

(19) World Intellectual Property Organization
International Bureau



(43) International Publication Date
22 January 2009 (22.01.2009)

PCT

(10) International Publication Number
WO 2009/010918 A2

(51) International Patent Classification:

G01S 15/89 (2006.01)

(21) International Application Number:

PCT/IB2008/052826

(22) International Filing Date: 14 July 2008 (14.07.2008)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:

60/950,371 18 July 2007 (18.07.2007) US

(71) Applicant (for all designated States except US): **KONINKLIJKE PHILIPS ELECTRONICS, N.V.** [NL/NL]; Groenewoudseweg 1, NL-5621 BA Eindhoven (NL).

(72) Inventors; and

(75) Inventors/Applicants (for US only): **GAUTHIER, Thomas** [FR/US]; P.O. Box 3003, Bothell, Washington 98041-3003 (US). **CRITON, Aline** [FR/FR]; PO Box 3003, Bothell, Washington 98041-3003 (US).

(74) Agent: **SCHOUTEN, Marcus, M.**; Philips Intellectual Property & Standards, High Tech Campus 44 PO Box 220, NL-5600 AE Eindhoven (NL).

(81) Designated States (unless otherwise indicated, for every kind of national protection available): AE, AG, AL, AM, AO, AT, AU, AZ, BA, BB, BG, BH, BR, BW, BY, BZ, CA, CH, CN, CO, CR, CU, CZ, DE, DK, DM, DO, DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, GT, HN, HR, HU, ID, IL, IN, IS, JP, KE, KG, KM, KN, KP, KR, KZ, LA, LC, LK, LR, LS, LT, LU, LY, MA, MD, ME, MG, MK, MN, MW, MX, MY, MZ, NA, NG, NI, NO, NZ, OM, PG, PH, PL, PT, RO, RS, RU, SC, SD, SE, SG, SK, SL, SM, ST, SV, SY, TJ, TM, TN, TR, TT, TZ, UA, UG, US, UZ, VC, VN, ZA, ZM, ZW.

(84) Designated States (unless otherwise indicated, for every kind of regional protection available): ARIPO (BW, GH, GM, KE, LS, MW, MZ, NA, SD, SL, SZ, TZ, UG, ZM, ZW), Eurasian (AM, AZ, BY, KG, KZ, MD, RU, TJ, TM), European (AT, BE, BG, CH, CY, CZ, DE, DK, EE, ES, FI, FR, GB, GR, HR, HU, IE, IS, IT, LT, LU, LV, MC, MT, NL, NO, PL, PT, RO, SE, SI, SK, TR), OAPI (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW, ML, MR, NE, SN, TD, TG).

Declaration under Rule 4.17:

— as to applicant's entitlement to apply for and be granted a patent (Rule 4.17(ii))

Published:

— without international search report and to be republished upon receipt of that report

(54) Title: SYSTEM AND METHOD FOR COINCIDENT B-MODE AND DOPPLER IMAGING USING CODED EXCITATION

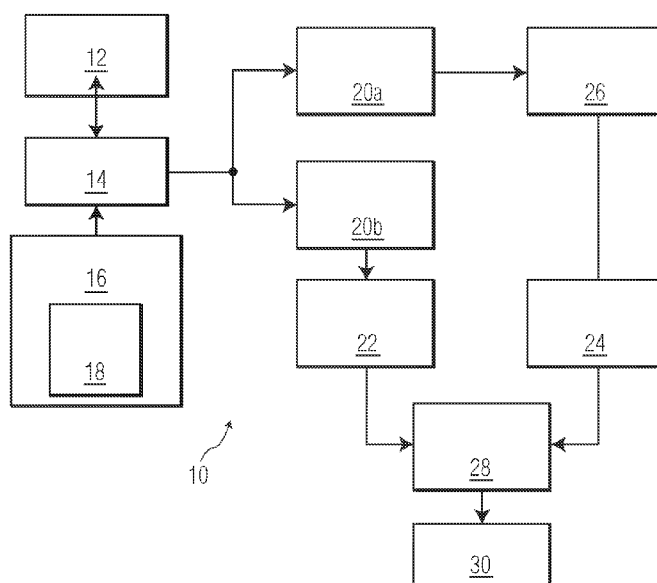


FIG. 1

(57) Abstract: An ultrasound system is disclosed wherein an encoded signal is emitted into a patient. A B-mode image is formed by decoding a harmonic of an echo signal and the Doppler image is formed by decoding the fundamental of the echo signal. Decoding includes convolving the echo signal with a time-reversed fundamental or harmonic of the code used to generate the encoded signal. The code may be a linear frequency modulated pulse.

**SYSTEM AND METHOD FOR COINCIDENT B-MODE
AND DOPPLER IMAGING USING CODED EXCITATION**

- [001]** This invention relates to ultrasound imaging systems and, more particularly to ultrasound imaging systems simultaneously forming B-mode and Doppler images.
- [002]** Ultrasonic pulse echo imaging of tissue structures, whereby received echo signals are amplitude detected and arranged in an image with depth displayed according to the round-trip transit time of transmitted ultrasound waves, is commonly referred to as B-mode imaging. B-mode imaging can be done at relatively high frame rates of display, since only one transmit pulse is needed to form one or more image lines (scanlines) of the display. Following transmission of an ultrasonic beam in a given direction, a sequence of echoes is received from along the beam direction, from the near field to the far field. Echoes from a number of such beams are amplitude detected and displayed adjacent to each other in relation to their transit time to form a two dimensional image of the structures that reflected the echoes.
- [003]** The frame rate of display for B-mode imaging is considerably higher than the frame rate of display for Doppler images such as power Doppler and colorflow images. This is because each Doppler image line must be interrogated a number of times in order to acquire a full line of Doppler data, which is needed to estimate the Doppler shift at points along the line. A set of lines of Doppler data acquired over time is referred to as an ensemble. The ensembles of data are needed to estimate the Doppler shift by fast Fourier transform or autocorrelation at each point along the line. The number of transmit pulses required to gather a full ensemble of samples at each scanline reduces the frame rate of display below that required to acquire the same image frame for B-mode display.
- [004]** The time required to form an ultrasonic image is even greater when the image is formed of two imaging modes. Colorflow images, for example are formed by acquiring both a B-mode image and a Doppler image, then presenting the final result as a composite of the two. In this way the flow of blood, displayed in the Doppler mode, is structurally depicted in its surrounding tissue and blood vessels by the B-mode display. The time required to form such an image is the time required to transmit each B-mode line and to receive echoes from along each line, plus the time required to transmit a plurality of Doppler pulses for each Doppler ensemble across the B-mode image, and to receive echoes in response to each Doppler transmission.

[005] Different types of transmit pulses are used for the B-mode and Doppler modes to optimize the information of each mode of imaging. For B-mode imaging, short transmit pulses are preferred because of the high axial resolution of the resulting echo samples. For Doppler imaging, where sensitivity and narrow transmit bands are generally high priorities, relatively long transmit pulses are employed. The time to produce one frame of a multi-mode image is thus the total of the transmit and receive times of both the B-mode and the Doppler signals, which is several multiples of the time required to scan the complete image field once.

[006] When the transmit pulses for B-mode and Doppler are time interleaved, the conventional technique to minimize beam steering changes, the time required to produce a single image frame is increased still further. This is because it is necessary to pre-condition the acoustic field for a particular type of pulse each time the type of pulse is changed to avoid image artifacts. For instance, suppose that a multi-mode image is to be produced using an eight-sample Doppler ensemble. Each line of the image must be scanned eleven times: a B-mode conditioning pulse is followed by a B-mode pulse, then a Doppler conditioning pulse is followed by the eight Doppler pulses. While the reduction in frame rate may be improved by using a small color box, thereby restricting Doppler scanning to only a portion of the full image width, the time required to form a complete image frame can become substantial when the color box occupies a substantial portion of the lateral dimension of the image. The frame rate of display is reduced correspondingly.

[007] The functionality of ultrasound systems concurrently forming Doppler and B-mode images is further reduced by peak-power limitations imposed by safety constraints. The voltages applied to the transducer can cause heating next to the patient's skin. Safety regulations therefore limit the amount of power that can be supplied to the transducer. This is a particularly significant drawback in the field of Tissue Harmonic Imaging (THI), wherein tissue density information is derived from a harmonic of the emitted signal resulting from non-linearities in the patient's tissues. Inasmuch as the harmonic band contains much less energy than the fundamental band, the excitation pulse may have a large amount of energy to improve the signal-to-noise ratio. Where THI is performed concurrently with Doppler imaging, the amount of energy of pulses used to form the Doppler image should be reduced to avoid excessive heating of the transducer. As a result, the quality of the Doppler images is reduced.

[008] In view of the foregoing, it would be advantageous to provide an improved system and method for concurrently forming Doppler and B-mode images.

- [009] In accordance with the principles of the present invention, a beamformer causes a transducer to emit an excitation signal encoded according to a code into a viewing area within a patient. The echo signal is received by the transducer, and the fundamental of the echo signal is decoded using the code. A harmonic of the echo signal is also decoded using the code. A Doppler image is formed using either the decoded fundamental or harmonic signal, and a B-mode image is formed using the other of the decoded fundamental or harmonic signals.
- [010] In another aspect of the invention, decoding the fundamental of the echo signal includes convolving the echo signal with a time-reversed fundamental of the code, and decoding the harmonic of the echo signal includes convolving the echo signal with a time-reversed harmonic of the code.
- [011] In another aspect of the invention, the code is a linear frequency modulated chirp.
- [012] In another aspect of the invention, decoding the fundamental and harmonic of the echo signal includes convolving the echo signal with multiple modulations of the coded signal and summing the convolutions. Where the code is a linear frequency modulated pulse, the modulations may be linear frequency modulated pulses having time varying frequencies bounded by different upper and/or lower frequencies and/or varying at a different rate than the code.
- [013] In the drawings:
- [014] Figure 1 is a schematic block diagram of an ultrasound system in accordance with an embodiment of the present invention.
- [015] Figure 2 is a plot illustrating a linear frequency modulated pulse in accordance with an embodiment of the present invention.
- [016] Figure 3A is a plot illustrating a time-reversed version of the linear frequency modulated pulse of Figure 2.
- [017] Figure 3B is a plot illustrating a time-reversed harmonic of the linear frequency modulated pulse of Figure 2.
- [018] Figure 4 is a schematic block diagram of a matched filter in accordance with an embodiment of the present invention.
- [019] Figures 5A-5C are plots illustrating segments of the time-reversed linear frequency modulated pulse of Figure 3B in accordance with an embodiment of the present invention.

[020] Figure 6 is a process flow diagram of a method for generating ultrasound images in accordance with an embodiment of the present invention.

[021] Figure 7 is a process flow diagram of a method for performing frequency compounding in accordance with an embodiment of the present invention.

[022] Referring to Figure 1, an ultrasound system 10 includes a transducer 12 for emitting and receiving ultrasonic signals. The transducer 12 may, for example, include piezoelectric transducer elements arranged in an array. The transducer 12 is driven by a beamformer 14. The beamformer 14 may apply signals to individual elements within the transducer 14 such that a single focused beam is emitted into a patient. The direction of the beam may be progressively varied to scan a viewing area within the patient. The beamformer 14 may be coupled to a beamformer controller 16 having a coding module 18. The controller 16 controls scanning of the beam across the viewing area by adjusting the phase and/or amplitude of signals applied to individual elements of the transducer 12. The coding module 18 generates a coded excitation signal that is adjusted as to duty cycle, phase and/or amplitude and applied to the elements of the transducer 12. The coded excitation signal may be a pulse having a duration longer than a typical ultrasound pulse and/or containing a wider range of frequencies. The coding used may be any known in the art such as Golay codes, Barker codes, or the like. In the illustrated embodiment, the coded excitation signal is a linear frequency modulated chirp as shown in Figure 2, in which the frequency increases or decreases with time. In the illustrated embodiment, the frequency increases at a constant rate. Although the coded signal and other waveforms referenced herein may include other frequencies due to their finite durations, frequency for purposes of this disclosure refers to the center or primary frequency of the coded signal or waveform.

[023] The transducer 12 converts the coded signals from the beamformer 14 into ultrasonic signals emitted into a viewing area within a patient along the ultrasound beam. The elements of the transducer 12 receive echo signals from the viewing area and convert them to electrical signals. The beamformer 14 receives the electrical signals and sums them. In some systems, the beamformer 14 may adjust the phase and/or amplitude of signals from individual transducer elements prior to summing them in order to bring them into phase coherence according to the origination of the echo signals along the ultrasound beam.

[024] The output of the beamformer 14 is processed by two or more matched filters 20a, 20b. The matched filters 20a, 20b extract information from the output of the beamformer 14, which is then used by a B-mode processor 22 and Doppler processor 24 to form B-mode and Doppler images, respectively, of the viewing area. The output of the

matched filter 20a may be stored in an ensemble store 26 before being processed by the Doppler processor 24. The ensemble store 26 stores the output of the matched filter 20a for multiple scans of the viewing area prior to processing by the Doppler processor until sufficient scans (preferably six to eight) have been performed to create a Doppler image. The output of the B-mode processor 22 and Doppler processor 24 are input to an image processor 28 that generates an image viewable on a display 30.

[025] Referring to Figures 3A and 3B, the matched filters 20a, 20b may perform a convolution of the output of the beamformer 14 with another waveform to extract information. In the illustrated embodiments, the matched filters 20a, 20b convolve the output of the beamformer 14 with a time-reversed version of the coded signal produced by the beamformer controller 16. Accordingly, where the coded signal is a linear frequency modulated chirp, the output of the beamformer 14 may be convolved with a linear frequency modulated chirp in which the frequency change with time is in a direction opposite the direction of frequency change in the coded signal. As shown in Figure 3A, a linear frequency modulated chirp having a frequency that decreases with time may be used where the emitted signal has the form of the linear frequency modulated chirp of Figure 2.

[026] In some embodiments, either the Doppler or the B-mode image is produced using a harmonic of the emitted signal that is generated by nonlinearities in the tissues from which the emitted signal reflects. Accordingly, one of the matched filters 20a, 20b convolves the output of the beamformer 14 with a time-reversed harmonic of the coded signal. For embodiments where the coded signal is a linear frequency modulated chirp, the output of the beamformer 14 may be convolved with the waveform of Figure 3B, which is a time-reversed second harmonic of the pulse of Figure 2. In a preferred embodiment, the B-mode image is formed using the time-reversed harmonic of the coded signal.

[027] The system and methods for ultrasound imaging disclosed herein facilitate the use of the same pulses to form both B-mode and Doppler images, thereby increasing the frame rate as compared to prior systems and methods that use separate pulses to form each type of image. In prior systems, different pulses are used to form the Doppler and B-mode images inasmuch as the B-mode image can achieve greater axial resolution with shorter pulses whereas the Doppler images use longer pulses to achieve a narrower transmit band. Using a coded pulse provides the benefits of a longer pulse while at the same time allowing for time compression of the echo signal, which maintains axial resolution in the B-mode image.

[028] As a result of the longer pulse duration, the amount of energy emitted into the viewing area during the pulse is greater, which improves penetration of the pulse and therefore improves far-field image quality. The longer pulse duration of the system and method disclosed also enables the transducer 12 to transmit greater energy during a pulse without increasing peak power. Inasmuch as the transducer 12 is positioned adjacent a patient's skin, the peak power supplied to the transducer 12 may need to be limited to avoid causing burns. In prior systems, high peak power pulses needed to be separated by lower power pulses to avoid causing excessive heating of the transducer. For example, tissue harmonic imaging (THI) generates a tissue density (grayscale) image using a harmonic of the reflected ultrasonic waves. Inasmuch as the reflected harmonic signals are weak relative to the fundamental, the peak power of the excitation pulse had to be high. As a result, the peak power of pulses used to generate the Doppler images had to be reduced. Coded excitation as described above avoids this problem and allows both tissue harmonic and Doppler imaging with low peak power pulses that still achieve good penetration while maintaining axial resolution.

[029] Referring to Figure 4, one or both of the matched filters 20a, 20b may convolve the output of the beamformer 14 with multiple waveforms in order to improve the resolution of a resultant image. For example, one or both of the matched filters 20a, 20b may include multiple filters 32a-32c each of which outputs a convolution of a waveform with the output of the beamformer 14. The outputs of the filters 32a-32c may then be summed, averaged, or otherwise combined, such as by a summer 34, to produce the output of the matched filter 20a, 20b. In some embodiments, the filters 32a-32c are quadrature bandpass (QBP) filters having coefficients 36 chosen such that the output is a convolution of the output of the beamformer 14 with a waveform.

[030] As ultrasound beams are scattered by a patient's tissues, the scattered waves tend to constructively and destructively interfere to form a speckle pattern in the resultant image. Convolution of the output of the beamformer with multiple distinct waveforms produces multiple distinct speckle patterns. Combining these speckle patterns tends to average out the speckle and improve the quality of the resultant image.

[031] In some embodiments, the filters 32a-32c convolve the output of the beamformer 14 with a waveform that is similar to the time-reversed coded signal or its harmonic but that has been modified as to one or more parameters to produce a slightly different waveform. For example where the coded signal is a linear frequency modulated chirp, the waveform convolved by the filter 32a-32c may be a linear frequency modulated

chirp that is bounded by different upper and/or lower frequencies and/or that changes in frequency at a faster or slower rate than the time-reversed coded signal or its harmonic. A major portion of each of the frequency ranges of the linear frequency modulated chirps of the filters 32a-32c preferably overlaps the frequency range of the fundamental or harmonic of the time-reversed coded signal. In other embodiments, the frequency ranges of the linear frequency modulated chirps of the filters 32a-32c lie within the frequency range of the fundamental or harmonic of the time-reversed coded signal.

[032] In some embodiments, one or more of the upper frequency, lower frequency, and rate of frequency change of the linear frequency modulated chirp of a filter 32a-32c differ from the corresponding parameters of the time-reversed coded signal and the linear frequency modulated chirps of the other filters 32a-32c by between five and fifty percent. In other embodiments, one or more of the upper frequency, lower frequency, and rate of frequency change of the linear frequency modulated chirp of a filter 32a-32c differ from the corresponding parameters of the time-reversed coded signal and the linear frequency modulated chirps of the other filters 32a-32c by between ten and twenty percent.

[033] For the matched filter 20a, 20b that extracts information from the harmonic of the coded signal, the filters 32a-32c may convolve the output of the beamformer 14 with linear frequency modulated chirps in which one or more of the upper frequency, lower frequency, and rate of frequency change differ from the corresponding parameters of the time-reversed harmonic of the coded signal and the linear frequency modulated chirps of the other filters 32a-32c by between five and fifty percent. In other embodiments, one or more of the upper frequency, lower frequency, and rate of frequency change of the linear frequency modulated chirps differ from the corresponding parameters of the time-reversed harmonic of the coded signal and linear frequency modulated chirps of the other filters 32a-32c by between ten and twenty percent.

[034] In a preferred embodiment, the linear frequency modulated chirps convolved by the filters 32a-32c with the output of the beamformer 14 are segments of the time-reversed fundamental or harmonic of the coded signal. For example, where the fundamental or harmonic of the time-reversed coded signal is a function of time $f(t)$ and has a pulse duration T , the filters 32a, 32b, and 32c may convolve the output of the beamformer 14 with segments of the coded signal or its harmonic such as from $f(A \times T)$ to $f(T)$, $f(B \times T)$ to $f(C \times T)$, or $f(0)$ to $f(D \times T)$, where A , B , C , and D are each less than one and C is greater than B .

[035] Figures 5A-5C illustrate segments of the time-reversed coded signal such as may be convolved by the filters 32a-32c with the echo signal. Segments of the harmonic of

the time-reversed coded signal may also be used. Figure 5A illustrates a pulse including the high frequency segment of the time-reversed coded signal. Figure 5B illustrates the low frequency segment of the time-reversed coded signal. Figure 5C illustrates the a middle portion of the time-reversed coded signal.

[036] In some embodiments, the filter 32a convolves the echo signal with from twenty to seventy percent (measured in terms of time) of the initial period of the time-reversed coded signal, and the filter 32b convolves the echo signal with the last twenty to seventy percent of the time-reversed coded signal, and the filter 32c convolves the echo signal with from thirty to one hundred percent of the middle portion of the time-reversed coded signal. Preferably the matched filter 20a coupled to the Doppler processor 24 has filters 32a-32c that convolve the echo signals with segments of the time-reversed signal whereas the matched filter 20b coupled to the B-mode processor 22 convolves the echo signals with segments of the harmonic of the time-reversed coded signal segmented as described above with respect to the time-reversed coded signal.

[037] The waveforms including the segments of the time-reversed coded signal or its harmonic, such as the segments of Figures 5A-5C, may have the same duration as the linear frequency modulated pulse of the coded signal or its harmonic. The segments may also have the same time position in the waveform convolved by the filter 32a-32c relative to the beginning of the waveform as the corresponding segment of the time-reversed coded signal or its harmonic has relative to the beginning of the time-reversed coded signal or its harmonic. The waveform of Figures 5A may therefore include a zero amplitude portion 38 following a segment of the time-reversed coded signal or the harmonic of the time-reversed coded signal and the waveform 5B may include zero amplitude portions 40 preceding a segment of the time-reversed coded signal or the harmonic of the time-reversed coded signal. The wave form of Figure 5C may include zero amplitude portions 42 on either side of a middle segment of the time-reversed coded signal or its harmonic.

[038] The zero amplitude portions may advantageously facilitate registration of the output of the filters 32a-32c. Registration compensates for the fact that due to the linear frequency modulation of the coded signal, one of the low and high frequency components will arrive at the transducer 12 earlier in time than the other, depending on the direction of frequency modulation. A filter 32a-32c convolving the echo signal with a waveform including only a portion of the time-reversed coded signal or its harmonic may therefore shift the output relative to a filter 32a-32c having a waveform corresponding to a different portion of the time-reversed coded signal or its harmonic. Accordingly, the portions of the coded

signal (or the harmonic of the coded signal) that are omitted from the waveforms of Figures 5A-5C may be replaced with zero amplitude portions of the same duration, as illustrated, such that the duration of the convolved waveforms are the same and the output of the filters 32a-32c are not shifted with respect to one another. Stated differently, the zero amplitude portions can introduce a delay into one or more of the filters 32a-32c such that the outputs are not shifted with respect to one another.

[039] Referring to Figure 6, a method 44 for producing ultrasound images may include generating coded signals, such as a linear frequency modulated pulses at step 46. At step 48, the coded signals are beamformed and, at step 50, the beamformed signals are transmitted into a viewing area of a patient by a transducer. At step 52, the echo signals are received and, at step 54, the echo signals are beamformed to bring the signals from individual elements of the transducer into phase coherence. At step 56, the echo signals are convolved with the time-reversed coded signal. At step 58, the result of the convolution is stored in an ensemble store.

[040] At step 60, the method 44 includes evaluating whether sufficient scans (preferably six to eight) have been stored to generate a Doppler image. If so, then a Doppler image of the viewing area is generated at step 62 using the convolutions of step 56.

[041] At step 64, the echo signals are convolved with the harmonic of the time-reversed coded signal and at step 66 the convolutions are used to generate a B-mode image of the viewing area. At step 68 the B-mode image is displayed. The Doppler image is also displayed at step 68 if sufficient echo signals have been received to generate a Doppler image.

[042] Steps 48-56 and 64 may be performed multiple times for each iteration of the method 44 for a plurality of scan lines to produce a two-dimensional scan of the viewing area. The method 44 may also be repeated to generate multiple images of a viewing area.

[043] Referring to Figure 7, convolving the echo signal with the coded signal or its harmonic at steps 56 and 64 may be performed according to a method 70. At step 72, the echo signal is convolved with a first modulation of the time-reversed coded signal or its harmonic, such as the segment shown in Figure 5A. At step 74, the echo signal is convolved with a second modulation of the time-reversed coded signal or its harmonic, such as the segment shown in Figure 5B. At step 76, the echo signal is convolved with a third modulation of the time-reversed coded signal or its harmonic, such as the segment shown in Figure 5C. At step 78, the convolutions of steps 72-76 are combined, such as by summing, weighted averaging, or the like.

WHAT IS CLAIMED IS:

1. A method for performing an ultrasound scan comprising:
generating a coded signal according to a code;
emitting the coded signal from a transducer into a viewing area of a patient;
receiving an echo signal from the viewing area;
decoding a harmonic signal of the echo signal according to the code;
decoding a fundamental signal of the echo signal according to the code;
generating a density image of the viewing area using one of the decoded harmonic signal and the decoded fundamental signal; and
generating a Doppler image of the viewing area using the other of the decoded harmonic signal and the decoded fundamental signal.
2. The method of claim 1, wherein decoding the harmonic signal of the echo signal comprises:
calculating a harmonic of the code
time reversing the harmonic of the code; and
convolving the echo signals with the time-reversed harmonic of the code.
3. The method of claim 2, wherein decoding the fundamental signal of the echo signal comprises time reversing the code and convolving the echo signals with the time-reversed code.
4. The method of claim 2, wherein the harmonic is the second harmonic.
5. The method of claim 1, wherein the code is a linear frequency modulated chirp.
6. The method of claim 1, wherein decoding the fundamental signal of the echo signal comprises:
convolving the echo signal with modulations of a time-reversed version of the code to generate a plurality of convolutions; and
summing the convolutions.

7. The method of claim 6, wherein the code is a linear frequency modulated chirp having a monotonically time varying frequency bounded by first and second frequencies and wherein the modulations of the time-reversed version of the code comprise linear frequency modulated chirps having a monotonically time varying frequency bounded by first and second frequencies, at least one of the first and second frequencies of the modulations being different from the first and second frequencies, respectively, of the code.

8. The method of claim 7, wherein at least one of the first and second frequencies of each of the modulations of the time-reversed version of the code differ from the first and second frequencies, respectively, of the code by between about five and fifty percent.

9. The method of claim 8, wherein at least one of the first and second frequencies of each of the modulations of the time-reversed version of the code differ from the first and second frequencies, respectively, of the code by between about ten and twenty percent.

10. A method for performing an ultrasound scan comprising:
generating a coded signal according to a first linear frequency modulated chirp, the linear frequency modulated chirp having a time varying frequency varying monotonically in a first direction within a first frequency range;
emitting a coded signal from a transducer into a viewing area of a patient;
receiving an echo signal from the viewing area;
generating a first convolution of the echo signal by convolving the echo signal with a second linear frequency modulated chirp having a time varying frequency varying monotonically in a second direction opposite the first direction within a second frequency range, the first and second frequency range overlapping over a major portion thereof;
generating a second convolution of the echo signal by convolving the echo signal with a third linear frequency modulated chirp having a time varying frequency varying monotonically in a second direction opposite the first direction within a third frequency range, the third frequency range overlapping the first frequency range over a major portion thereof;
summing the first and second convolutions;

generating an ultrasound image according to the sum of the first and second convolutions.

11. The method of claim 10, wherein the second and third linear frequency modulated pulses are segments of the first frequency modulated pulse.

12. The method of claim 11, wherein the second linear frequency modulated pulse includes a segment of a beginning portion of the first frequency modulated pulse and the third linear frequency modulated pulse includes a segment of an ending portion of the first frequency modulated pulse.

13. The method of claim 11, wherein the second and third frequency modulated pulses have a length substantially equal that of the first linear frequency modulated pulse and wherein second linear frequency modulated pulse includes a trailing zero amplitude portion and wherein the third linear frequency modulated pulse includes a leading zero amplitude portion.

14. An ultrasound system comprising:
a signal generator operable to generate a coded signal according to a code;
a transducer coupled to the signal generator and operable to emit an ultrasonic signal according to the coded signal and to receive an echo signal;
a first matched filter coupled to the transducer and operable to generate a decoded harmonic of the echo signal;
a second matched filter coupled to the transducer and operable to generate a decoded fundamental of the echo signal; and
an image generator coupled to the first and second matched filter and operable to generate Doppler and tissue images according to the decoded fundamental and harmonic of the echo signal.

15. The ultrasound system of claim 14 wherein the image generator is operable to generate the Doppler images from the decoded fundamental of the echo signal.

16. The ultrasound system of claim 14 wherein the image generator is operable to generate the tissue images from the decoded harmonic of the echo signal.

17. The ultrasound system of claim 14 wherein the first matched filter convolves the echo signal with a time-reversed harmonic of the code to generate the decoded harmonic of the echo signal.

18. The ultrasound system of claim 14 wherein the time-reversed harmonic of the code is a time-reversed second harmonic of the code.

19. The ultrasound system of claim 14 wherein the second matched filter convolves the echo signal with a time-reversed version of the code to generate the decoded fundamental of the echo signal.

20. The ultrasound system of claim 14 wherein the code is a linear frequency modulated chirp.

21. The ultrasound system of claim 14 wherein the second matched filter is operable to convolve the echo signal with modulations of a time-reversed version of the code to generate a plurality of convolutions and to sum the convolutions to generate the decoded fundamental of the echo signal.

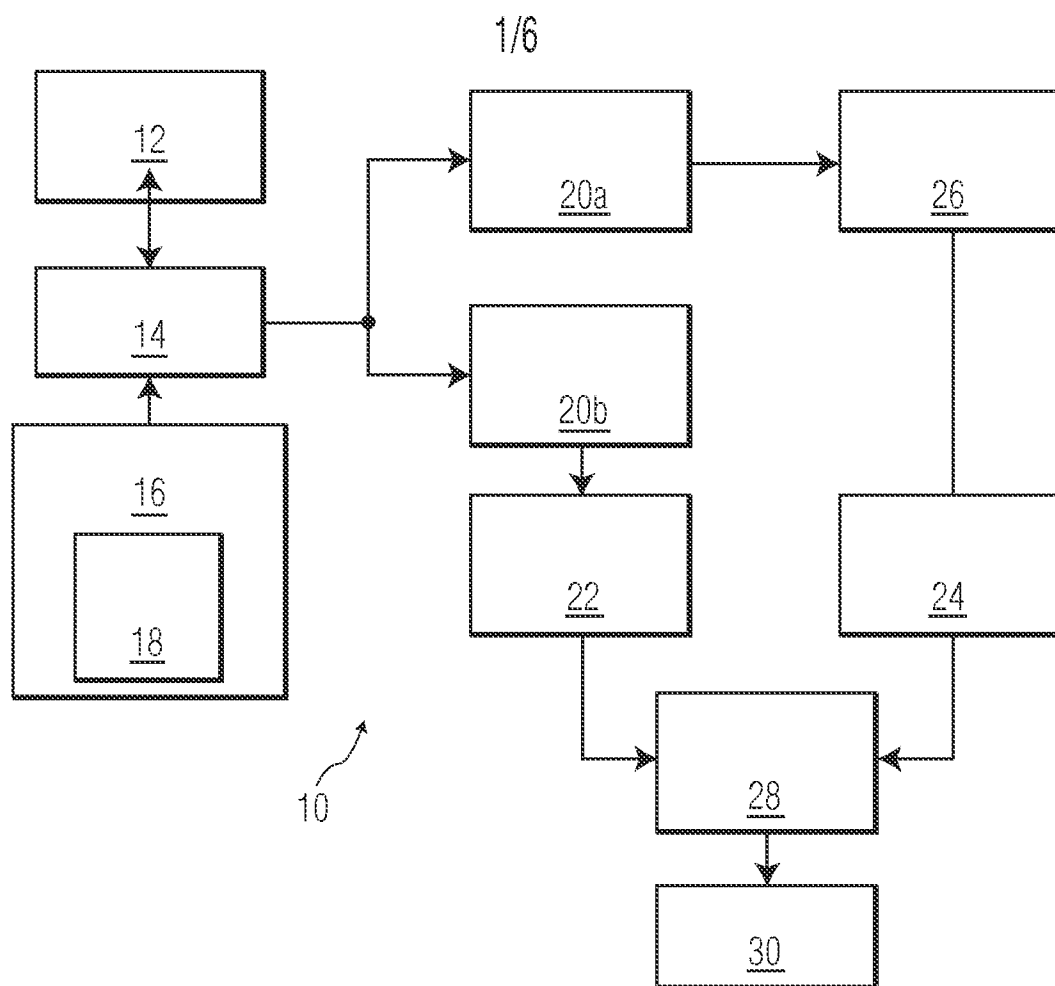


FIG. 1

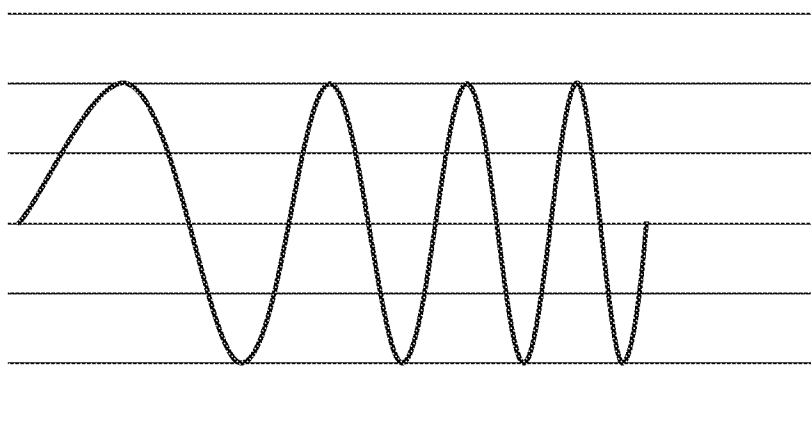


FIG. 2

2/6

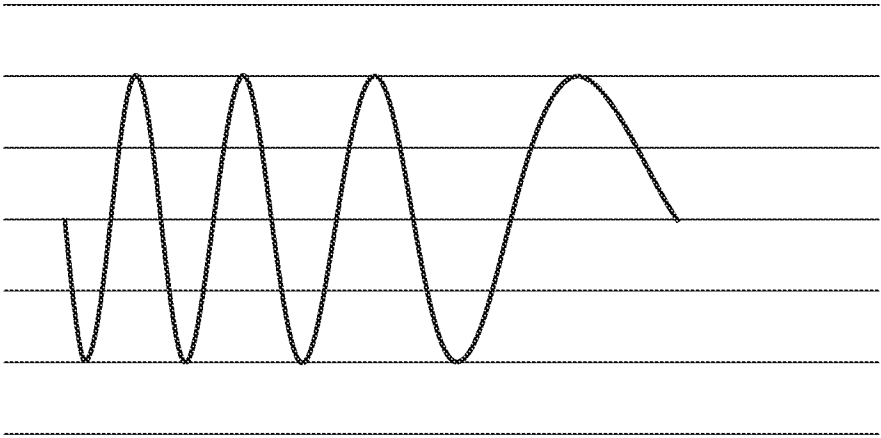


FIG. 3A

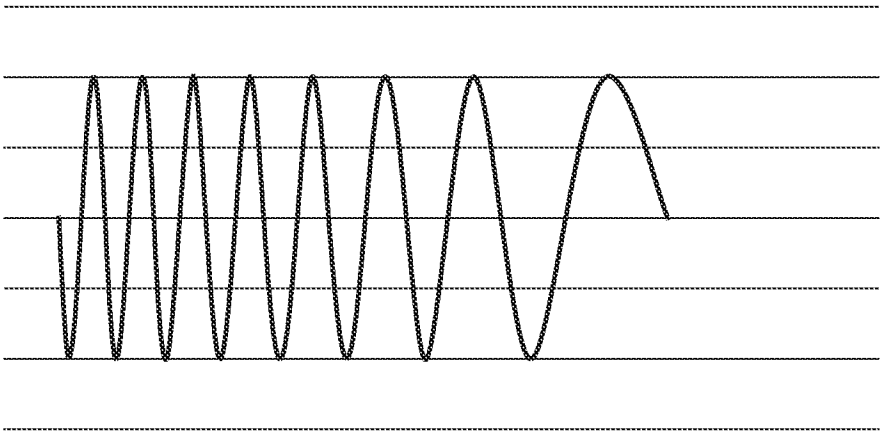


FIG. 3B

3/6

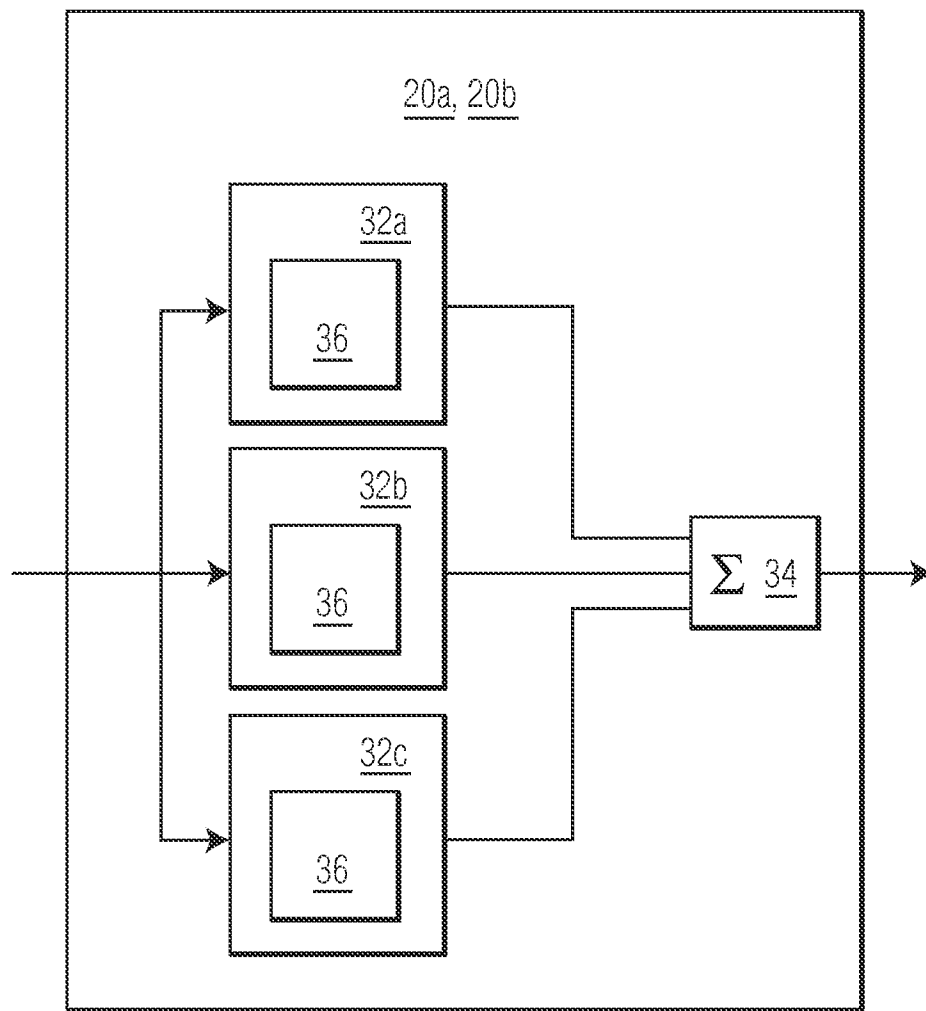


FIG. 4

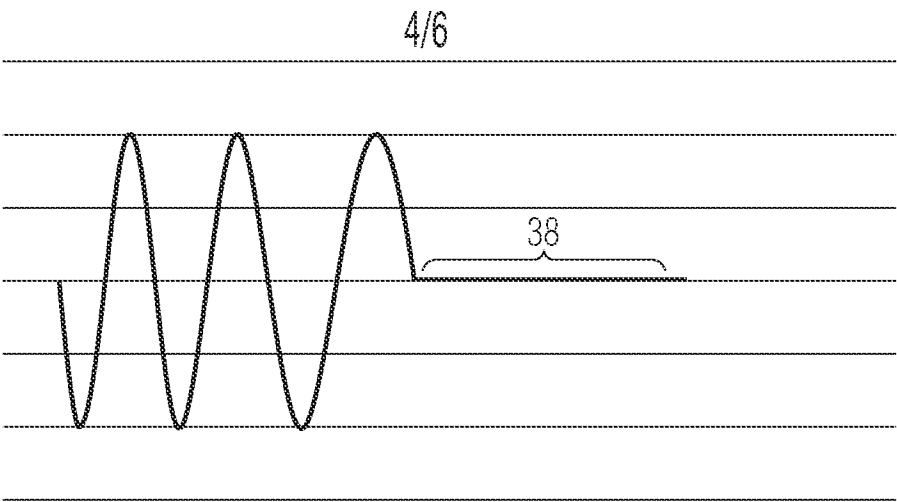


FIG. 5A

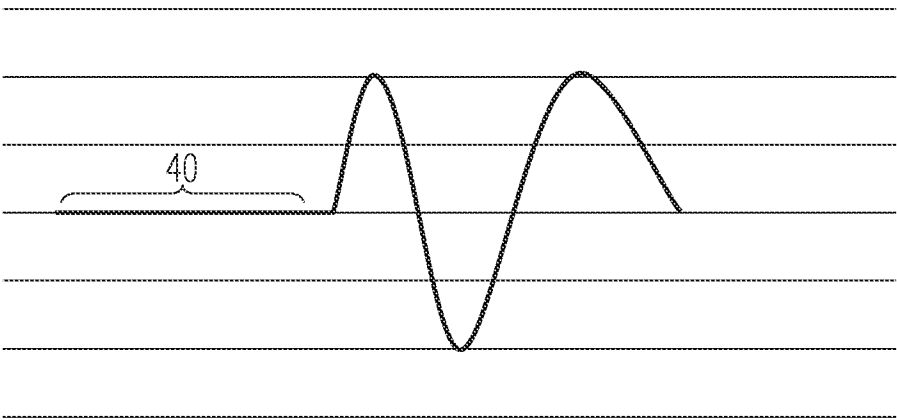


FIG. 5A

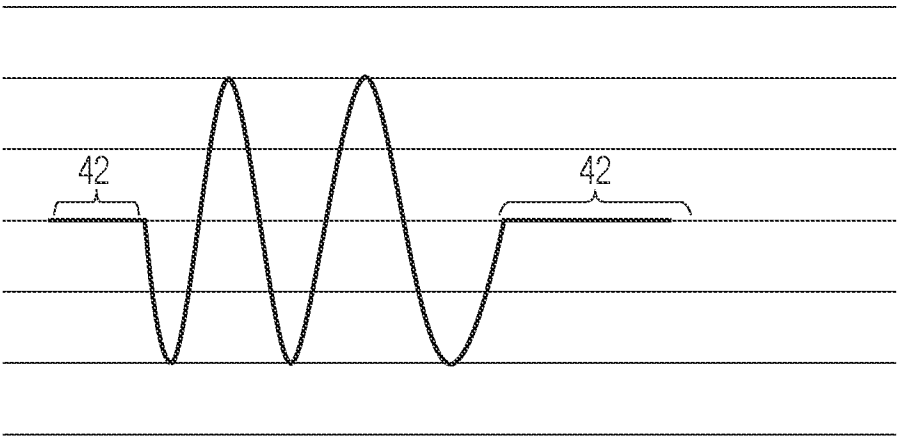
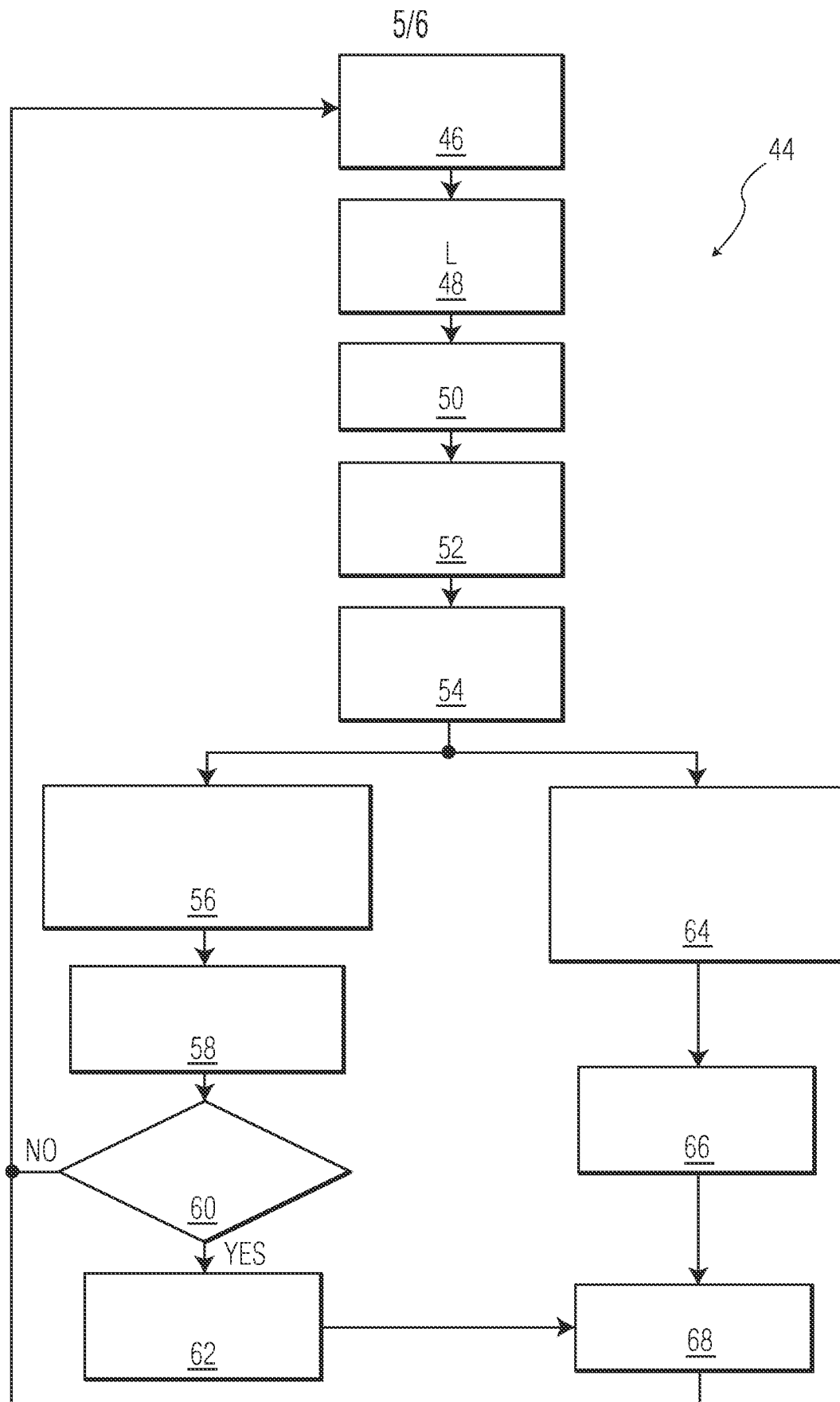


FIG. 5C



6/6

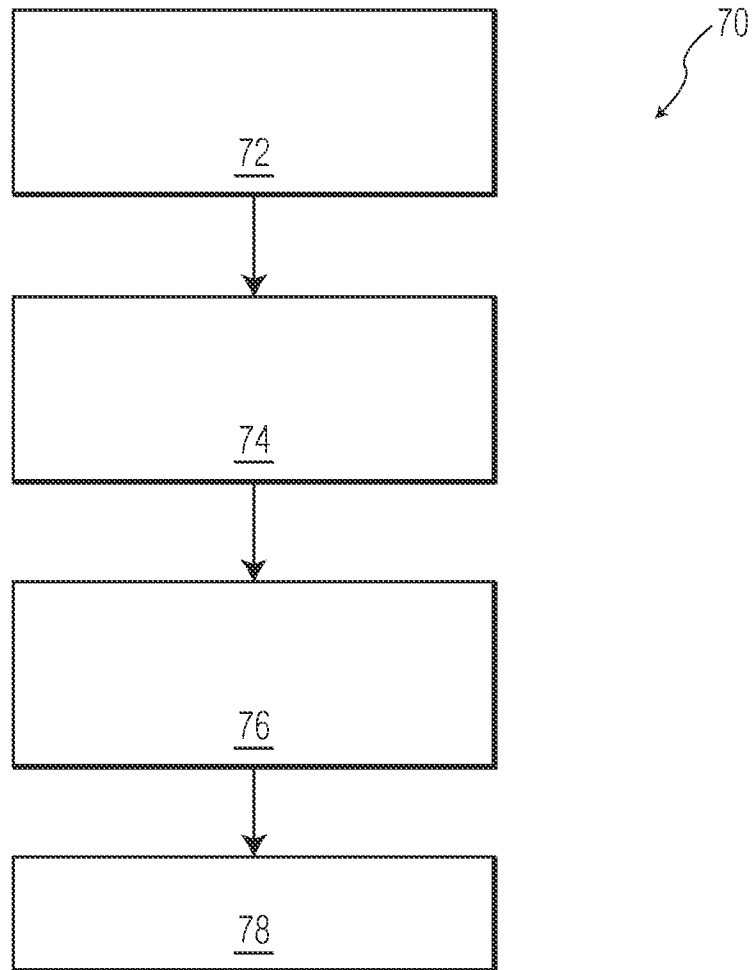


FIG. 7