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(54) **MULTIPLE SCAN-PLANE ULTRASOUND IMAGING OF OBJECTS**

ULTRASCHALLDARSTELLUNG VON OBJEKTEN IN MEHREREN SCANEbenen

FORMATION D'IMAGES ULTRASONORES A PLANS DE BALAYAGE MULTIPLES D'OBJETS

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Description

1. Field of the invention

[0001] The present invention is directed to real time ultrasound imaging of an object, like a biological structure, in multiple scan planes. Designs of efficient ultrasound phased array bulk wave transducers for two-dimensional ultrasound imaging, with several, electronically selectable 2D scan planes, are given. The electronic selection of the 2D scan plane direction allows so rapid switching of the direction that practically real time observation of moving objects like the heart, can be obtained simultaneously in several scan planes. The invention also addresses real time ultrasound monitoring of the cardiac function.

2. Background of the invention

[0002] In medical ultrasound imaging, one often examines an object through a variety of two-dimensional (2D) scan plane directions to observe the form, regional variations, and volume of the object. In particular one uses such multiple scan plane imaging of the heart to assess regional variations in the wall motion and myocardial contraction/relaxation, as well as to calculate the time varying volume of the heart, or the temporal variation of the fiber stress and strain in the heart. Other situations are observations of fetuses, tumors, etc., especially for calculating the volume to study growth.

[0003] In practical clinical examinations, such multiple 2D scan planes are generally obtained through manual movement of the 2D ultrasound probe, where for example with transesophageal measurements of the heart with a 2D phased array probe, special mechanisms have been designed for rotation of the ultrasound array through remote control. A phased array transducer for electronic selection of the ultrasound scan planes in two angular directions has been presented in [1], but this solution has found limited practical use, mainly because one wants to observe the object in more than two planes, generally at least three or four angular directions of the 2D scan plane.

[0004] The present invention presents a solution to this problem, with the design of a phased array transducer that allows free electronic selection of the 2D scan plane in more than 2 angular directions, for example 3 or 4 angular directions.

3. Summary of the invention

[0005] The invention is defined in independent claims 1,3 and 11.

[0006] The invention presents methods for practically real time ultrasound imaging of objects in 3 or more 2D ultrasound scan planes, where the 2D scan-planes are rotated around a common axis. With practically real time is here meant that the image data is collected along a set of sample beams within said 2D scan planes, and the sample beam data collection occurs so frequently of the object that all the sample beam data defining the image is collected within so short time period in relation to movements of the object, that the collected data captures a for practical purposes frozen frame the moving object. The image data can in addition be displayed on an image screen with so slow delay that one for practical purposes can view the object as it moves.

[0007] Such multiple 2D real time imaging is useful to study regional variations in cardiac wall motion. Using image analysis for automatic selection of the edges of for example a heart cavity or a tumor, one can study dimension and volume variations of the object to study tumor growth or ventricular volumes and ejection fraction. Based on ventricular dimensions, one can calculate myocardial strain, and with Doppler analysis one can quantitate regional wall velocity and strain velocity of the myocardium in each beam direction. Doppler analysis can also be used with forced deformation of tissue to study regional variations in tissue elasticity with methods referred to as elastography. With added input of the left ventricular pressure, as for example during anesthesia or critical care monitoring, the invention also presents methods for real time calculation of the left ventricular fiber stress, for accurate observation of the physiological function of the myocardium. The invention hence devices the use of such imaging for monitoring of cardiac performance, for example during anaesthesia or other critical care monitoring.

[0008] The invention further presents detailed designs of ultrasound transducer arrays that allows sector scanning of an ultrasound beam in three or four 2D scan planes that are rotated around a common axis. The design is composed of at least two active piezoelectric, phased array transducer layers mounted face to face in a sandwich structure that is mounted on a backing material. The piezoelectric sandwich front face is connected to the load material through a set of elastic impedance interfacing layers for improving the acoustic power coupling between the piezoelectric layers and the load material. In a special wide band design, the impedance interfacing layer closest to the piezoelectric layers has the same characteristic impedance as the piezoelectric layers, as described in [5].

[0009] Both the front and back side electrodes of each array layer are formed as a set of isolated, parallel finger/element electrodes that are connected to a switching circuit. The finger/element directions on the front and back side of each layer form angles to each other. The finger/element directions of the different piezoelectric layers are arranged so that the finger directions of at least one set of electrodes of one layer form an angle to the finger directions of at least one

set of electrodes of other layers.

[0010] The switching circuit is electronically controlled so that the full set of finger electrodes of each piezoelectric layer surface can selectively be connected either to the hot wires of a phased array ultrasound imaging instrument, or connected to signal ground. This allows electronic selection of one of a set of 2D scan plane directions for each transmitted pulse. With limited number of beams in each 2D image, one can get so high frame rate that when switching the scan planes sequentially, the images in the different scan planes show practically real time display of the object. Hence, the imaging instrument can show real time display of 3 or 4 scan plane directions, for example to observe regional contraction abnormalities of the myocardium:

[0011] Transmitting a wide beam and covering it with several narrow receive beams in parallel, one can increase the frame rate by a factor of order 2 - 6. Other increase in the frame rate is obtained by reducing the beam density where the scan planes intersect, and where the beams have limited information about the object, for example internal in the left ventricle while the ventricular walls are covered with high density of beams for ventricular wall motion analysis. For special high frame rate observation of the heart in the different scan planes, one can keep the same scan plane direction for a whole cardiac cycle, and change direction of the scan plane right before the onset of the myocardial contraction, for example triggered by the ECG signal from the heart.

[0012] The invention also shows a special embodiment where the multi-scan plane phased array is mounted at the tip of an endoscope for imaging of objects from internal in the body. In particular, such a method is useful for transesophageal imaging of the heart in many scan planes, for example to assist anesthesia monitoring during surgery. The transesophageal array can further be rotated mechanically in the endoscope for detailed alignment of the scan planes with interesting areas, or to increase the number of scan planes that are imaged.

[0013] Ultrasound contrast agents can be used to observe regional variations of myocardial blood perfusion. Such contrast agent can also be administered for efficient determination of the ventricular volumes. An example of an array embodiment is given that allows imaging in multiple frequency bands. This embodiment is well adapted for transmission of ultrasound pulses within one band of frequencies, and reception in sub, 2nd, 3rd, or 4th harmonic component of the transmit band. Such harmonic imaging is especially useful for the detection of ultrasound contrast agent, both to assess regional variations in the myocardial blood perfusion, and for automatic detection of the volume of ventricular cavities, and for detection of regions of tumors with increased vascularization.

4. Summary of the drawings

[0014]

Figure 1 shows in Figure 1a an example of an imaging system that allows real time imaging of an object simultaneously in three 2D scan planes, while Figure 1b shows an example display of 3 real time 2D images, and Figure 1c illustrates a spiral scanning of ultrasound beam for observation of the myocardium with minimal lag between beams across the myocardium.

Figure 2 shows a cross section in the thickness direction of a typical layer structure of the transducer array, according to the invention,

Figure 3 shows a two-layer array with finger electrode structure that allows for electronic selection of three 2D scan planes as in Figure 1a,

Figure 4 shows transmit transfer functions that can be obtained with the transducer structure in Figure 3,

Figure 5 shows yet another transducer array structure with two piezoelectric layers according to the invention that allows for electronic selection of 4 scan planes,

Figure 6 shows example scan planes that can be selected with the transducer array structure in Figure 5,

Figure 7 shows how the scan planes can be selected with an electronic switching circuit located in the ultrasound probe close to the array, to minimize the number of cables connecting the probe to the imaging instrument,

Figure 8 shows a transesophageal probe with an ultrasound array and electronic switching circuit mounted at a tip of a gastroscope according to the invention,

Figure 9 shows transmission of the ultrasound beam within three selectable 2D scan planes rotated around a common axis with a 2D matrix array,

Figure 10 shows by example displays of the image data, where Figure 10a shows example display of 4 simultaneous, real time 2D images, and Figure 10b shows a bullet display of myocardial strain data in three 2D images,

Figure 11 shows temporal displays of cardiac physiological variables obtained with a cardiac monitoring instrument

according to the invention, showing both a fast time which displays variations within a heart beat, and a slow time frame that shows variation of parameters from beat to beat.

5. Detailed description of the invention

[0015] Several example embodiments according to the invention is presented in the following. It is clear that this presentation is meant for illustration purposes only, and by no means represents limitations of the invention, which in its broadest aspect is defined by the claims appended hereto.

[0016] As an example embodiment of the invention, Figure 1a shows a transducer array 100 that is capable of transmitting pulsed ultrasound beams with directions freely and electronically steerable within for example three 2D sector scan planes 101, 102, and 103, that are rotated with different angular directions around a common axis 104. The transducer array is connected via a cable 105 to an ultrasound imaging instrument 106 with image outputs given to a display screen 107.

[0017] In a typical imaging situation, the ultrasound imaging instrument 106 directs signals to the transducer that transmit ultrasound beams in selected directions within the 2D scan planes 101 - 103. The back scattered signal is for each beam direction analyzed in the instrument to generate a set of image parameters for a set of range samples along the beam directions. Typical image parameters can be a compressed version of the backscattered amplitude for imaging of tissue structures, or Doppler frequency parameters for imaging of blood velocities, velocities of tissue movement, or strain of tissue regions, for example myocardial strain, or enforced strain on a tumor for elastography imaging, all according to known methods.

[0018] The beam scanning can be done in many ways, for example sequentially for each 2D scan plane in turn, or in a spiral pattern as described below. The image parameters for the beam directions that belong to each scan plane are then grouped together to form 2D images of the image parameters for the scan planes. These 2D images are then shown in real time on a display screen, for example as the images 111, 112, 113 in Figure 1b. This Figure also illustrates a time trace 114 of a cardiac ECG to be used as timing reference, with an area 115 for display of alphanumeric data extracted from the images or other measurements.

[0019] With 64 beams per 2D image with an image range of 15 cm, one typically obtains 72 2D images per second. With 3 sequential 2D scan planes one hence gets 24 full updates of the three 2D scan-planes per second. Increasing to 4 sequential scan planes, one gets 18 full updates per second of the four 2D scan-planes. 64 beams with a 16 mm aperture array at 3 MHz give a total opening angle of the 2D image around ~ 60 deg with full angular sampling of the 2D image.

[0020] Transmitting a wide transmit beam and covering it with several (for example 2 - 6) narrow receive beams in parallel, allows an increase of the image rate by a factor 2 - 6. This also gives time for expanding the scan angle of each 2D scan-plane, for example to 90 deg. The image rate and/or the width of the 2D scan can also be increased by reduced beam density, especially in the overlap region of the planes close to the axis 104, and in regions with little information about the object, like the middle of the ventricle with apical long axis imaging of the heart. If faster 2D image rate is needed, for example to study details in the onset of cardiac contraction and relaxation, one can image a full cardiac cycle for each scan plane, and change the scan plane at the end of the cardiac cycle, for example triggered by the ECG signal.

[0021] For assessment of regional wall motion abnormalities of the heart, it is important that the time delay is minimal between the beams that samples the wall in different regions. A useful cardiac scanning is then obtained by apical imaging of the ventricle, locating the array center axis 104 along the left ventricle long axis, which in Figure 1c is shown as normal to the Figure in the center. A typical myocardium cross section is shown as the area 125 in this image. Cross sections of the scan planes 101, 102, 103 are shown as 121, 122, 123. In this particular beam scanning pattern, the scan plane direction is switched in a circle for each sequential beam direction, so that the beam scanning occurs in a spiral cone with steadily increasing/decreasing angle of the beam to the axis 104. The crosses indicates some of the interrogating beams, where a selection 126, 127, 128, indicate a subset of the beams that follow each other in a sequence in the direction 129. At the last beam 130 with this beam angle to the axis 104, the beam angle is increased so that the next beam cross section is 131 in the same scan plane 121 as the beam 126, but with larger (or possibly smaller) beam angle to the center axis 104. The beam direction 130 is in this example followed by the beam direction 132 in 2D scan plane 122 and so on as indicated by the arrows. As the myocardium 125 with this scanning pattern is imaged with beams of close to the same opening angle between the beam and axis 104, one can cover the whole myocardium in a limited time interval. Typically, the myocardium is covered with around $10^6 \sim 60$ consecutive beams that can be collected in

a time interval of 10 - 15 msec with a frame rate of ~ 80 frames per second. This provides a highly adequate volume frame rate to study regional variations in the contraction, relaxation pattern of the myocardium.

[0022] For display, the image data from the beam directions belonging to each 2D scan plane are grouped together to form 2D images that are displayed in real time as exemplified in Figure 1b. The multiple scan planes can then be used for real time observation of regional wall motion in the heart and volume calculations of the heart cavities, both for diagnosis and for peri and post surgical monitoring, and monitoring in other critical care situations, as discussed below.

[0023] Two embodiments of transducer arrays that provide the beam scanning according to the invention, is now presented.

[0024] Figure 2 shows a cross section in the thickness direction of a phased array according to the invention, where 201 and 202 show two piezoelectric array layers mounted on a backing structure 203 with so high acoustic power absorption that reflected waves in the backing material can be neglected. The piezoelectric layers are covered with phased array finger/element electrodes 208, 209, 210, and 211. The front electrodes 209 of the back layer and the back electrodes 210 of the front layer are electrically isolated from each other with the thin layer 212, for example a glue film. Between the piezoelectric layers and the acoustic load material 204, the structure contains a set of elastic layers 205 to interface the characteristic impedance of the piezoelectric layers (201, 202) to that of the load material 204.

[0025] A method for such impedance interfacing that gives a particularly wide bandwidth, is given in [5], where the first layer of structure 205 in contact with the piezoelectric layers contains an elastic layer with close to the same characteristic impedance as the piezoelectric layers, followed by layers with falling characteristic impedance towards the load. Such an impedance interfacing is particularly useful when using the two piezoelectric layers in electrical parallel coupling to transmit a low frequency pulse, and then receive higher harmonic components (2nd to 4th) of this pulse on for example the back layer 201, as described below. One might also transmit a pulse on layer 201 and receive a sub-harmonic pulse on the electrical parallel of layers 201 and 202, as described below. In other situations one can use more narrow-band impedance interfacing layers 205, according to standard methods.

[0026] An example of dicing of the piezoelectric plates to obtain a reduced characteristic impedance ceramic/polymer composite, with an arrangement of electrodes that allows phased steering of an ultrasound beam in three 2D planes, is shown in Figure 3, which shows the piezoelectric layers and electrodes from the faces. In Figure 3a, 301 shows dicing grooves of the ceramic piezoelectric layers, the same for both layers, with the remaining ceramic posts/islands as 302. The dicing grooves 301 are filled with polymer to produce a ceramic/polymer composite structure with characteristic impedance reduced from that of the bulk ceramic material by a factor around 0.5, depending on the relative volume fill of ceramic.

[0027] In a particular embodiment according to the invention, the back face of the piezoelectric layer 201 is covered with a set of divided finger/element electrodes 303 as shown in Figure 3b, while the front face of the back layer 201 and the back face of the front layer 202 are covered with divided finger electrodes 304 and 305 that have the same shape and direction, as shown in Figure 3c. The front face of piezoelectric layer 202 is covered with divided finger electrodes 306 shown in Figure 3d. The electrode sets 303 - 306 corresponds to the electrodes 208 - 211 in Figure 2, where the finger electrodes 304 and 305 are electrically isolated from each other by the layer 212 as illustrated in Figure 2.

[0028] A principle drawing of the 2D scan planes that can be obtained with the array in Figure 3, is shown in Figure 1a, where the array is illustrated as 100. Grounding electrode sets 304/209, 305/210, and 306/211, and operating the back electrodes 303/208 as the hot electrodes of the elements of a linear phased array transducer, one will obtain a 2D image scan plane shown as 101 in Figure 1a. Grounding electrode sets 303/208, 305/210, and 306/211, and operating the electrodes 304/209 as the hot element electrodes of a phased array, one obtains a 2D image scan plane shown as 102 in Figure 1a. Grounding electrode sets 303/208, 304/209, and 305/210, and operating the front electrodes 306/211 as the hot element electrodes of a phased array, one obtains a 2D image scan plane shown as 103 in Figure 1a.

[0029] Choosing the dicing and electrode directions at 60 deg angle to each other as shown in Figure 2a - 2d, one gets an equilateral triangle of the ceramic islands, and 60 degrees angle between the 2D scan planes in Figure 1a, which gives an even angular sampling of the object. With such angles, one gets equal dimensions of the ceramic islands of the composite allowing equal width of the electrodes/elements of the array. Other angular divisions of the array/electrodes is also possible, for example to sample the left ventricle closer to other, wanted image planes.

[0030] The transduction capabilities of the array are conveniently described by the transmit transfer functions $H_{tt}(\omega)$ of the array elements, defined as the transfer function from the transmit voltage on the electrodes to the vibration velocity on the element front surface. Typical transmit transfer functions for the back layer 201 and the front layer 202, with the other layer electrically shorted, are shown as 401 and 402 in Figure 4, respectively. Both scan planes 101 and 102 in Figure 1a are using the back piezoelectric layer 201 of Figure 2 with the transfer function 401 with close to flat response in a wide frequency range of 1.5 - 4.2 MHz. Scan plane 103 of Figure 1a is using the front piezoelectric layer 202 of Figure 2 with the transfer function 402 of Figure 4. We see that both layers have similar transfer function (401 and 402) in the 2.8 - 4.2 MHz frequency range. Hence, within this frequency band, the pulses for the third scan plane 103 obtained with the front layer can be made similar to the pulses for the scan planes 101 and 102 obtained with the back layer.

[0031] In Figure 2, the polarizations of the piezoelectric layers are shown as P1 (206) and P2 (207). With opposite

directions of the polarizations as shown in this example, one obtains electrical parallel coupling of the layers by grounding the front (211/306) and back (208/303) electrodes and operating the mid electrodes (209/304 and 210/305) as the hot electrodes of a phased array. The transmit transfer function for such electrical parallel coupled operation of the piezo-electric layers, is shown as 403 in Figure 4. We notice that this transfer function covers a lower band of frequencies from 0.8 - 2.2 MHz.

[0032] Hence, aside from the ability to electronically select three scan planes, the structure allows for operations in multiple frequency bands in scan plane 102, both for regular 1st harmonic imaging in multiple frequency bands and for harmonic imaging. For example, with electric parallel coupling of the layers as described above, one can transmit in a low frequency band 403. Grounding electrodes 208/303, 210/305, and 211/306 while operating 209/304 as the hot element electrodes of a phased array, one can receive 2nd, 3rd, or 4th harmonic components of the transmit band 401 with the back layer. Similarly, one can transmit in a high frequency band with grounded electrodes 208/303, 210/305, 211/306 and operating electrodes 209/304 as the hot element electrodes of a phased array, and receive sub harmonics of the transmit band by grounding electrodes 208/303 and 211/306, and connecting electrodes 209/304 and 210/305 to operate the two layers electrically in parallel for a low frequency band.

[0033] A dual piezoelectric layer structure that provides free selection of 4 scan planes, is shown in Figure 5, which shows the piezoelectric layers and electrodes from the faces as in Figure 3. In the thickness direction, the structure contains two piezoelectric layers, with an impedance interface of elastic layers to the load, as shown in Figure 2. In Figure 5a is shown an example of dicing of the ceramic piezoelectric layers, where 501 shows the dicing grooves and 502 shows the ceramic posts/island. The difference to the structure in Figure 3 is found in the ceramic dicing to form the ceramic/polymer composite, and the directions of the finger/element electrodes. In particular, for the structure in Figure 5 the two sets of intermediate electrodes 504/209 and 505/210 between layers 201 and 202 have different directions as shown in Figures 5c and 5d. The back side of layer 201 is furnished with a set of finger electrodes 503/208 shown in Figure 5b, and the front side of layer 202 is furnished with a set of finger electrodes 506/211 shown in Figure 5e.

[0034] We see that by grounding the electrodes 504/209, 505/210 and 506/211, and operating the back electrodes 503/208 as the hot element electrodes of a linear phased array, we obtain a 2D scan plane shown as 601 in Figure 6. Grounding the electrodes 503/208, 505/210 and 506/211, and operating the electrodes 504/209 as the hot element electrodes of the phased array, we get the 2D scan plane 602 in Figure 6. Grounding the electrodes 503/208, 504/209 and 505/210, and operating the electrodes 506/211 as the hot element electrodes of the phased array, we get the 2D scan plane 603 in Figure 6. Grounding the electrodes 506/211, 503/208 and 504/209, and operating the electrodes 505/210 as the hot electrodes of a phased array, we get the 2D scan plane 604 in Figure 6. The transfer function for the back and the front layers will be the same as 401 and 402 shown in Figure 4, while this structure will not allow phased array scanning with electric parallel coupling of the layers in the low frequency band 403.

[0035] Electronic selection of the illustrated scan planes can be done with an integrated circuit as for example illustrated in Figure 7. This Figure shows by example as 701 the multi layer/multi electrode transducer array of Figure 3. Wire sets 703, 704, 705, and 706 connects the electrode sets 303, 304, 305, and 306 to the electronic switching circuit 702, which by the control signals 707 selects the set of electrodes to be connected to the hot wires of the set of coaxial cables 708 that are fed to the imaging instrument 710, and the set of electrodes that are connected to the signal ground level 709. Such an electronic circuit can be mounted close to the array, so that only the cables 708 and the control signals 707 for the scan plane selection is connecting to the imaging instrument.

[0036] The transducer array and the scan plane selection circuit is then typically mounted at the tip of a hand held probe for imaging from the body surface, or at the tip of an endoscope, like a gastroscope, for imaging of objects from inside the body, for example transesophageal imaging of the heart or endoscopic imaging of a tumor.

[0037] Figure 8 shows by way of example a transesophageal probe 800 with an array 801 according to this invention mounted at the tip of the gastroscope that is inserted into the body. The angular directions of the steerable tip 802 of the probe are controlled via wires by the rotation handles 803 and 804, according to standard methods for wire control of the tip of endoscopes. This particular embodiment contains an additional control 805 for rotation as indicated by the arrow 806 of the array 801 within the gastroscope tip 802, for example to adjust one of the scan planes to follow the long axis of the heart. The other scan planes are then grouped around this long axis according to the design of the array. This adjustment can be used for standardized observation of regional wall motion of the heart or accurate calculation of the volume of heart cavities or other objects like tumors. The endoscope probe is connected to the imaging instrument through the cable 807.

[0038] Electronic steering of the beam within a selected set of 2D scan planes, can also be done with a two-dimensional matrix array illustrated as 900 in Figure 9. This array is composed of a set of small elements 905 located in a 2D matrix, and where the signal for each element can be individually delayed. By delaying the signals of the individual elements, the beam can be steered and focused in a selected directions within a cone 906, according to known principles. In particular one can select beam directions within a set of 2D scan planes 901, 902, 903 which allows scanning of the ultrasound beam as described above. However, the 2D matrix array severely increases the complexity of the ultrasound

scanning system, compared to the arrays in Figures 3 and 5.

[0039] Electronic selection of the 2D scan planes from a set, allows practically simultaneous imaging of the heart in 3 or 4 scan planes with the above described arrays, as discussed in relation to Figure 1. Here, Figure 1b shows by way of example a display of real time imaging with three 2D scan planes that for example could be obtained with the array in Figure 3. Similarly Figure 10a shows the display of 4 real time scan planes for example obtained with the array in Figure 5. In this example the 2D image displays 1001-1004 could correspond to the 2D scan planes 601 - 604 in Figure 6. A time trace 1005 showing the ECG for timing references, is included in the display together with a region 1006 for display of alphanumeric data, like the end diastolic and end systolic ventricular volumes, the ejection fraction, etc. The images in Figure 1a and 10 can typically be used for wall motion scoring, where the score data can be presented in a bullet image, according to well known principles.

[0040] Through automatic detection of the ventricular cavities in the 2D images, one can do real time calculation of the volumes of the heart cavities, especially the left ventricle as shown in [6], to study filling and ejection patterns, and also the ejection fraction of the left ventricle. Ultrasound contrast agent can be used to study regional perfusion variations in the myocardium, where harmonic imaging of the contrast agent improves the contrast to tissue power ratio. Ultrasound contrast agent can also be used in difficult to image patients for improved determination of the cavity regions.

[0041] In many monitoring situations, a measured left ventricular pressure $p_{LV}(t)$ is available throughout the whole cardiac cycle. In critical situations, like peri and post surgery, $p_{LV}(t)$ can be obtained with a catheter in the left ventricle. In other situations, one can obtain $p_{LV}(t)$ from the aortic pressure throughout systole when the aortic valves are open. The aortic pressure can be obtained from a catheter in a central vessel, or through diameter measurements of the subclavian or carotid vessels. Combined with ultrasound measurements of the left ventricular diameter, $D(t)$, and wall thickness, $H(t)$, one can calculate the fiber stress $\sigma_f(t)$ in the myocardium according to known methods [7] as

$$\sigma_f(t) = \frac{H(t)}{(D(t) + H(t)) \left[3 - \left(\frac{D(t) + H(t)}{2L(t)} \right)^2 \right]} p_{LV}(t) \quad (1)$$

where $L(t)$ is the length of the left ventricular cavity. $L(t)$ can in most situations be approximated by a constant average value. As the myocardial muscle volume is constant, one can also obtain the temporal variations of the wall thickness $H(t)$ from $D(t)$ and $L(t)$. The left ventricular volume V_{LV} can then be approximated by the truncated ellipsoid and calculated based on $D(t)$ and $L(t)$.

[0042] There also exists a formula for calculating the fiber stress from the left ventricular cavity volume, V_{LV} , and the myocardial wall volume V_w [8] as

$$\sigma_f(t) = \frac{3}{\ln \left(1 - \frac{V_w}{V_{LV}(t)} \right)} p_{LV}(t) \quad (2)$$

[0043] Such calculations hence allows, according to the invention, real time displays of advanced physiological contraction/relaxation parameters of the myocardium. Temporal differentiation of the parameter wave-forms, including the volume traces, enhances information about the physiological function during the rapid changes at the onset and the end of the contraction.

[0044] The average relative strain in the myocardial fibers in a ventricle with close to uniform contraction of the myocardium in all regions, can be approximated as equal to the relative variation in the cardiac diameter, $\epsilon_a(t) = D(t)/D_{max}$. With regional variations in the strain, as found with ischemic heart disease, the average strain is still an interesting measure of the cardiac performance, but in this situation one would also like to see an image of the regionally variable strain $\epsilon(\underline{r}, t)$, where $\underline{r}$ is a vector coordinate of the of the myocardial surface. Based on Doppler measurements along each beam direction, one can estimate the local strain rate in the myocardium, which can be displayed in different ways, for example as color or grey scale indications in the bullet image 1007 in Figure 10b, where the angular direction in the bullet image indicates the angular direction of the scan planes of 101, 102, and 103 Figure 1, and the radial direction in the bullet image indicates the distance from the apex of the heart.

[0045] To further illustrate the monitoring situation, Figure 11 shows an example display according to the invention, where the dimensional measurement of the heart obtained with the arrays according to the invention is used to calculate contractile parameters of the left ventricle. The display shows from the top to bottom, the ECG traces 1101, the left ventricular pressure 1102, the fiber stress 1103 and average, relative fiber strain 1104, and the left ventricular volume traces 1105. For trend information over a longer time period, one can show in a slower time scale the end systolic volume, ESV (1106), as the minimum of V_{LV} per beat, the end diastolic volume, EDV (1107), as the maximum of V_{LV} per beat, the stroke volume, SV (1108), cardiac output, EF (1109), and the ejection fraction, CO (1110), calculated as

$$SV = EDV - ESV$$

$$CO = SV \cdot HR$$

$$EF = SV / EDV$$

(3)

where $HR = 60/THR$ is the number of beats per minute calculated from the duration THR of the cardiac cycle per beat in seconds, for example measured for each heart beat from the ECG. These parameters together with HR (1111) and other parameters as for example $\max(dV/dt)$, $\max(\sigma_f)$, $\max(d\sigma_f/dt)$, $\max(\epsilon_a)$, $\max(d\epsilon_a/dt)$, etc. can conveniently be displayed as numbers for each heart beat along each trace or in an alphanumeric area 1112.

[0046] Thus, while there have been shown and described and pointed out fundamental novel features of the invention as applied to a preferred embodiment thereof, it will be understood that various omissions and substitutions and changes in the form and details of the devices illustrated, and in their operation, may be made by those skilled in the art without departing from the invention.

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[0047]

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[2] US Pat 5,724,976, Mar. 10, 1998: "Ultrasound Imaging Preferable to Ultrasound Contrast Echography" Inventors: Yoshitaka Mine, Makoto Hiram, Hiroyuki Tsujino, Shiro Saito. Assignee: Kabushiki Kaisha Toshiba, Kawasaki, Japan

[3] US Pat 5,825,117 Oct. 20, 1998: "Second Harmonic Imaging Transducers" Inventors: William J. Ossmann, Larry Pendergass, Martin K. Mason. Assignee: Hewlett-Packard Company, Palo Alto Calif.

[4] US Pat 5,957,851 Sep. 28, 1999: "Extended Bandwidth Ultrasonic Transducer" Inventor: John Hossack. Assignee: Acuson Corporation, Mountain View, Calif.

[5] US Pa Appl 60/260,023 Filed Jan 05 2001: "Wide or Multiple Frequency Band Transducer and Transducer Arrays". Inventors: Bjørn A. J. Angelsen and Tonni F. Johansen

[6] Aakhus S., Mæhle J., Bjørnstad K.: "A new method for echocardiographic computerized three-dimensional reconstruction of left ventricular endocardial surface: In vitro accuracy and clinical repeatability of volumes." J Am Soc Echocardi, Vol 7, No6, 1994:571-581

[7] Rabben S.I., Irgens F., Angelsen B.: "Equations for estimating muscle fiber stress in the left ventricular wall", Heart and Vessels 1999, 14:189-196

[8] Arts T, Bovendeerd PHM, Prinzen FW, Reneman RS: "Relation between left ventricular cavity pressure and volume and systolic fiber stress and strain in the wall." J Biophys 59, 1991:93-102

Claims

1. An ultrasound imaging system for practically real time ultrasound imaging of an object in at least 3 different 2D image planes, including:

a display screen for displaying ultrasound images of an object in at least three different 2D image planes, an ultrasound transducer array arranged to electronically scan an ultrasound beam with electronically and freely selectable beam directions within at least three different 2D sector scan planes that are located with different angular directions around a common axis, the beam direction being switchable either sequentially for each 2D sector scan plane, or both within and between the 2D sector scan planes from pulse to pulse; means for detecting and using the back scattered signal from pulses in each beam direction to form image data as a function of depth along image beam directions; and means for grouping the image data from each image beam direction together to form 2D images of the object from each 2D sector scan plane that are obtained with a frame rate that, for practical purposes in relation to object movements, displays real time 2D images from the at least three different 2D sector scan planes.

2. An ultrasound imaging system according to Claim 1, arranged such that the 2D scan plan is cyclically changed for each beam sampling, so that a 3D scanning of the object in a spiral cone is obtained for observation of the object with minimal delays between sampling beams in neighbouring 2D scan planes.
3. A transducer array for an ultrasound imaging system according to Claim 1 or Claim 2, in which the transducer array is built up of at least two piezoelectric array layers mounted face to face in a sandwich structure with a front face and a back face, the back face of the piezoelectric sandwich being mounted on a backing material, the front face being connectable to the acoustic load material through at least one elastic layer for acoustic impedance interfacing between the piezoelectric layers and the load material, a set of parallel finger electrodes attached on each face of the piezoelectric layer where the electrodes are electrically isolated from each other, the directions of the fingers on the front and back face for each piezoelectric layer forms a non-zero angle to each other, each set of finger electrodes on the faces being arranged to selectively be connected as the hot element electrodes to a phased array beam former, or connected to the beam former signal ground so that selectively connecting the whole set of finger element electrodes on one side of at least one piezoelectric layer as the hot element electrodes to the beam former, and grounding all other sets of finger electrodes on the other faces of the piezoelectric layers gives a linear phased array transducer where the angular direction of the 2D scan plane around a common axis is determined by the set of the finger electrodes that are selected as the hot electrodes of the array.
4. A transducer array according to Claim 3, in which the piezoelectric array layers are made as a composite of ferroelectric ceramic and polymer, where the composite structure is the same through all piezoelectric layers in the sandwich and the ceramic posts of the composite are separated from each other with at least some linear grooves, some of the linear grooves conforming with the separations between the finger electrodes on the face of the piezoelectric layers, and hence also the active phased array elements defined by the electrodes connected to the hot element wires of the beam former.
5. A transducer array according to Claim 4, wherein the ultrasound transducer array includes two piezoelectric array layers, in which the ceramic posts in the composite structure have a triangular cross section, the triangles being arranged so that the dicing of the ceramic can be done to form the linear grooves, in which some of the linear grooves define the separation between the finger electrodes and hence also the selected, active phased array elements, the finger electrodes of one face of each piezoelectric layer having the same, first direction while the direction of the finger electrodes of the other faces are different from each other and different from said first direction.
6. A transducer array according to Claim 5, where the piezoelectric layers have opposite polarising directions, in which for operation in a lower frequency band in one 2D scan plane normal to the first direction, the two sets of finger electrodes having the same first direction are connected to each other to form the hot electrodes of a phased array, while the other set of electrodes are coupled to signal ground so that electric parallel operation of the layers is obtained for operation in a lower frequency band, and for operation in a higher frequency band with the 2D scan plane also normal to the first direction, only the electrodes of the front or the back layer with the first direction are used as the hot electrodes of the phased array, while all the other sets of finger electrodes are connected to signal ground.
7. An ultrasound imaging system including a transducer array according to any one of Claims 3 to 6, wherein the ultrasound array is mounted at the tip of an endoscope to be inserted into an object for ultrasound imaging of internal structures of the object, and where optionally the array can be rotated in the endoscope tip controlled by means at the external end of the endoscope or in the imaging instrument.
8. An ultrasound cardiac function monitoring system including an ultrasound imaging system according to Claim 1 or

Claim 2, and means for measuring a number of cardiac physiological measurements, in which the display means is arranged to visualise simultaneously the multiple scan planes of the heart on one part of the display or on a first display, and temporal variation of parameters derived from the ultrasound images and/or other physiological measurements shown as a function of time on other parts of the display screen or on a second display screen respectively.

9. An ultrasound cardiac function monitoring system according to Claim 8, including a means for edge detection in the ultrasound images for calculation of the volume of heart cavities, and/or calculation of relative strain in the myocardial fibres.
10. An ultrasound cardiac function monitoring system according to Claim 8 or Claim 9, in which ventricular dimensions or volume are combined with ventricular pressure measurements to calculate the fibre stress in the cavity walls,
11. An ultrasound imaging method for practically real time ultrasound imaging of an object in at least three different 2D image planes, the method including:

scanning an ultrasound beam direction with electronically and freely selectable beam directions within at least three different 2D sector scan planes that are located with different angular directions around a common axis, the beam direction being switchable either sequentially for each 2D sector scan plane or both within and between the 2D sector scan planes from pulse to pulse;

detecting and using the back scattered signal from pulses in each beam direction to form image data as a function of depth along image beam directions; and

grouping the image data from each image beam direction together to form 2D images of the object from each 2D sector scan plane that are obtained with a frame rate that, for practical purposes in relation to object movements, displays real time 2D images from the at least three different 2D sector scan planes.

Patentansprüche

1. Ultraschall-Bildgebungssystem für Ultraschallbildgebung in praktisch Echtzeit eines Objekts in mindestens 3 verschiedenen zweidimensionalen Bildebenen, enthaltend:

- einen Anzeigebildschirm zum Anzeigen von Ultraschallbildern eines Objekts in mindestens drei verschiedenen zweidimensionalen Bildebenen,

- eine Ultraschall-Umformeranordnung, die angeordnet ist, um elektronisch einen Ultraschallstrahl mit elektronisch und frei wählbaren Strahlrichtungen innerhalb von mindestens drei verschiedenen zweidimensionalen Sektor-Abtastebenen, die sich mit verschiedenen Winkelrichtungen um eine gemeinsame Achse befinden, abzutasten, wobei die Strahlrichtung entweder sequenziell für jede zweidimensionale Sektor-Abtastebene oder sowohl innerhalb als auch zwischen den zweidimensionalen Sektor-Abtastebenen von Impuls zu Impuls umschaltbar ist.

- Mittel zum Erkennen und Nutzen des Rückstreuungssignals von Impulsen in jeder Strahlrichtung, um Bilddaten als eine Funktion der Tiefe entlang von Bildstrahlrichtungen zu bilden, und

- Mittel zum Zusammengruppieren der Bilddaten von jeder Bildstrahlrichtung, um zweidimensionale Bilder des Objekts von jeder zweidimensionalen Sektor-Abtastebene zu bilden, die mit einer Rahmenrate erfasst werden, die für praktische Zwecke in Beziehung auf Objektbewegungen zweidimensionale Bilder in Echtzeit von den mindestens drei verschiedenen zweidimensionalen Sektor-Abtastebenen anzeigt.

2. Ultraschall-Bildgebungssystem nach Anspruch 1, derart angeordnet, dass die zweidimensionale Abtastebene für jede Strahlabtastung zyklisch geändert wird, so dass eine dreidimensionale Abtastung des Objekts in einem Spiralkegel zur Beobachtung des Objekts mit minimalen Verzögerungen zwischen Abtaststrahlen in angrenzenden zweidimensionalen Abtastebenen erhalten wird.
3. Umformeranordnung für ein Ultraschall-Bildgebungssystem nach Anspruch 1 oder Anspruch 2, wobei die Umformeranordnung aus mindestens zwei piezoelektrischen Anordnungsschichten aufgebaut ist, die Fläche zu Fläche in einer Mehrschichtstruktur mit einer Vorderfläche und einer Rückfläche montiert sind, wobei die Rückfläche der piezoelektrischen Mehrschichtstruktur auf einem Grundmaterial montiert ist, die Vorderfläche mit dem akustischen Füllmaterial durch mindestens eine elastische Schicht für akustische Impedanzkopplung zwischen den piezoelektrischen Schichten und dem Füllmaterial verbindbar ist, ein Satz von parallelen Fingerelektroden an jeder Fläche der piezoelektrischen Schicht angebracht ist, wobei die Elektroden elektrisch voneinander isoliert sind, die Rich-

tungen der Finger an der Vorder- und Rückfläche für jede piezoelektrische Schicht einen Winkel ungleich null miteinander bilden, jeder Satz von Fingerelektroden auf den Flächen angeordnet ist, um selektiv als die Spannung führenden Elementelektroden mit einem phasengesteuerten Anordnungsstrahlformer verbunden zu werden oder mit der Strahlformer-Signalerde verbunden zu werden, so dass selektives Verbinden des gesamten Satzes von Fingerelement-Elektroden an einer Seite von mindestens einer piezoelektrischen Schicht als die Spannung führenden Elementelektroden mit dem Strahlformer und Erdung aller anderen Sätze von Fingerelektroden an den anderen Flächen der piezoelektrischen Schichten einen linearen phasengesteuerten Anordnungs-Umformer ergibt, wobei die Winkelrichtung der zweidimensionalen Abtastebene um eine gemeinsame Achse durch den Satz der Fingerelektroden, die als die Spannung führenden Elektroden der Anordnung ausgewählt sind, bestimmt wird.

4. Umformeranordnung nach Anspruch 3, wobei die piezoelektrischen Anordnungsschichten als eine Zusammensetzung aus ferroelektrischer Keramik und Polymer erstellt werden, wobei die Zusammensetzungsstruktur die gleiche durch alle piezoelektrischen Schichten in der Mehrschichtstruktur ist und die keramischen Ständer der Zusammensetzung durch mindestens einige lineare Rillen voneinander getrennt sind, wobei einige der linearen Rillen mit den Trennungen zwischen den Fingerelektroden an der Fläche der piezoelektrischen Schichten übereinstimmen, und somit auch die aktiven phasengesteuerten Anordnungselemente, die durch die Elektroden definiert sind, die mit den Spannung führenden Elementdrähten des Strahlformers verbunden sind.
5. Umformeranordnung nach Anspruch 4, wobei die Ultraschall-Umformeranordnung zwei piezoelektrische Anordnungsschichten enthält, wobei die keramischen Ständer in der Zusammensetzungsstruktur einen dreieckigen Querschnitt haben, die Dreiecke so angeordnet sind, dass die Zerteilung der Keramik durchgeführt werden kann, um die linearen Rillen zu bilden, wobei einige der linearen Rillen die Trennung zwischen den Fingerelektroden definieren und somit auch die ausgewählten, aktiven phasengesteuerten Anordnungselemente, wobei die Fingerelektroden einer Fläche jeder piezoelektrischen Schicht die gleiche erste Richtung haben, während die Richtung der Fingerelektroden der anderen Flächen voneinander verschieden und von der ersten Richtung verschieden ist.
6. Umformeranordnung nach Anspruch 5, wobei die piezoelektrischen Schichten entgegengesetzte Polarisationsrichtungen haben, wobei für Operation in einem niedrigeren Frequenzband in einer zweidimensionalen Abtastebene, die senkrecht zu der ersten Richtung ist, die zwei Sätze von Fingerelektroden, die dieselbe erste Richtung haben, miteinander verbunden sind, um die Spannung führenden Elektroden einer phasengesteuerten Anordnung zu bilden, während die Elektroden des anderen Satzes an Signalerde gekoppelt sind, so dass elektrisch parallele Operation der Schichten für Operation in einem niedrigeren Frequenzband erhalten wird, und für Operation in einem höheren Frequenzband, wobei die zweidimensionale Abtastebene auch senkrecht zu der ersten Richtung ist, nur die Elektroden der vorderen oder der hinteren Schicht mit der ersten Richtung als die Spannung führenden Elektroden der phasengesteuerten Anordnung verwendet werden, während alle anderen Sätze von Fingerelektroden mit Signalerde verbunden sind.
7. Ultraschall-Bildgebungssystem, das eine Umformeranordnung nach einem der Ansprüche 3 bis 6 enthält, wobei die Ultraschallanordnung an der Spitze eines Endoskops montiert ist, das in ein Objekt für Ultraschall-Bildgebung von internen Strukturen des Objekts einzuführen ist, und wobei die Anordnung optional in der Endoskopspitze durch Mittel an dem externen Ende des Endoskops oder in dem Bildgebungsinstrument gedreht werden kann.
8. Ultraschall-Herzfunktions-Überwachungssystem, enthaltend ein Ultraschall-Bildgebungssystem nach Anspruch 1 oder Anspruch 2 und Mittel zum Messen einer Zahl von kardialen physiologischen Messungen, wobei das Anzeigemittel angeordnet ist, um gleichzeitig die mehreren Abtastebenen des Herzens auf einem Teil der Anzeige oder auf einer ersten Anzeige und zeitliche Variation von Parametern, die von den Ultraschallbildern und/oder anderen physiologischen Messungen abgeleitet werden und als eine Funktion der Zeit angezeigt werden, jeweils auf anderen Teilen des Anzeigebildschirms bzw. auf einem zweiten Anzeigebildschirm sichtbar zu machen.
9. Ultraschall-Herzfunktions-Überwachungssystem nach Anspruch 8, enthaltend ein Mittel für Randerkennung in den Ultraschallbildern zur Berechnung des Volumens von Herzhohlräumen und/oder Berechnung von relativer Dehnung in den Herzmuskelfasern.
10. Ultraschall-Herzfunktions-Überwachungssystem nach Anspruch 8 oder Anspruch 9, wobei Ventrikelabmessungen oder -volumen mit Ventrikeldruckmessungen kombiniert werden, um die Faserspannung in den Hohlraumwänden zu berechnen.
11. Ultraschall-Bildgebungsverfahren für Ultraschallbildgebung in praktisch Echtzeit eines Objekts in mindestens drei

verschiedenen zweidimensionalen Bildebenen, das Verfahren enthaltend:

- Abtasten einer Ultraschallstrahlrichtung mit elektronisch und frei wählbaren Strahlrichtungen innerhalb von mindestens drei verschiedenen zweidimensionalen Abtastebenen, die sich mit verschiedenen Winkelrichtungen um eine gemeinsame Achse befinden, wobei die Strahlrichtung entweder sequenziell für jede zweidimensionale Sektor-Abtastebene oder sowohl innerhalb als auch zwischen den zweidimensionalen Sektor-Abtastebenen von Impuls zu Impuls umschaltbar ist.
- Erkennen und Nutzen des Rückstreuungssignals von Impulsen in jeder Strahlrichtung, um Bilddaten als eine Funktion der Tiefe entlang den Bildstrahlrichtungen zu bilden; und
- Zusammengruppieren der Bilddaten von jeder Bildstrahlrichtung, um zweidimensionale Bilder des Objekts von jeder zweidimensionalen Sektor-Abtastebene zu bilden, die mit einer Rahmenrate erfasst werden, die für praktische Zwecke in Beziehung auf Objektbewegungsgert zweidimensionale Bilder in Echtzeit von den mindestens drei verschiedenen zweidimensionalen Sektor-Abtastebenen anzeigt.

Revendications

1. Système de formation d'image ultrasonore pour la formation d'image ultrasonore pratiquement en temps réel d'un objet dans au moins trois plans image 2D différents, comprenant :

- un écran d'affichage pour afficher les images ultrasonores d'un objet dans au moins trois plans image 2D différents,
- un réseau de transducteur ultrasonore agencé pour balayer électroniquement un faisceau ultrasonore avec des directions de faisceau électroniquement et librement sélectionnables dans au moins trois plans de balayage secteur 2D différents qui sont situés avec différentes directions angulaires autour d'un axe commun, la direction de faisceau étant commutable soit séquentiellement pour chaque plan de balayage secteur 2D soit à la fois à l'intérieur et entre les plans de balayage secteur 2D d'impulsion à impulsion;
- un moyen pour détecter et utiliser le signal rétrodiffusé à partir d'impulsions dans chaque direction de faisceau pour former des données image en fonction de la profondeur le long des directions de faisceau d'image ; et
- un moyen pour grouper les données image de chaque direction de faisceau d'image ensemble pour former des images 2D de l'objet à partir de chaque plan de balayage secteur 2D qui sont obtenues avec un taux de trame qui, à des fins pratiques en rapport avec les mouvements d'objet, affiche les images 2D en temps réel des au moins trois plans de balayage secteur 2D différents.

2. Système de formation d'image ultrasonore selon la revendication 1. agencé de sorte que le plan de balayage 2D est cycliquement modifié pour chaque échantillonnage de faisceau, de sorte qu'un balayage 3D de l'objet dans un cône en forme de spirale est obtenu pour observation de l'objet avec un minimum de retard entre les faisceaux d'échantillonnage dans les plans de balayage 2D voisins.

3. Réseau de transducteur pour un système de formation d'image ultrasonore selon la revendication 1 ou la revendication 2, dans lequel le réseau de transducteur est constitué d'au moins deux couches de réseau piézoélectrique montées face à face dans une structure en sandwich avec une face avant et une face arrière, la face arrière du sandwich piézoélectrique étant montée sur un matériau de support, la face avant pouvant être connectée au matériau de charge acoustique par au moins une couche élastique pour l'interfaçage d'impédance acoustique entre les couches piézoélectriques et le matériau de charge, un ensemble d'électrodes en forme de doigt parallèles attachées sur chaque face de la couche piézoélectrique où les électrodes sont électriquement isolées les unes des autres, les directions des doigts sur la face avant et la face arrière pour chaque couche piézoélectrique forment un angle non nul les unes par rapport aux autres, chaque ensemble d'électrodes en forme de doigt sur les faces étant agencé pour être sélectivement connecté en tant qu'électrodes d'élément chaudes à un système de formation de faisceau de réseau phasé, ou connecté à la terre de signalisation du système de formation de faisceau de sorte à connecter sélectivement l'ensemble des électrodes d'élément en forme de doigt sur un côté d'au moins une couche piézoélectrique comme les électrodes d'élément chaudes au système de formation de faisceau, et la mise à la terre de tous les autres ensembles d'électrodes en forme de doigt sur les autres faces des couches piézoélectriques donne un transducteur de réseau phasé linéaire où la direction angulaire du plan de balayage 2D autour d'un axe commun est déterminée par l'ensemble des électrodes en forme de doigt qui sont choisies comme les électrodes chaudes du réseau.

4. Réseau de transducteur selon la revendication 3, dans lequel les couches de réseau piézoélectriques sont formées

comme un composite de céramique ferroélectrique et de polymère, où la structure composite est la même dans toutes les couches piézoélectriques dans le sandwich et les montants en céramique du composite sont séparés les uns des autres avec au moins certaines rainures linéaires, certaines des rainures linéaires se conformant aux séparations entre les électrodes en forme de doigt sur la face des couches piézoélectriques, et de ce fait également les éléments de réseau phasé actif définis par les électrodes connectées aux fils d'élément chauds du système de formation de faisceau.

5. Réseau de transducteur selon la revendication 4, dans lequel le réseau de transducteur ultrasonore comprend deux couches de réseau piézoélectriques, dans lequel les montants en céramique dans la structure composite ont une coupe transversale triangulaire, les triangles étant agencés de sorte que le découpage en dés de la céramique peut être effectué pour former les rainures linéaires, où certaines des rainures linéaires définissent la séparation entre les électrodes en forme de doigt et de ce fait également les éléments de réseau phasé actifs sélectionnées, les électrodes en forme de doigt d'une face de chaque couche piézoélectrique ayant la même première direction alors que la direction des électrodes en forme de doigt des autres faces est différente d'une électrode à l'autre et différente de ladite première direction.
6. Réseau de transducteur selon la revendication 5, où les couches piézoélectriques ont des directions de polarisation opposées, où pour un fonctionnement dans une bande de fréquence inférieure dans un plan de balayage 2D normal par rapport à la première direction, les deux ensembles d'électrodes en forme de doigt ayant la même première direction sont connectées les uns aux autres pour former les électrodes chaudes d'un réseau phasé, alors que l'autre ensemble d'électrodes est couplé à la terre de signalisation de sorte que le fonctionnement parallèle électrique des couches est obtenu pour un fonctionnement dans une bande de fréquence inférieure, et pour un fonctionnement dans une bande de fréquence supérieure avec le plan de balayage 2D également normal par rapport à la première direction, seules les électrodes de la couche avant ou de la couche arrière avec la première direction sont utilisées comme les électrodes chaudes du réseau phasé, alors que tous les autres ensembles d'électrodes en forme de doigt sont connectés à la terre de signalisation.
7. Système de formation d'image ultrasonore comprenant un réseau de transducteur selon l'une quelconque des revendications 3 à 6, dans lequel le réseau ultrasonore est monté au niveau de l'embout d'un endoscope inséré dans un objet pour formation d'image ultrasonore de structures internes de l'objet et où facultativement le réseau peut être mis en rotation dans l'embout d'endoscope commandé par un moyen au niveau de l'extrémité externe de l'endoscope ou dans l'instrument de formation d'image.
8. Système de surveillance de la fonction cardiaque ultrasonore comprenant un système de formation d'image ultrasonore selon la revendication 1 ou la revendication 2, et un moyen pour mesurer un certain nombre de mesures physiologiques cardiaques, où le moyen d'affichage est agencé pour visualiser simultanément les multiples plans de balayage du coeur sur une partie de l'affichage ou sur un premier affichage, et la variation temporelle de paramètres dérivés des images ultrasonores et/ou d'autres mesures physiologiques montrées en fonction du temps sur d'autres parties de l'écran d'affichage ou sur un second écran d'affichage respectivement.
9. Système de surveillance de la fonction cardiaque ultrasonore selon la revendication 8, comprenant un moyen pour la détection de contour dans les images ultrasonores pour le calcul du volume des cavités cardiaques et/ou le calcul de la tension relative dans les fibres myocardiques.
10. Système de surveillance de la fonction cardiaque ultrasonore selon la revendication 8 ou la revendication 9, dans lequel les dimensions ventriculaires ou le volume ventriculaire sont combinés avec les mesures de pression ventriculaire pour calculer la contrainte de fibre dans les parois de cavité.
11. Procédé d'imagerie ultrasonore pour la formation d'image ultrasonore pratiquement en temps réel d'un objet dans au moins trois plans image 2D différents, le procédé comprenant :

- le balayage d'une direction de faisceau ultrasonore avec des directions de faisceau électroniquement et librement sélectionnables dans au moins trois plans de balayage 2D différents qui sont situés avec différentes directions angulaires autour d'un axe commun, la direction de faisceau étant commutable soit séquentiellement pour chaque plan de balayage secteur 2D soit à la fois à l'intérieur et entre les plans de balayage secteur 2D d'impulsion à impulsion;
- la détection et l'utilisation du signal rétrodiffusé à partir d'impulsions dans chaque direction de faisceau pour former des données image en fonction de la profondeur le long des directions de faisceau d'image ; et

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- le groupement des données image de chaque direction de faisceau d'image ensemble pour former des images 2D de l'objet à partir de chaque plan de balayage secteur 2D qui sont obtenues avec un taux de trame qui, à des fins pratiques en rapport avec les mouvements d'objet, affiche les images 2D en temps réel des au moins trois plans de balayage secteur 2D différents.

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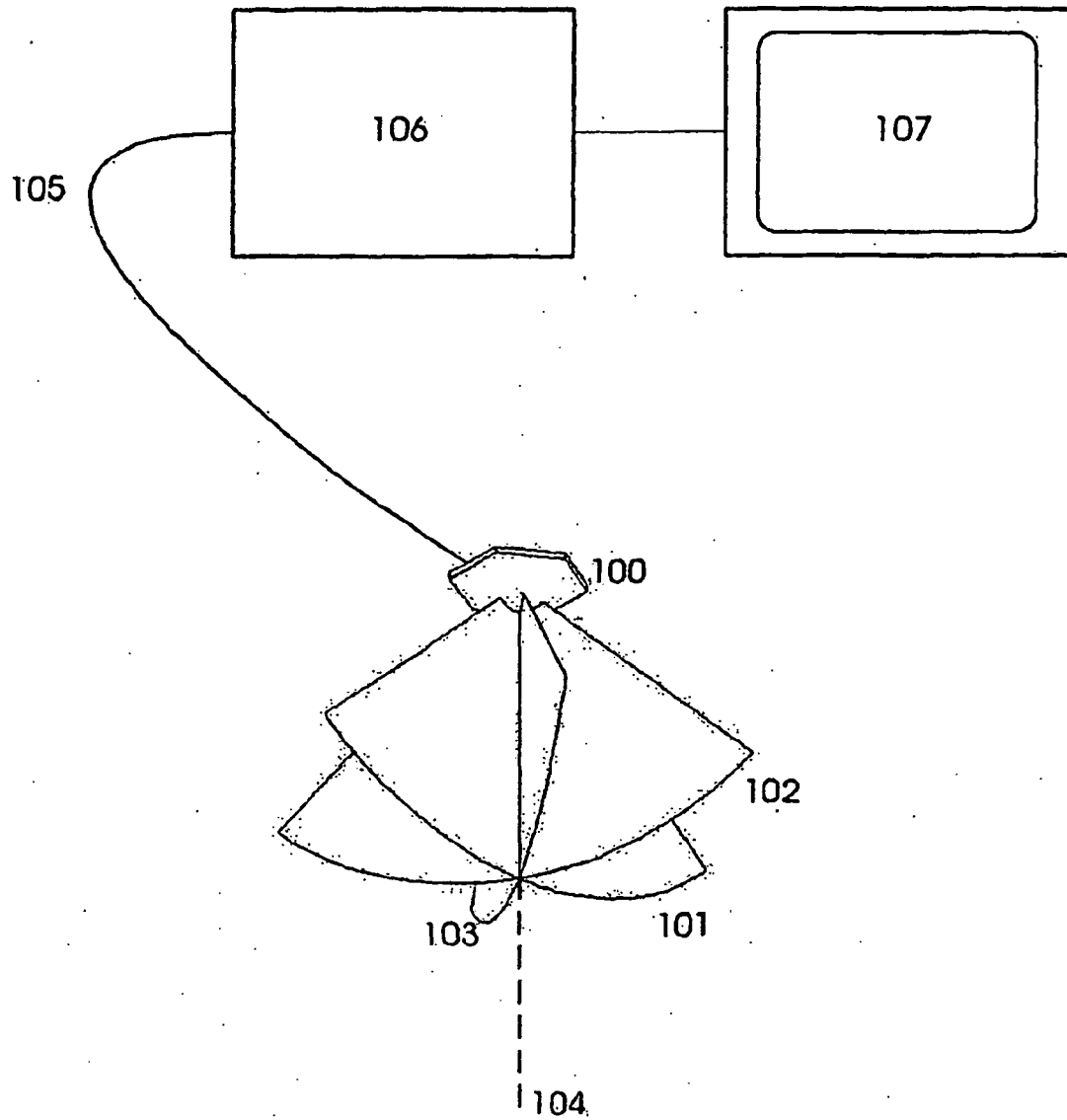


Figure 1a

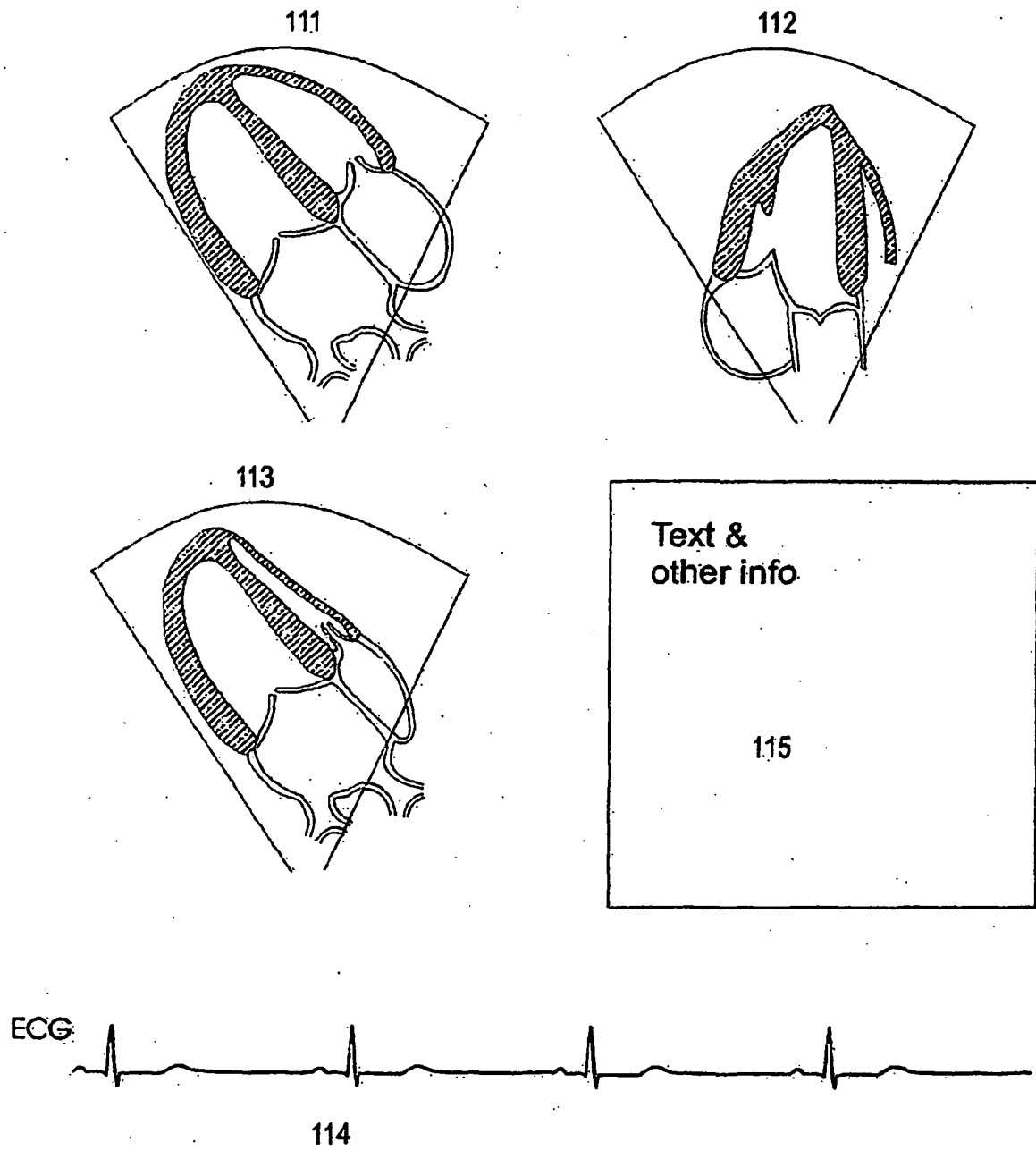


Figure 1b

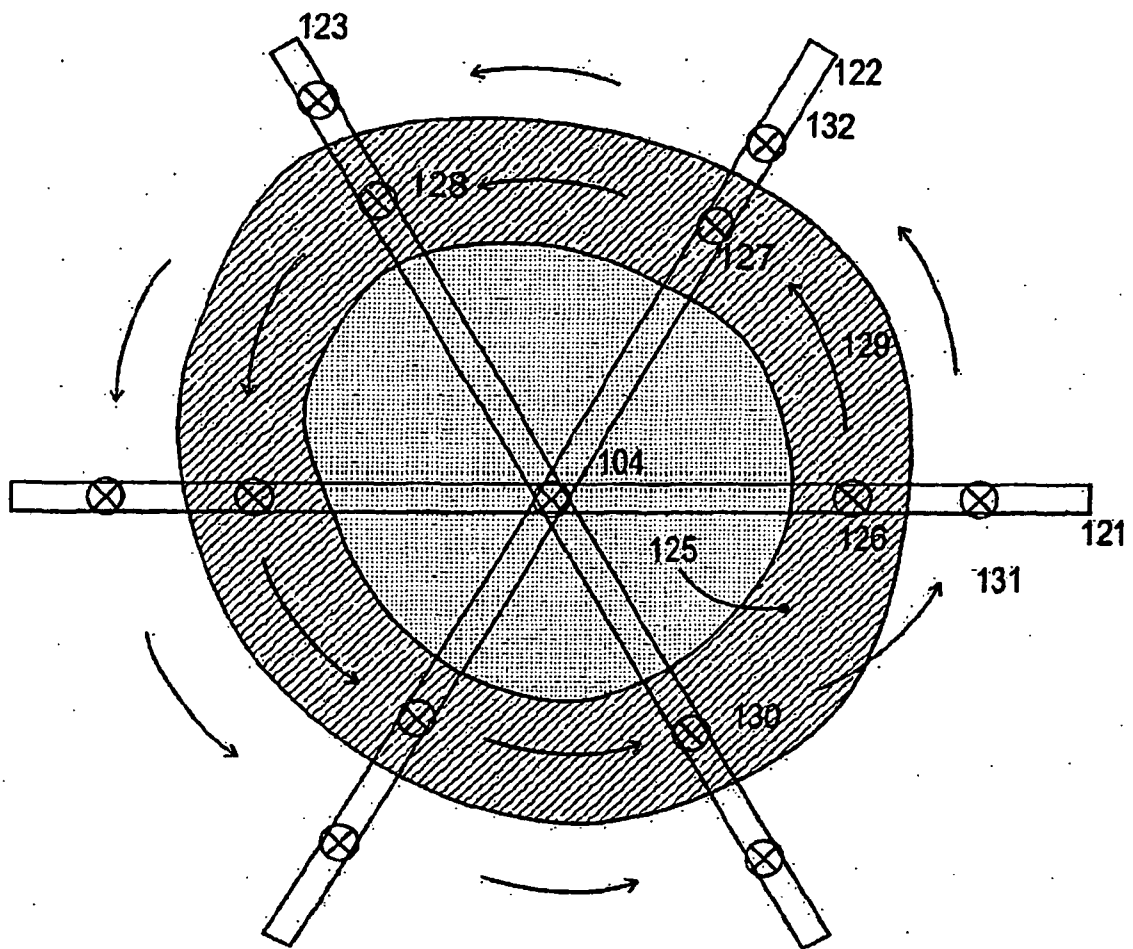


Figure 1c

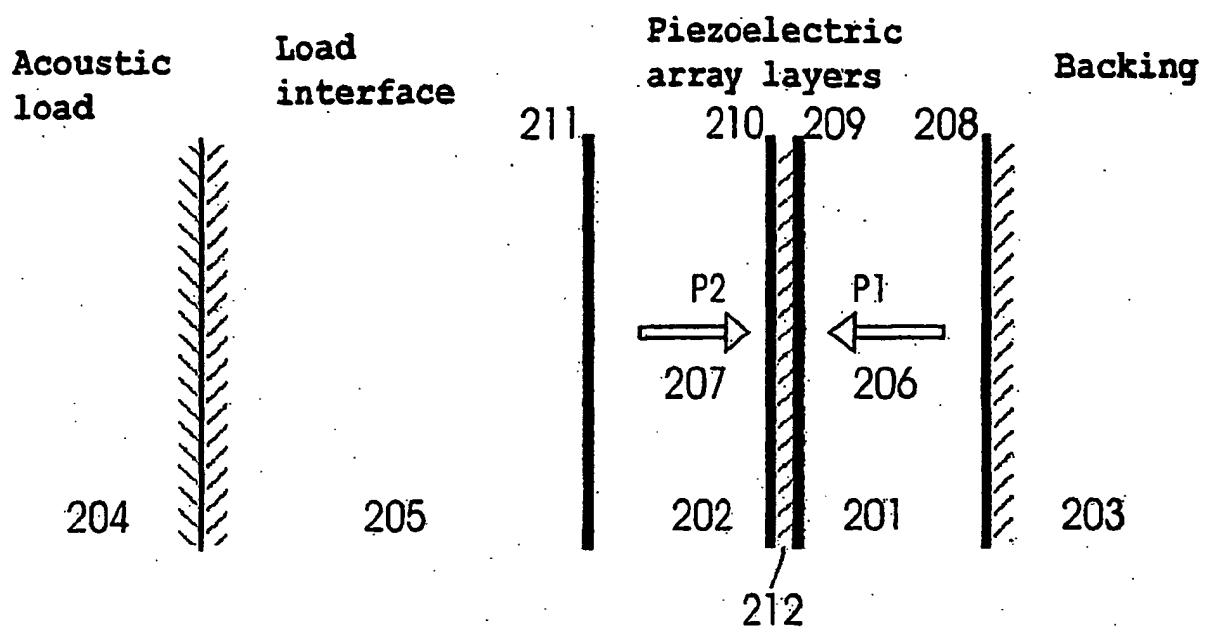


Figure 2

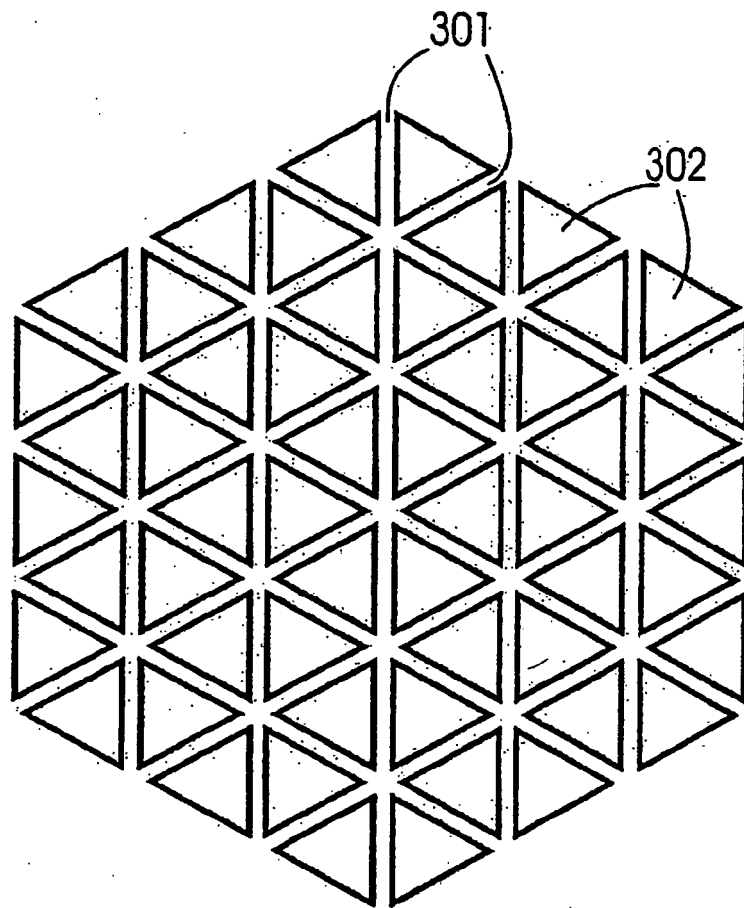


Figure 3a

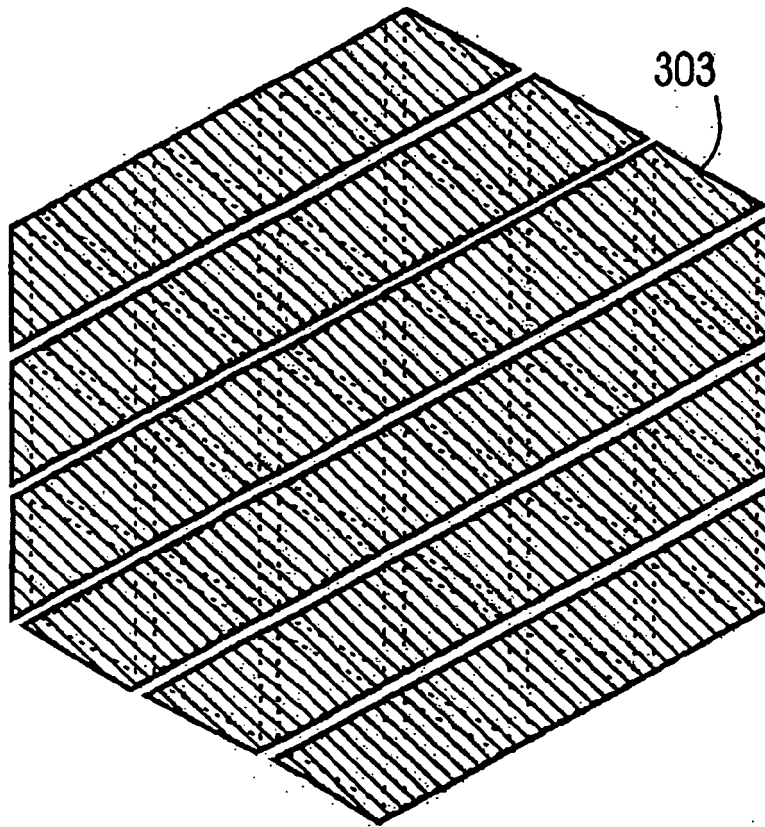


Figure 3b

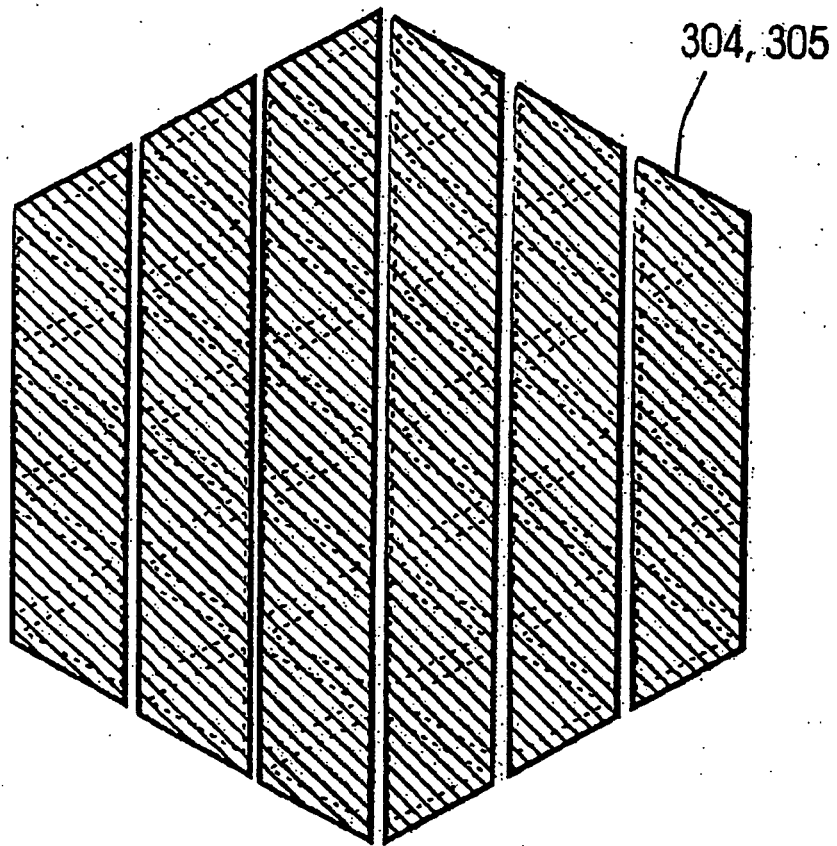


Figure 3c

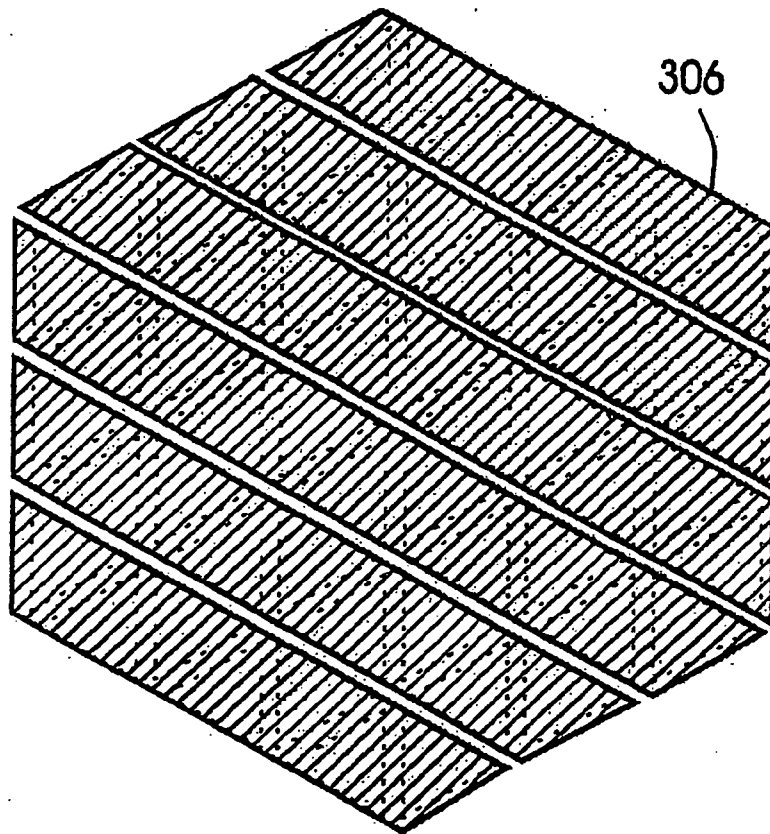


Figure 3d

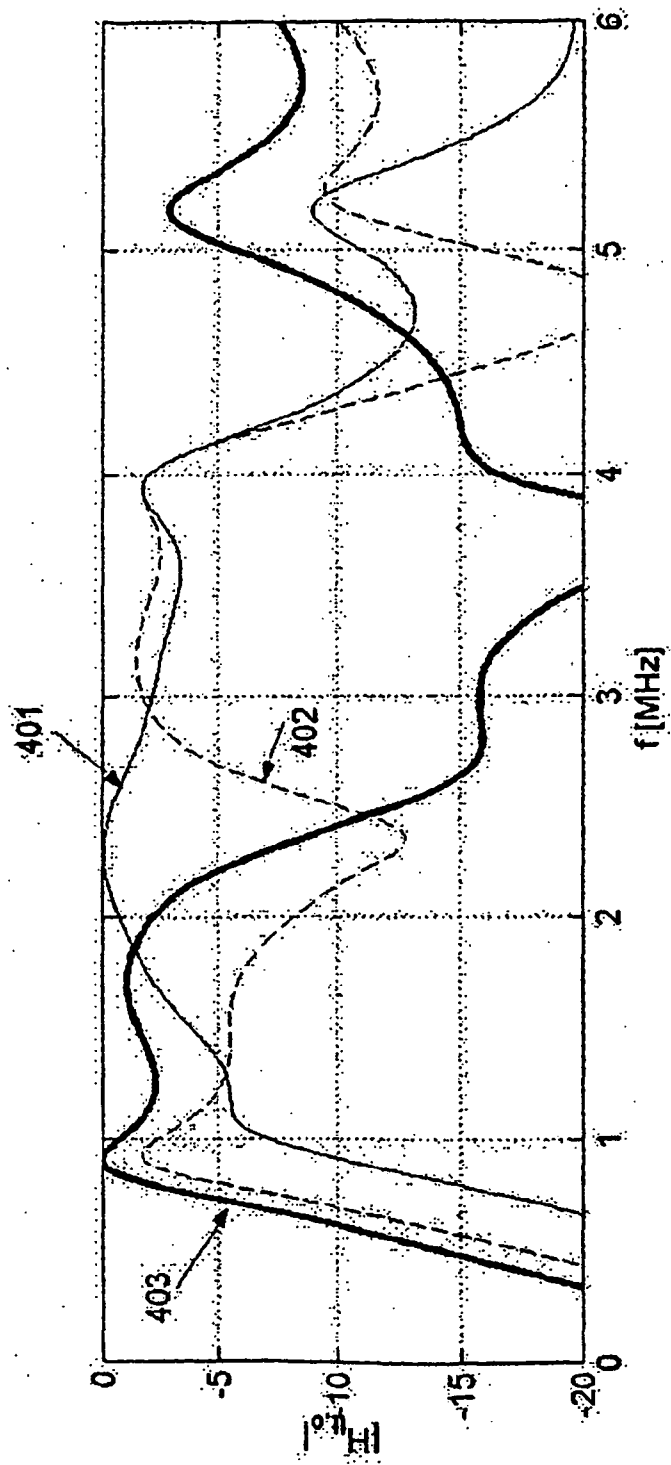


Figure 4

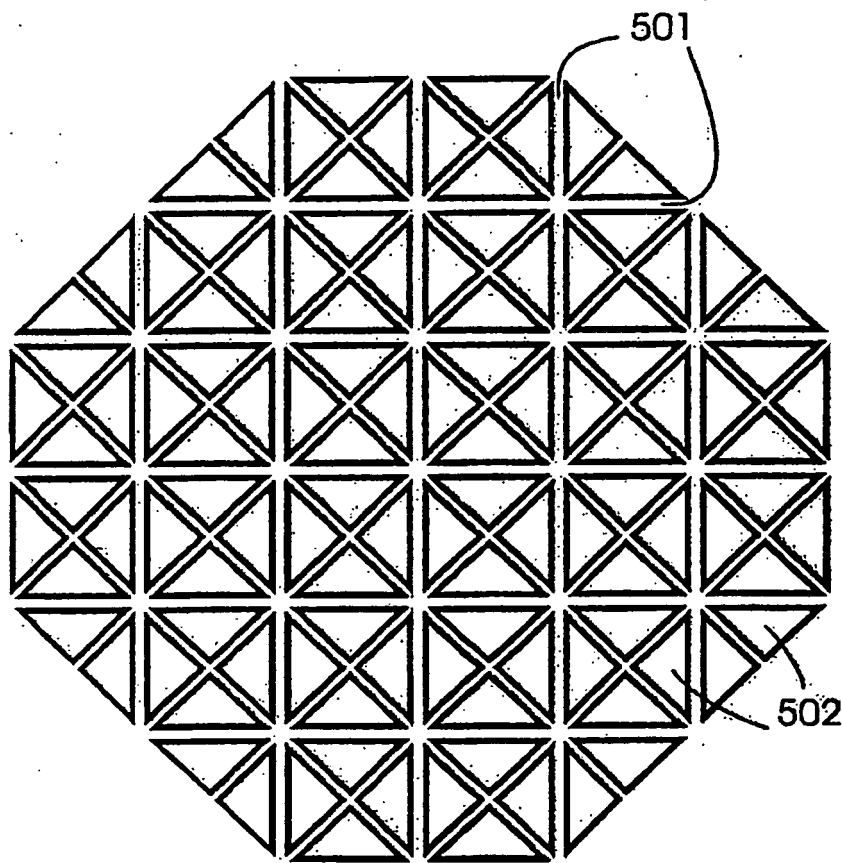


Figure 5a

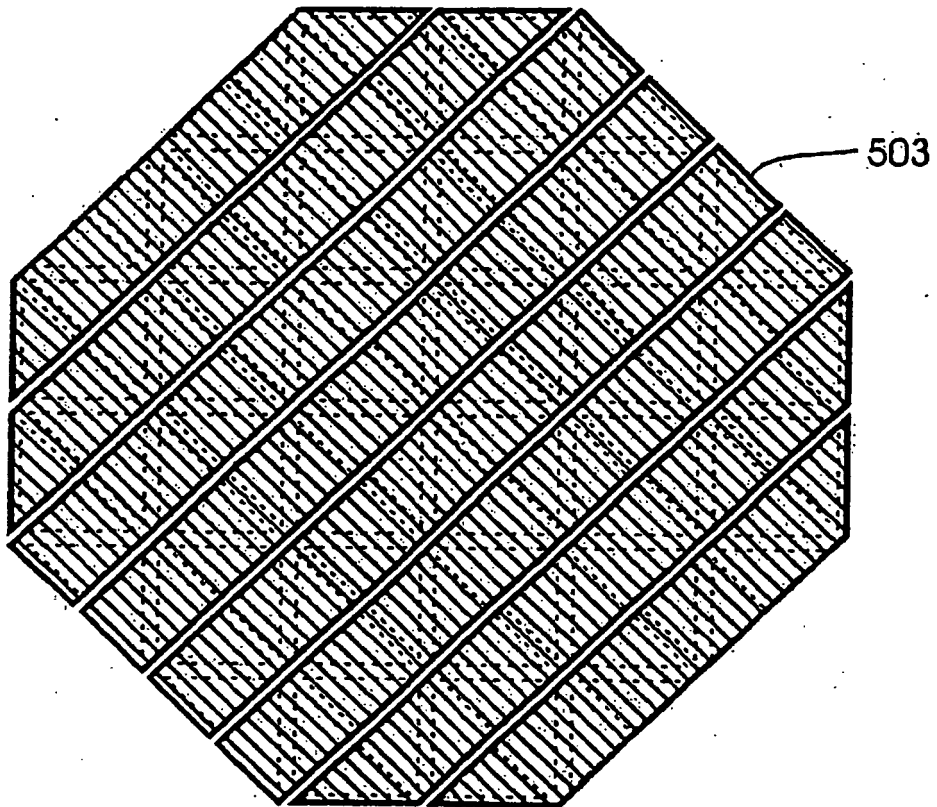


Figure 5b

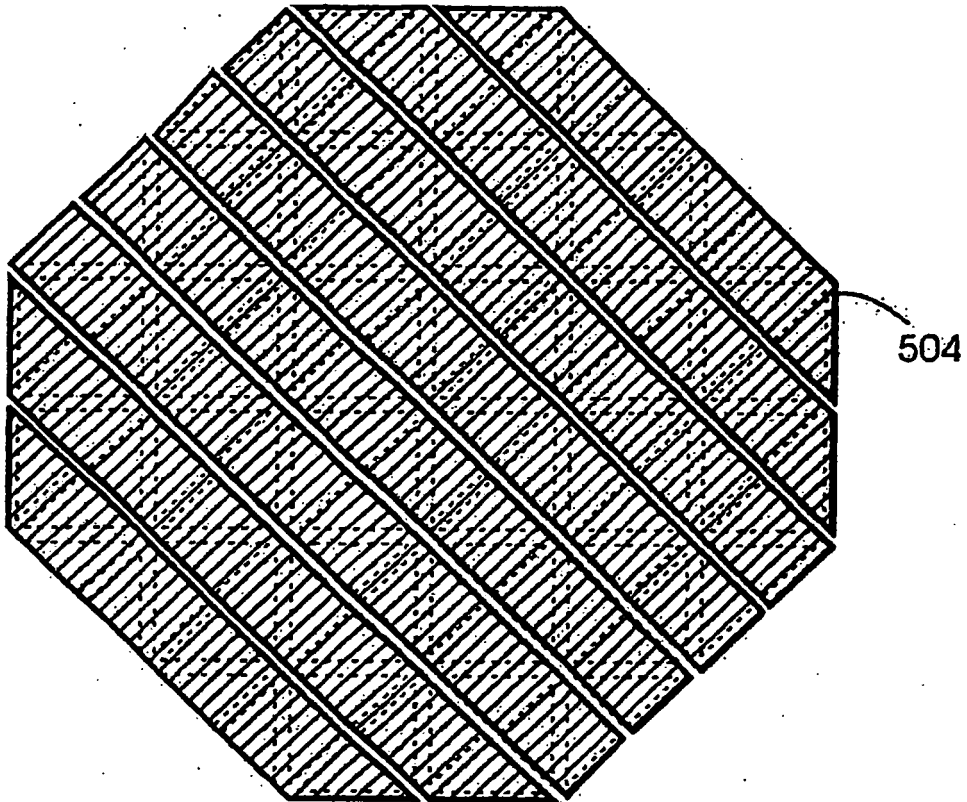


Figure 5c

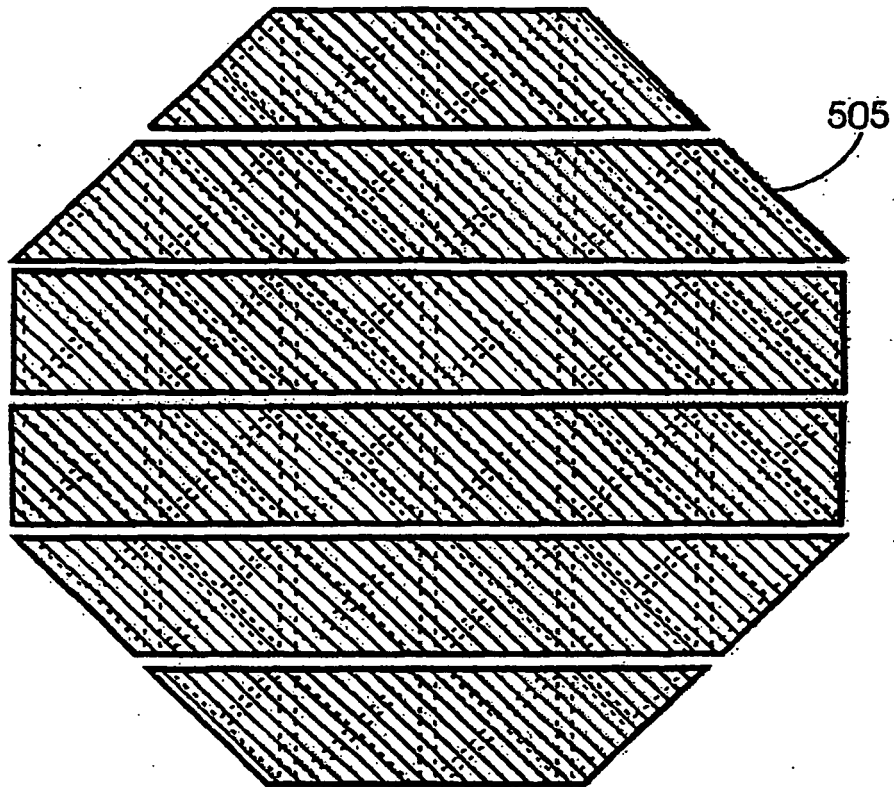


Figure 5d

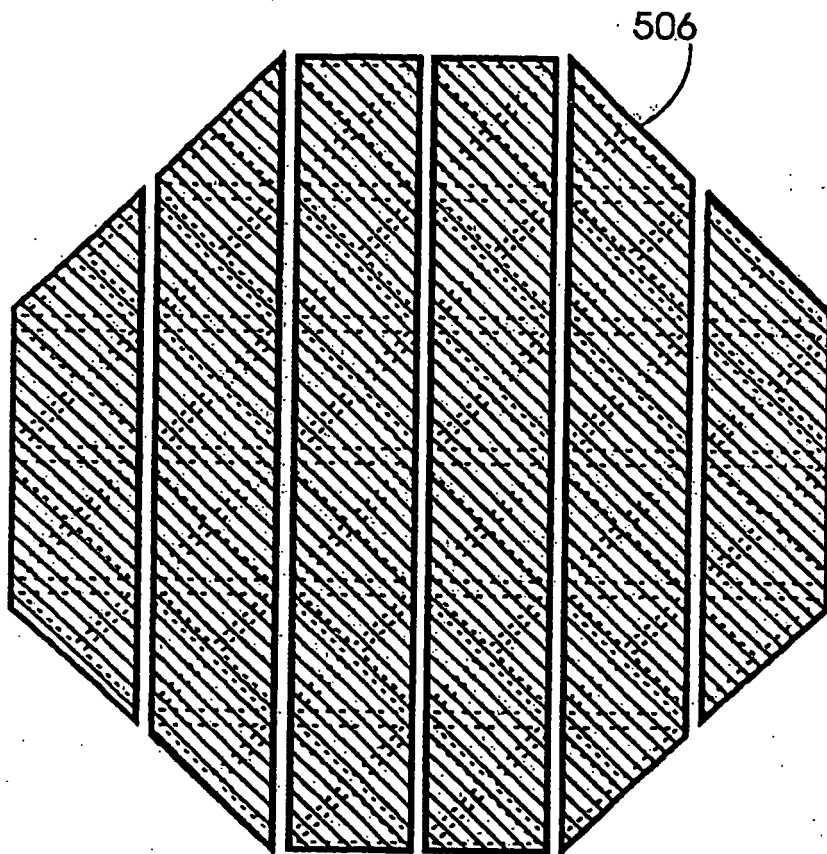


Figure 5e

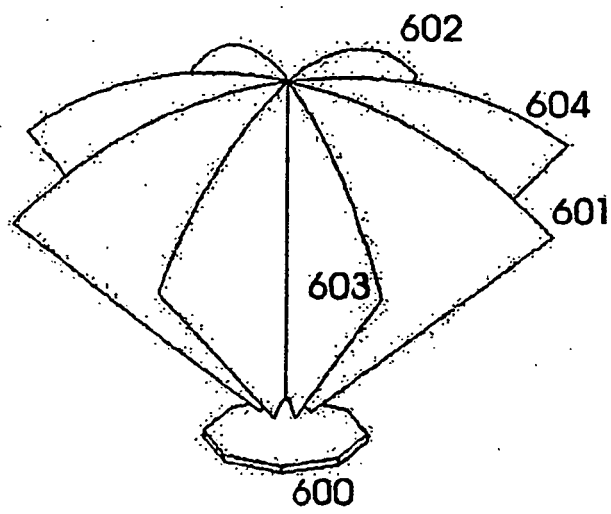


Figure 6

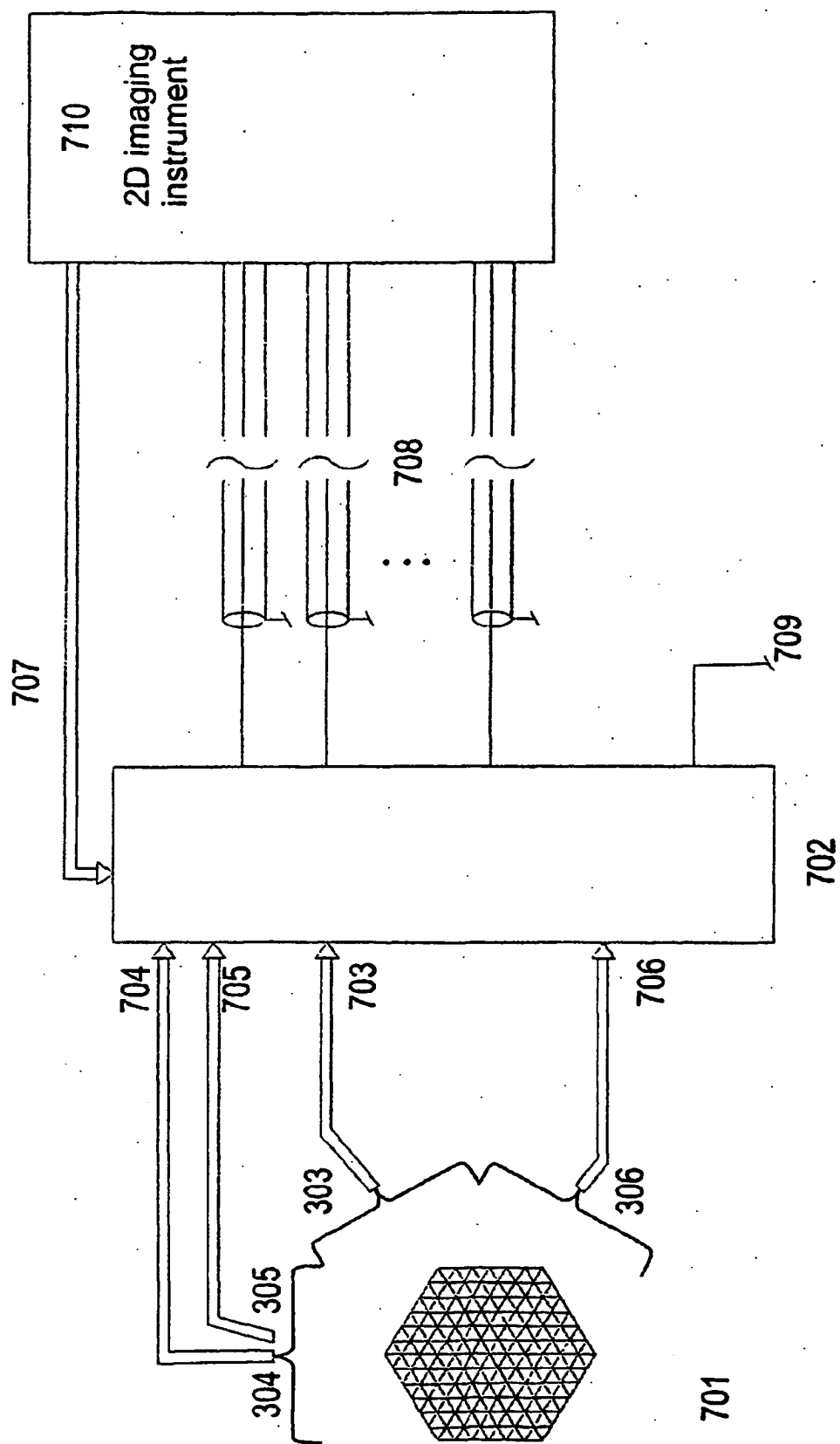


Figure 7

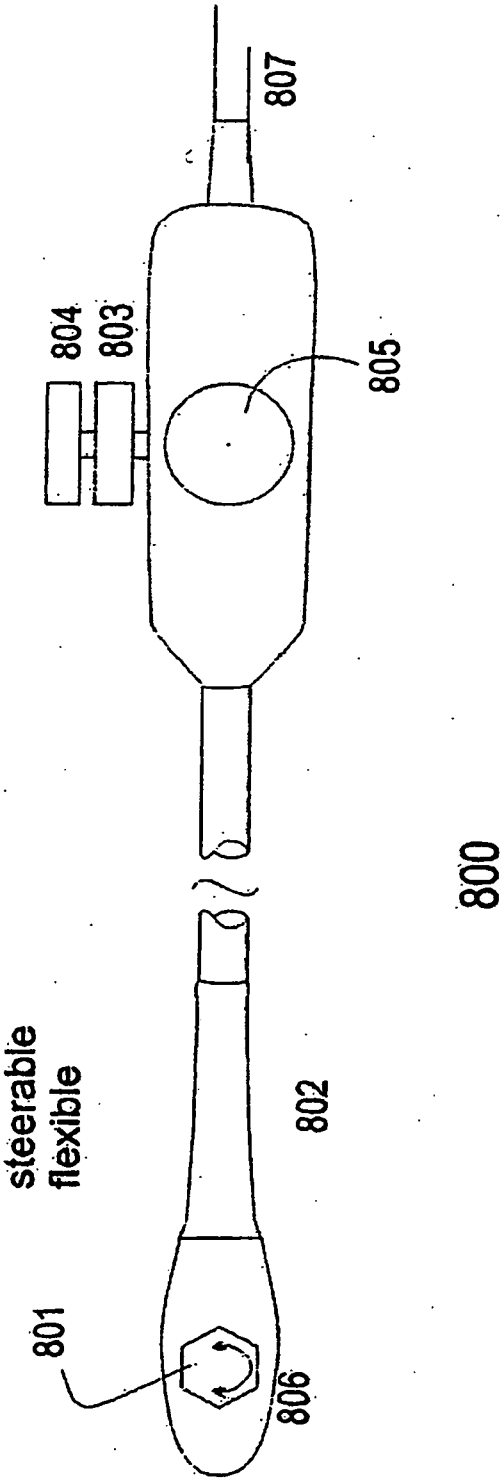


Figure 8

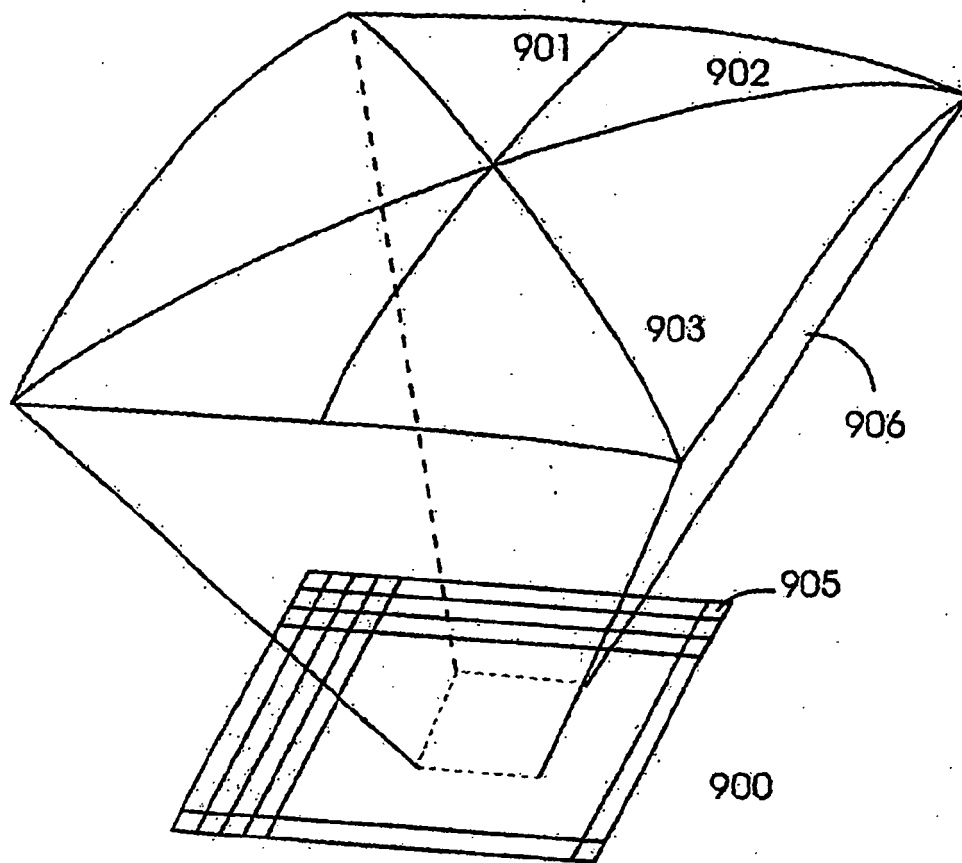


Figure 9

Text & other info

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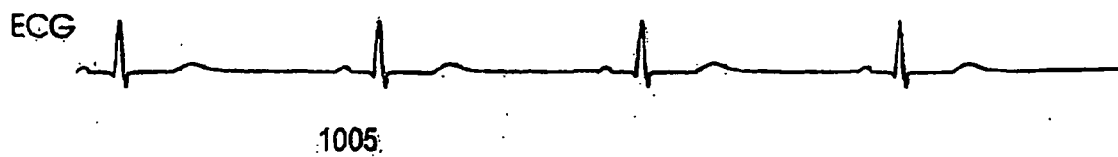
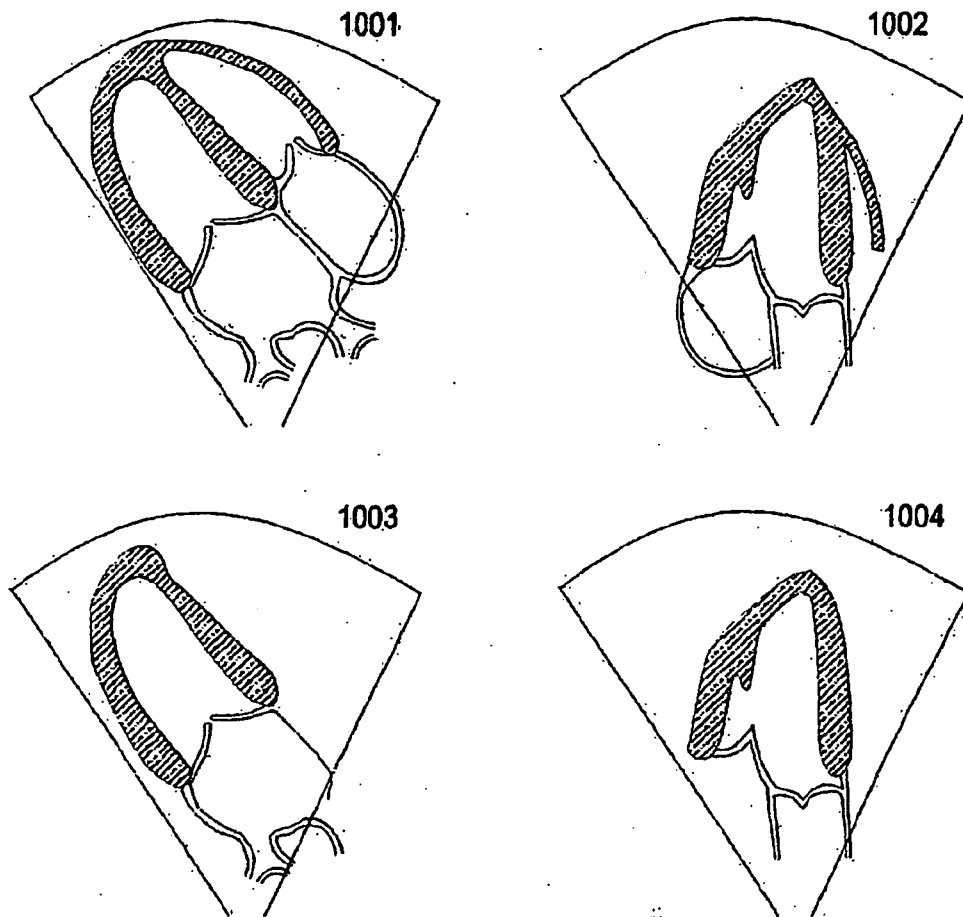


Figure 10a

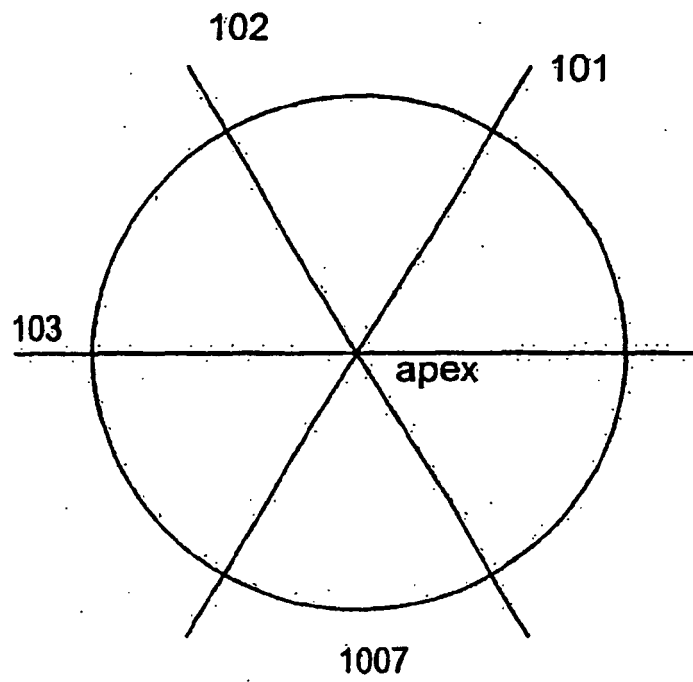


Figure 10b

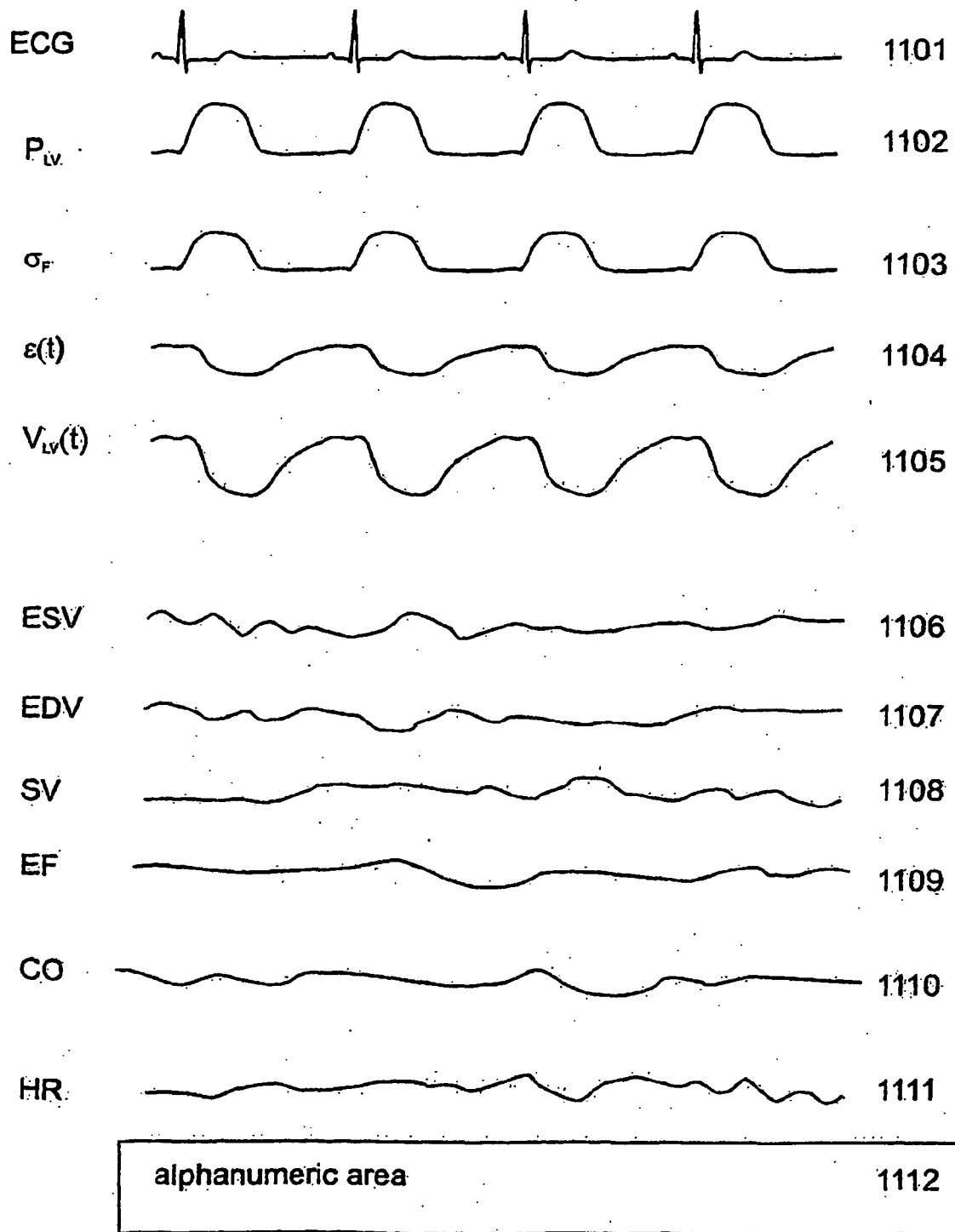


Figure 11

REFERENCES CITED IN THE DESCRIPTION

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专利名称(译)	物体的多个扫描平面超声成像		
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代理机构(译)	PROCK , THOMAS		
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摘要(译)

给出了围绕公共轴旋转的至少三个二维扫描平面中的对象的实时超声成像方法，以及允许这种成像的超声换能器阵列的设计。该方法还被引入到心脏功能的监测情况中，其中结合其他测量结果例如LV压力，计算诸如射血分数和肌肉纤维应力的生理参数。

$$\sigma_f(t) = \frac{H(t)}{(D(t)+H(t)) \left(3 - \left(\frac{D(t)+H(t)}{2L(t)} \right)^2 \right)} P_{LV}(t)$$