(19) Europäisches Patentamt
European Patent Office
Office européen des brevets

(11) EP 1 292 847 B1

(12) **EUROPEAN PATENT SPECIFICATION**

(45) Date of publication and mention of the grant of the patent:
17.04.2013 Bulletin 2013/16

(21) Application number: **01933918.3**

(22) Date of filing: **02.05.2001**

(51) Int Cl.:
G01S 15/89 (2006.01) ***G01S 7/52*** (2006.01)

(86) International application number:
PCT/EP2001/004951

(87) International publication number:
WO 2001/090776 (29.11.2001 Gazette 2001/48)

(54) **ULTRASONIC SPATIAL COMPOUNDING WITH CURVED ARRAY SCANHEADS**

RÄUMLICHE KOMPOUNDIERUNG MIT VERWENDUNG VON BOGENFÖRMIGER ANORDNUNG VON ULTRASSCHALLKÖPFEN

MISE EN COMPOSITION SPATIALE ULTRASONORE PAR DES TETES DE BALAYAGE A RESEAU INCURVE

(84) Designated Contracting States:
AT BE CH CY DE DK ES FI FR GB GR IE IT LI LU MC NL PT SE TR

(30) Priority: **23.05.2000 US 577021**

(43) Date of publication of application:
19.03.2003 Bulletin 2003/12

(73) Proprietor: **Koninklijke Philips Electronics N.V.**
5621 BA Eindhoven (NL)

(72) Inventor: **JAGO, James, R.**
NL-5656 AA Eindhoven (NL)

(74) Representative: **Roche, Denis et al**
Philips IP&S France
Société Civile SPID
33 rue de Verdun
BP 313
92156 Suresnes Cedex (FR)

(56) References cited:
WO-A-00/20884 **WO-A1-00/20890**
US-A- 5 538 004 **US-A- 5 653 235**
US-A- 6 057 632

- **ENTREKIN R ET AL: "Real time spatial compound imaging in breast ultrasound: technology and early clinical experience" MEDICAMUNDI, PHILIPS MEDICAL SYSTEMS, SHELTON, CT,, US, vol. 43, no. 3, pages 35-43, XP002181821 ISSN: 0025-7664**

EP 1 292 847 B1

Printed by Jouve, 75001 PARIS (FR)

Description

[0001] This invention relates to ultrasonic diagnostic imaging systems and, in particular, to ultrasonic diagnostic imaging systems which produce spatially compounded images of Doppler signal information.

[0002] U.S. patent 6,210,328 describes apparatus and methods for performing real time spatial compounding of ultrasonic diagnostic images. Spatial compounding is an imaging technique in which a number of ultrasound images of a given target that have been obtained from multiple vantage points or angles are combined into a single compounded image by combining the data (e.g., by linearly or nonlinearly averaging or filtering) at each point in the compound image which has been received from each angle. The compounded image typically shows lower speckle and better specular reflector delineation than conventional ultrasound images from a single viewpoint.

[0003] In a constructed implementation an ultrasonic transducer scans a target from a number of different perspectives. For example, several sector images can be sequentially acquired by a phased array transducer, each with an apex located at a different point along the array. As a second example a steered linear array can be used to image the target with a sequence of groups of beams, each group steered at a different angle with respect to the axis of the array. In either case the received images are processed in the usual way by beamforming and detection and stored in a memory. To form the compound image the images are spatially aligned (if not already aligned by a common beam steering reference, for instance) by spatially correlating the image data. The common spatial locations in the images are then compounded by averaging or summing and the resultant compound image is displayed.

[0004] The improvements in image quality afforded by spatial compounding are a function of the number of echoes from different look directions which are compounded at the various points in the image field. As my aforementioned patent application describes, a scanhead array of finite size will produce an image region in which maximal spatial compounding occurs, referred to therein as the region of maximum image quality (RMIQ). The RMIQ is in turn a function of the number of look directions and the angular range traversed by the different look directions. It is desirable to have a RMIQ which is as large as possible with the RMIQ covering a substantial portion of the image field.

[0005] U.S. Patent 5,653, 235 describes a system for generating an ultrasound image of a region in an object with a two-dimensional array of transducer elements. The two-dimensional array produces a beam that illuminates the region at different orientations. The aperture of the two-dimensional array is at least substantially equal to the aperture generated by a linear array of transducer elements extending across the substantially shortest distance between two opposite edges on the two-dimensional array. The ultrasound power emitted from the transducer elements is not spatially uniform and multiple echoes reduce speckle in the image.

[0006] The article entitled "Real Time Spatial Compound Imaging in breast ultrasound: technology and early clinical experience" by R. Entrekin et al., published in MEDICAMUNDI, Philips Medical Systems, Shelton, CT, US, vol. 43, no. 3, pages 35-43, describes that compound imaging starts by acquiring multiple frames from different viewing angles. These overlapping frames are then combined to form a real-time compound image on the display.

[0007] The independent claims, which are appended to the description, define various aspects of the invention. The dependent claims define additional features, which may be used to advantage. In accordance with the principles of the present invention, spatial compounding is performed by use of a curved array scanhead.

[0008] In one embodiment the beams of the different look directions are steered in parallel groups, affording an ease in image scan conversion. In other words, in this embodiment, beams are steered in parallel beam paths.

[0009] In another embodiment the beams of the different look directions are radially steered by taking advantage of the array curvature, affording an ease in beamforming and image scan conversion, improved sampling uniformity, an improvement in the RMIQ size, and improved signal to noise ratio. In more detail, a first group of beams may be steered from different points along the face of the array transducer, whereby each beam of the first group exhibits a first angle with respect to the face of the array transducer at its respective point. A second group of beams is then steered from different points along the face of the array transducer, whereby each beams the second group exhibits a second angle with respect to the face of the array transducer at its respective point.

[0010] In an embodiment, the echoes of more than one scanline are beamformed with the same set of coefficient magnitude values. Groups of beams may be steered with uniform angular spacing. Beamforming may form echoes from multiple look directions at points along a scanline using apertures with centers which separate linearly as a function of depth and said apertures may increase with depth.

[0011] In an embodiment, echoes at different depths in the image field may be registered, wherein points at a common depth utilize the same registration coefficient values.

[0012] In the drawings:

Fig. 1 illustrates in block diagram form an ultrasonic diagnostic imaging system constructed in accordance with the principles of the present invention;

Fig. 2 illustrates in block diagram form a preferred implementation of the spatial compounding processor of Fig. 1;

Fig. 3 illustrates a first compound scanning format for a curved array transducer in accordance with the principles of the present invention;

Fig. 4 illustrates a second compound scanning format for a curved array transducer in accordance with the principles of the present invention;

Fig. 5a and 5b compare the beam angles used in the illustrated compound scanning formats of Figs. 3 and 4;

Fig. 6 illustrates the separation of apertures with image field depth; and

Fig. 7 illustrates the registration processing efficiency realized by use of the compound scanning format of Fig. 4.

[0013] Referring first to Fig. 1, an ultrasonic diagnostic imaging system constructed in accordance with the principles of the present invention is shown. A scanhead 12 including a curved array transducer 10 transmits beams at different angles over an image field denoted by the dashed rectangle and parallelograms. Three groups of scanlines are indicated in the drawing, labeled A, B, and C with each group being steered at a different angle relative to the image field. Within each group, each beam exhibits a different angles of incidence to a line tangential to the point of origin on the array face from which the beam center extends. These angles progressively increase or decrease (depending upon one's point of reference) from one edge of the group to the other. The transmission of the beams is controlled by a transmitter 14 which controls the phasing and time of actuation of each of the elements of the array transducer so as to transmit each beam from a predetermined origin along the array and at a predetermined angle. The echoes returned from along each scanline are received by the active elements of the array aperture for that beam, digitized as by analog to digital conversion, and coupled to a digital beamformer 16. Generally the active aperture of each beam will dynamically expand as echoes are received from increasing depths and/or transmitted to increasing depths by zone focusing. The digital beamformer delays and sums the echoes from the array elements to form a sequence of focused, coherent digital echo samples along each scanline. The transmitter 14 and beamformer 16 are operated under control of a system controller 18, which in turn is responsive to the settings of controls on a user interface 20 operated by the user of the ultrasound system. The system controller controls the transmitter to transmit the desired number of scanline groups at the desired angles, transmit energies and frequencies. The system controller also controls the digital beamformer to properly delay and combine the received echo signals for the apertures and image depths used.

[0014] The scanline echo signals are filtered by a programmable digital filter 22, which defmes the band of frequencies of interest. When imaging harmonic contrast agents or performing tissue harmonic imaging the passband of the filter 22 is set to pass harmonics of the transmit band. The filtered signals are then detected by a detector 24. In a preferred embodiment the filter and detector include multiple filters and detectors so that the received signals may be separated into multiple passbands, individually detected and recombined to reduce image speckle by frequency compounding. For B mode imaging the detector 24 will perform amplitude detection of the echo signal envelope. For Doppler imaging ensembles of echoes are assembled for each point in the image and are Doppler processed to estimate the Doppler shift or Doppler power intensity.

[0015] In accordance with the principles of the present invention the digital echo signals are processed by spatial compounding in a processor 30. The digital echo signals are initially pre-processed by a preprocessor 32. The preprocessor 32 can preweight the signal samples if desired with a weighting factor. The samples can be preweighted with a weighting factor that is a function of the number of component frames used to form a particular compound image. The pre-processor can also weight edge lines that are at the edge of one overlapping image so as to smooth the transitions where the number of samples or images which are compounded changes. The pre-processed signal samples may then undergo a resampling in a resampler 34. The resampler 34 can spatially realign the estimates of one component frame to be in registration with those of another component frame or to the pixels of the display space.

[0016] After resampling the image frames are compounded by a combiner 36. Combining may comprise summation, averaging, peak detection, or other combinational means. The samples being combined may also be weighted prior to combining in this step of the process. Finally, post-processing is performed by a post-processor 38. The post-processor normalizes the combined values to a display range of values. Post-processing can be most easily implemented by look-up tables and can simultaneously perform compression and mapping of the range of compounded values to a range of values suitable for display of the compounded image.

[0017] The compounding process may be performed in estimate data space or in display pixel space. In a preferred embodiment scan conversion is done following the compounding process by a scan converter 40. The compound images may be stored in a Cineloop memory 42 in either estimate or display pixel form. If stored in estimate form the images may be scan converted when replayed from the Cineloop memory for display. The scan converter and Cineloop memory may also be used to render three dimensional presentations of the spatially compounded images as described in U.S. Patents 5,485,842 and 5,860,924, or displays of an extended field of view by overlaying successively acquired, partially overlapping images in the lateral dimension. Following scan conversion the spatially compounded images are processed for display by a video processor 44 and displayed on an image display 50.

[0018] Fig. 2 illustrates a preferred implementation of the spatial compounding processor 30 of Fig. 1. The processor 30 is preferably implemented by one or more digital signal processors 60 which process the image data in various ways. The digital signal processors 60 can weight the received image data and can resample the image data to spatially align pixels from frame to frame, for instance. The digital signal processors 60 direct the processed image frames to a plurality of frame memories 62 which buffer the individual image frames. The number of image frames capable of being stored by the frame memories 62 is preferably at least equal to the maximum number of image frames to be compounded such as sixteen frames. As described in the previously referenced US patent 6,210,328, the digital signal processors are responsive to changes in system control parameters including image display depth, number of scanlines or line density, number of transmit focal zones, amount of deadtime per pulse repetition interval (PRI), number of transmissions per image line, depth of region of greatest compounding, clinical application, number of simultaneous modes, size of region of interest, mode of operation, and acquisition rate for determining the number of component frames to compound at a given point in time. The digital signal processors select component frames stored in the frame memories 62 for assembly as a compound image in accumulator memory 64. The compounded image formed in the accumulator memory 64 is weighted or mapped by a normalization circuit 66, then compressed to the desired number of display bits and, if desired, remapped by a lookup table (LUT) 68. The fully processed compounded image is then transmitted to the scan converter for formatting and display.

[0019] The digital signal processors 60 determine the number of frames which are to be compounded to improve image quality while still providing an acceptable realtime compound image frame rate. Increasing the number of component frames does not lead to a proportional or unlimited increase in the image quality of the compound image. There is, therefore, a practical maximum number of frames, each steered by a minimum angle, that can be usefully employed to improve image quality in spatial compound scanning. This number can vary widely depending on the transducer design and size of the active aperture, but can be as large as 16 component frames per compound image for an array with a large acceptance angle and small active apertures. The maximum useful number of frames will also depend on the mixture of speckle and anisotropic scatterers in the tissue of interest, and therefore on the clinical application.

[0020] Fig. 3 illustrates a first compound scanning format for a curved array transducer in accordance with the principles of the present invention. The curved array transducer is shown as a 1D array for ease of illustration, but may also be al.5D array, a 1.75D array, or a 2D array. The curved array transducer 10 consists of n element extending in an arc from element 10_1 on the left to element 10_n on the right. The arc may be tightly curved or gently curved, depending upon the desired application. In this embodiment the curved array transmits and receives groups of beams in three look directions. The group of beams consisting of beams A1 through AN are directed at an angle of $+\theta$ with respect to a line drawn normal to the center of the face of the array. The group of beams consisting of beams B1 through BN are directed at an angle of $-\theta$ with respect to the same line, and the group of beams consisting of beams C1 through CN are directed at an angle of 0° with respect to the same line. The number of beams N of each group may be the same or different from group to group, and may be greater, equal to, or less than the number of array elements n. The beams of each group are parallel and of increasing separation proceeding from the innermost beam of each group (B1 of group B, AN of group A, and C(N/2) of group C) by reason of the progressively changing angle of incidence of each beam with its origin on the face of the array. The time delay and weighting coefficients used in transmit and receive beamforming will thus progressively change from line to line across each group. This embodiment affords an ease in beamforming because the symmetry of the scanning format allows coefficients and weights to be reused for multiple lines. Beams AN and B1 can use the same beamforming weights and coefficients but in reverse order as their apertures are the mirror image of each other. Likewise, beams C1 and Cn are mirrored and can use the same coefficients and weights. In a constructed embodiment the mirroring relationship may mean that the same magnitude values may be used for the different scanlines but with a sign difference. This symmetrical advantage exists across each of groups A and B with respect to each other, and for the two halves of group C.

[0021] The compound scanning format of Fig. 3 is seen to have a RMIQ which is scanned by all three look directions in the roughly triangular region 70 in the center of the image field. This region is bounded by scanline B1 on the left, scanline AN on the right, and by the curved array 10 at the top. This triangular area will thus exhibit the greatest effects of spatial compounding.

[0022] Fig. 4 illustrates a second compound scanning format in accordance with the principles of the present invention. This embodiment uses look directions having a common orientation to their points of origin on the face of the array. The beams in Fig. 4 are grouped in three groups: A1 through AN including illustrated intermediate beams Ai, Aj, and Ak; B1 through BN including illustrated intermediate beams Bi, Bj, and Bk; and C1 through CN. Each scanline in the group A1 through AN has the same angle of incidence to its point of origin on the face of the array; in the illustrated example the angle is drawn to be approximately +30° to a line normal to the face of the array. Each scanline in the group B1 through BN exhibits a -30° degree angle of incidence with respect to the same reference line, and each scanline in the group C1 through CN exhibits a 0° angle of incidence to the reference line.

[0023] This second compound scanning format affords numerous advantages. One advantage is that the same beamforming time delay and weighting coefficients can be used for each beam of a given group as the beam apertures are translated across the array for different beams, given customary allowance for edge effects at the ends of the finite array. Each beam of group A is steered on transmit and reception at a +30° angle, for example. The symmetry of the groups provides an advantage even with edge effects; the coefficients and weights used for beam A1 can be mirrored and used for beam BN, for instance.

[0024] Another advantage is the wider field of view which is realized by this scanning format. This is evident from a comparison of the scanned image fields of Figs. 3 and 4.

[0025] Another advantage is improved signal to noise performance or, stated another way, the larger useful apertures as compared with the embodiment of Fig. 3. The most acute angle of incidence in the embodiment of Fig. 4 is the most acute look direction angle, which is 30°. In the embodiment of Fig. 3 these angles are made even more acute by the arc of the curved array, as illustrated in Fig. 5a. This drawing shows three beams or scanlines emanating from array element 10_1 at the edge of the array of Fig. 3. The three beams are steered to be parallel to the other beams of groups A, B, and C, respectively. As the drawing shows, beam A extends almost normal (12°) to the face of the element 10_1 and will therefore be relatively effectively steered and strongly received. But beam C has an angle of incidence of 42° to the normal and beam B has an angle of incidence of 72° to the normal, and will only contribute weakly to beam steering, focusing and reception. The steeper the angle of incidence, the poorer the reception of the beam echoes or, alternatively stated, the smaller the contribution of the received echo by that element to the beamformed signal and/or the poorer the signal to noise. This is compared to the beams A, B, and C for element 10_1 of Fig. 4, shown in Fig. 5a, where the steepest angle of incidence is 30°. Furthermore, these illustrations are only for the beam center. The angle of incidence will be even steeper at elements at one of the ends of the active transmit and receive apertures. The signal strength may be so weak or noisy at these high angles of incidence that elements will be excluded from the transmit or receive aperture, either because their contributions to beamforming are so minimal or noisy or both, thereby limiting the active aperture, particularly at shallow depths of field.

[0026] Another advantage of the embodiment of Fig. 4 which is illustrated in Figs. 5a and 5b is grating lobe performance. The same grating lobes are drawn for beam B in each of Figs. 5a and 5b, consisting of a main lobe 80 and side lobes 81 and 82. It is desirable that the grating lobes not extend into the image field in any significant way. As Fig. 5b shows, at an angle of 30° the grating lobes 80,81,82 do not extend into the image field in front of element 10_1. However beam B of Fig. 5a, with its 72° angle of incidence, brings grating lobe 81 into the image field, degrading the image performance. The embodiment of Fig. 4 is thus seen to have better grating lobe performance.

[0027] Another advantage of the embodiment of Fig. 4 is the greatly enlarged RMIQ where the image field is scanned at all of the look directions. The RMIQ 90 in Fig. 4 is seen to extend from the position of beam B1 on the left to the position of beam AN on the right. As the drawing illustrates, the RMIQ is not a small triangular area as in the case of Fig. 3, but is a much larger trapezoidal, almost rectangular area. The RMIQ can be made even larger by more tightly curving the array or by the use of shallower look direction angles.

[0028] Another advantage of the embodiment of Fig. 4 is more uniform spatial sampling. As explained above, the scanline density is highest in the center of the image field of Fig. 3 and decreases as the beams and scanlines become more widely separated toward the edges of the field in the case where beam origins are uniformly spaced across the face of the array. In the embodiment of Fig. 4, uniform spacing of the beam origins results in uniform separation from scanline to scanline across each scanline group, as illustrated by the angular spacing of beams Ai,Aj,Ak, Bi,Bj,Bk, and Ci,Cj,Ck. While the beams diverge in the far field due to the radial nature of the scan, they are uniformly angularly spaced and hence provide uniform spatial sampling of the image field.

[0029] Another advantage of the embodiment of Fig. 4 is that the apertures of the beams separate uniformly as a function of depth, providing better compounding and more uniform speckle reduction over the depth of field, as illustrated by Fig. 6. This drawing shows a scanline SL from along which echoes are received from different look directions by a curved array transducer 10. Close to the transducer array at depth D_1 along the scanline a point is interrogated from three different look directions by beams b_1, b_2, and b_3. Beam b_1 is received from point D_1 by elements of an aperture A_1, beam b_2 is received from point D_1 by an aperture A_2, and beam b_3 is received from point D_1 by an aperture A_3. Points at greater depths along scanline SL are also interrogated by beams from multiple look directions, and the receive apertures of the beams grow (*i.e.*, include more elements) as echoes are received from increasing depths. Thus, beam b_4 of larger aperture A_4 and beam b_6 of larger aperture A_6 interrogate deeper point D_2. Beam b_2 can also be used to interrogate point D_2, but echoes from this greater depth along the scanline SL will be received by a larger aperture A_5.

[0030] It is seen in this example that near field point D_1 is interrogated by beams from three relatively small apertures A_1, A_2, and A_3, which grow to larger sizes A_4, A_5, and A_6 as deeper points such as D_2 are interrogated. The aperture centers of the interrogating beams separate uniformly with depth as deeper points D_n are interrogated, and this uniformity provides a uniformity over the image of the compounding effect and the resulting speckle reduction. This can be show mathematically starting with the expression

$$C = R\left(\phi_B - \sin^{-1}\left(\frac{R\sin\phi_B}{R+r}\right)\right)$$

where C is the separation of the aperture centers along the face of the array, R is the radius of curvature of the curved array, ϕ_B is the beam steering angle in radians relative to its point of intersection at the face of the array, and r is the range from the array to the point on the scanline being interrogated. For relatively small beam steering angles

$$C \cong R\left(\phi_B - \left(\frac{R\sin\phi_B}{R+r}\right)\right),$$

and therefore

$$C \cong R\phi_B - \frac{R^2\sin\phi_B}{R+r}$$

$$C \cong R^2\phi_B + Rr\phi_B - R^2\phi_B$$

and

$$C \cong Rr\phi_B$$

which shows that the aperture separation C is linearly related to depth r. This linear relationship means that more uniform speckle reduction will result from this implementation of the present invention.

[0031] A benefit which is realized by both curved array scanning embodiments is the use of the same sequence of registration coefficients for each scanline of each group, as illustrated in Fig. 7. US patent 6,135,956 describes a two stage scan conversion process for spatial compounding by which echoes from the different look directions are first registered and then compounded in estimate space, then scan converted to the desired image format in display space. An advantage of this technique is that no reprogramming of the scan converter is needed when switching between the spatial compounding mode of operation and uncompounded imaging. From the perspective of the scan converter, the incoming image data appears the same and is processed the same way. Since no reprogramming of the scan converter is needed, the user can switch between modes more rapidly. This two stage scan conversion process may be used with all of the embodiments of the present invention. Fig. 7 illustrates registration in estimate space, the first step in the two stage process, for the compound scanning technique of Fig. 4. In this illustration the scanlines of group C (C1,C2,C3,...CN) which are directed normal to the points of origin on the curved array have been acquired and are sampled by a digital beamformer at the circled points of equal range (time) increments. Scanlines of group A (A1,A2,A3,...AN) are acquired at a common angle of incidence across the image field and sampled at the same equal range increments as shown by Xs. The samples (Xs) of group A are now to be registered with the samples (Os) of group C. A preferred way to do this is four point interpolation. For example, the C2 sample 102 is within a quadrilateral area delineated by the four A samples 202, 204, 206, and 208. An A sample at the location of C sample 102 may be interpolated from the four acquired A samples by four point or up-down, right-left interpolation, by which the four A samples are combined in proportion to their proximity to the location of the C sample 102. This proportionate combining is achieved by weighting the A samples, then combining them.

[0032] Moving down the C2 scanline, an A sample is interpolated at the location 106 of the next C sample on the scanline. Sample location 106 is bounded by the four A samples at 202, 204, 214 and 216, which are used to interpolate an A sample at location 106. Because the relationship of all of the locations has changed from the location of sample 102 to that of location 106, a new set of four weighting factors is used to interpolate the A sample at location 106. This process continues down the scanline, each time using an appropriate set (location proportionate) of weighting factors for the interpolation.

[0033] However, when the registration process is done for the next scanline, the sequence of weighting factors used for the previous scanline can be reused. This is due to the recurring spatial relationships of the samples across the image field. For instance, the weights applied to A samples 208, 210, 204, and 212 to interpolate an A sample at location 108 of scanline C3 have the same values as the weights previously used to interpolate an A sample at location 102 of the previous scanline C2. The interpolation coefficients for registration are depth-dependent but not line-dependent, and hence can be the same from line to line. Thus, the same set of interpolation weights can be used for registration of each scanline, obviating the need to recalculate weighting factors for each scanline.

[0034] A computer program product comprising a set of instructions may be used for the ultrasonic diagnostic imaging system, to produce spatially compounded ultrasonic images with beams transmitted in multiple look directions from a curved array scanhead as above-described.

Claims

1. An ultrasonic diagnostic imaging system which produces spatially compounded ultrasonic images comprising:

   - an array transducer (10);
   - a transmitter (14), coupled to said array transducer, that drives said array transducer to produce electronically steered beams;
   - a beamformer (16), coupled to said array transducer, which forms coherent echo signals in response to said electronically steered beams;
   - a signal processor (30), coupled to said beamformer, which spatially registers echoes of multiple look directions by means of interpolation and then compounds the spatially registered echoes to form spatially compounded image signals; and
   - a display device (50), responsive to said compounded image signals, which displays a spatially compounded image

   characterized in that

   - the array transducer is a curved array transducer

   and **in that**

   - the transmitter (14) is coupled to said curved array transducer to drive said curved array transducer to produce electronically steered beams (A, B, C), each having a different angle of incidence to a line tangential to a point of origin on the face of the curved array transducer from which the center of the beam extends so as to interrogate points in an image field from multiple look directions ($-\theta$, 0, $+\theta$).

2. The ultrasonic diagnostic imaging system of Claim 1, wherein said curved array transducer (10) transmits a first group of beams (A1-AN) in a first look direction ($+\theta$), a second group of beams (B1-BN) in a second look direction (0), and a third group of beams (C1-CN) in a third look direction ($-\theta$).

3. The ultrasonic diagnostic imaging system of Claim 2, wherein the beams of each group (A1-AN, B1-BN, C1-CN) are substantially parallel.

4. The ultrasonic diagnostic imaging system of Claim 2, wherein the beams of each group (A1-AN, B1-BN, C1-CN) exhibit a predetermined steering angle ($-\theta$, 0, $+\theta$) with respect to their points of origin on the face of the curved array transducer (10).

5. The ultrasonic diagnostic imaging system of Claim 3, wherein the first group of beams (A1-AN) are steered at an angle of θ_1 relative to a line extending normal to the face of the curved transducer array (10), the second group of beams (B1-BN) are steered at an angle of θ_2 relative to said line extending normal to the face of the curved transducer array, and the third group of beams (C1-CN) are steered at an angle of θ_3 relative to said line extending normal to the face of the curved transducer array.

6. The ultrasonic diagnostic imaging system of Claim 5, wherein angle θ_2 is equal to $-\theta_1$.

7. The ultrasonic diagnostic imaging system of Claim 6, wherein θ_3 is equal to zero degrees.

8. The ultrasonic diagnostic imaging system of Claim 7, wherein said beamformer (16) utilizes coefficients to form the coherent echo signals of a scanline, and wherein the same set of coefficient values is utilized to form the coherent echo signals of more than one scanline.

9. The ultrasonic diagnostic imaging system of Claim 4, wherein the first group of beams (A1-AN) are each steered at an angle of θ_1 relative to a line extending normal to the face of the curved transducer array (10) at the origin of a beam, the second group of beams (B1-BN) are steered at an angle of θ_2 relative to said line extending normal to the face of the curved transducer array at the origin of a beam, and the third group of beams (C1-CN) are steered at an angle of θ_3 relative to said line extending normal to the face of the curved transducer array at the origin of a beam.

10. The ultrasonic diagnostic imaging system of Claim 2, wherein said beamformer (16) utilizes coefficients to form the coherent echo signals of a scanline, and wherein the same set of coefficient magnitude values is utilized to form the coherent echo signals of each of the scanlines of a group of scanlines formed in response to a group of beams (A1-AN, B1-BN, C1-CN).

11. The ultrasonic diagnostic imaging system of Claim 2, wherein each of the beams of a group (A1-AN, B1-BN, C1-CN) exhibits substantially the same angle of incidence with the face of said curved array transducer (10).

12. The ultrasonic diagnostic imaging system of Claim 2, wherein the beams of each group (A1-AN, B1-BN, C1-CN) are uniformly angularly spaced.

13. The ultrasonic diagnostic imaging system of Claim 1, wherein said beamformer (16) forms echo signals with apertures of said curved array transducer (10) along scanlines (SL), and wherein the centers of apertures (A1, ...A6) used to interrogate points (D1, D2) on said scanlines separate linearly as a function of the distance between the point of interest and the curved array transducer.

14. The ultrasonic diagnostic imaging system of Claim 13, wherein said apertures (A1, ...A6) increase in size as a function of depth.

15. The ultrasonic diagnostic imaging system of Claim 1, wherein said signal processor (30) uses coefficients to spatially register echoes of multiple look directions, and wherein said coefficients depend on the distance between a point of interest and the curved array transducer but do not depend on the scanline.

16. A method of ultrasonic imaging to produce spatially compounded ultrasonic images comprising:

- interrogating points in an image field from multiple look directions ($-\theta$, 0, $+\theta$) with beams (A, B, C) steered from a curved array transducer (10), said beams being steered by a combination of the curvature of said curved array transducer and electronic beam steering, whereby each beam has a different angle of incidence to a line tangential to a point of origin on the face of the curved array transducer from which the center of the beam extends;
- beamforming echoes received from said points from multiple look directions;
- spatially registering echoes of multiple look directions by means of interpolation and then compounding the spatially registered echoes to form spatially compounded image signals; and
- displaying a spatially compounded image in response to said spatially compounded image signals.

17. Computer program product comprising a set of instructions for using the ultrasonic diagnostic imaging system as claimed in one of Claims 1 to 15, to produce spatially compounded ultrasonic images with beams transmitted in multiple look directions from a curved array scanhead.

Patentansprüche

1. Diagnostisches Ultraschall-Bildgebungssystem, das räumlich zusammengesetzte Ultraschallbilder erzeugt (engl. Spatial Compounding) und Folgendes umfasst:

- einen Array-Wandler (10);
- einen mit dem genannten Array-Wandler gekoppelten Sender (14), der den genannten Array-Wandler zum Erzeugen von elektronisch gelenkten Strahlenbündeln ansteuert;
- einen mit dem genannten Array-Wandler gekoppelten Strahlformer (16), der in Reaktion auf die genannten elektronisch gelenkten Strahlenbündel kohärente Echosignale formt;
- einen mit dem genannten Strahlformer gekoppelten Signalprozessor (30), der Echos aus mehreren Blickrichtungen mithilfe von Interpolation räumlich registriert und anschließend die räumlich registrierten Echos zusammensetzt, um räumlich zusammengesetzte Bildsignale zu formen; und
- eine auf die genannten zusammengesetzten Bildsignale reagierende Anzeigevorrichtung (50), die ein räumlich zusammengesetztes Bild anzeigt, **dadurch gekennzeichnet, dass**
- der Array-Wandler ein Konvexwandler ist und dass
- der Sender (14) mit dem genannten Konvexwandler gekoppelt ist, um den genannten Konvexwandler zum Erzeugen elektronisch gelenkter Strahlenbündel (A, B, C) anzusteuern, die jeweils einen anderen Einfallswinkel auf eine Linie tangential zu einem Ursprungspunkt auf der Stirnfläche des Konvexwandlers haben, von dem ausgehend sich die Mitte des Strahlenbündels erstreckt, um Punkte in einem Bildfeld aus mehreren Blickrichtungen (-•, 0, +•) abzufragen.

2. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 1, wobei der Konvexwandler (10) eine erste Gruppe von Strahlenbündeln (A1-AN) in einer ersten Blickrichtung (+•), eine zweite Gruppe von Strahlenbündeln (B1-BN) in einer zweiten Blickrichtung (0) und eine dritte Gruppe von Strahlenbündeln (C1-CN) in einer dritten Blickrichtung (-•) aussendet.

3. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 2, wobei die Strahlenbündel jeder Gruppe (A1-AN, B1-BN, C1-CN) im Wesentlichen parallel sind.

4. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 2, wobei die Strahlenbündel jeder Gruppe (A1-AN, B1-BN, C1-CN) einen vorgegebenen Lenkungswinkel (-• , 0, +•) in Bezug auf ihre Ursprungspunkte auf der Stirnfläche des Konvexwandlers (10) aufweisen.

5. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 3, wobei die erste Gruppe von Strahlenbündeln (A1-AN) in einem Winkel $\bullet_1$ relativ zu einer Linie gelenkt wird, die senkrecht zur Stirnfläche des Konvexwandlers (10) verläuft, die zweite Gruppe von Strahlenbündeln (B1-BN) in einem Winkel $\bullet_2$ relativ zu der genannten Linie gelenkt wird, die senkrecht zur Stirnfläche des Konvexwandlers verläuft, und die dritte Gruppe von Strahlenbündeln (C1-CN) in einem Winkel $\bullet_3$ relativ zu der genannten Linie gelenkt wird, die senkrecht zur Stirnfläche des Konvexwandlers verläuft.

6. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 5, wobei der Winkel •$_2$ gleich -•$_1$ ist.

7. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 6, wobei •$_3$ gleich null Grad ist.

8. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 7, wobei der genannte Strahlformer (16) Koeffizienten nutzt, um die kohärenten Echosignale einer Scanlinie zu formen, und wobei der gleiche Satz von Koeffizientenwerten genutzt wird, um die kohärenten Echosignale von mehr als einer Scanlinie zu formen.

9. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 4, wobei die erste Gruppe von Strahlenbündeln (A1-AN) jeweils in einem Winkel •$_1$ relativ zu einer Linie gelenkt wird, die senkrecht zur Stirnfläche des Konvexwandlers (10) am Ursprung eines Strahlenbündels verläuft, die zweite Gruppe von Strahlenbündeln (B1-BN) in einem Winkel •$_2$ relativ zu der genannten Linie gelenkt wird, die senkrecht zur Stirnfläche des Konvexwandlers am Ursprung eines Strahlenbündels verläuft und die dritte Gruppe von Strahlenbündeln (C1-CN) in einem Winkel •$_3$ relativ zu der genannten Linie gelenkt wird, die senkrecht zur Stirnfläche des Konvexwandlers am Ursprung des Strahlenbündels verläuft.

10. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 2, wobei der genannte Strahlformer (16) Koeffizienten nutzt, um die kohärenten Echosignale einer Scanlinie zu formen, und wobei der gleiche Satz von Koeffizientenmagnitudenwerten genutzt wird, um die kohärenten Echosignale von jeder der Scanlinien einer Gruppe von Scanlinien zu formen, die in Reaktion auf eine Gruppe von Strahlenbündeln (A1-AN, B1-BN, C1-CN) geformt wurden.

11. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 2, wobei jedes der Strahlenbündel einer Gruppe (A1-AN, B1-BN, C1-CN) im Wesentlichen den gleichen Einfallswinkel zu der Stirnfläche des genannten Konvexwandlers (10) aufweist.

12. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 2, wobei die Strahlenbündel jeder Gruppe (A1-AN, B1-BN, C1-CN) einen gleichmäßigen Winkelabstand zueinander haben.

13. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 1, wobei der genannte Strahlformer (16) Echosignale mit Aperturen des genannten Konvexwandlers (10) entlang Scanlinien (SL) formt, und wobei die Mittelpunkte von Aperturen (A1, ..., A6) genutzt werden, um Punkte (D1, D2) auf den genannten Scanlinien linear separat als eine Funktion des Abstands zwischen dem interessierenden Punkt und dem Konvexwandler abzufragen.

14. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 13, wobei die genannten Aperturen (A1, ..., A6) in der Größe als eine Funktion der Tiefe zunehmen.

15. Diagnostisches Ultraschall-Bildgebungssystem nach Anspruch 1, wobei der genannte Signalprozessor (30) Koeffizienten nutzt, um Echos aus mehreren Blickrichtungen räumlich zu registrieren, und wobei die genannten Koeffizienten von dem Abstand zwischen einem interessierenden Punkt und dem Konvexwandler abhängen, aber nicht von der Scanlinie.

16. Verfahren der Ultraschall-Bildgebung zum Erzeugen von räumlich zusammengesetzten Ultraschallbildern (Spatial Compounding), wobei das Verfahren Folgendes umfasst:

- Abfragen von Punkten in einem Bildfeld aus mehreren Blickrichtungen (-•, 0, +•) mit von einem Konvexwandler (10) aus gelenkten Strahlenbündeln (A, B, C), wobei die genannten Strahlenbündel durch eine Kombination der Krümmung des genannten Konvexwandlers und der elektronischen Strahlenbündellenkung gelenkt werden, wobei jedes Strahlenbündel einen anderen Einfallswinkel auf eine Linie tangential zu einem Ursprungspunkt auf der Stirnfläche des Konvexwandlers hat, von dem ausgehend sich die Mitte des Strahlenbündels erstreckt;
- Strahlenformen der von den genannten Punkten aus mehreren Blickrichtungen empfangenen Echos;
- räumliches Registrieren der Echos aus mehreren Blickrichtungen mithilfe von Interpolation und anschließendes Zusammensetzen der räumlich registrierten Echos, um räumlich zusammengesetzte Bildsignale zu formen; und
- Anzeigen eines räumlich zusammengesetzten Bildes in Reaktion auf die genannten räumlich zusammengesetzten Bildsignale.

17. Computerprogrammprodukt mit einem Satz von Anweisungen zur Nutzung des diagnostischen Ultraschall-Bildgebungssystems nach einem der Ansprüche 1 bis 15 zum Erzeugen von räumlich zusammengesetzten Ultraschallsignalen mit von einem Konvexwandler in mehreren Blickrichtungen gesendeten Strahlenbündeln.

Revendications

1. Système d'imagerie diagnostique ultrasonique qui produit des images ultrasoniques spatialement composées, comprenant :

- un transducteur en réseau (10) ;
- un transmetteur (14), couplé audit transducteur en réseau, qui excite ledit transducteur en réseau pour produire des faisceaux orientés électroniquement ;
- un formeur de faisceaux (16), couplé audit transducteur en réseau, qui forme des signaux d'échos cohérents en réponse auxdits faisceaux orientés électroniquement ;
- un processeur de signaux (30), couplé audit formeur de faisceaux, qui enregistre spatialement des échos de multiples directions de visée au moyen d'interpolation et puis compose les échos spatialement enregistrés pour former des signaux d'images spatialement composées ; et
- un dispositif d'affichage (50), répondant auxdits signaux d'images composées, qui affiche une image spatialement composée,

caractérisé en ce que

- le transducteur en réseau est un transducteur en réseau incurvé,

et **en ce que**

- le transmetteur (14) est couplé audit transducteur en réseau incurvé pour exciter ledit transducteur en réseau incurvé pour produire des faisceaux orientés électroniquement (A, B, C), chacun possédant un angle différent d'incidence par rapport à une ligne tangentielle à un point d'origine sur la face du transducteur en réseau incurvé à partir duquel le centre du faisceau s'étend afin d'interroger des points dans un champ d'image à partir de multiples directions de visée (-•, 0, +•).

2. Système d'imagerie diagnostique ultrasonique selon la revendication 1, dans lequel ledit transducteur en réseau incurvé (10) transmet un premier groupe de faisceaux (A1-AN) dans une première direction de visée (+•), un deuxième groupe de faisceaux (B1-BN) dans une deuxième direction de visée (0), et un troisième groupe de faisceaux (C1-CN) dans une troisième direction de visée (-•).

3. Système d'imagerie diagnostique ultrasonique selon la revendication 2, dans lequel les faisceaux de chaque groupe (A1-AN, B1-BN, C1-CN) sont sensiblement parallèles.

4. Système d'imagerie diagnostique ultrasonique selon la revendication 2, dans lequel les faisceaux de chaque groupe (A1-AN, B1-BN, C1-CN) présentent un angle d'orientation prédéterminé (-•, 0, +•) par rapport à leurs points d'origine sur la face du transducteur en réseau incurvé (10).

5. Système d'imagerie diagnostique ultrasonique selon la revendication 3, dans lequel le premier groupe de faisceaux (A1-AN) est orienté à un angle de $\bullet_1$ par rapport à une ligne s'étendant de façon normale à la face du réseau transducteur incurvé (10), le deuxième groupe de faisceaux (B1-BN) est orienté à un angle de $\bullet_2$ par rapport à ladite ligne s'étendant de façon normale à la face du réseau transducteur incurvé, et le troisième groupe de faisceaux (C1-CN) est orienté à un angle de $\bullet_3$ par rapport à ladite ligne s'étendant de façon normale à la face du réseau transducteur incurvé.

6. Système d'imagerie diagnostique ultrasonique selon la revendication 5, dans lequel l' angle $\bullet_2$ est égal à -• 1.

7. Système d'imagerie diagnostique ultrasonique selon la revendication 6, dans lequel $\bullet_3$ est égal à zéro degré.

8. Système d'imagerie diagnostique ultrasonique selon la revendication 7, dans lequel ledit formeur de faisceaux (16) utilise des coefficients pour former les signaux d'échos cohérents d'une ligne de balayage, et dans lequel le même jeu de valeurs de coefficient est utilisé pour former les signaux d'échos cohérents de plus d'une ligne de balayage.

9. Système d'imagerie diagnostique ultrasonique selon la revendication 4, dans lequel le premier groupe de faisceaux (A1-AN) est orienté à un angle de $\bullet_1$ par rapport à une ligne s'étendant de façon normale à la face du réseau transducteur incurvé (10) à l'origine d'un faisceau, le deuxième groupe de faisceaux (B1 -BN) est orienté à un angle de $\bullet_2$ par rapport à ladite ligne s'étendant de façon normale à la face du réseau transducteur incurvé à l'origine d'un faisceau, et le troisième groupe de faisceaux (C1-CN) est orienté à un angle de $\bullet_3$ par rapport à ladite ligne s'étendant de façon normale à la face du réseau transducteur incurvé à l'origine d'un faisceau.

10. Système d'imagerie diagnostique ultrasonique selon la revendication 2, dans lequel ledit formeur de faisceaux (16) utilise des coefficients pour former les signaux d'échos cohérents d'une ligne de balayage, et dans lequel le même jeu de valeurs d'amplitude de coefficient est utilisé pour former les signaux d'échos cohérents de chacune des lignes de balayage d'un groupe de lignes de balayage formé en réponse à un groupe de faisceaux (A1 -AN, B1-BN, C1-CN).

11. Système d'imagerie diagnostique ultrasonique selon la revendication 2, dans lequel chacun des faisceaux d'un groupe (A1-AN, B1-BN, C1-CN) présente sensiblement le même angle d'incidence avec la face dudit transducteur en réseau incurvé (10).

12. Système d'imagerie diagnostique ultrasonique selon la revendication 2, dans lequel les faisceaux de chaque groupe (A1-AN, B1-BN, C1-CN) sont espacés de façon uniformément angulaire.

13. Système d'imagerie diagnostique ultrasonique selon la revendication 1, dans lequel ledit formeur de faisceaux (16) forme des signaux d'échos avec des ouvertures dudit transducteur en réseau incurvé (10) le long de lignes de balayage (SL), et dans lequel les centres d'ouvertures (A1, ...A6) utilisés pour interroger des points (D1, D2) sur lesdites lignes de balayage se séparent linéairement en fonction de la distance entre le point d'intérêt et le transducteur en réseau incurvé.

14. Système d'imagerie diagnostique ultrasonique selon la revendication 13, dans lequel lesdites ouvertures (A1, ...A6) augmentent en taille en fonction de la profondeur.

15. Système d'imagerie diagnostique ultrasonique selon la revendication 1, dans lequel ledit processeur de signaux (30) utilise des coefficients pour enregistrer spatialement des échos de multiples directions de visée, et dans lequel lesdits coefficients dépendent de la distance entre un point d'intérêt et le transducteur en réseau incurvé mais ne dépendent pas de la ligne de balayage.

16. Procédé d'imagerie ultrasonique pour produire des images ultrasoniques spatialement composées, comprenant :

- l'interrogation de points dans un champ d'image à partir de multiples directions de visée (-•, 0, +•) avec des faisceaux (A, B, C) orientés à partir d'un transducteur en réseau incurvé (10), lesdits faisceaux étant orientés par une association de la courbure dudit transducteur en réseau incurvé et de l'orientation de faisceau électronique, moyennant quoi chaque faisceau présente un angle différent d'incidence par rapport à une ligne tangentielle à un point d'origine sur la face du transducteur en réseau incurvé à partir duquel le centre du faisceau s'étend ;
- la formation de faisceau d'échos reçus desdits points à partir de multiples directions de visée ;
- l'enregistrement spatialement d'échos de multiples directions de visée au moyen d'interpolation et puis la composition des échos spatialement enregistrés pour former des signaux d'images spatialement composées ; et
- l'affichage d'une image spatialement composée en réponse auxdits signaux d'images spatialement composées.

17. Produit programme d'ordinateur comprenant un jeu d'instructions pour utiliser le système d'imagerie diagnostique ultrasonique selon une des revendications 1 à 15, pour produire des images ultrasoniques spatialement composées avec des faisceaux transmis dans de multiples directions de visée à partir d'une tête de balayage en réseau incurvé.

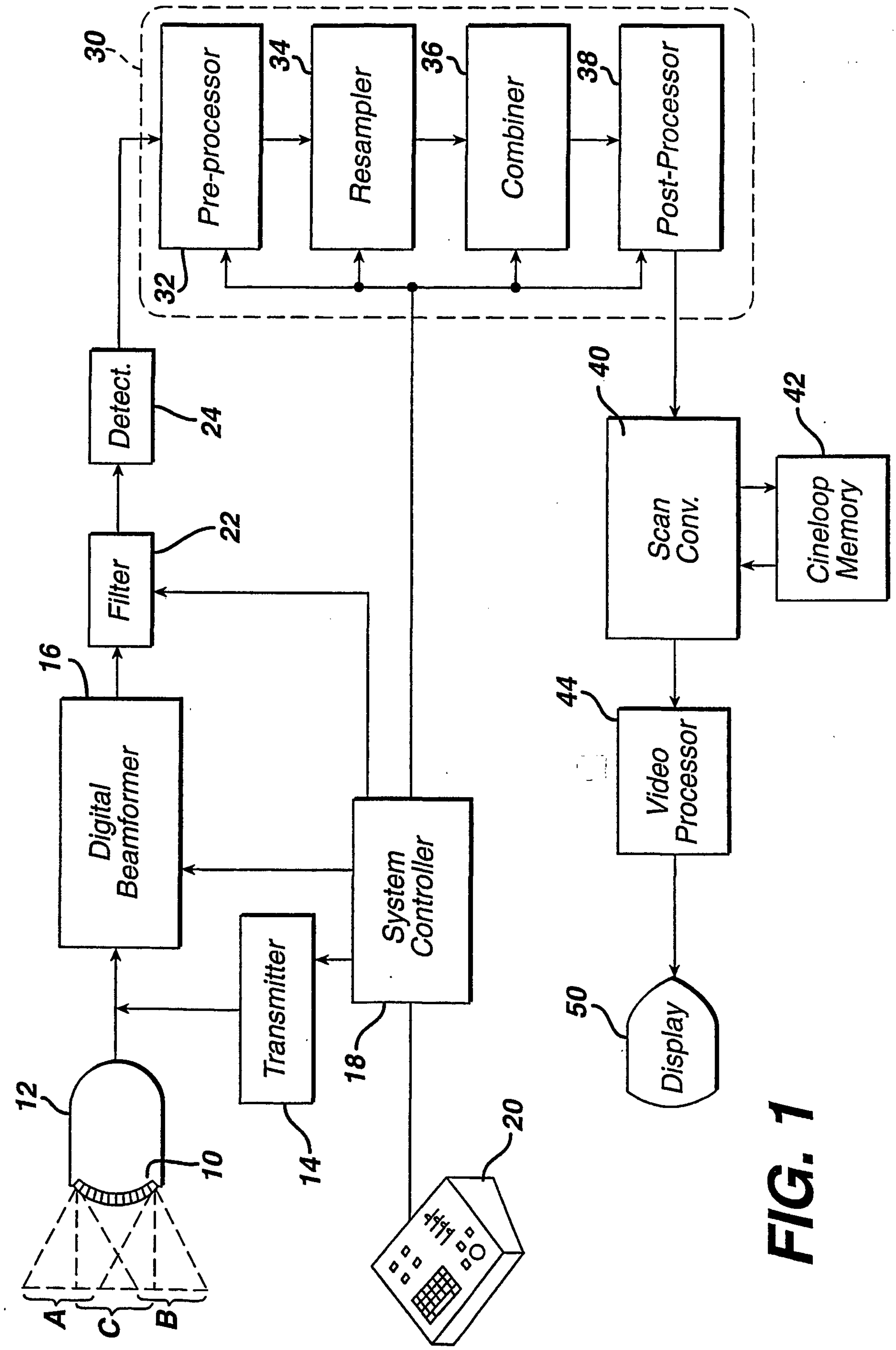
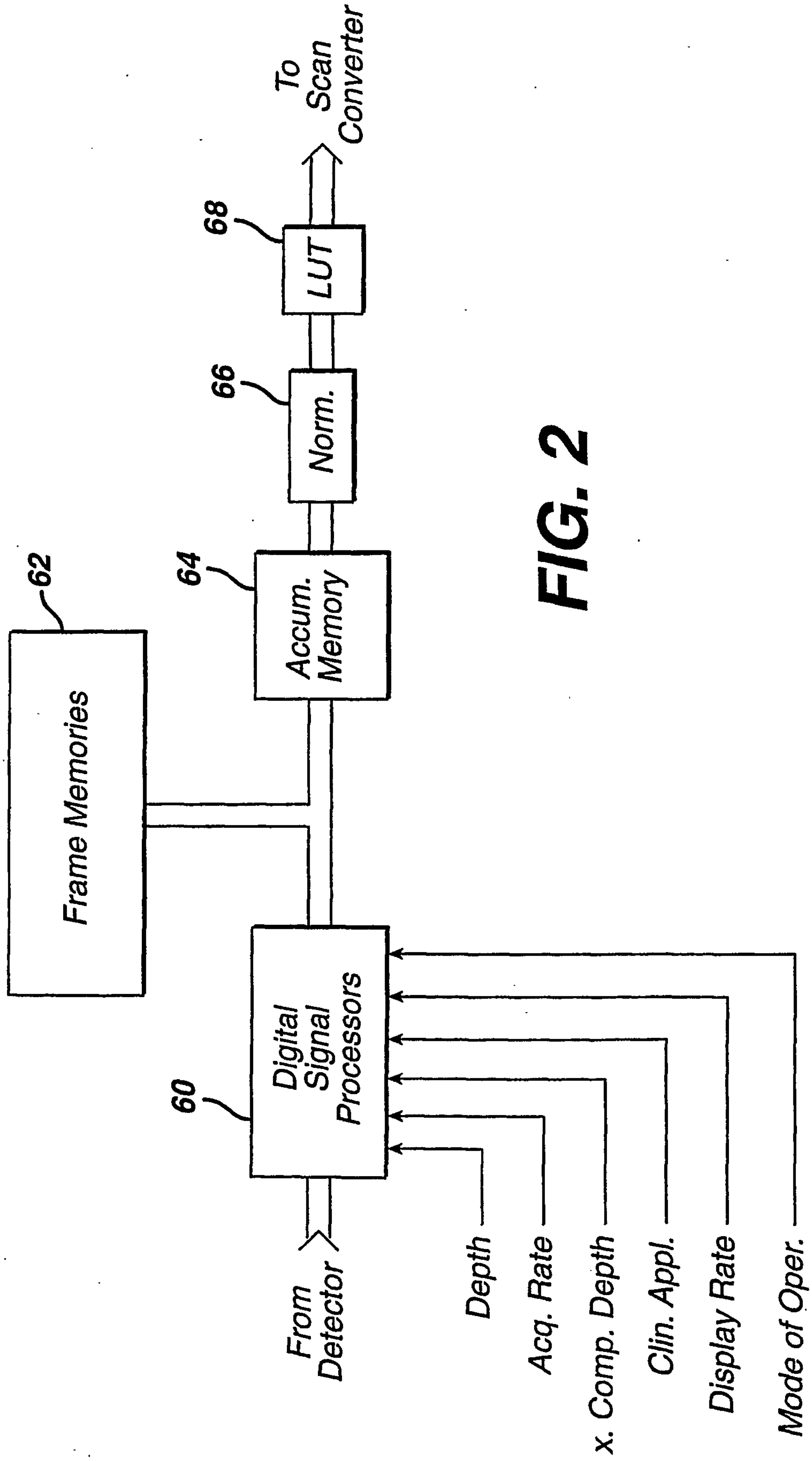

30
Pre-processor
34
Resampler
36
Combiner
38
Post-Processor
32
Detect.
24
Filter
22
16
Digital Beamformer
12
10
A
C
B
Transmitter
14
System Controller
18
20
40
Scan Conv.
42
Cineloop Memory
44
Video Processor
50
Display

FIG. 1



To Scan Converter
68
LUT
66
Norm.
62
64
Accum. Memory
Frame Memories
60
Digital Signal Processors
From Detector
Depth
Acq. Rate
x. Comp. Depth
Clin. Appl.
Display Rate
Mode of Oper.

FIG. 2

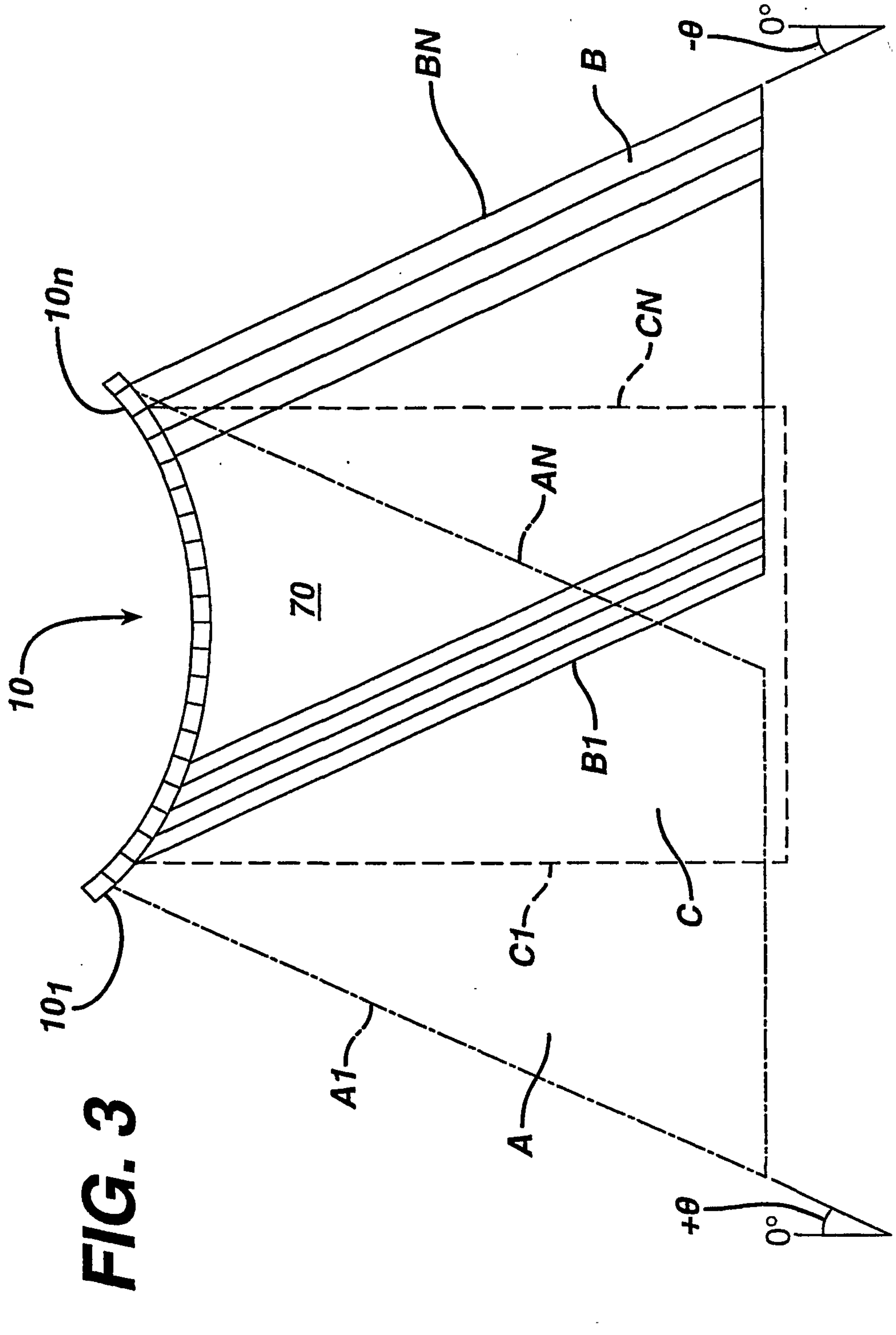


FIG. 3

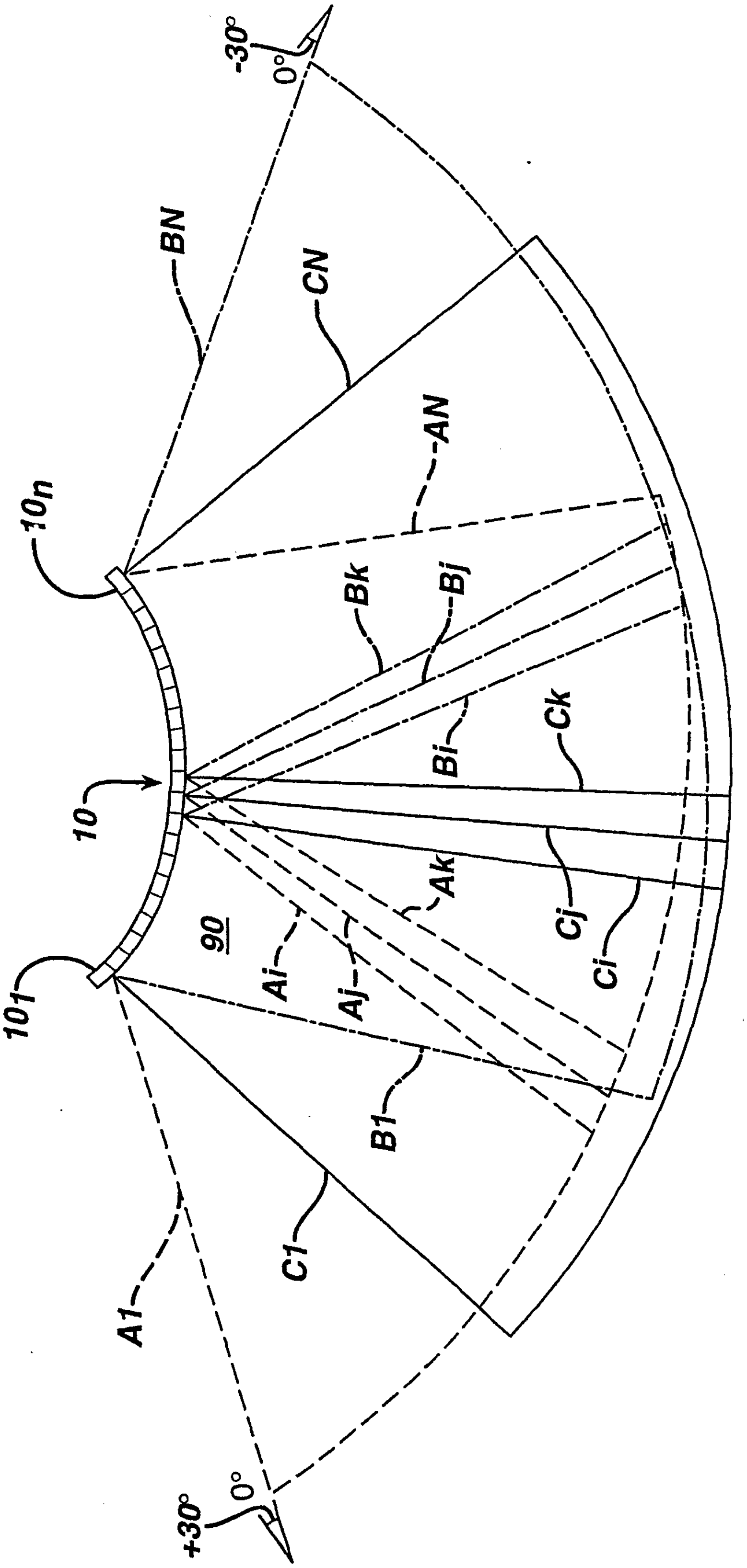


FIG. 4

FIG. 5a

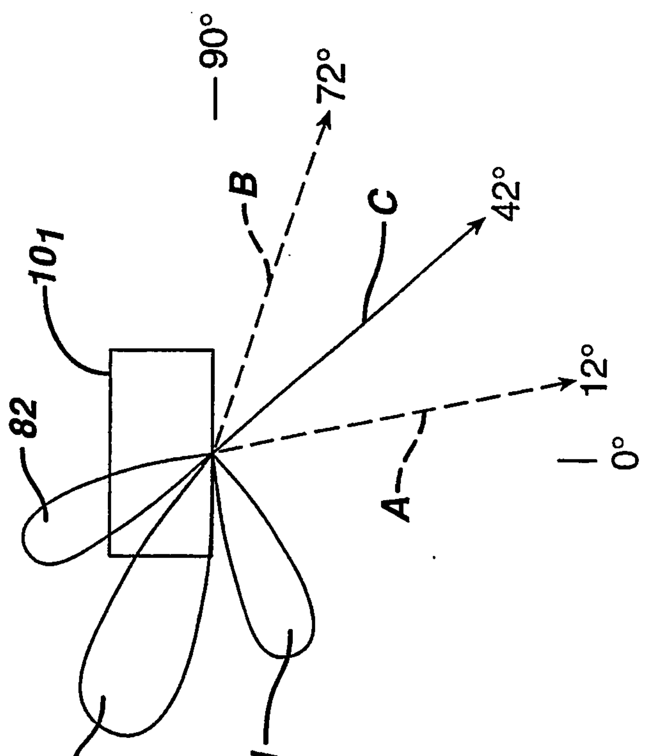


FIG. 5b

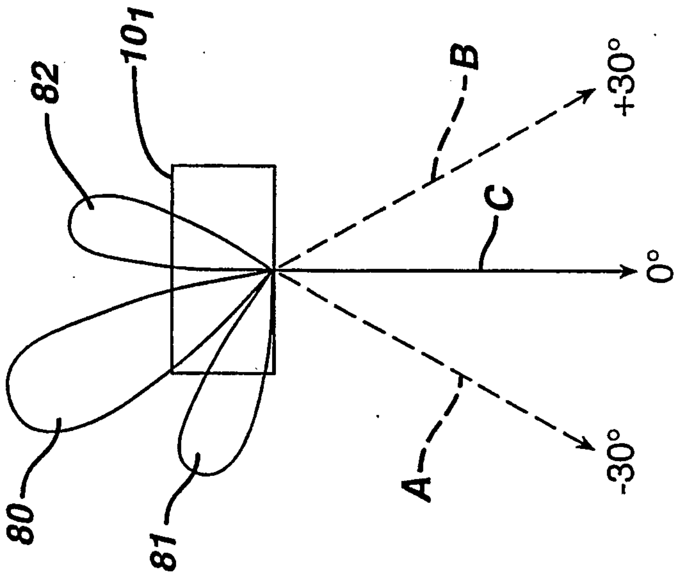
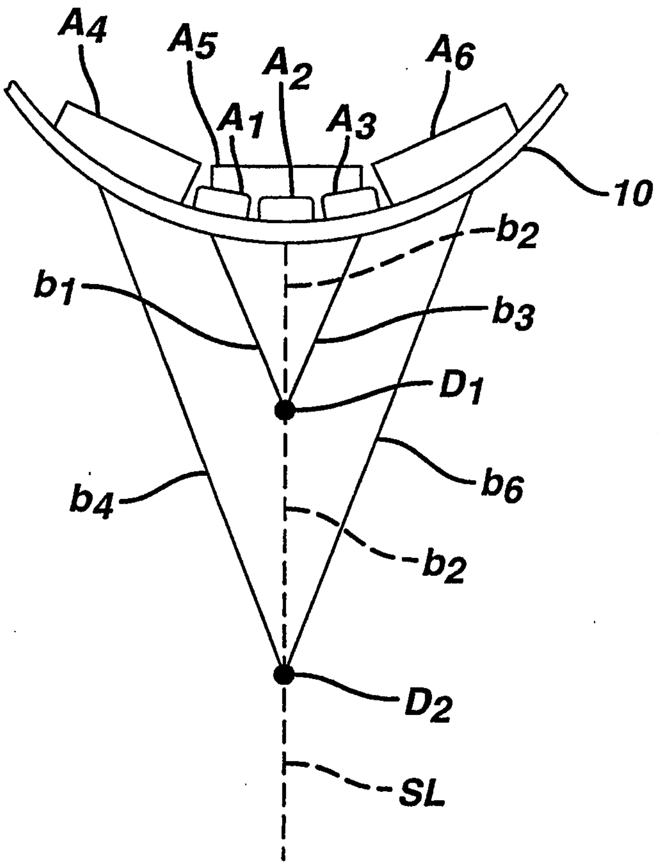


FIG. 6



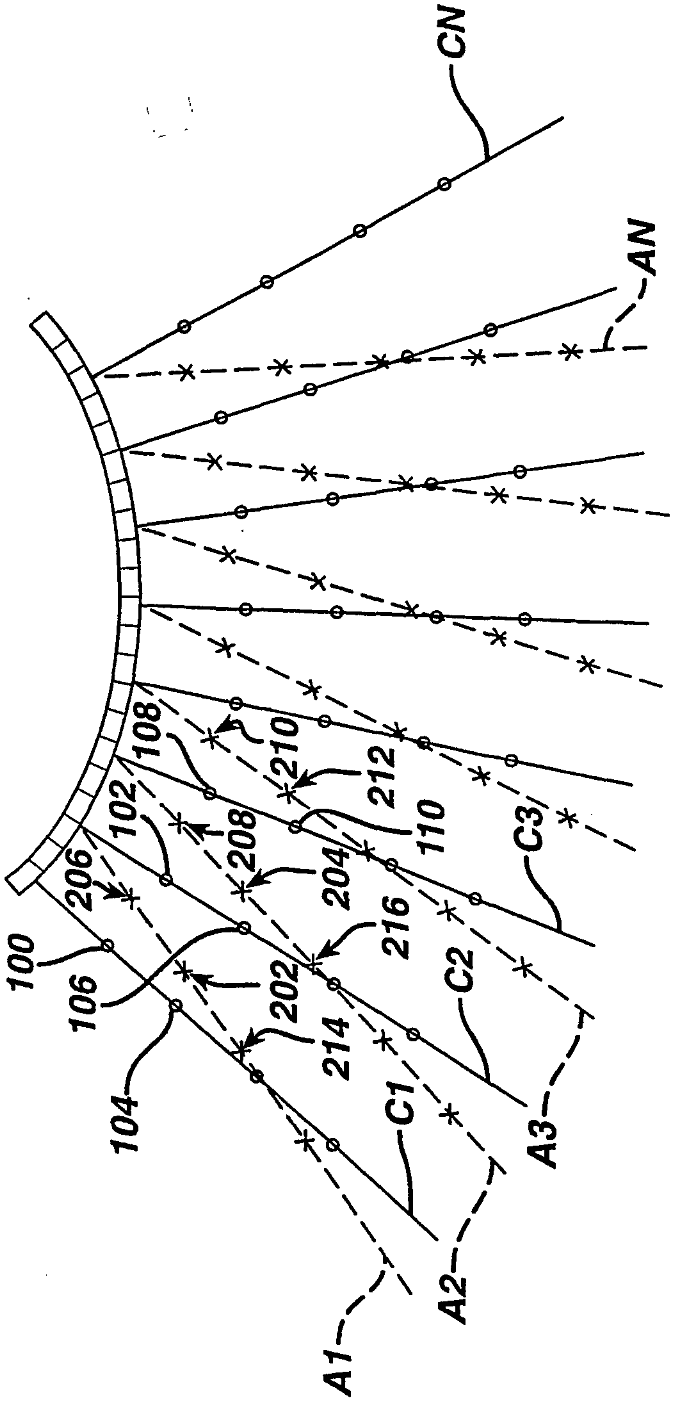

CN
AN
108
102
210
212
208
110
206
204
100
216
C3
106
202
214
C2
104
C1
A3
A2
A1

FIG. 7

REFERENCES CITED IN THE DESCRIPTION

This list of references cited by the applicant is for the reader's convenience only. It does not form part of the European patent document. Even though great care has been taken in compiling the references, errors or omissions cannot be excluded and the EPO disclaims all liability in this regard.

Patent documents cited in the description

- US 6210328 B **[0002] [0018]**
- US 5653235 A **[0005]**
- US 5485842 A **[0017]**
- US 5860924 A **[0017]**
- US 6135956 A **[0031]**

Non-patent literature cited in the description

- **R. ENTREKIN et al.** Real Time Spatial Compound Imaging in breast ultrasound: technology and early clinical experience. *MEDICAMUNDI, Philips Medical Systems,* vol. 43 (3), 35-43 **[0006]**

专利名称(译)	具有弯曲阵列扫描头的超声空间复合		
公开(公告)号	EP1292847B1	公开(公告)日	2013-04-17
申请号	EP2001933918	申请日	2001-05-02
[标]申请(专利权)人(译)	皇家飞利浦电子股份有限公司		
申请(专利权)人(译)	皇家飞利浦电子N.V.		
当前申请(专利权)人(译)	皇家飞利浦N.V.		
[标]发明人	JAGO JAMES R		
发明人	JAGO, JAMES, R.		
IPC分类号	G01S15/89 G01S7/52 A61B8/00		
CPC分类号	G01S15/892 G01S7/52077 G01S15/8995		
代理机构(译)	ROCHE，DENIS		
优先权	09/577021 2000-05-23 US		
其他公开文献	EP1292847A2		
外部链接	Espacenet		

摘要(译)

超声诊断成像系统用于产生空间复合的超声图像，其中光束从弯曲阵列扫描头沿多个观察方向传输。光束转向是弯曲阵列的曲率和电子束转向的函数。光束可以以平行光束组的方式操纵，或者以沿着阵列具有与其原点相同的角度方向的组来操纵。所描述的实施例提供了波束形成和配准系数使用，采样均匀性，散斑减少均匀性以及最大复合效果的更大区域的优点。

$$C = R\left(\phi_B - \sin^{-1}\left(\frac{R\sin\phi_B}{R+r}\right)\right)$$