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(54) Title: ULTRASONIC DIAGNOSTIC IMAGING SYSTEM AND METHOD WITH HARMONIC SPATIAL COMPOUNDING

(57) Abstract: An ultrasonic diagnostic imaging system and method uses harmonic imaging to acquire a plurality of images at a variety of look angles. The number of acquired ultrasonic images that are compounded to form a spatially compounded image is varied in response to changes in system operating parameters initiated directly or indirectly by the system user.

Ultrasonic diagnostic imaging system and method with harmonic spatial compounding

Technical Field

The invention relates to ultrasonic diagnostic imaging systems and methods and, in particular, to ultrasonic diagnostic imaging systems that produce spatially compounded images and harmonic ultrasonic imaging systems.

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Background of the Invention

Spatial compounding is an imaging technique in which a number of ultrasonic images of a given target that have been obtained from multiple vantage points or angles (look directions) are combined into a single compounded image by combining the data received from each point in the compound image target which has been received from each angle. Examples of spatial compounding are described in U.S. Patents Nos. 4,649,927; 4,319,489; and 4,159,462. Real time spatially compound imaging is performed by rapidly acquiring a series of partially overlapping component image frames from substantially independent spatial directions, and utilizing an array transducer to implement electronic beam steering and/or electronic translation of the component frames. The component frames are combined into a compounded image by summation, averaging, peak detection, or other combinational means. The acquisition sequence and formation of compounded images are repeated continuously at a rate limited by the acquisition frame rate, that is, the time required to acquire the full complement of scanlines over the selected width and depth of imaging.

The compounded image typically shows lower speckle and better specular reflector delineation than conventional ultrasonic images from a single viewpoint. Speckle is reduced (i.e., speckle signal to noise ratio is improved) by the square root of N in a compound image with N component frames, provided that the component frames used to create the compounded image are substantially independent and are averaged. Several criteria can be used to determine the degree of independence of the component frames (see, e.g., O'Donnell et al. in IEEE Trans. UFFC v.35, no.4, pp. 470-76 (1988). In practice, for spatially compound imaging with a steered linear array, this implies a minimum steering angle between component frames that is typically on the order of several degrees.

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The second way that spatially compound scanning improves image quality is by improving the acquisition of specular interfaces. For example, a curved bone-soft tissue interface produces a strong echo when the ultrasonic beam is exactly perpendicular to the interface, and a very weak echo when the beam is only a few degrees off perpendicular.

5   These interfaces are often curved, and with conventional scanning only a small portion of the interface is visible. Spatially compound scanning acquires views of the interface from many different angles, making the curved interface visible and continuous over a larger field of view. Greater angular diversity generally improves the continuity of specular targets. However, the angular diversity available is limited by the acceptance angle of the transducer

10   array elements. The acceptance angle depends on the transducer array element pitch, frequency, and construction methods.

Another ultrasonic imaging modality is harmonic ultrasonic imaging. It has been known for some time that tissue and fluids have inherent nonlinear properties. Tissue and fluids will, even in the absence of a contrast agent, develop and return their own non-

15   linear echo response signals, including signals at harmonics of the fundamental. Muir and Carstensen explored these properties of water beginning in 1980, and Starritt *et al.* looked at these properties in human calf muscle and excised bovine liver.

While these non-linear echo components of tissue and fluids are generally not as great in amplitude as the harmonic components returned by harmonic contrast agents, they

20   do exhibit a number of characteristics that have been recognized as being advantageous in conventional ultrasonic imaging. In particular, it has been recognized that negligible harmonic signals are generated very close to the transducer, which allows for clutter reduction when imaging through narrow orifices such as the ribs since fundamental signal reverberations are not being used for imaging. Additionally, it has been recognized that the

25   levels of a harmonic beam side lobe are lower than the corresponding levels of the side lobes of the fundamental beam, which has implications for off-axis clutter reduction. Finally, it has been recognized that the main lobe of the harmonic is narrower than that of its fundamental, which allows for improved lateral resolution.

Although each of these modalities – spatially compounded imaging and

30   harmonic imaging – has advantages in certain situations, each has certain performance limitations. In particular, the performance of ultrasonic imaging systems using spatial compounding is limited because grating lobes generated by a transducer array may cause false returns to be generated. With reference to Figure 1 which shows a narrowband example, an ultrasonic array 10 transmits and receives an ultrasonic signal having a main

lobe 14 and a plurality of pairs of grating lobes, only one of which 18 is shown in Figure 1.

The grating lobes 18 are shown with an amplitude that is significantly less than the amplitude

of the main lobe 14 because of the limited angular response of the transducer elements. The

main lobe 14 has a higher amplitude because the main lobe 14 is transmitted by the array 10

5   with a higher sensitivity, and the main lobe 14 is received by the array 10 with a higher

sensitivity. As is well known in the art, the angle θ between the main lobe 14 and the grating

lobes 18 is given by the formula: $\theta = \mathrm{Sin}^{-1} \lambda/P$, where P is the pitch of the array 10, *i.e.*, the

center-to-center distance between elements of the array 10, and λ is the wavelength of the

ultrasonic signal. When the wavelength λ of the transmitted ultrasonic signal is equal to the

10   pitch P of the array 10, the angle θ between the main lobe 14 and the grating lobes 18 is 90

degrees. As a result, an ultrasonic signal is not transmitted into tissues T positioned adjacent

the array 10 through the grating lobes 18 so that the only image generated is an image

resulting from insonification by the main lobe 14. Therefore, the grating lobes 18 do not

present any problem when the main lobe 18 is directed straight into the tissues T and the

15   angle θ between the main lobe 14 and the grating lobes 18 is 90 degrees or more, as shown in

Figure 1. If, however, the angle θ between the main lobe 14 and the grating lobes 18 is 45

degrees, as shown in Figure 2, ultrasonic signals are transmitted through the grating lobes 18

into the tissues, and echoes are returned from the tissues T through the grating lobes 18. As a

result, the image generated is an image resulting from the main lobe 14 as well as clutter and

20   possibly a "false image" resulting from the grating lobes 18.

In spatial compounding, the tissue must be imaged from a variety of beam

steering angles or look directions, as shown in Figure 3. A spatially compounded image of

an object O in the tissues T is the result of returns from the array 10 at a first angle ϕ_1, returns

from the array 10 at a second angle ϕ_2, returns from the array 10 at a third angle ϕ_3, etc. For

25   the array 10 to image at each of these angles ϕ, it is necessary for the array 10 to steer the

main lobe 14 to such angle ϕ_1, for example, as shown in Figure 4. When the main lobe 14 is

steered to an angle ϕ_1, the grating lobes 18 are positioned at an angle of $\phi_1 \pm 90$ degrees so

that one of the grating lobes 18a extends into the tissues T at an angle at which the array has

greater sensitivity. Under these circumstances, ultrasonic returns are received from both the

30   main lobe 14, and the grating lobe 18, and these returns are indistinguishable and thus both

contribute to the ultrasonic image. The returns from the grating lobe 18 therefore result in

clutter or a false image because they do not emanate from the object O being imaged. The

result is an image of the object O cluttered by returns from tissues imaged by the grating lobe

18. This clutter can make it very difficult to view the object O being imaged. This problem might be alleviated to some extent by reducing the pitch P of the array 10 and/or increasing the wavelength of the transmitted ultrasonic, but there are limitations on the practical ability to either increase pitch P or decrease the wavelength beyond certain limits.

5   The problem of grating lobe clutter is even more acute as the angle ϕ of the main lobe 14 is increased, which is highly desirable for beam steering during spatially compound imaging. As shown in Figure 5, the sensitivity A (both transmit and receive) of the array 10 is a function of the angle ϕ of a lobe, whether the lobe is a main lobe 14 or a grating lobe 18. When the main lobe 14 has an angle ϕ_{M1} equal to 90 degrees, as shown in

10   Figure 1, the sensitivity A of the main lobe 14 is relatively large, as shown in Figure 5. At the same time, the sensitivity A of the grating lobe 18 is relatively small because the angle ϕ_{G1} of the grating lobe 18 is -90 degrees, as also shown in Figure 5. However, when the main lobe 14 is steered to a relatively large angle ϕ_{M2} of 80 degrees, one of the grating lobes 18 is at an angle ϕ_{G2} of -10 degrees. As a result, the amplitude of the fundamental lobe 14 is

15   relatively small and the amplitude of the grating lobe 18 is relatively large, as also shown in Figure 5. The problem of grating lobe clutter is therefore more severe with spatially compound imaging where beams are steered over a wide range of steering angles.

          There is therefore a need for a system and method for spatially compound imaging that does not suffer from image clutter resulting from grating lobe returns.

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Summary of the Invention

          A harmonic ultrasonic compounded imaging system and method uses a transducer array to transmit ultrasonic signals at a fundamental frequency. The transmitted ultrasonic signals have a main lobe that is directed at a target in a plurality of look directions.

25   The transducer array then receives ultrasonic echo signals at a harmonic frequency by suitable means, such as by using a receive beamformer coupled to the transducer array and a filter or pulse inversion processor. The ultrasonic signals at the harmonic frequency have main lobes that are aligned in each of the look directions as the fundamental frequency main lobes. A spatially compounded image is then generated from the received ultrasonic signals

30   at the harmonic frequency using, for example, a compound image processor. The ultrasonic signals at the fundamental frequency may have a grating lobe in addition to the main lobe. The ultrasonic signals in the grating lobe are transmitted in a direction that is different from the direction the ultrasonic signals in the main lobe are transmitted. Similarly, the received ultrasonic signals at the harmonic frequency may have a grating lobe in addition to the main

lobe. The direction of the grating lobe of the received harmonic signals is different from the
direction of the grating lobe of the transmitted fundamental signals so that the amplitude of
any ultrasonic signals received through the grating lobe at the harmonic frequency is
relatively low. In addition for steering angles at which the main lobe still has appreciably
5   higher amplitude than the grating lobe, the non-linear nature of harmonic generation results
in an increase in the relative amplitudes of the main and grating lobes at the harmonic
frequency.

Brief Description of the Drawings

10   Figure 1 is a schematic diagram illustrating the manner in which a
fundamental ultrasonic signal having a first frequency is conventionally transmitted by a
transducer array.
                    Figure 2 is a schematic diagram illustrating the manner in which a
fundamental ultrasonic signal having a second frequency is conventionally transmitted by a
15   transducer array.
                    Figure 3 is a schematic diagram illustrating the manner in which a transducer
array conventionally images a target at multiple look angles to provide a spatially
compounded ultrasonic image.
                    Figure 4 is a schematic diagram illustrating the manner in which a transducer
20   array conventionally images a target at one look angle during spatially compound imaging.
                    Figure 5 is a graph showing the gain and sensitivity of a conventional
transducer array as a function of look angle.
                    Figure 6 is a schematic diagram illustrating the manner in which a transducer
array images a target at one look angle during spatially compound imaging in accordance
25   with an embodiment of the invention.
                    Figure 7 is a block diagram of a system for generating a spatially compounded
harmonic ultrasonic image in accordance with one embodiment of the invention.
                    Figure 8 is a block diagram of one embodiment of a processor used in the
system of Figure 7.
30   Figures 9a-c are diagrams schematically illustrating the effects of increasing
and decreasing the number of acquired frames that are compounded to form a spatially
compounded harmonic ultrasonic image using the system of Figure 7.

Detailed Description of the Invention

The manner in which the problem of grating lobe clutter in spatial compounding can be reduced in accordance with one embodiment of the invention is illustrated in Figure 6. Figure 6 shows a main lobe 20 transmitted at a fundamental frequency by an ultrasonic array 22 at a "look angle" of ϕ_1. Also shown is a grating lobe 28

5   transmitted at the fundamental frequency at an angle of ϕ_2. The array 22 receives ultrasonic signals at higher and sub harmonics of the fundamental frequency, in this example drawn for the second harmonic. A main lobe 30 of the harmonic signals has the same look angle ϕ_1 as the main lobe 20 of the fundamental signals so that the array 22 will receive harmonic signals from the same direction in which the fundamental signals were transmitted. However, since

10   the frequency of the harmonic signals is twice the frequency of the fundamental signals, the wavelength λ of the harmonic signals is half the wavelength λ of the fundamental signals. The ratio λ/P, and a resulting look angle ϕ_3, of a harmonic grating lobe 38 is therefore different from the ratio λ/P, and the resulting look angle ϕ_2, of the fundamental grating lobe 28. The array 22 thus receives ultrasonic signals through the harmonic grating lobe 38 in a

15   direction that is different from the direction that ultrasonic signals are transmitted through the fundamental grating lobe 28. Since there are no ultrasonic signals transmitted in the direction from which the ultrasonic signals are received through the harmonic grating lobe 38, the harmonic grating lobe 38 does not cause appreciable clutter of an image generated using the harmonic main lobe 30. The use of harmonic imaging for spatially compound imaging thus

20   produces results that are not possible when using the fundamental to generate an image through spatial compounding.

In practice, when a broadband transmit signal is used, both main and grating lobes are spread out as a function of the frequencies (and wavelengths) of the transmit signal. Under these circumstances there is opportunity for overlap between the fundamental and

25   harmonic grating lobes. However, for steering angles at which the fundamental main lobe is greater than the fundamental grating lobe, an additional mechanism helps reduce the amplitude of the harmonic grating lobe. That mechanism is that the generation of harmonic signal is by nature non-linear, meaning that, for a case where the fundamental main lobe has twice the amplitude of its grating lobe, there will be more than a factor of two between the

30   harmonic signals generated by the fundamental main and grating lobes.

Harmonic spatially compound imaging has the additional advantage of using a lower fundamental transmit frequency (e.g., 2/3 f_0) for an array of a given pitch, thus increasing the angle between the main lobe 20 of the fundamental and the grating lobe 28 of

the fundamental. This means that the grating lobe at the fundamental frequency is even
lower than it would be for operation at a frequency f_0 in the conventional case.

One embodiment of a system 90 and method in accordance with the invention
is shown in Figure 7. The system 90 includes a scanhead 100 having a transducer array 102

5   that 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 scanhead 100. The
transmission of the beams is controlled by a transmitter 104, which controls the phasing and
time of actuation of each of the elements of the array 102 so as to transmit each beam from a

10   predetermined origin along the array and at a predetermined angle. The echoes returned from
along each scanline are received by the elements of the array 102, digitized as by analog to
digital conversion, and coupled to a digital beamformer 106. The digital beamformer 116
delays and sums the echoes from elements of the array 102 to form a sequence of focused,
coherent digital echo samples along each scanline. The transmitter 104 and beamformer 106

15   are operated under control of a system controller 108, which in turn is responsive to the
settings of controls on a user interface 120 operated by the user of the ultrasonic system. The
system controller 108 controls the transmitter 104 to transmit the desired number of scanline
groups at the desired angles, transmit energies and frequencies. The system controller 108
also controls the digital beamformer 106 to properly delay and combine the received echo

20   signals for the apertures and image depths used.

The scanline echo signals are filtered by a programmable digital filter 122,
which defines the band of frequencies of interest. The passband of the filter 122 is set to pass
harmonics of the signal that is transmitted by the transmitter 104. As an alternative to
separating the harmonic signals by filtering, pulse inversion processing may be employed as

25   described in U.S. Patents 5,706,819 (for harmonic contrast agents) and 5,951,478 (for tissue
harmonic signals), by which echoes received from a target in response to multiple, differently
modulated transmit pulses are combined to cancel linear signals while emphasizing non-
linear (e.g., second harmonic) signals. The transmitted signal may be a single fundamental
frequency, two or more fundamental frequencies, or a band of frequencies centered at a

30   nominal fundamental frequency. In any case, the signal passed by the filter 122 will be a
single harmonic frequency, two or more harmonic frequencies, or a band of frequencies
centered at a nominal harmonic frequency, respectively. The filtered harmonic signals are
then detected by a detector 124. The filter 122 and detector 124 preferably 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 124 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.

5   The digital echo signals are processed by spatial compounding in a processor 130. A pre-processor 132 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 132 can also weight edge lines that are at the edge of one overlapping image so as to smooth

10   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 134. The resampler 134 can spatially realign the estimates of one component frame to those of another component frame or to the pixels of the display space.

        After resampling, the image frames are compounded by a combiner 136.

15   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 138. The post-processor 138 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

20   compression and mapping of the range of compounded values to a range of values suitable for display of the compounded image.

        The compounding process may be performed in estimate data space or in display pixel space. In one embodiment, scan conversion is done following the compounding process by a scan converter 140. The compound images may be stored in a Cineloop

25   memory 142 in either estimate or display pixel form. If stored in estimate form, the images may be scan converted when replayed from the Cineloop memory 142 for display. The scan converter 140 and Cineloop memory 142 may also be used to render three dimensional presentations of the spatially compounded images as described in U.S. Patents Nos. 5,485,842 and 5,860,924, or displays of an extended field of view by overlaying successively

30   acquired, partially overlapping images in the lateral dimension. Following scan conversion, the spatially compounded images are processed for display by a video processor 144 and displayed on an image display 150.

        Figure 8 illustrates one embodiment of the spatial compounding processor 130 of Figure 7. The processor 130 is preferably implemented by one or more digital signal

processors 160, which process the image data in various ways. The digital signal processors
160 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 160 direct the
processed image frames to a plurality of frame memories 162, which buffer the individual

5   image frames. The number of image frames capable of being stored by the frame memories
162 is preferably at least equal to the maximum number of image frames to be compounded,
such as sixteen frames. The digital signal processors 160 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),

10   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 162 for assembly as a compound image in accumulator memory 164. The

15   compounded image formed in the accumulator memory 164 is weighted or mapped by a
normalization circuit 166, then compressed to the desired number of display bits and, if
desired, remapped by a lookup table (LUT) 168. The fully processed compounded image is
then transmitted to the scan converter 140 for formatting and display.

         The digital signal processors 160 determine the number of frames that are to

20   be compounded to improve image quality while still providing an acceptable real-time
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 spatially compound scanning. This

25   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.

30   The system 90 has particular utility when the clinician is changing between
"survey" and "target" modes of operation. During the survey mode of operation, the
clinician is rapidly maneuvering the scanhead 100 to quickly ascertain the presence of
prominent physiological landmarks or features. The compounding of a large number of
component frames when the scanhead 100 is in motion will result in blurred images.

Consequently the number component frames being compounded is reduced during the survey mode. As the clinical operator identifies potential abnormalities, the scanhead motion is slowed down or stopped completely to image the features of interest ("targeted" or "study" mode). At this point, the image features within the component frames are more highly

5   correlated, and motion blurring in the compound image is substantially reduced or completely eliminated as long as the scanhead 100 is substantially stationary. During the target mode, the number of component frames being compounded is increased to produce high quality images without the consequence of blurring.

It is well known that the frame rate of an image with a given line density is

10   dependent upon the maximum display depth of the image, because the speed of sound in tissue (~1.54 millimeters per microsecond) imposes a minimum round trip propagation delay of 13 microseconds for every centimeter of image depth. Typically, additional delay time is also added to prevent reverberation artifacts, $i.e.$, the receipt of echoes at the beginning of one image line which are returning from deep depths of a previous image line. For an image

15   consisting of 192 ray lines and a depth of 2 centimeters, the acquisition frame rate can be 100 frames per second or greater, but for an 8 centimeter image depth the frame rate can drop to 25 frames per second. While 25 frames per second is adequate for real time examination, a compound image frame rate for 7 component frames at this depth is less than 4 frames per second. A frame rate this slow would generally be considered inadequate for real time

20   examination. Conversely, a compound image with 3 component frames with an image depth of 2 centimeters would have a compound frame rate of 33 frames per second, which is higher than necessary for real time examinations.

Compound harmonic scanning with a steered linear array results in a pattern of overlapping component frames such that the region of maximum image quality (RMIQ)

25   where all N frames overlap is a trapezoidal or an inverted triangular region with its base at the top of the compound image. For a small number of component frames steered at the minimum angle, this region of maximum image quality extends deep into the compound image. For a large number of component frames, the RMIQ is relatively shallow. This is illustrated by Figures 9a-9c, which show three different compound scan geometries, each

30   consisting of several partially overlapping steered linear component frames scanned from a linear array scanhead 100. For visual clarity, the minimum steering angle was chosen as 15° between component frames. Figure 9a shows two component frames, which are compounded, and the RMIQ extends the full 4.0 cm height of the image. Figure 9b shows an image that compounds four component frames, and the RMIQ now extends only 2.3 cm from

the top of the compound frame. Similarly, Figure 9c shows that, for seven component

frames, the RMIQ of the compound image only extends 1.3 centimeters in depth. These

drawings show that increasing the number of frames in the compound image decreases the

size of the RMIQ. Thus, spatial compounding using component frames with large steering

5   angles contributes very little additional image quality at depths deeper than the RMIQ.

It is apparent from the foregoing illustrations that that at shallow depths the

frame rate is relatively high, which allows the use of more component frames (larger value of

N) for spatially compound imaging while still maintaining an adequate frame rate of display.

This is also consistent with the decrease in the depth of the RMIQ associated with more

10   component frames (larger value of N) for spatially compound imaging. Therefore, the

relationships between image depth, frame rate, the number of component frames in a

spatially compounded image, the size and depth of the RMIQ, and image quality can be

exploited to optimize overall performance. Table 1 below shows how these tradeoffs can

advantageously be made for different depths, while always maintaining a compound frame

15   rate of 10 Hz or greater.

Table 1.

Image display depth, cm	# of frames in compound image	Acquisition frame rate, Hz	Compound display rate, Hz	Max. steering angle, degrees	Depth of RMIQ, cm
2	7	100	14	45	1.3
4	5	50	10	30	2.3
6	3	37.5	12.5	15	4.0
8	2	25	12.5	9	8.0

Thus it is seen that as the image display depth increases, the system controller

20   108 responds by decreasing the number of frames that are acquired and compounded to form

the displayed compounded image. When the user selects a greater display depth for the

scanhead 100, the ultrasonic system 90 responds by decreasing the number of frames of the

compounded image. As the acquisition frame rate declines the number of frames

compounded is also decreased. If the user reduces the frame rate, for example, by increasing

25   the number of lines of an image, the ultrasonic system 90 will respond by decreasing the

number of compounded frames in a displayed image. When the look directions of the

acquired image frames are varied by steering the transmit beams in a number of different

look directions, the angle of the sides of the trapezoidal-shaped image decreases with increases in the number of images that are compounded. These adaptive changes in the number of frames being compounded maintain the display rate of the compounded image at more than ten frames per second, or some other rate which is acceptable for the particular

5   clinical application being performed.

As the number of frames compounded in the displayed image decreases, the depth of the region of greatest compounding preferably increases, but is comprised of fewer compounded image frames. As the image depth is decreased, the maximum steering angle of the steered transmit beams also preferably increases. A comparison of Figures 9a-9c

10   illustrates how a greater maximum steering angle will more effectively cover a shallow imaging depth, whereas a lesser maximum steering angle is more effective for greater imaging depths.

Thus it is seen that the use of harmonic spatially compound imaging allows high quality of compounded images to be produced at a wide variety of look angles. The

15   system 90 preferably operates by decreasing the number of look directions and the maximum steering angle with increasing image depth. The number of look directions is also preferably decreased when the number of lines or line density of the image is increased; when the number of transmit focal zones is increased; when the amount of deadtime per PRI is increased; when the number of transmissions per image line is increased (*e.g.*, for synthetic

20   aperture, pulse inversion harmonic imaging), when the number of simultaneous modes is increased (*e.g.*, spectral Doppler together with 2D imaging), the size of the region of interest is increased (*e.g.*, image zoom is decreased or turned off), the clinical application (*e.g.*, changing from abdominal or peripheral vascular imaging to cardiac imaging), or changing from target to survey mode. When the number of simultaneously acquired lines is increased

25   by increasing multiline acquisition, the number of look directions can be increased.

From the foregoing it will be appreciated that, although specific embodiments of the invention have been described herein for purposes of illustration, various modifications may be made without deviating from the spirit and scope of the invention. Accordingly, the invention is not limited except as by the appended claims.

CLAIMS:

1.   A harmonic ultrasonic spatially compounded diagnostic imaging system, comprising:

-       a transducer array;

-       a transmitter coupled to the transducer array, the transmitter applying transmit signals
5   to the transducer array at a fundamental frequency, the transmit signals causing the
          transducer array to direct ultrasonic signals toward a target from a plurality of
          different look directions;

-       a receive beamformer coupled to the transducer array, the beamformer generating
          echo signals from the target at each of the plurality of different look directions
10   responsive to the ultrasonic signals being directed toward the look direction;

-       a filter receiving the echo signals and passing echo signals having a frequency that are
          a harmonic of the fundamental frequency; and

-       a compound image processor coupled to the filter to receive the echo signals passed
          by the filter, the image processor spatially compounding the received echo signals
15   from each of the different look directions to form a spatially compounded image.

2.   The harmonic ultrasonic spatially compounded diagnostic imaging system of
Claim 1, wherein:

-       said transmitter applies transmit signals to the transducer array to cause the transducer
20   array to direct a main lobe of ultrasonic signals at the fundamental frequency toward a
          target from a first set of different look directions and to produce a grating lobe of the
          ultrasonic signals at the fundamental frequency in a second set of directions
          corresponding to respective look directions in the first set, each of the directions in
          the second set being different from the respective look directions in the first set; and
25 -   wherein said receive beamformer generates echo signals from the target through a
          main lobe of ultrasonic signals at a harmonic of the fundamental frequency at each of
          the plurality of different look directions responsive to the ultrasonic signals at the
          fundamental frequency in the main lobe being directed toward each look direction,
          the receiver further receiving echo signals through a grating lobe of ultrasonic signals

at the harmonic of the fundamental frequency from a third set of directions corresponding to respective look directions in the first set, each of the directions in the third set being different from respective directions in the second set and from respective look directions in the first set.

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3.   The ultrasonic diagnostic imaging system of claim 1 or 2, wherein the compound image processor forms spatially compounded images in real time while the transducer array is operated to acquire target echoes at a plurality of look directions.

10 4.   The ultrasonic diagnostic imaging system of claim 1 or 2, wherein the spatially compounded image comprises a three dimensional image.

5.   The ultrasonic diagnostic imaging system of claim 1 or 2, wherein the spatially compounded image comprises a spatially compounded Doppler processed image indicative

15 of moving ultrasound scatterers in the target.

6.   The ultrasonic diagnostic imaging system of claim 1 or 2, wherein the harmonic comprises a non-linear signal returned in response to the transmission of a linear fundamental frequency signal.

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7.   The ultrasonic diagnostic imaging system of claim 6, wherein the harmonic comprises a second harmonic of the fundamental frequency.

8.   The ultrasonic diagnostic imaging system of claim 1 or 2, wherein the

25 harmonic echo signals comprise tissue harmonic echo signals.

9.   The ultrasonic diagnostic imaging system of claim 1 or 2, wherein the filter comprises a pulse inversion processor.

30 10.   The ultrasonic diagnostic imaging system of claim 1 or 2, wherein the array transducer transmits multiple times per image line during at least one of synthetic aperture operation and pulse inversion harmonic imaging operation.

11.   The ultrasonic diagnostic imaging system of claim 1 or 2, wherein the array transducer produces multiple image lines in response to one pulse transmission.


12.   A method of forming a harmonic ultrasonic spatially compounded image,
5   comprising:

- acquiring a plurality of ultrasonic images of a region of the body which exhibit different look directions by means of an array transducer transmitting ultrasound signals into the body at a fundamental frequency and receiving ultrasound signals from the body at a frequency that is a harmonic of the fundamental frequency;

10 - compounding a number of the ultrasonic images to form a spatially compounded ultrasonic image; and

- displaying the spatially compounded ultrasonic image at a display frame rate, wherein the number of images which are compounded to form the spatially compounded image is varied in response to the setting of one or more of the parameters of: image

15   display depth, frame acquisition rate, number of scanlines, image line density, number of transmit focal zones, amount of deadtime per PRI, number of transmissions per image line, depth of region of greatest compounding, clinical application, number of simultaneous modes, size of region of interest, and mode of operation.
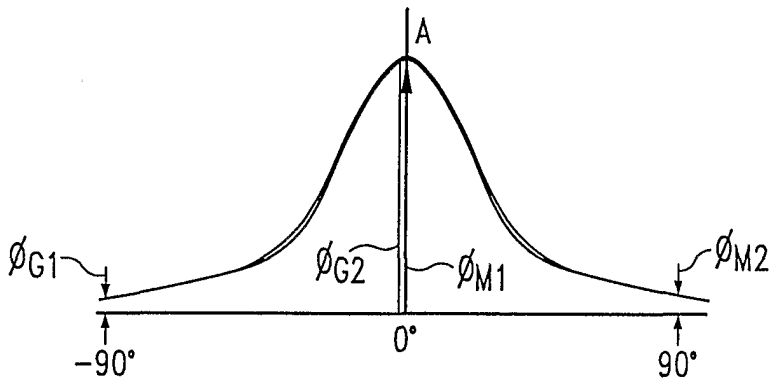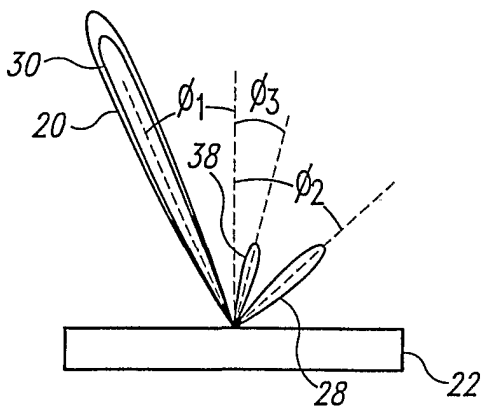
1/5
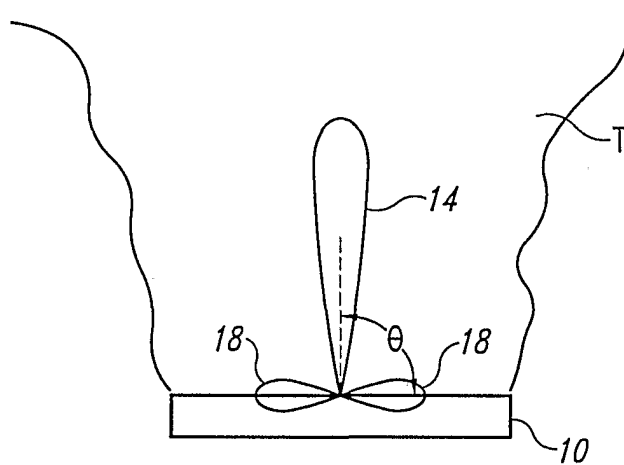
Fig. 1
(Prior Art)
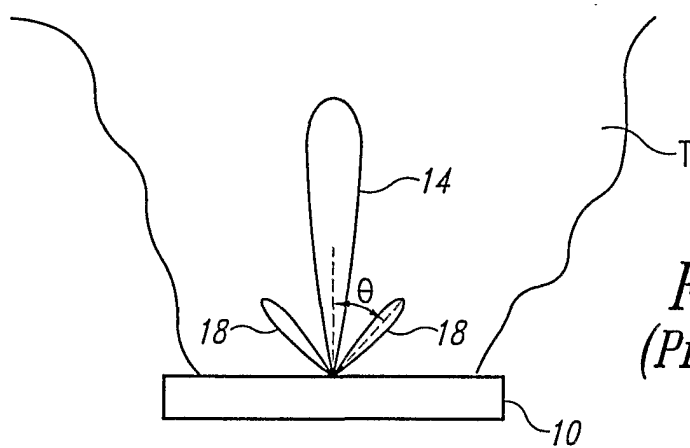
Fig. 2
(Prior Art)
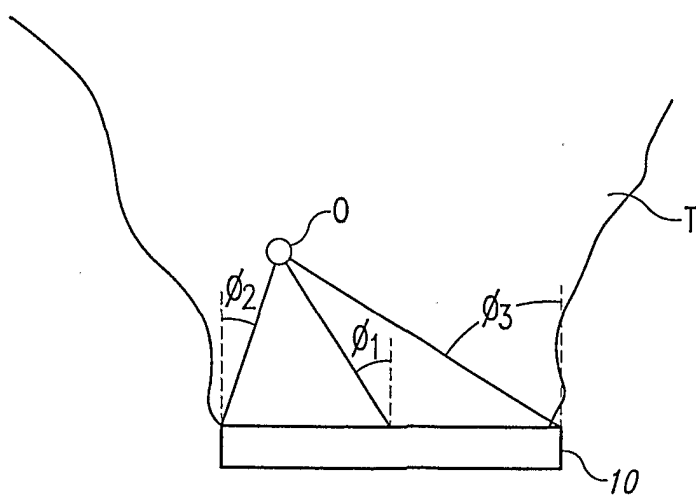
Fig. 3
(Prior Art)
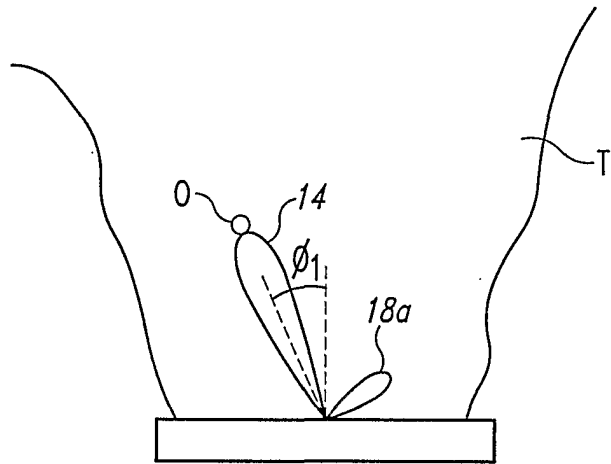
Fig. 4
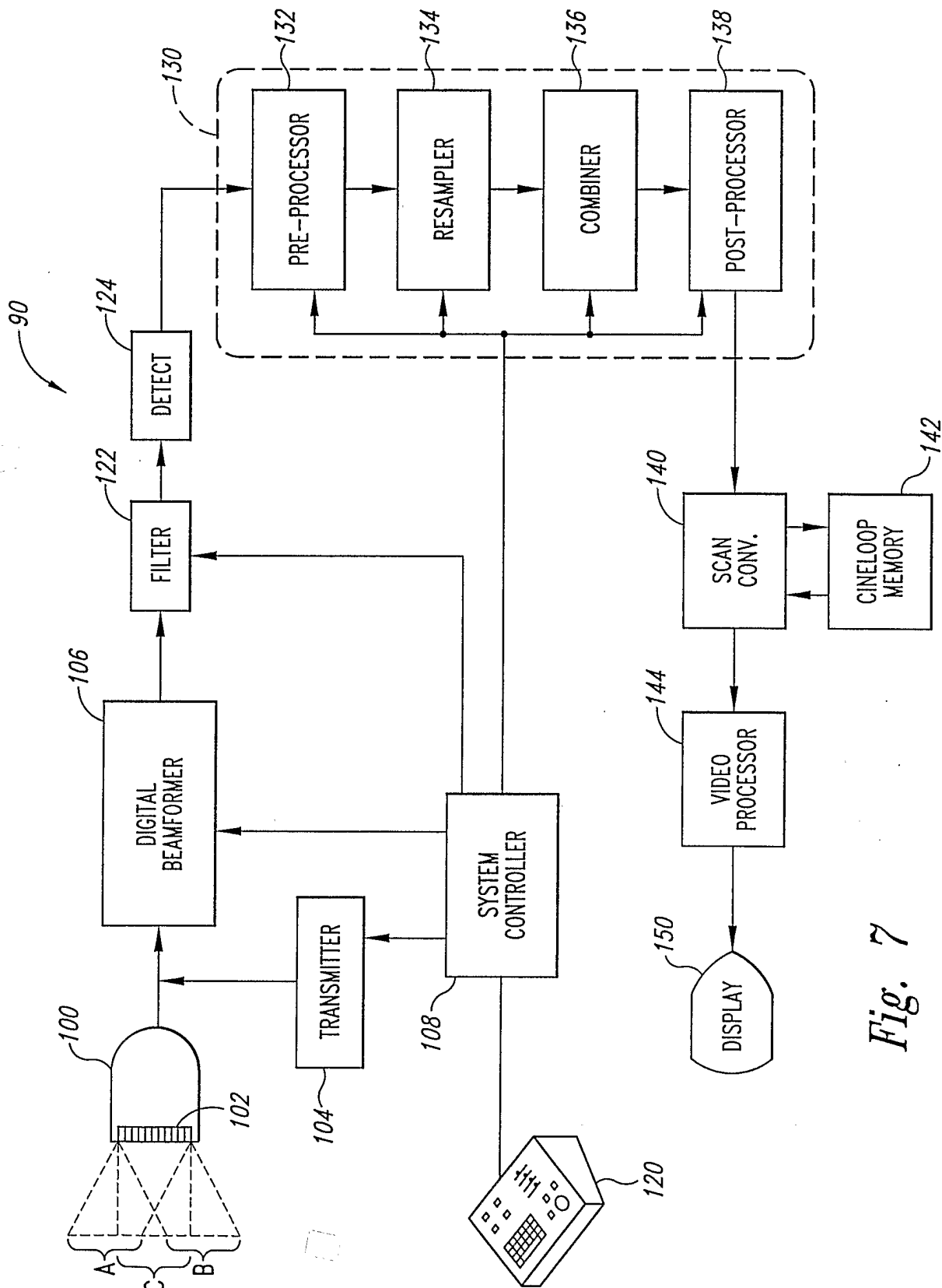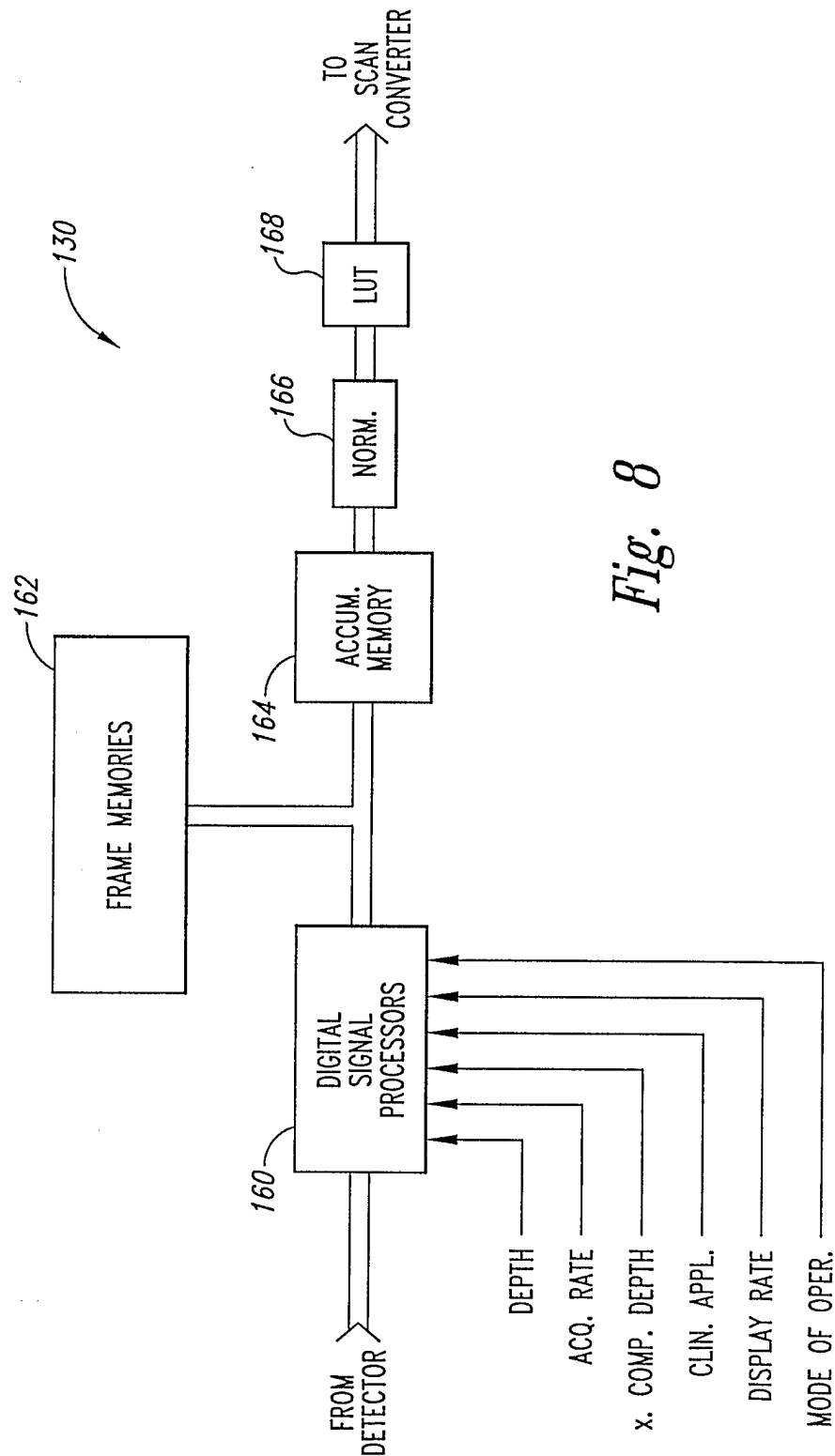*(Prior Art)*

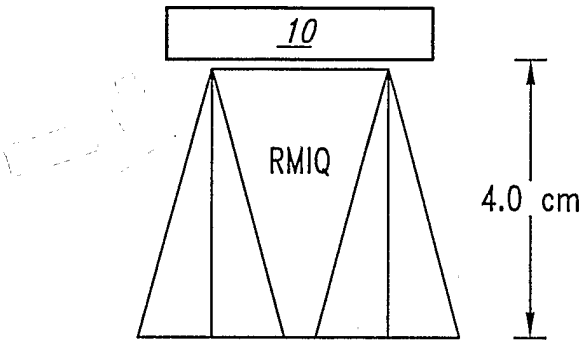*Fig. 5*
(Prior Art)

Fig. 6

Fig. 7

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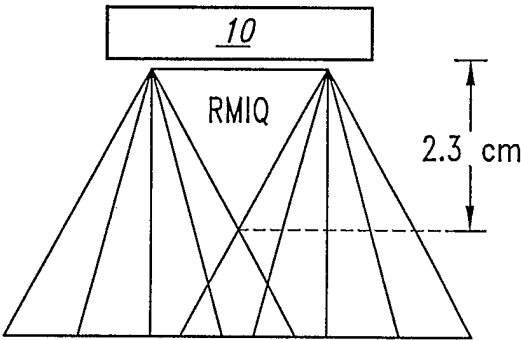*Fig. 8*

Fig. 9A

Fig. 9B
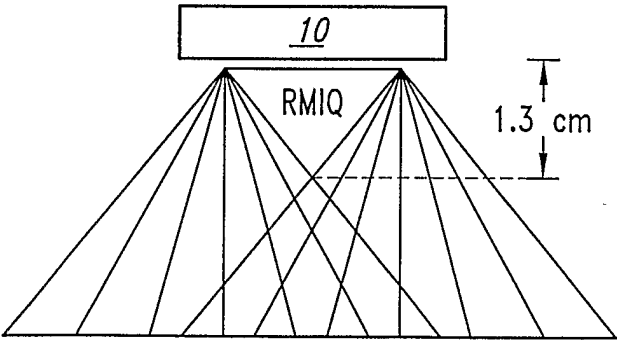
Fig. 9C

专利名称(译)	具有谐波空间复合的超声诊断成像系统和方法		
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摘要(译)

超声诊断成像系统和方法使用谐波成像来以多种视角获取多个图像。复合以形成空间复合图像的采集的超声图像的数量响应于由系统用户直接或间接启动的系统操作参数的变化而变化。