on each of said transmit transducer and said receiver transducer for determining the position of said transmit transducer in relation to said receiver transducer.

10 39. The apparatus of any one of claims 21 through 38, inclusive, 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 reconstructs each two dimensional image, assembles the 2D images into a 3D volume in computer memory using interpolation
15 algorithms, aligns and performs 3D segmentation on the volume, and displays the 3D volume as a projection on a plane, or a 2D slice of arbitrary orientation

20 40. The apparatus of any one of claims 21 through 38, inclusive, 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 reconstructs each two dimensional image, assembles the 2D images into a 3D volume in computer memory using interpolation
25 algorithms, aligns and performs 3D segmentation on the volume, and displays the 3D volume as a projection on a plane, or a 2D slice of arbitrary orientation.

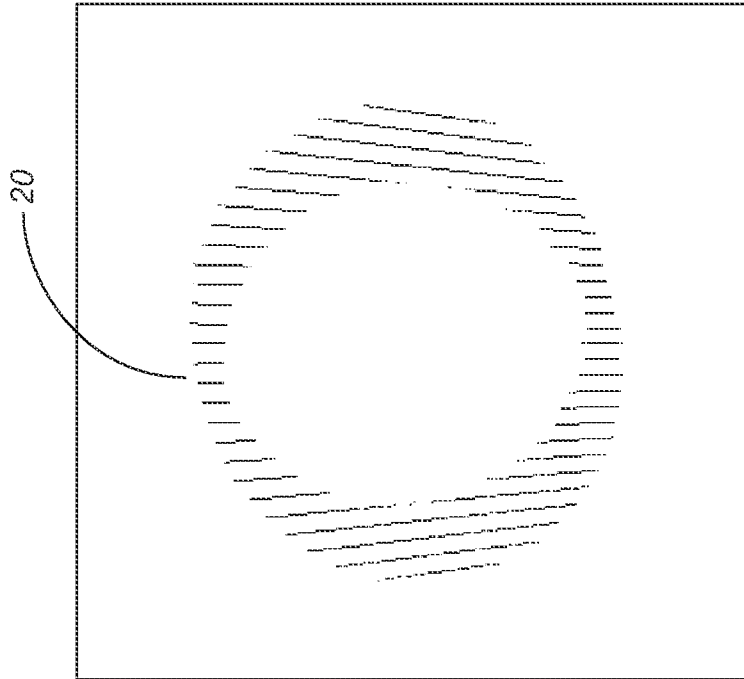


FIG. 5

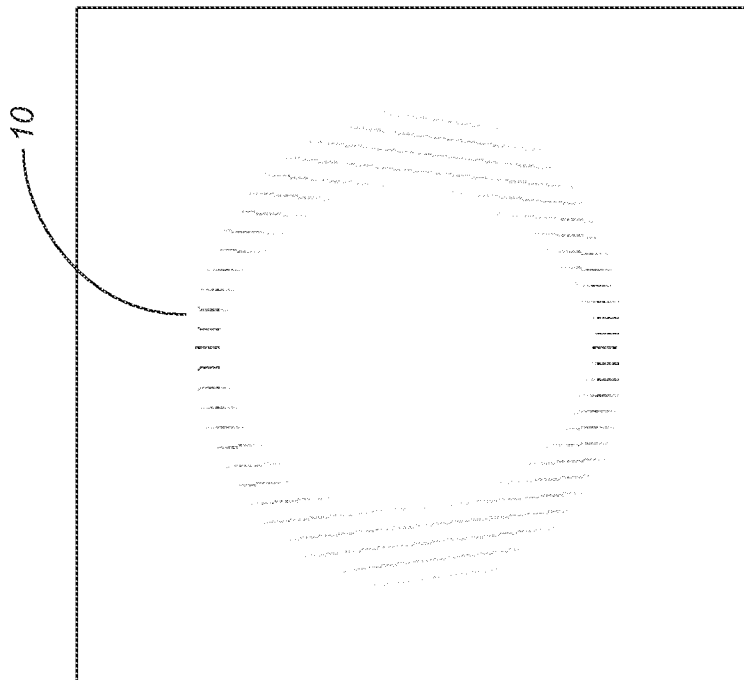
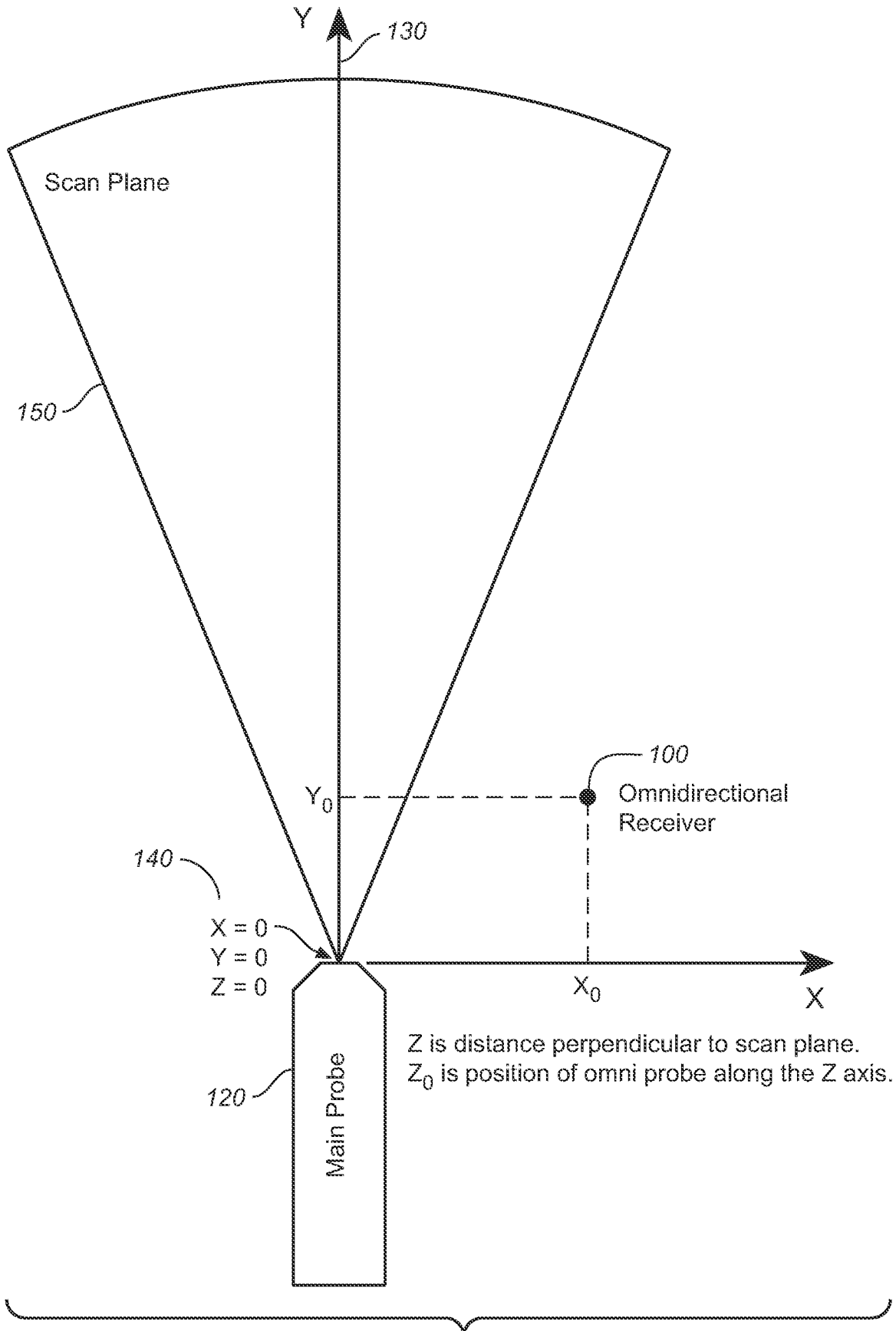


FIG. 1



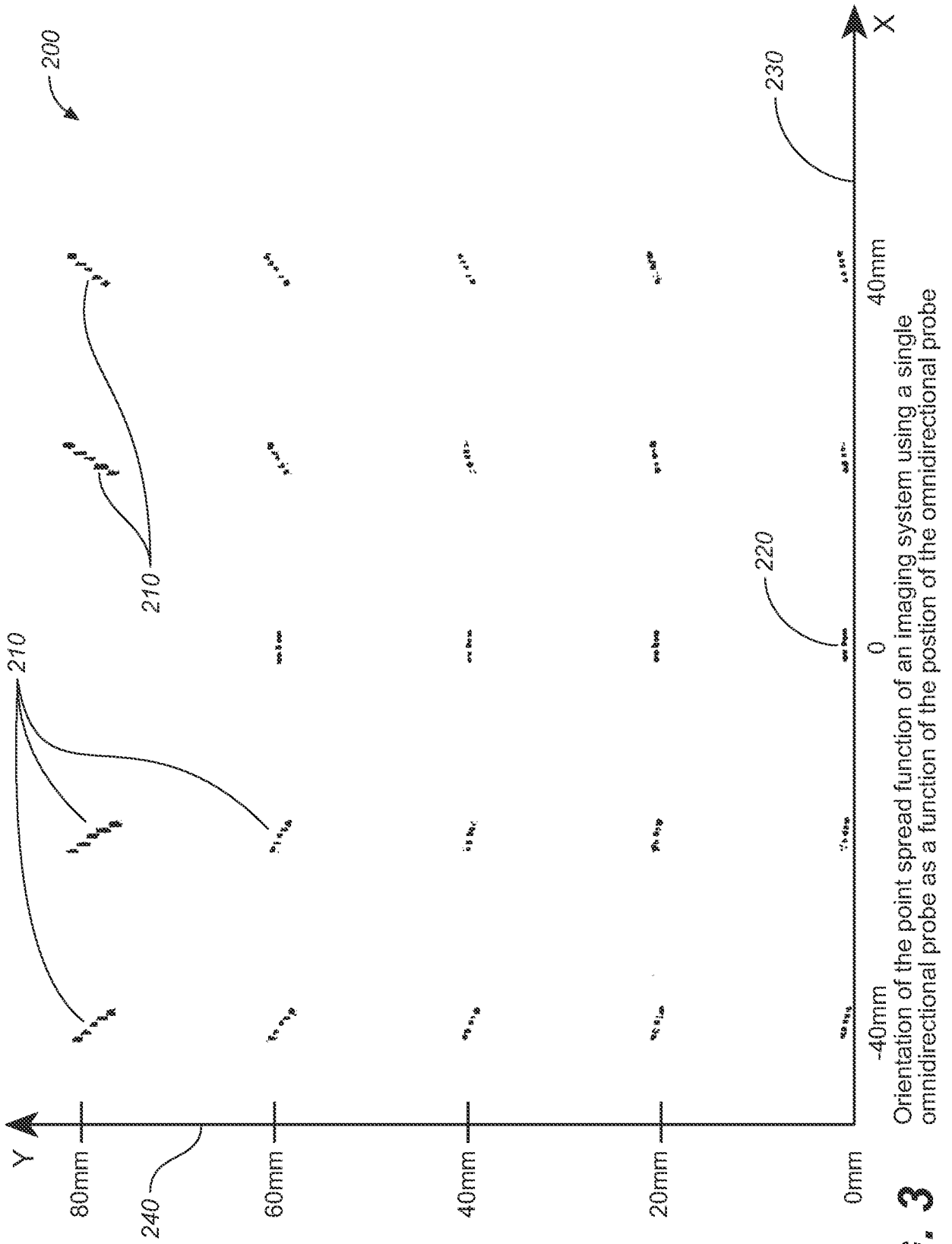


FIG. 3 Orientation of the point spread function of an imaging system using a single omnidirectional probe as a function of the position of the omnidirectional probe

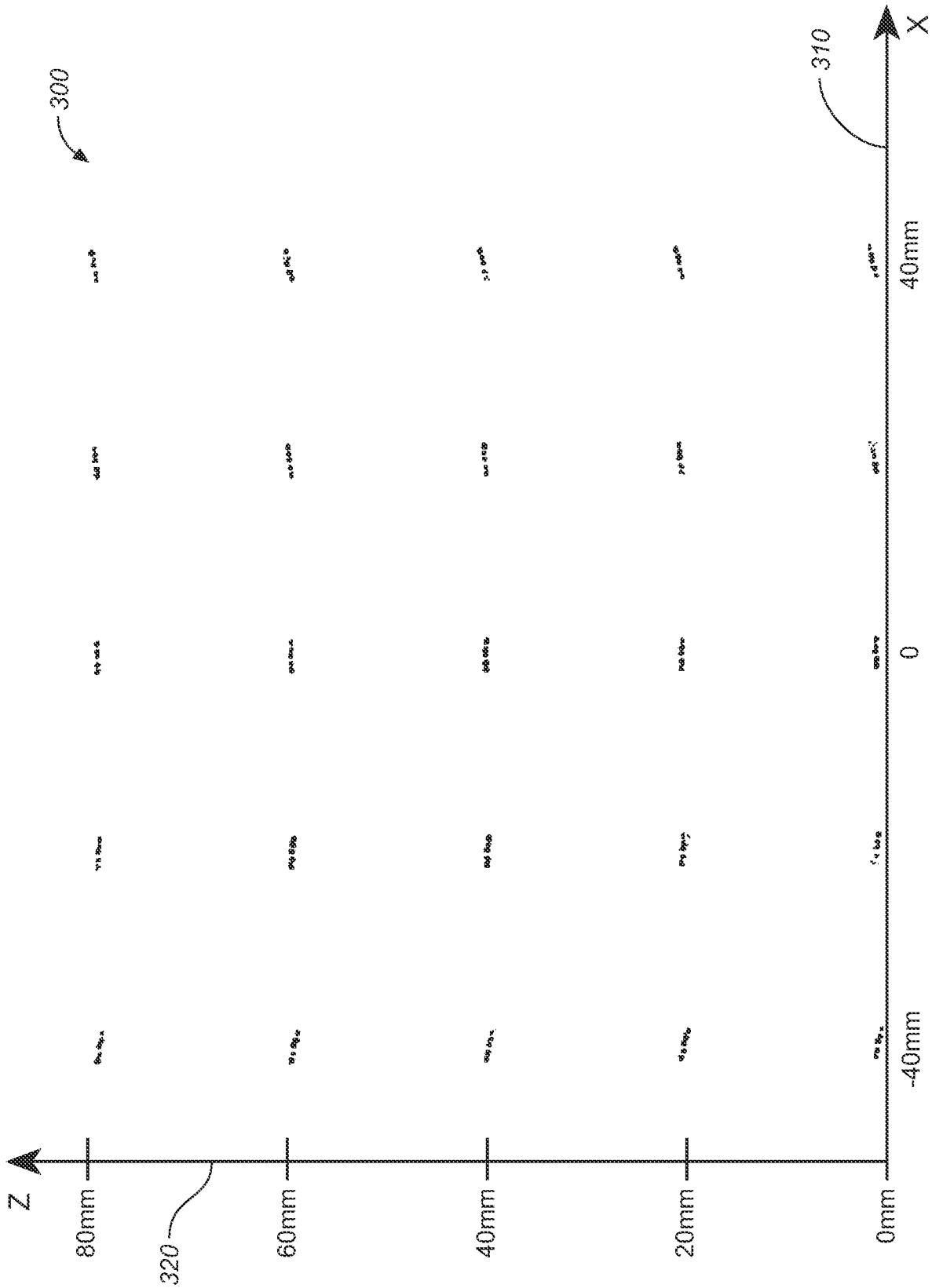


FIG. 4 Orientation of the point spread function of an imaging system using a single omnidirectional probe as a function of the position of the omnidirectional probe when it is not in the scan plane $y=0$

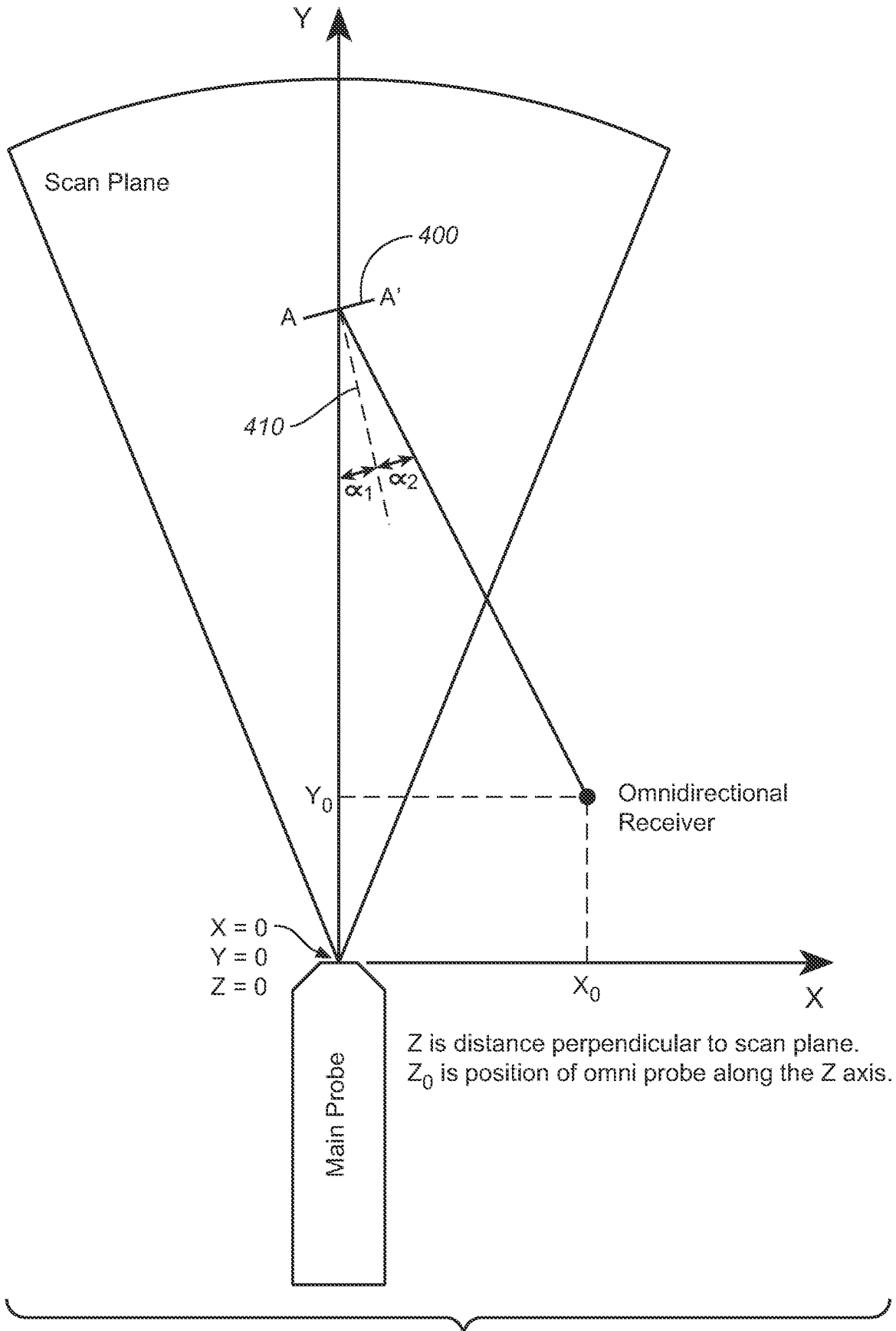
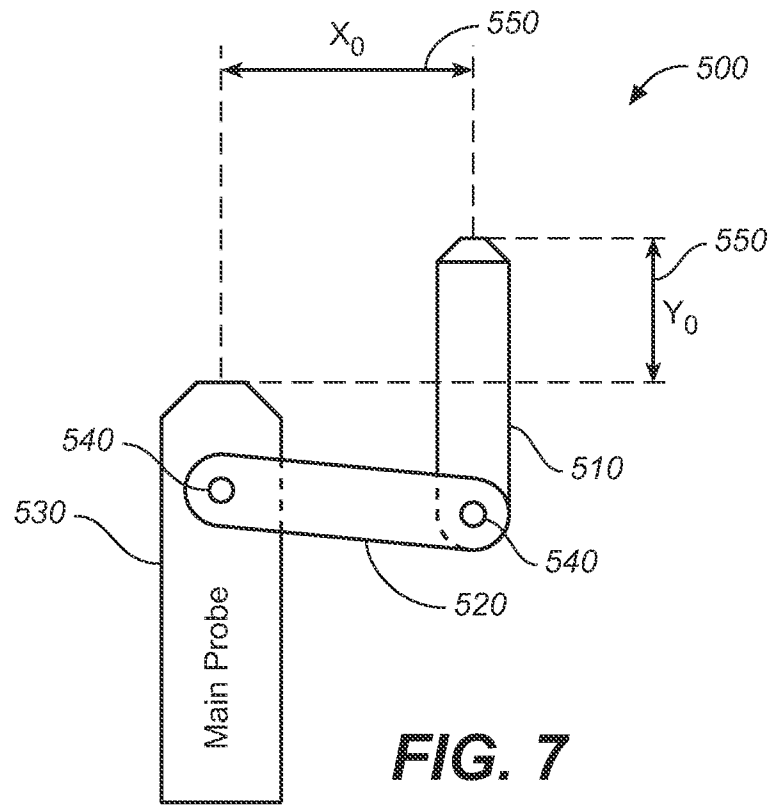
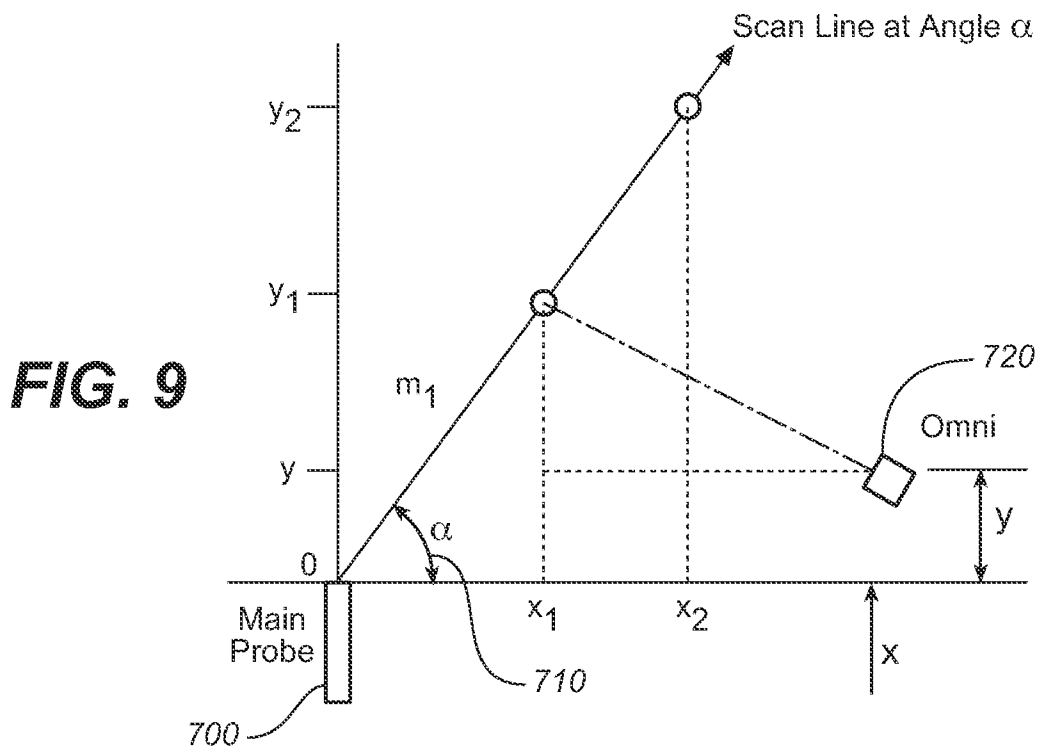
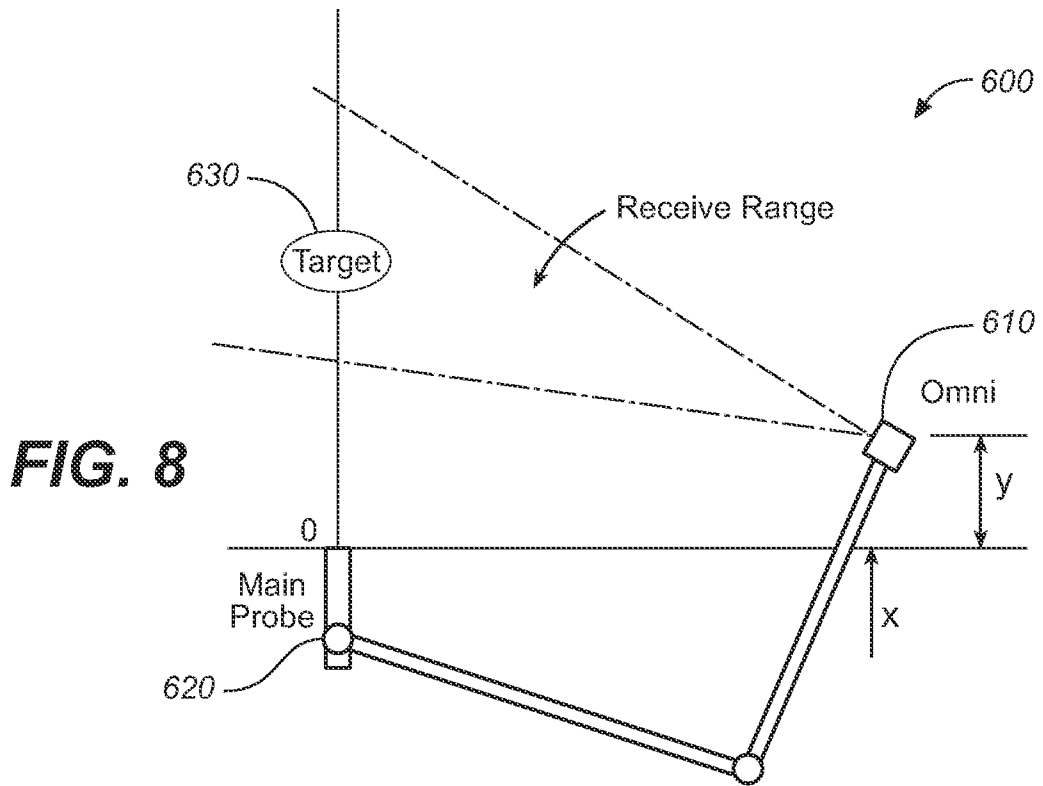


FIG. 6





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800

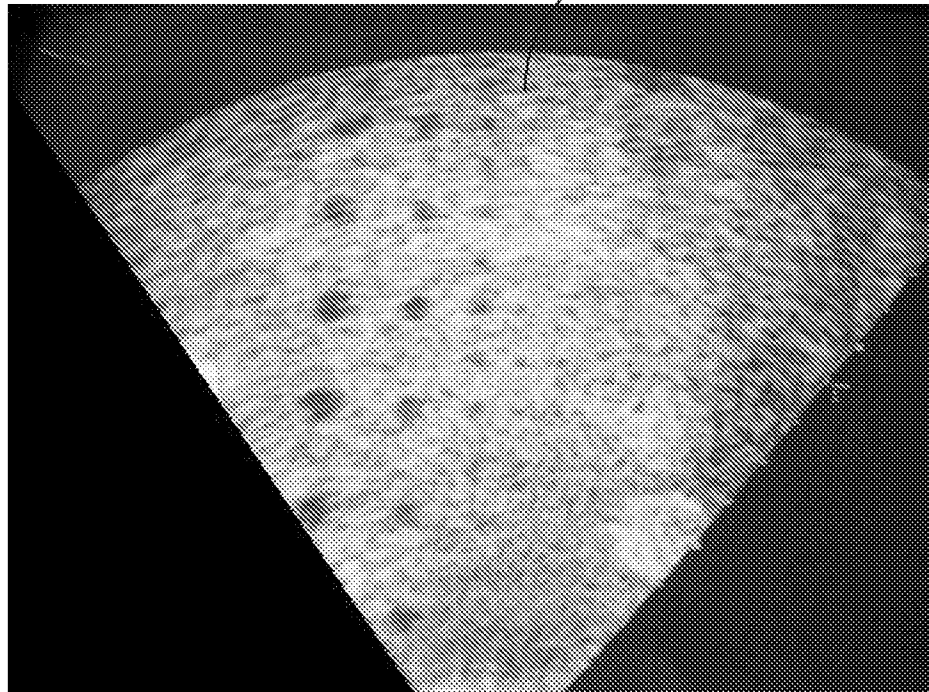


FIG. 10A

900

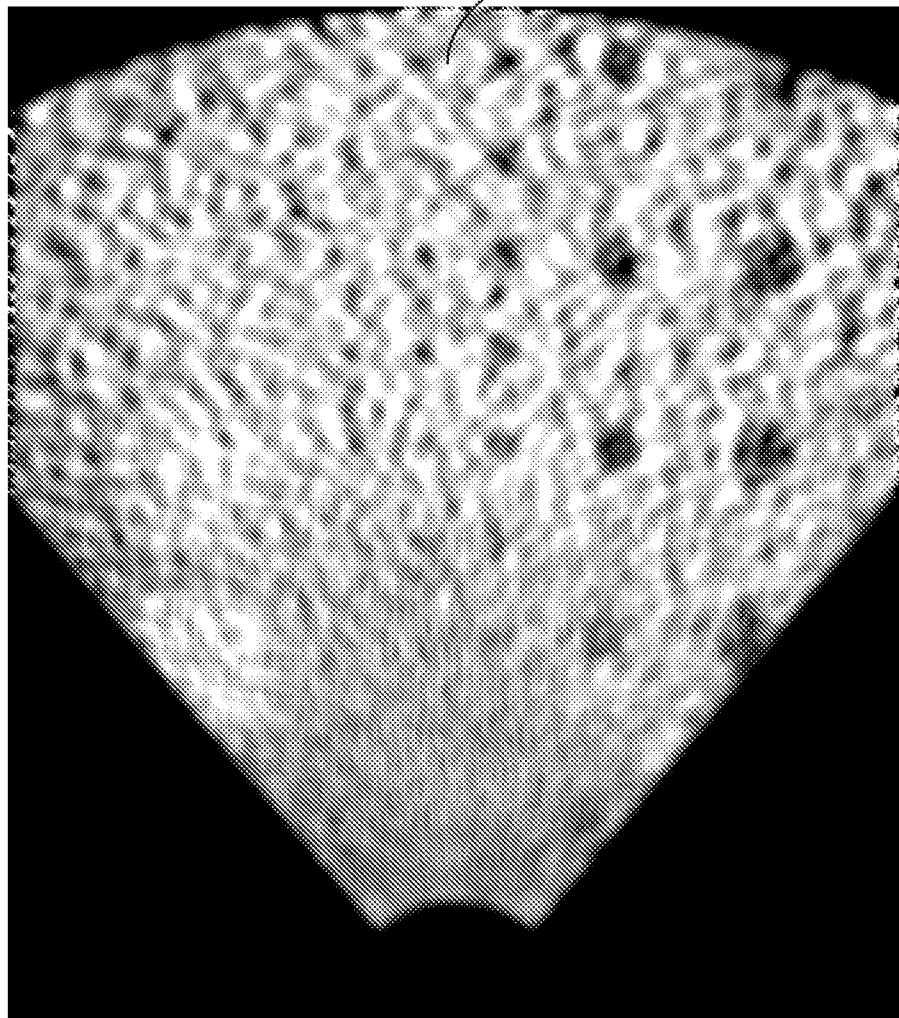


FIG. 10B

9 / 11

1000

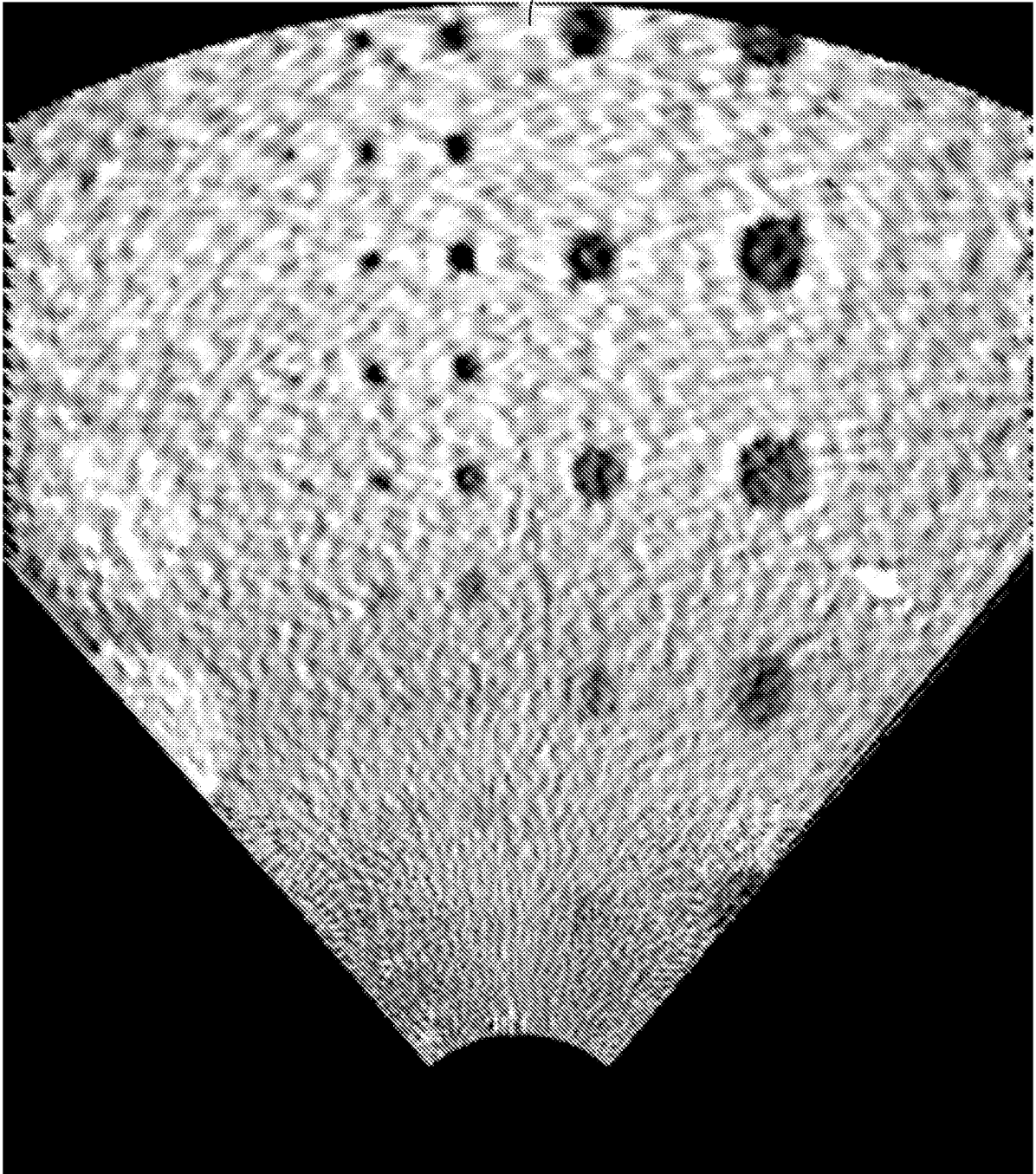


FIG. 11

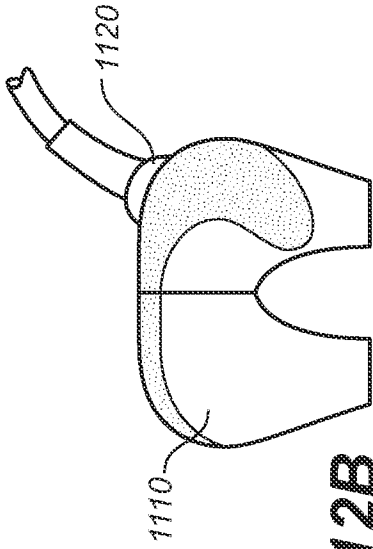


FIG. 12B

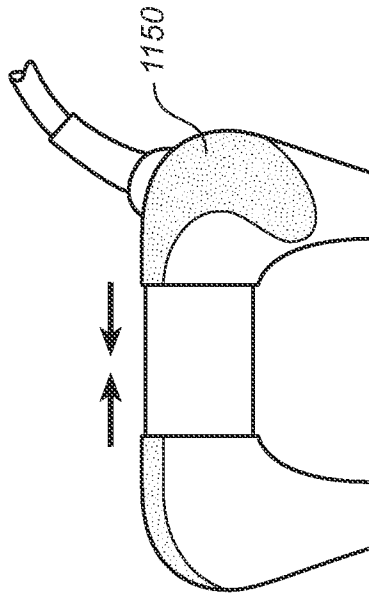


FIG. 12C

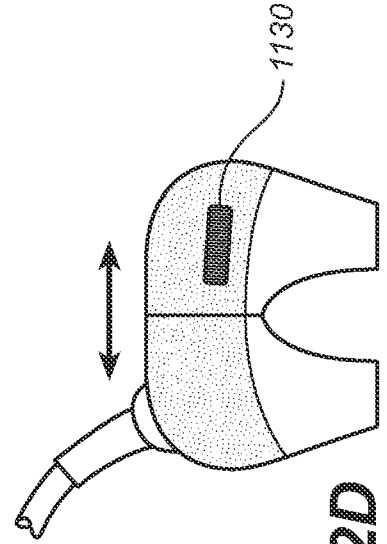


FIG. 12D

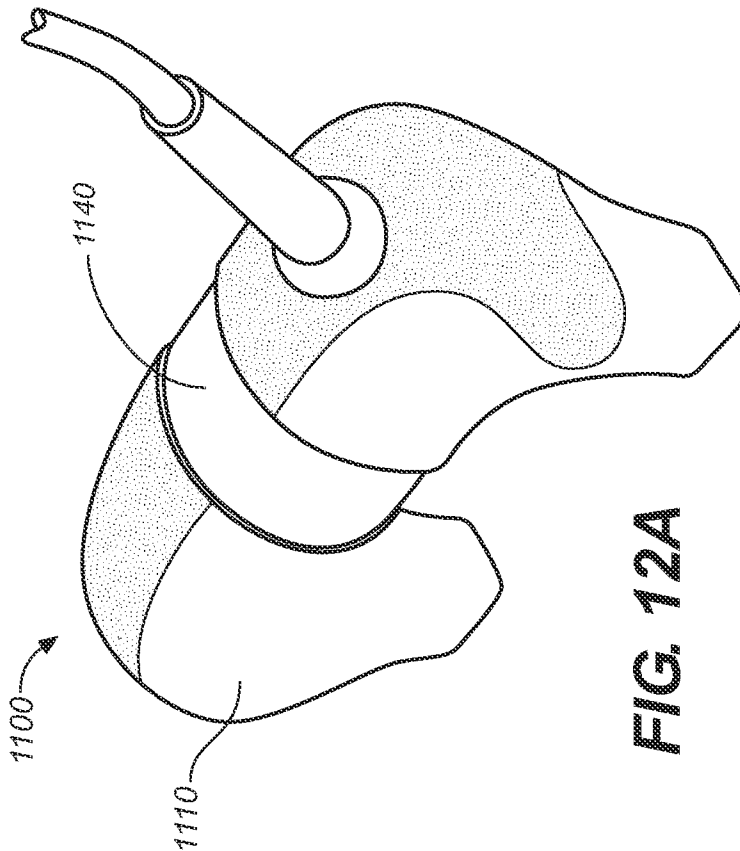


FIG. 12A

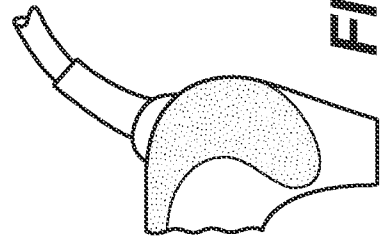


FIG. 12E

