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(54) **Title:** ULTRASONIC DIAGNOSTIC IMAGING OF NONLINEARLY INTERMODULATED AND HARMONIC FREQUENCY COMPONENTS

(57) **Abstract:** An ultrasonic diagnostic imaging system and method are described for performing nonlinear echo signal imaging with harmonic and intermodulation product (sum or difference frequency) components. Both the harmonic and the intermodulation products are produced by nonlinear effects of tissue or contrast agents and both are advantageously separated from the fundamental transmit components of the echo signals by pulse inversion processing. The use of both nonlinear components can improve the signal to noise ratio of the ultrasonic images, and the two types of components can be blended or used in different regions of an image to offset the effects of depth dependent attenuation.

Ultrasonic diagnostic imaging of nonlinearly intermodulated and harmonic frequency components

This invention relates to ultrasonic diagnostic imaging systems and, in particular, to ultrasonic diagnostic imaging systems which produce images using component frequencies which have been produced by a nonlinear transmission medium or target within the body.

5   Harmonic imaging is in widespread use in ultrasonic imaging today because of the advantages in signal segmentation and image clarity which it provides. Harmonic imaging at present has two applications: harmonic contrast imaging and tissue harmonic imaging, both of which involve nonlinear signal components. Examples of these two types of harmonic imaging may be found in U.S. Patents 5,833,613 (for harmonic contrast) and

10   5,879,303 (tissue harmonic) of which I am a co-inventor. Harmonic contrast imaging provides the advantage of sharp signal segmentation and signal-to-clutter improvement due to the fact that harmonic contrast agents return relatively strong harmonic signals in response to fundamental frequency insonification. These relatively strong harmonic contrast signals are readily distinguished from the fundamental frequency and relatively low level harmonic

15   signals returned from tissue and other substances in the body. Tissue harmonic imaging, while having a signal-to-noise deficit as compared to fundamental frequency imaging, provides an advantage in image clarity through a reduction in image clutter. Since the distortion of acoustic waves passing through tissue which gives rise to the harmonic components only begins to develop and build as the waves travel deeper into the body, near

20   field scatterers which are a source of image clutter can scatter only the low or insignificant levels of harmonic energy present at shallow depths. Thus, tissue harmonic images will exhibit reduced clutter as compared to fundamental frequency images, although at reduced signal levels due to the lower level of the harmonic components and by reason of depth dependent attenuation of the higher frequency harmonic signals. It would be desirable to

25   utilize nonlinear signals which afford the foregoing advantages but with greater signal levels, better signal-to-clutter ratios, broader bandwidths, and reduced depth dependent attenuation.

In accordance with the principles of the present invention, ultrasonic imaging is performed by transmitting an ultrasound beam with two or more different frequency components. When the beam passes through a nonlinear transmission medium or encounters

a nonlinear scatterer the different frequency components intermodulate and develop sum and difference frequencies, as well as multiple frequencies of the fundamental, which are contained in returning echo signals, detected and used to form an ultrasonic image. In accordance with one aspect of the present invention, the transmitted frequency components

5   are located on opposite sides of the transducer peak response (center) frequency, with a received difference frequency signal located in the vicinity of the peak response point of the transducer characteristic. In accordance with another aspect of the present invention, the amplitudes of the transmitted frequencies are chosen in consideration of the effects of depth dependent attenuation and/or the properties of contrast agent microbubbles. In accordance

10   with a further aspect of the present invention, the sum or difference frequencies are matched with a harmonic or subharmonic of one of the transmitted frequencies, thereby providing a receive signal containing nonlinear signal energy from both harmonic and intermodulation effects. In accordance with yet another aspect of the present invention, the sum and difference signals, being produced by nonlinear effects, are separated from the linear transmit

15   signals by the pulse inversion process. In accordance with yet a further aspect of the present invention, a transmit beam includes multiple frequency components for the production of a broadband echo signal of multiple sum and difference frequency components.

20   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;

Figs. 2a and 2b illustrate two beam transmitters suitable for use in the embodiment of Figure 1;

25   Fig. 3 illustrates the transmit and receive bands of a transducer operated in accordance with the present invention;

Fig. 4 illustrates some of the variable transmit parameters of a multifrequency transmitter of the present invention;

Fig. 5 illustrates another set of transmit and receive bands of a transducer

30   operated in accordance with the present invention;

Figs. 6a-6d illustrate beam patterns of another embodiment of the present invention;

Fig. 7 illustrates another embodiment of the present invention which uses both intermodulation product signals and second harmonic signals;

Fig. 8 illustrates an embodiment of the present invention which uses nonlinear sum and harmonic components to reduce the problem of depth dependent attenuation;

Fig. 9 illustrates an embodiment of the present invention which uses nonlinear difference and harmonic components to reduce the problem of depth dependent attenuation;

5　　　　　　Figs. 10a and 10b illustrate a multifrequency transmit pulse waveform and a spectrum of that pulse;

Figs. 11a and 11b illustrate a low amplitude (low nonlinearity) microbubble response to the pulse of Figure 10a and the spectrum of the echo returned by that response;

Figs. 12a and 12b illustrate echoes from a nonlinear microbubble response to 10　the pulse of Figure 10a and its spectrum;

Fig. 13 illustrates the nonlinear portion of the spectrum of Figure 12b which is separated by pulse inversion processing;

Figs. 14a and 14b illustrate the spectra of transmit pulses designed for microbubbles of two different sizes;

15　　　　　　Figs. 15a and 15b illustrate a pulse waveform for a three frequency transmit beam and the spectrum of the pulse;

Figs. 16a and 16b illustrate a frequency modulated multifrequency transmit pulse and the spectrum of the pulse;

Fig. 17 illustrates the use of pulse inversion to separate sum or difference 20　signal components from a multiline receiver; and

Fig. 18 illustrates the use of pulse inversion to separate sum or difference signal components using spatially distinct transmit beams.

25　　　　　　Referring first to Figure 1, an ultrasound system constructed in accordance with the principles of the present invention is shown in block diagram form. This system operates by scanning a region of the body being imaged with ultrasonic transmit beams each having two or more frequency components f_1, f_2, etc. As each beam is transmitted along its steered path through the body, the multiple frequency components of the beam are 30　intermodulated by the nonlinear effects of the tissue through which the beam passes or the nonlinear response of a contrast agent microbubble encountered by the beam. This intermodulation of the multiple frequency components of the beam produces signal components which are at the sum and difference frequencies of the transmit frequencies, that

is, f_1+f_2 and f_1-f_2. The sum or difference signal components are contained in echoes returned from the body and are used to produce the ultrasonic image of the scanned region.

Sum and difference frequencies have been used in ultrasound systems in the past. These applications have generally involved the simultaneous transmission of two or more beams by separate transducers to a specific target location at which the beams interact with each other to produce sum and difference signals which are then usually detected by a third transducer. In U.S. Pat. 3,987,673 an arrangement of this type is used for a Doppler flowmeter to produce Doppler shifted signals at the intersection point with reduced susceptibility to angle of incidence variation. In U.S. Pat. 5,601,086 the beat frequencies at the interaction volume of two intersecting beams are used to measure the flow rate of a contrast agent, and in U.S. Pat. 5,903,516 the beat force generated at the intersection of two beams generates an acoustic wave that can be detected by a receiver. All of these arrangements require precise aiming of the transmitting transducers so that the beams will intersect at the desired location. They all require multiple transducers, usually two for transmitting and a third for receiving. Furthermore, each transmission only elicits information from a single area in the image field, the area in which the multiple beams intersect. The intersection point must be steered over the entire image field in order to scan the full field. By contrast, an embodiment of the present invention can utilize only a single transmit beam which is steered through the image region in the normal manner and does not have to intersect any other beam. Since only a single beam is necessary, only a single mechanically steered transducer or electronically steered array transducer is needed, although the present invention can also be practiced with multiple transducers if desired. Furthermore, as the beam travels through the body along its beam path, the sum and difference signal components of the different frequency components of the beam are reflected from points in the body all along the beam path, enabling a full scanline of image information to be acquired by each transmission, rather than just information from a single area of the image field.

An embodiment of the present invention can make efficient use of the finite bandwidth of the scanhead transducer used for imaging. In harmonic imaging of tissue and contrast agents, the transmit and receive bands must both be contained within the transducer passband so that the single transducer scanhead can be used for both transmission and reception of the differing transmit and receive bands. As shown in my U.S. Pat. 5,879,303 in Figs. 7 and 8, this generally results in the transmit band being located at the lower end of the transducer passband and the receive band being located at the upper end of the transducer passband. Unlike fundamental frequency imaging, where the identical transmit and receive

bands are both located in the center of the transducer passband where response is optimal, neither of the operative bands is at the center of the transducer band.

An embodiment of the present invention can overcome this problem of suboptimal response. Figure 3 illustrates a representative transducer passband 60. In this example the transducer is transmitting a beam with two transmit bands 62 and 64, one located at 1.4 MHz and another located at 4.2 MHz. As the beam passes through the tissue of the body, and when it encounters a contrast agent microbubble, the nonlinear effects of these substances cause the two frequencies of the transmit beam to mix or intermodulate and produce sum and difference frequencies in the manner of a parametric array, which are returned in echoes of the transmit beam. In the example of Figure 3, the difference frequency of the two transmit frequencies is (4.2-1.4), which is 2.8 MHz as shown by the receive band 66. As the passband 60 shows, this echo frequency falls at or around the center of the transducer passband, where the transducer exhibits its most sensitive response. Thus, the transmit frequencies are located at the upper and lower portions of the transducer passband, but for transmission the passband rolloff is acceptable due to the power level of the transmit beam. The received echoes are many dB lower in intensity than the transmit beam, and are advantageously received at the most sensitive region of the transducer passband.

Figure 1 utilizes a transmitter 16 which transmits multifrequency beams for the nonlinear generation of sum and difference signals within the body. The transmitter is coupled by a transmit/receive switch 14 to the elements of an array transducer 12 of a scanhead 10. The transmitter is responsive to a number of control parameters which determine the characteristics of the transmit beams, as shown in the drawing. Figure 4 illustrates these parameters spectrally in relation to the passband 60 of the transducer 12. The two frequencies f_1 and f_2 of the multifrequency beam are controlled, which determines the frequency differential Δf between the two frequency components of the beam, as well as the frequencies at which the sum (f_1+f_2) and difference (f_1-f_2) frequency components will fall. Also controlled are the amplitudes or intensities a and b of the two transmitted frequency components, causing the transmit beam to be of the form $(b\sin(2\pi f_1 t)+a\sin(2\pi f_2 t))$. The received difference signal component (f_1-f_2) will have an amplitude c as illustrated by the size of band 66 in Figure 4. This amplitude c is not a linear product of the a and b intensities, however, as the sum and difference signals result from nonlinear effects.

Figure 5 illustrates an embodiment of the present invention which utilizes sum signal components. In this example the transmit beam includes two frequency bands 72 and 74 located at 1.2 MHz and 1.6 MHz. These transmit frequency components are nonlinearly

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intermodulated or mixed to result in an echo sum signal band 76 located at 2.8 MHz. The transmit frequencies will also have a difference component of 0.4 MHz which is below the lower cutoff of the transducer passband 60. The sum components 76 are seen to fall at the center of the transducer passband 60 and are used for imaging by the ultrasound system.

5   Returning to Figure 1, the transducer array 12 receives echoes from the body containing sum and difference frequency components which are within the transducer passband. These echo signals are coupled by the switch 14 to a beamformer 18 which appropriately delays echo signals from the different elements then combines them to form a sequence of sum and/or difference signals along the beam from shallow to deeper depths.

10   Preferably the beamformer is a digital beamformer operating on digitized echo signals to produce a sequence of discrete coherent digital echo signals from a nearfield to a farfield depth of field. The beamformer may be a multiline beamformer which produces two or more sequences of echo signals along multiple spatially distinct receive scanlines in response to a single transmit beam. The beamformed echo signals are coupled to a nonlinear signal

15   separator 20. The separator 20 may be a bandpass filter which passes a sum or difference passband 66,76 to the relative exclusion (attenuation) of the transmit bands 62,64 or 72,74. In the illustrated embodiment the separator 20 is a pulse inversion processor which separates the nonlinear signals including the sum and difference components by the pulse inversion technique. Since the sum and difference frequency signals are developed by nonlinear effects,

20   they may advantageously be separated by pulse inversion processing. For pulse inversion the transmitter has another variable transmit parameter which is the phase of the transmit pulse as shown in the drawing. The ultrasound system transmits two or more beams of different transmit polarities or phases. For the illustrated two pulse embodiment, the scanline echoes received in response to the first transmit pulse are stored in a Line1 buffer 22. The scanline

25   echoes received in response to the second transmit pulse are stored in a Line2 buffer 24 and then combined with spatially corresponding echoes in the Line1 buffer by a summer 26. Alternatively, the second scanline of echoes may be directly combined with the stored echoes of the first scanline without buffering. As a result of the different phases or polarities of the transmit pulses, the out of phase fundamental (linear) echo components will cancel and the

30   nonlinear sum and difference components, being in phase, will combine to reinforce each other, producing enhanced and isolated nonlinear sum and/or difference frequency signals. The sum and difference frequency signals may be further filtered by a filter 30 to remove undesired signals such as those resulting from operations such as decimation. The signals are then detected by a detector 32, which may be an amplitude or phase detector. The echo

signals are then processed by a signal processor 34 for subsequent grayscale, Doppler or other ultrasound display, then further processed by an image processor 36 for the formation of two dimensional, three dimensional, spectral, parametric, or other display. The resultant display signals are displayed on a display 38.

5   Figures 2a and 2b illustrate two detailed embodiments of the transmitter 16 of Figure 1. In the embodiment of Figure 2a, the waveforms for the different frequency components of the transmit pulse are formed separately in digital operations, then combined to form a composite multifrequency transmit signal for application to a transducer element. An f_1 generator 42 produces the f_1 transmit signal component and an f_2 generator 44 produces

10   the f_2 transmit signal component. The generators produce their respective transmit waveforms in response to input control parameters such as f_1 Sel. and f_2 Sel. shown in the drawing which determine the f_1 and f_2 frequency components of the transmitted beam. Other variable input parameters (not shown) may be intensity parameters a and b, and phase or polarity parameters for pulse inversion transmit signals. Alternatively, the output waveforms

15   produced by the generators 42 and 44 may be varied in amplitude and phase or polarity before or after being combined by a combiner 46 into a composite transmit pulse which contains the multiple transmit frequency components. In Figure 2a the waveforms produced by the generators are weighted by digital weighting processor circuits 43 and 45 which apply the weighting factors a and b to the generated waveforms. The weighting circuits can take the

20   form of digital multipliers and the sign of the weighting factor (+1,-1) can be used to control the polarity of the output waveform. The composite transmit pulse is applied to a D/A converter 48 for conversion to an analog signal, which may be further amplified and filtered as desired and used to drive a transducer element 12'.

Figure 2b illustrates a second transmitter embodiment in which the composite

25   multifrequency transmit signals are produced in advance, then stored in a waveform library 50, which may be in the form of a digital memory. When a certain multifrequency transmit pulse is desired it is selected from the library 50, weighted by a weighting circuit 51, and stored in a transmit register 52. When the transmitter is triggered to transmit the beam the composite multifrequency waveform is shifted out of the transmit register 52 by a clock

30   signal CLK, converted to an analog signal by the D/A converter 48 and applied to the transducer element 12'. The amplitude of the transmit pulse may be varied by either a digital multiplier preceding the A/D converter such as one used in the weighting circuit, or by an analog amplifier following the A/D converter, and may be filtered in either the analog or digital domain as desired. The individual frequency components may not be separately

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adjusted in amplitude following the transmit register 52 as the waveform is already a composite at that point in this embodiment.

It will be appreciated that the beam transmitted by the transducer array is steered in a desired beam direction and focused at a desired depth of focus, both of which are effected by the timing of application of transmit waveforms to different elements of the transducer array. Accordingly a number of transmit channels such as those shown in Figures 2a and 2b are employed in the transmitter, one for each differently timed transmit waveform. The transmit channels may use the same transmit waveform but the times at which the waveforms of the channels are applied to the transducer elements are varied by the time delay profile needed for steering and focusing from the active elements of the transmit aperture.

An embodiment of the present invention can provide improved lateral resolution and reduced clutter by virtue of the differing beam profiles of the various frequency components. This is illustrated by the beam profiles of Figures 6a-6d. Figure 6a illustrates a beam profile for a lower f_2 frequency component having a relatively broad main lobe 80 and sidelobes 82. The sidelobes, which are a source of clutter as explained in my '303 patent, are x_2 dB below the peak of the main lobe as shown in the drawing. Figure 6b illustrates a beam profile for a higher f_1 frequency component. The main lobe 90 of the f_1 frequency component is relatively narrower than that of the f_2 frequency component by virtue of the higher f_1 frequency. In this illustration the sidelobes 92 are x_1 dB below the main lobe level and for purposes of this example it is assumed that $x_1=x_2$. Figure 6c illustrates a beam profile for a resultant sum or difference signal component, in this case, the difference frequency component of f_1-f_2. Since this beam profile is a product of the two components of Figures 6a and 6b and is for a frequency greater than the f_2 frequency in this example, the main lobe 100 of the difference frequency is relatively narrower than the f_2 main lobe 80. Furthermore the sidelobes 102 which are a source of clutter are x_3 dB below the level of the main lobe 100 where $x_3>x_1=x_2$. Thus there is an improvement in resolution and clutter performance over the use of the low frequency beam component alone. There is also an improvement in penetration over the high frequency component due to the reception at the lower difference frequency.

Figures 7-9 illustrate embodiments which combines the benefits of both parametric array difference imaging, harmonic imaging and sum frequency imaging. In the embodiment of Figure 7 the transmit beam includes a high frequency component f_1 exhibiting a band BW_1, and a low frequency component f_2 exhibiting a band BW_2. The frequencies of the transmit beam are chosen so that two types of nonlinearly developed

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components will be located in the vicinity of the center of the transducer passband 60: difference frequency components (f_1-f_2) and second harmonic components $2f_2$ of the lower transmit frequency f_2 as shown in this example, and/or a subharmonic of the upper transmit frequency such as $\frac{1}{2}f_1$. While each of these nonlinear components will exhibit its own

5　relatively narrow band, careful selection of the transmit frequencies can result in a combined receive band BW_c which is broader than the transmitted bands. This means that the echo passband will be more broadband in nature for better signal resolution, and that the echo signals will exhibit an improved signal-to-noise level since they are a combination of both nonlinear effects. The received echo signals of this embodiment will also exhibit enhanced

10　resolution and clutter performance as illustrated by Figure 6d, which shows a main lobe 110 and sidelobes 112 for a harmonic echo component $2f_2$. As my '303 patent explains the harmonic beam profile will exhibit improved clutter and signal-to-clutter performance as the sidelobes 112 are x_4 dB below the level of main lobe 110, where $x_4 > x_1 = x_2$, which is especially beneficial in cases of highly inhomogeneous aberrating tissue layers. Thus, the

15　resultant echo signals can exhibit improved performance where $x_3 \geq x_4 > x_1, x_2$ and the signal-to-noise performance of the difference beam profile is x_3 dB. It is seen therefore that the embodiment of Figure 7 combines the performance benefits of both parametric and harmonic imaging in a broadband imaging embodiment.

　　　　　One of the problems which harmonic imaging presents is the attenuation of

20　high frequency harmonics as the high frequency signals pass through tissue. This depth dependent attenuation limits the signal penetration and hence the image depth during harmonic imaging. In the embodiment of Figure 8, a combination of harmonic and sum frequency imaging is used to address this problem. The transmit beam includes transmit frequency components f_1 and f_2 as shown in the drawing. These components are at the lower

25　portion of the transducer passband 60 and their low frequencies provide good depth penetration in tissue. During reception three echo signal bands are received in the central, more sensitive portion of the transducer passband 60: a harmonic $2f_2$ of the f_2 transmit component, a sum signal $f_1 + f_2$, and a harmonic $2f_1$ of the f_1 transmit component. Each of these bands is progressively lower in frequency than the previous band, resulting in

30　progressively better depth penetration with each band. The three echo signal bands can be received throughout the depth of the image, or they can be selectively used over different depths to produce an image which is a blend of harmonic and sum echo signal components. For example, echo signals in the $2f_2$ band can be received to form the image at shallow depths, the sum signals in band $f_1 + f_2$ can be used to produce the image an intermediate depth,

and the $2f_1$ signal band can be used to form the image at the deepest image depths, thereby taking advantage of the relative penetration performance of each band. To reduce noise a tracking filter can be employed which moves from a passband encompassing the $2f_2$ band as echoes are initially received at shallow depths, then moves progressively lower as echoes are

5   received from a beam at greater and greater depths. An example of frequencies which may be employed for f_1 and f_2 in a constructed embodiment of Figure 8 are 1.2 MHz and 1.6 MHz, resulting in receive signal bands at 3.2 MHz, 2.8 MHz, and 2.4 MHz.

       Figure 9 illustrates an embodiment which addresses the problem of depth dependent attenuation by using a combination of harmonic and difference frequency echo

10   signals. The transmit beam in this drawing contains frequency components f_1 and f_2 as shown. The received echo signals include signals in band $2f_1$, the harmonic of the transmit frequency f_1, and signals in band $f_2 - f_1$, the difference frequency band. The $2f_1$ high frequency signals are used to produce the near field (shallow depth) portion of the image, providing the clutter reduction which is characteristic of harmonic imaging. Since the $2f_1$ high frequencies

15   are more greatly affected by depth dependent attenuation, the far field (greater depth) portion of the image is formed from the low frequency $f_2 - f_1$ difference frequency signals. A variation is to use a blend of both the high and low frequencies in an intermediate range between the near and far field depths. As in the previous embodiment a tracking filter could be used to move the receive signal passband from higher to lower frequencies as the echo signals are

20   received from ever increasing depths to reduce noise in the image. An example of frequencies which may be employed for f_1 and f_2 in a constructed embodiment of Figure 9 are 3.0 MHz and 5.0 MHz, resulting in the harmonic signals being located in a 6.0 MHz band and the difference frequency signals in a 2.0 MHz band.

       Furthermore, the parametric and harmonic echo components, both being

25   formed by nonlinear processes, may both be advantageously separated from the transmitted signal components in the previous embodiments by either bandpass filtering or, preferably, the pulse inversion technique. This is shown by echoes resulting from two multifrequency transmit beams which differ in phase by 180° (π):

       (+): $\sin \omega_1 t + \sin \omega_2 t$ and

30   (-): $\sin(\omega_1 t + \pi) + \sin(\omega_2 t + \pi)$

       where the two transmit frequencies are $\omega 1 t$ and $\omega 2 t$ and (+) and (-) indicate the phase difference between the two beams. When the echo signals from the two beams are combined

by the pulse inversion technique to separate the second harmonic components the following results:

$\sin(\omega_1 t + \pi) = -\sin \omega_1 t$, and therefore the combination of both echoes is $\sin \omega_1 t - \sin \omega_1 t = 0$, which cancels the fundamental transmit component $\omega_1 t$. The harmonic components of the

5   two echoes are $\sin 2\omega_1 t$ and $\sin(2\omega_1 t + 2\pi) = \sin 2\omega_1 t$, which combine to produce $2\sin 2\omega_1 t$, the separated second harmonic of $\omega_1 t$. When the two echoes are combined to produce difference signals, the following results:

$\sin(\omega_1 t - \omega_2 t)$ and

$\sin(\omega_1 t + \pi - \omega_2 t - \pi) = \sin(\omega_1 t - \omega_2 t)$ which combine to produce the result of

10   $2\sin(\omega_1 t - \omega_2 t)$. Thus, the parametric difference signal is, like the pulse inversion separated harmonic, enhanced to twice the level of the desired component in a single echo signal. The embodiments of Figures 7, 8, and 9, when utilized with pulse inversion separation, thus provide the enhanced benefits of both parametric and harmonic imaging.

An exemplary acoustic pressure pulse 200 which may be employed in an

15   embodiment of the present invention is shown in Figure 10a. The spectrum of this transmit pulse is shown in Figure 10b and has two components f_1 and f_2 as shown by bands 202 and 204, located about frequencies of 1.4 MHz and 3.6 MHz in this example. When the multifrequency pulse 200 is transmitted in a nonlinear medium such as microbubbles or tissue the nonlinear response will have components that are harmonics of the fundamental

20   transmit bands: f_1, $2f_1$, $3f_1$, etc. and f_2, $2f_2$, $3f_2$, etc. The nonlinear response will also have components of the sum and difference frequencies of the transmit pulse and their harmonics: f_1+f_2, $2(f_1+f_2)$, $3(f_1+f_2)$, etc. and f_2-f_1, $2(f_2-f_1)$, $3(f_2-f_1)$, etc. The harmonic beam has the characteristics described above which make it more desirable than the fundamental frequency beam for many applications, including narrower beam-width, lower sidelobes, and less

25   aberration artifacts due to the cumulative generation process which gives rise to the harmonic components. The difference frequency beam is also formed by the cumulative generation process and enjoys many of the same benefits as the harmonic beam, and in addition generally exhibits lower sidelobes and a lower absorption characteristic than the harmonic beam.

30   When the acoustic pulse 200 is used to insonify microbubbles at low amplitudes which are insufficient to cause nonlinear bubble oscillation, the oscillation of the bubble radius will appear similar to that shown by curve 210 in Figure 11a. This example assumes that the microbubble is a 3.2 μm microbubble in blood which is insonified by a

pressure of 0.01 MPa. The echo returned by the microbubble will be linear in nature and have a frequency content with two spectral peaks 212,214 as shown in Figure 11b, which closely resembles the spectrum of the transmit pulse shown in Figure 10b, and also the bubble linear resonance characteristic.

5   When the amplitude of the transmitted acoustic pulse is increased so that nonlinear bubble behavior results, the radius of the microbubble will oscillate nonlinearly as illustrated by the radius-time curve 216 in Figure 12a. In this example the acoustic pressure is assumed to be 0.1 MPa. The spectrum of the echo returned by the microbubble is shown in Figure 12b. This spectrum includes the two transmitted frequency components in bands 212

10   (where f_1=1.4 MHz) and 214 (where f_2=3.6 MHz). The spectrum also includes a peak 222 at the second harmonic of f_1 (where $2f_1$=2.8 MHz) and a peak 220 at the difference frequency between f_1 and f_2 (where f_2-f_1=(3.6-1.4)=2.2 MHz). When two echoes are obtained from transmit pulses of opposite phase or polarity and combined, the odd harmonics including fundamental frequencies are eliminated, leaving only the even harmonics and sum and

15   difference frequencies in bands 220,222 as shown in Figure 13.

        Another advantage of the present invention when used with contrast agents is that the multiple transmit frequencies will excite microbubbles of multiple sizes. Most contrast agents in present use have microbubbles with diameters of 1-10 μm. Smaller bubbles respond better to higher frequency excitation pulses and larger bubbles respond better to

20   lower frequencies. The amplitudes of the respective frequency components of the transmit pulse can therefore be chosen to preferentially excite microbubbles of a given size or range of sizes. This is done by the choice of coefficients *a* and *b* in the expressions shown above. For example, Figures 14 show spectra of echoes from microbubbles with different sizes when insonified with pulses containing two frequency components as shown in Figures 10a and

25   10b. In this example small (linear) amplitude relationships are used for simplicity of illustration. In Figure 14a the response of a 1 μm microbubble is shown, whereas in Figure 14b a 4 μm bubble response is illustrated. The smaller bubble of Figure 14a has a larger response 214' at the higher frequency and a smaller response 212' at the lower frequency. The larger bubble of Figure 14b has a larger response 212" at the lower frequency and a

30   smaller response 214" at the higher frequency. When a broad range of frequencies is transmitted, a broad range of bubble sizes will be excited. When used for tissue harmonic imaging rather than contrast agents, where there are no bubbles, the higher band 214' can be of a greater transmit intensity than the lower band 212' to offset the effects of depth dependent frequency attenuation, which attenuates higher frequencies to a greater degree than

lower frequencies. Different ratios of coefficients *a* and *b* for the f_1 and f_2 transmit components can be used for different depth zones or a multiple transmit zone focus embodiment. For example the *a* coefficient for the high frequency component can remain constant or decline with depth while the *b* coefficient for the low frequency component is

5   increased with depth or deeper zones. These amplitude variations for high and low frequency components can be used individually or together in a particular embodiment.

Figures 15a and 15b illustrate a transmit pulse waveform 230 and spectrum for a multifrequency transmit beam having three frequency components. As the spectrum of Figure 15b illustrates, the pressure pulse of Figure 15a has three frequency peaks 232, 234,

10   and 236, located at f_1=1.4 MHz, f_2=2.0 MHz, and f_3=3.6 MHz. These transmit frequency components will produce nonlinear echo components including:

$$f_3-f_2 = 1.6 \text{ MHz}$$

$$f_3-f_1 = 2.2 \text{ MHz}$$

$$2f_1 = 2.8 \text{ MHz}$$

15     $$f_1+f_2 = 3.4 \text{ MHz and}$$

$$2f_2 = 4.0 \text{ MHz},$$

all of which can be encompassed within the transducer passband 60. Thus, it is seen that a multifrequency transmit pulse of three constituent frequency components can result in a received echo signal of broad bandwidth. As above, the linear components of the received

20   echo signals can be removed by pulse inversion processing.

Another embodiment of the present invention is illustrated by Figures 16a and 16b. In this embodiment the transmit pulse is frequency modulated (an FM pulse, also known as a chirp pulse.) An acoustic pressure waveform 240 which is frequency modulated from a lower to a higher frequency is shown in Figure 16a. The spectrum of the illustrated waveform

25   has a nominal center frequency of 2.1 MHz as shown in Figure 16b. The nonlinear results of interaction of the different frequency components (harmonics, sum and difference frequencies) of such a pulse can fill the complete transducer passband with echo components ranging from 1.6 to 4.0 MHz.

Figure 17 illustrates an embodiment employing multiline reception, that is,

30   reception of multiple receive lines in response to a single transmit event, with pulse inversion separation of sum or difference signals. In this embodiment an array transducer 12 transmits a beam T_1 with two frequency components which is expressed as $\sin\omega_1 t + \sin\omega_2 t$. In response to this transmit beam two receive beams are produced, R_A and R_B, which are located laterally on either side of the transmit beam T_1. A second multifrequency beam T_2 is

transmitted which is out of phase with the first transmit beam and is expressed as

$\sin(\omega_1 t + \pi) + \sin(\omega_2 t + \pi)$. Again, two receive beams pare produced in response to this

second transmit beam which are laterally located on either side of the T_2 transmit beam. In

the illustrated embodiment the two transmit beams are laterally spatially aligned and the

receive beams are aligned on either side of the transmit beam locations. To separate the sum

or difference signal components the aligned receive beams are combined by pulse inversion,

that is, $R_A + R_C$ and $R_B + R_D$. Due to the phase difference of the transmit waveforms the

fundamental components will cancel and the sum or difference signals will reinforce each

other. In the embodiment of Figure 17 it is seen that two spatially distinct receive scanlines of

sum or difference signals have been created from two transmit events, a 1:1 correspondence

and an improvement over conventional two (or higher) transmit event pulse inversion. Higher

order multiline can result in even greater numbers of received scanlines per transmit event

and hence even greater frame rates of display.

Figure 18 illustrates an embodiment where laterally spatially distinct transmit

beams are employed to produce spatially intermediate sum or difference receive beams. A

transmit beam T_1 of the form $\sin \omega_1 t + \sin \omega_2 t$ is first transmitted and a beam R_A is received

which is in alignment with the T_1 transmit beam. A second transmit beam T_2 of the form

$\sin(\omega_1 t + \pi) + \sin(\omega_2 t + \pi)$ is transmitted at a spatially distinct location along the aperture of

the transducer 12 and a beam R_B is received in alignment with the second transmit. The two

receive beams R_A and R_B are combined by pulse inversion ($R_A + R_B$) to form a receive beam

of sum or difference signal components at an intermediate spatial location indicated at 300. A

succession of laterally spaced transmit beams which alternate in phase as shown by T_1 and T_2

can be used to produce separated sum or difference scanlines between each pair of transmit

beam locations, thereby resulting in approximately a 1:1 ratio of transmit beams to separated

sum or difference scanlines.

Instead of receiving beams R_A and R_B in alignment with the transmit beams as

previously discussed, the received beams can be steered to the same intermediate location as

shown by co-aligned beams R_A' and R_B' in Figure 18. The two receive beams are then

combined by pulse inversion to produce the sum or difference signal scanline at the

intermediate location. It will further be appreciated that multiline reception can be used to

receive two or more beams on either side of transmit beams T_1 and T_2 in Figure 18 followed

by pulse inversion separation of sum or difference signals, resulting in an even greater ratio

of received scanlines to transmit beams and an even greater display frame rate.

I have described a nonlinear imaging technique which uses both the high and low end of the transducer bandwidth for excitation and listens (receives) at the center of the passband where sensitivity is greatest. The received bandwidth is effectively increased by listening at both the second harmonic and the difference frequency. Narrower bandwidths may be used for excitation so that more energy is delivered to a target for better signal-to-noise characteristics without sacrificing the resolution afforded by multi-component received echo signals. The principles of the present invention are fully applicable to three dimensional imaging, where sum or difference signal components and/or harmonic components are produced from insonification of a volumetric region.

CLAIMS:

1.   An ultrasonic imaging method utilizing nonlinear signal components comprising:

transmitting ultrasonic waves into a region to be imaged, the waves containing at least two different frequency components;

5   receiving echo signals in response to the waves containing harmonic and intermodulated signal components produced by nonlinear effects of materials in the region to be imaged; and

producing a two or three dimensional ultrasonic image utilizing the harmonic and intermodulated signal components.

10

2.   The ultrasonic imaging method of Claim 1, wherein transmitting and receiving are performed by an array of transducer elements exhibiting a transducer passband having a center frequency;

wherein transmitting comprises transmitting a beam with at least one of the

15   transmitted different frequency components being more remotely located from the center of the transducer passband than at least one of the received signal components.

3.   The ultrasonic imaging method of Claim 2, wherein transmitting comprises transmitting a beam with two different frequency components which are more remotely

20   located from the center of the transducer passband than the two received signal components.

4.   The ultrasonic imaging method of Claim 1, wherein receiving comprises receiving harmonic and difference frequency signal components.

25   5.   The ultrasonic imaging method of Claim 1, wherein receiving comprises receiving harmonic and sum frequency signal components.

6.   The ultrasonic imaging method of Claim 1, wherein transmitting comprises transmitting a chirp signal.

7.   The ultrasonic imaging method of Claim 1, wherein transmitting comprises transmitting ultrasonic waves exhibiting two different frequency components of different relative intensities.

5

8.   The ultrasonic imaging method of Claim 1, wherein receiving comprises receiving two or more spatially different beams of echo signals in response to the transmitted beam, the received beams containing harmonic and intermodulated signal components produced by nonlinear effects of materials in the region to be imaged

10

9.   The ultrasonic imaging method of Claim 1, further comprising separating harmonic and intermodulated signal components of received echo signals from fundamental frequency components by pulse inversion processing.

15   10.   The ultrasonic imaging method of Claim 9, wherein transmitting comprises transmitting two differently phased ultrasonic beams into the region to be imaged, each beam containing two or more different frequency components; and

           wherein receiving comprises receiving echo signals in response to each beam which contain fundamental transmit frequency components and harmonic and intermodulated
20   signal components.

11.   The ultrasonic imaging method of Claim 10, wherein transmitting further comprises transmitting two spatially aligned, differently phased ultrasonic beams.

25   12.   The ultrasonic imaging method of Claim 11, wherein receiving further comprises receiving a beam of echo signals in response to each transmitted beam which is spatially aligned with the transmit beam location.

13.   The ultrasonic imaging method of Claim 11, wherein receiving further
30   comprises receiving a plurality of beams of echo signals in response to each transmitted beam, at least one of which received beams is laterally offset from the transmit beam location.

14. The ultrasonic imaging method of Claim 13, wherein receiving further comprises receiving a plurality of beams of echo signals in response to a second transmitted beam at locations which are aligned with the locations of beams received in response to a first transmitted beam.

5

15. The ultrasonic imaging method of Claim 10, wherein transmitting further comprises transmitting two laterally separate, differently phased ultrasonic beams.

16. The ultrasonic imaging method of Claim 15, wherein receiving further comprises receiving a beam of echo signals in response to each transmitted beam which is spatially aligned with the transmit beam location.

10

17. The ultrasonic imaging method of Claim 15, wherein receiving further comprises receiving a plurality of beams of echo signals in response to each transmitted beam, at least one of which received beams is laterally offset from the transmit beam location.

15

18. The ultrasonic imaging method of Claim 17, wherein receiving further comprises receiving a plurality of beams of echo signals in response to a second transmitted beam, at least one of which is at a location which is aligned with the location of a beam received in response to a first transmitted beam.

20

19. The ultrasonic imaging method of Claim 15, wherein receiving further comprises receiving a beam of echo signals in response to each transmitted beam which is laterally offset from the transmit beam location.

25

20. The ultrasonic imaging method of Claim 19, wherein receiving further comprises receiving a beam of echo signals in response to a second transmitted beam at a location which is aligned with the location of a beam received in response to a first transmitted beam.

30

21. An ultrasonic diagnostic imaging system comprising:

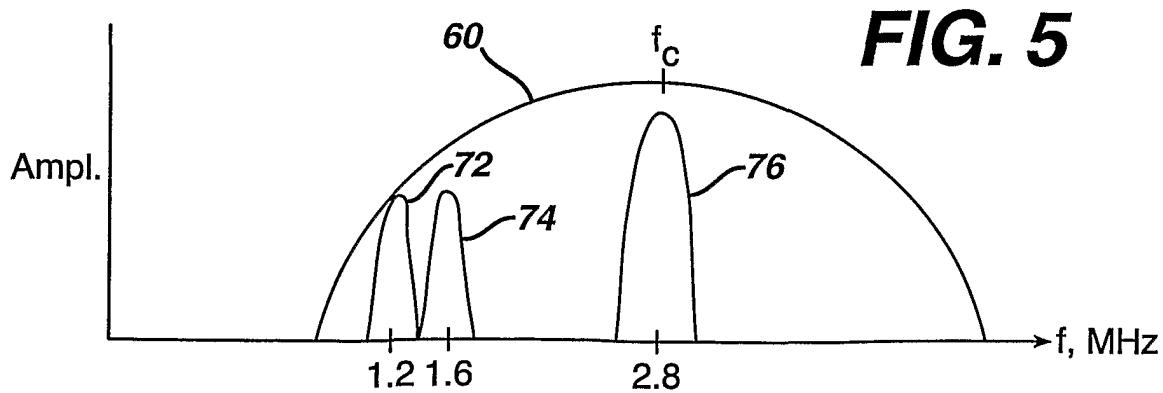a transducer array exhibiting a passband encompassing the frequencies of two different transmit frequency components, a harmonic receive frequency, and the frequency of an intermodulation product of the two different transmit frequency components;

a transmitter, coupled to the transducer array which actuates the transducer array to transmit a beam containing two different transmit frequency components;

a receiver coupled to the transducer array which produces coherent echo signals at a harmonic of a transmit frequency component and of an intermodulation product of the two different transmit frequency components;

an image processor, coupled to the receiver, which produces ultrasonic image signals from a harmonic of a transmit frequency component and of an intermodulation product of the two different transmit frequency components; and

a display.

22.   The ultrasonic diagnostic imaging system of Claim 21, wherein the image processor further comprises a pulse inversion processor which separates harmonic and intermodulated signal components of received echo signals from fundamental frequency components.

23.   The ultrasonic diagnostic imaging system of Claim 22, wherein the ultrasonic image signals produced by the image processor exhibit different proportions of harmonic to intermodulated signal components in different regions of an ultrasonic image field.

24.   The ultrasonic diagnostic imaging system of Claim 23, wherein the ultrasonic image signals produced by the image processor exhibit a greater proportion of harmonic signal components at shallower depths and a greater proportion of intermodulated signal components at deeper depths.

25.   The ultrasonic diagnostic imaging system of Claim 23, further comprising a tracking filter, responsive to harmonic and intermodulated signal components of received echo signals which act to pass different proportions of harmonic and intermodulated signal components for echo signals of different depths.

26.   The ultrasonic diagnostic imaging system of Claim 21, wherein the frequencies of the different transmit frequency components are more centrally located in the

passband than the harmonic receive frequency and the frequency of the intermodulation
product.

27.   The ultrasonic diagnostic imaging system of Claim 21, wherein the harmonic
receive frequency and the frequency of the intermodulation product are more centrally
located in the passband than the frequencies of the different transmit frequency components.

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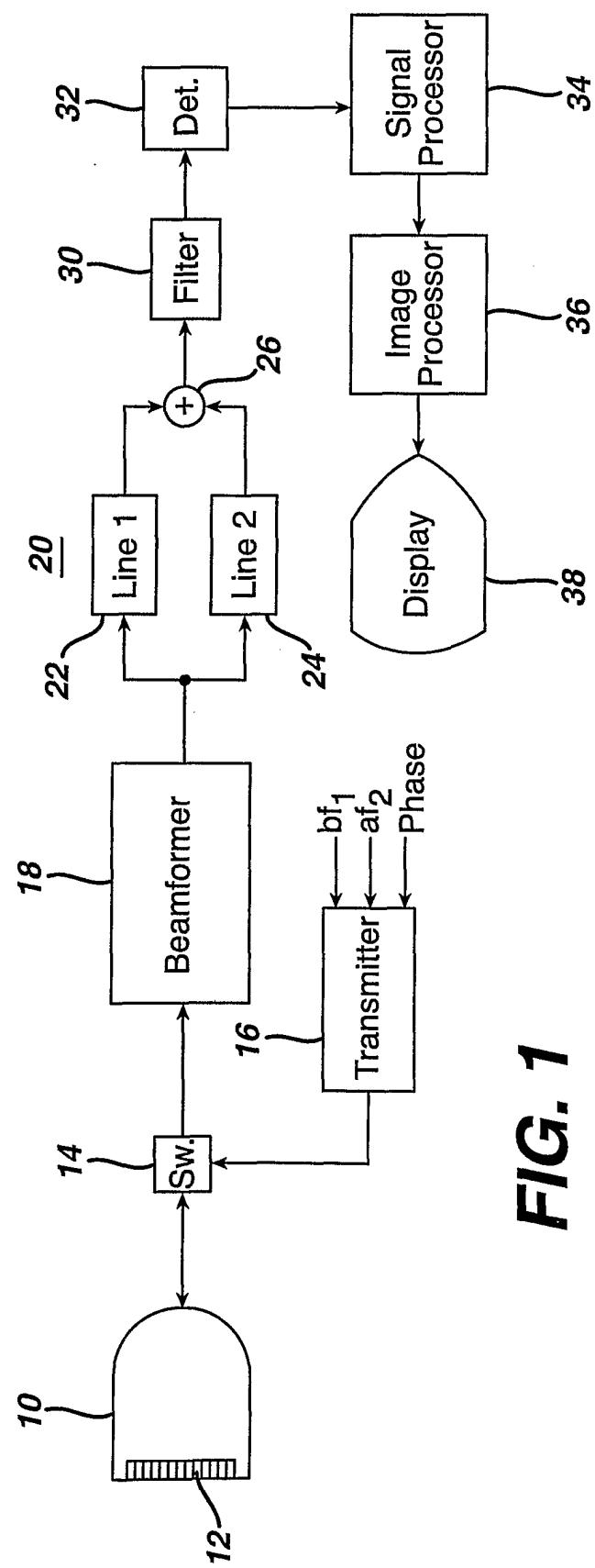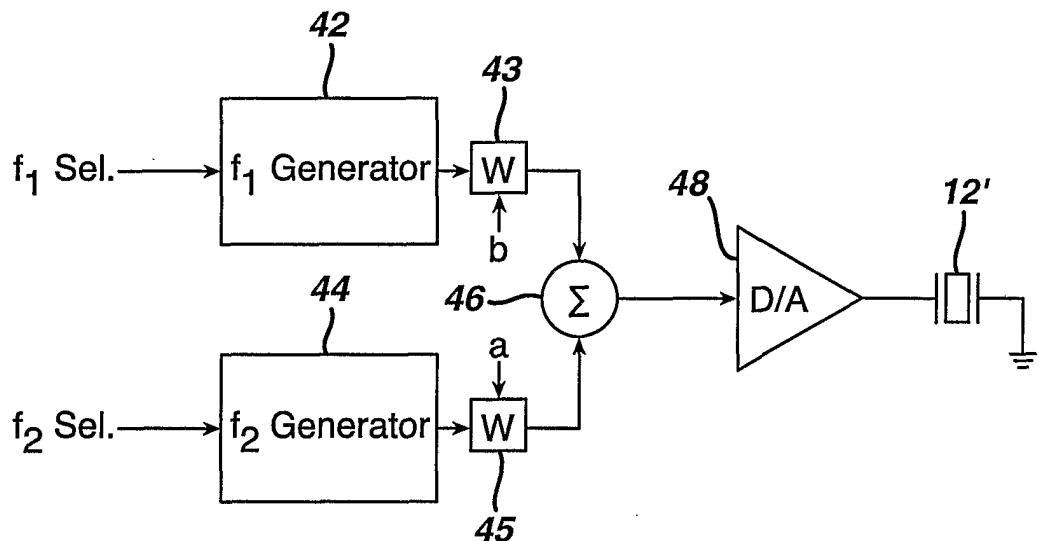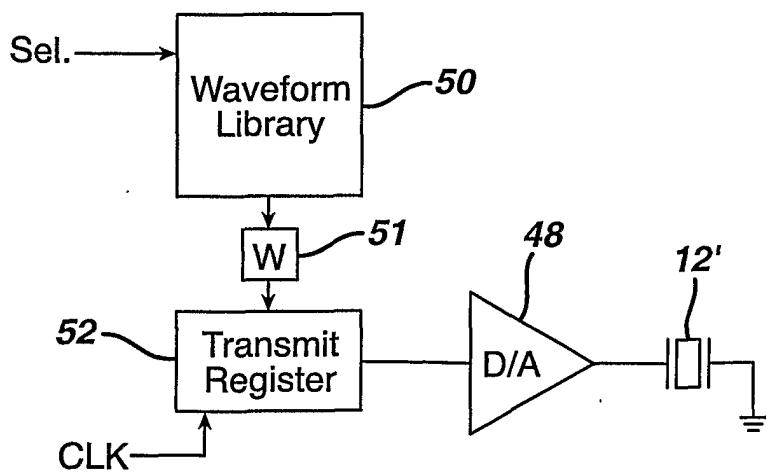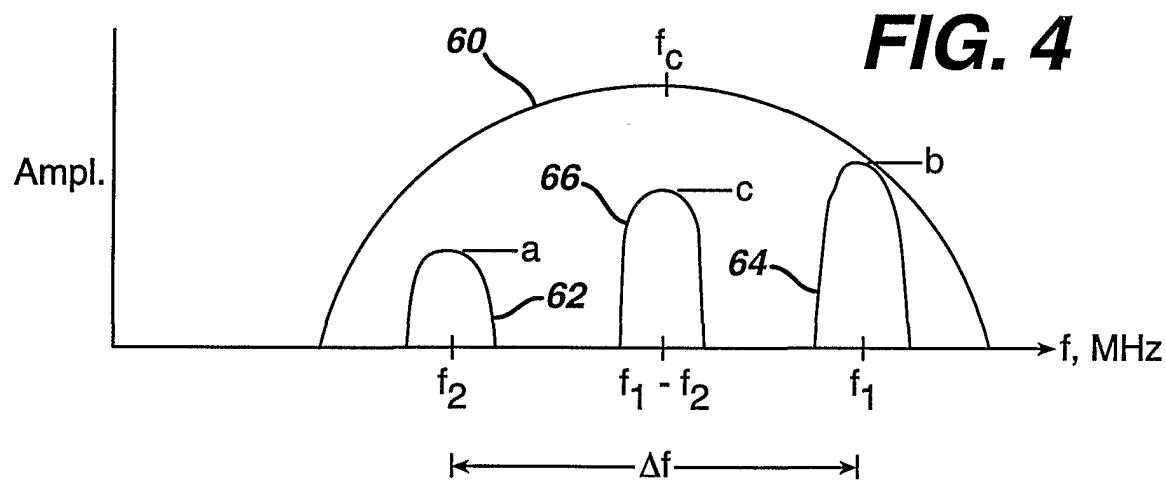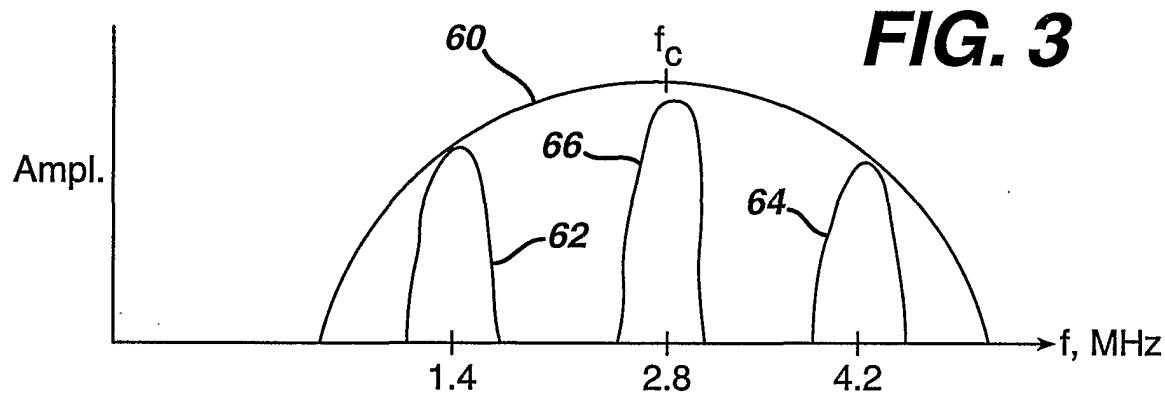
FIG. 1
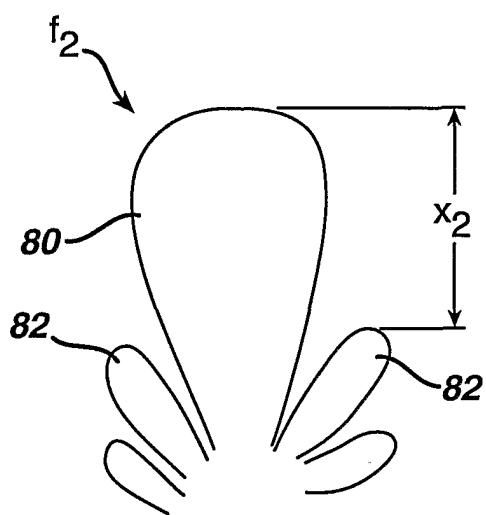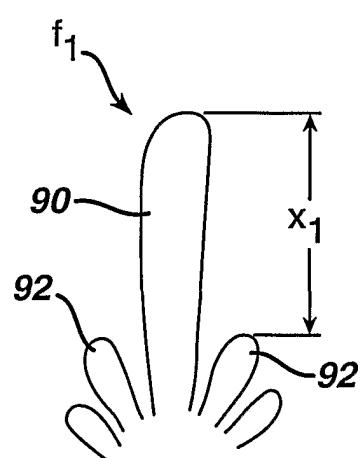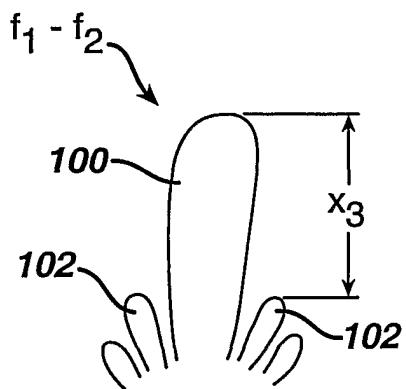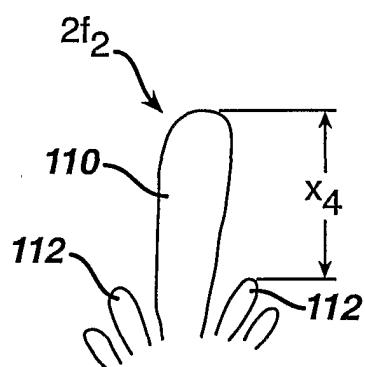
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FIG. 2a

FIG. 2b
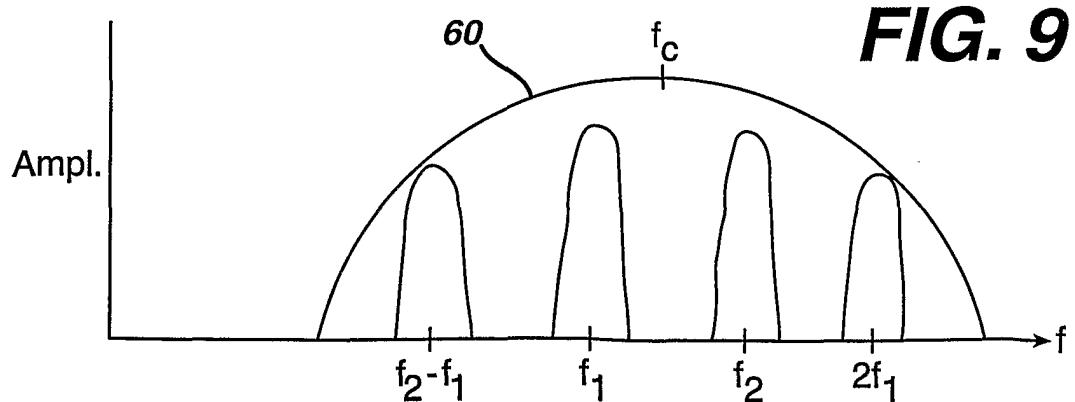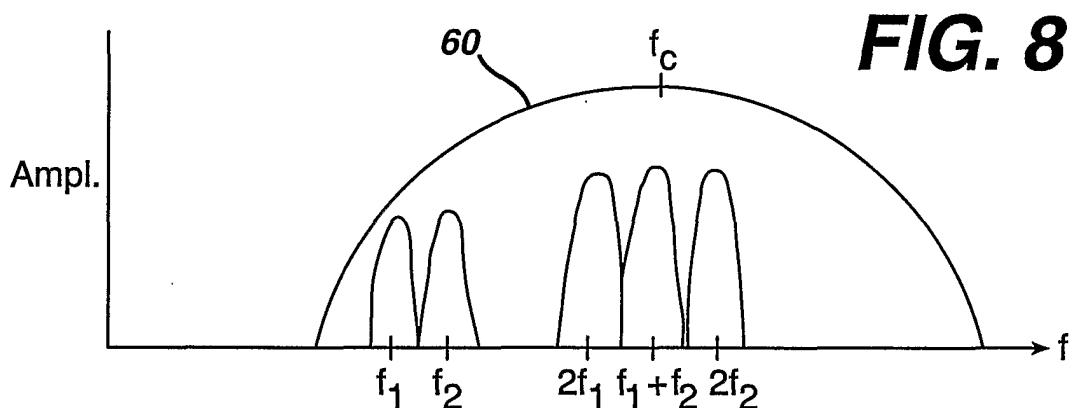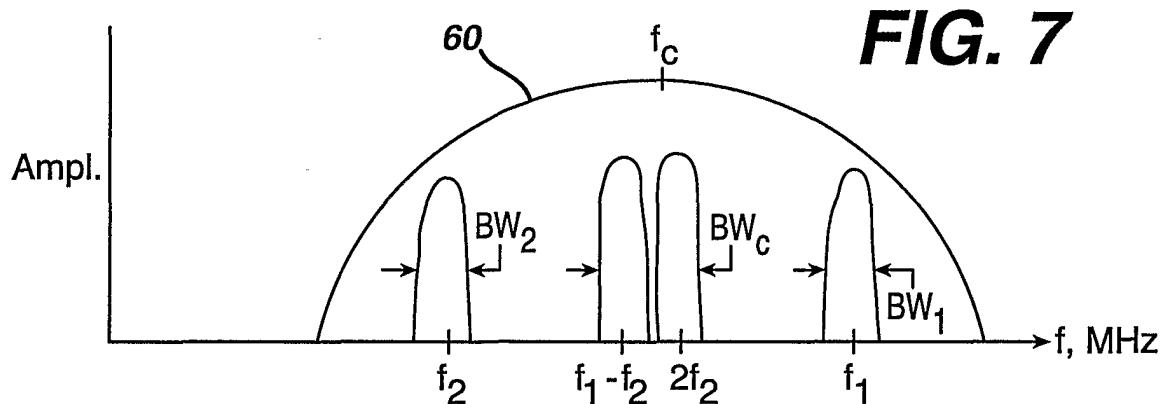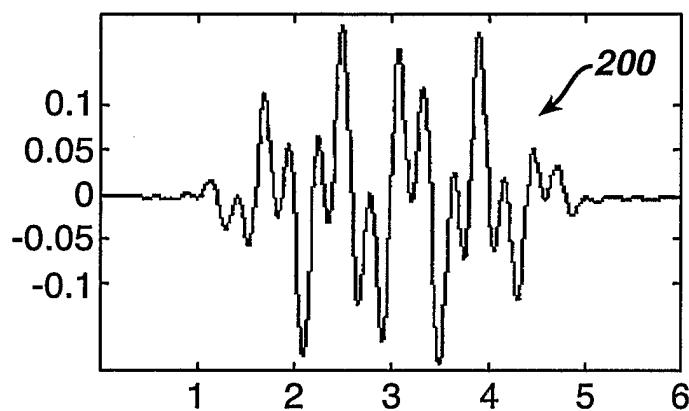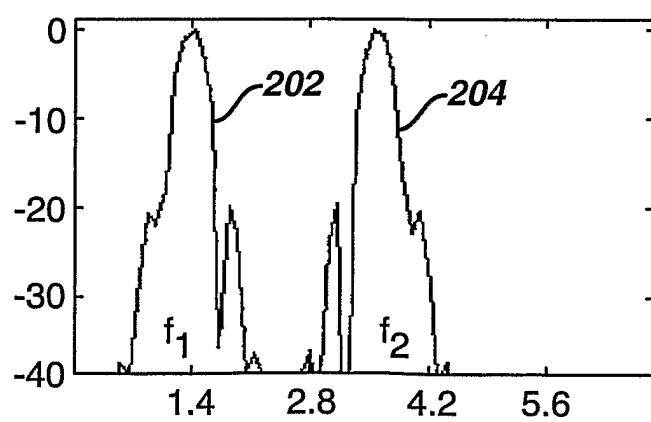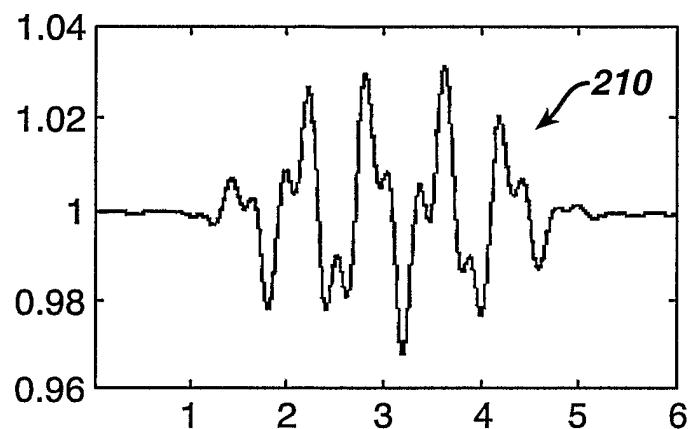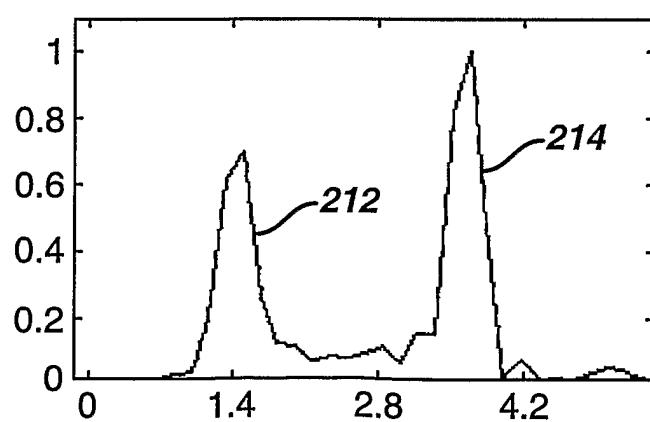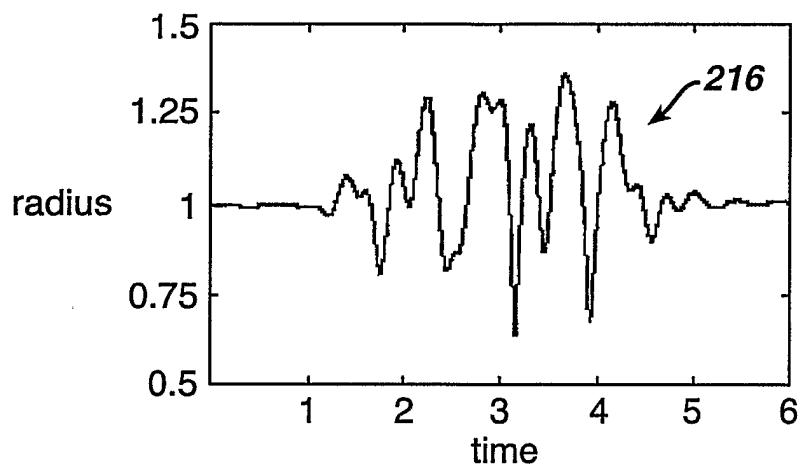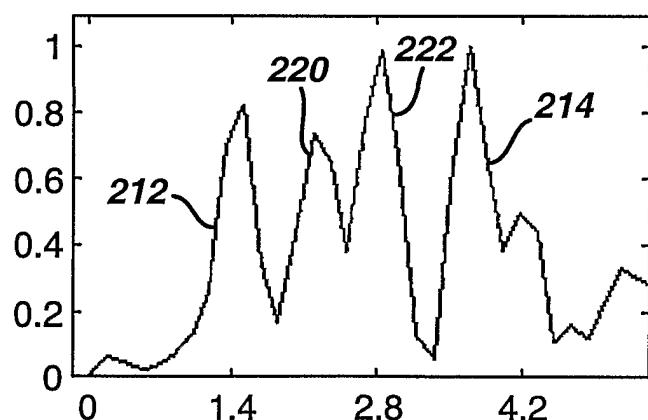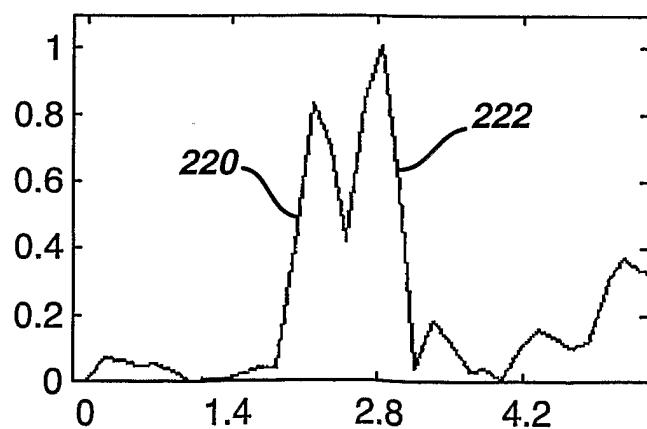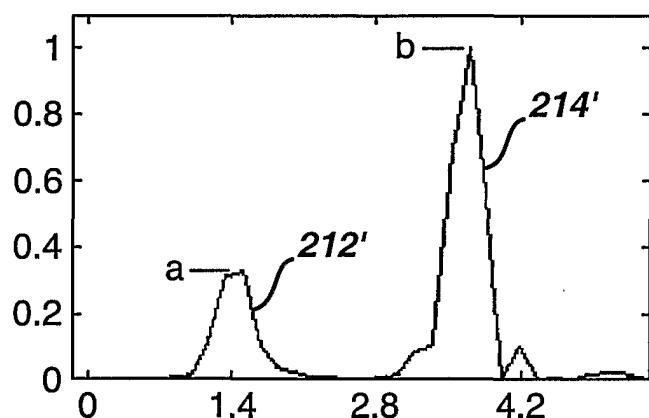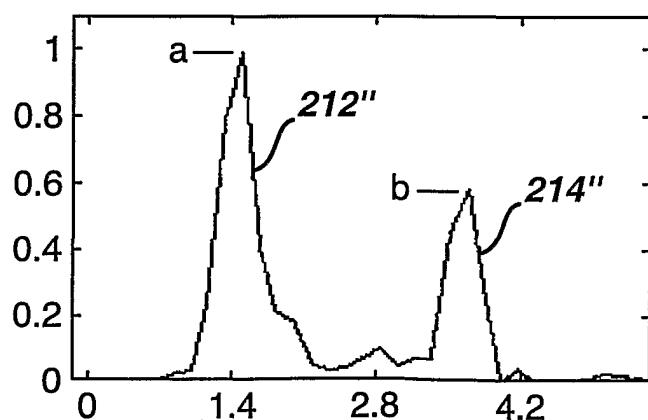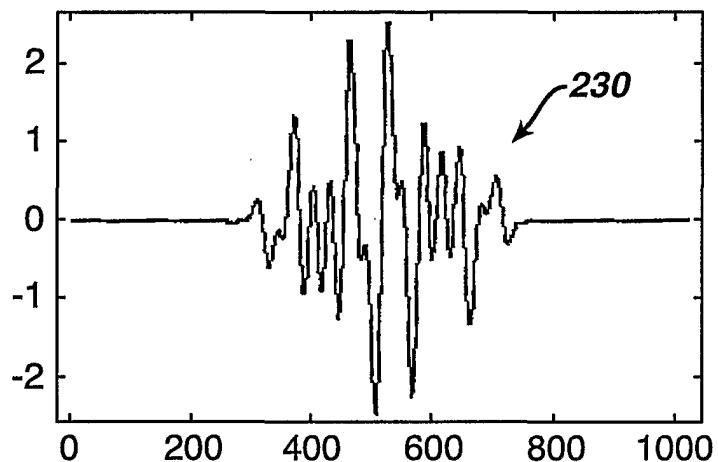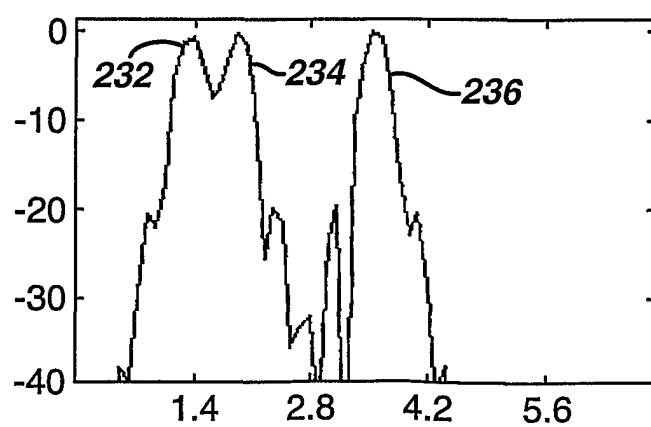
FIG. 3

FIG. 4

FIG. 5

FIG. 6a

f$_2$

80

82 x_2

82

FIG. 6b

f$_1$

90

92 x_1

92

FIG. 6c

f$_1$ - f$_2$

100

102 x_3

102

FIG. 6d

2f$_2$

110

112 x_4

112

FIG. 7

FIG. 8

FIG. 9

FIG. 10a

FIG. 10b

FIG. 11a

FIG. 11b

FIG. 12a

FIG. 12b

FIG. 13

FIG. 14a

FIG. 14b

FIG. 15a

FIG. 15b

FIG. 16a

FIG. 16b

FIG. 17

$$\sin \omega_1 t + \sin \omega_2 t$$

R_A T_1 R_B

$$\sin(\omega_1 t + \pi) + \sin(\omega_2 t + \pi)$$

R_C T_2 R_D

12

FIG. 18

T_1 T_2

$$\sin \omega_1 t + \sin \omega_2 t \qquad 300 \qquad \sin(\omega_1 t + \pi) + \sin(\omega_2 t + \pi)$$

R_A R_B

R'_A R'_B

12
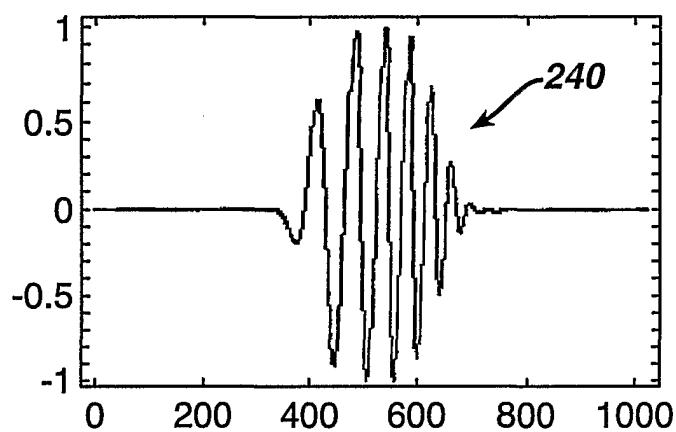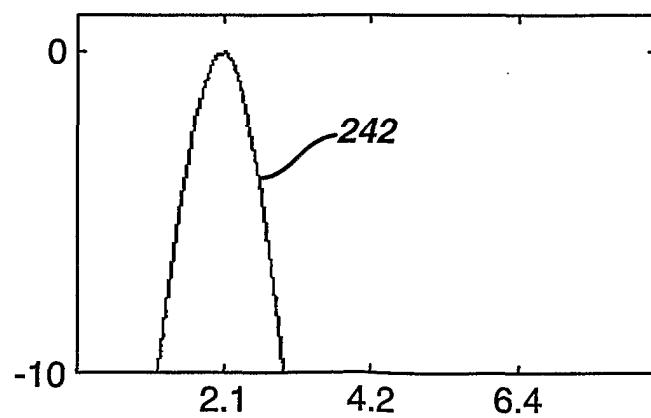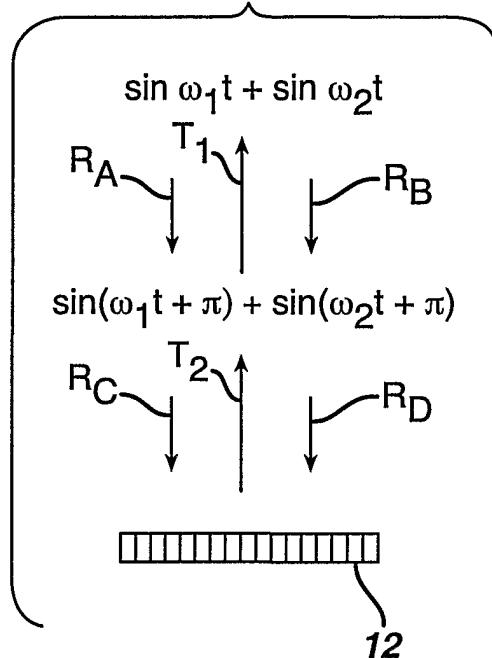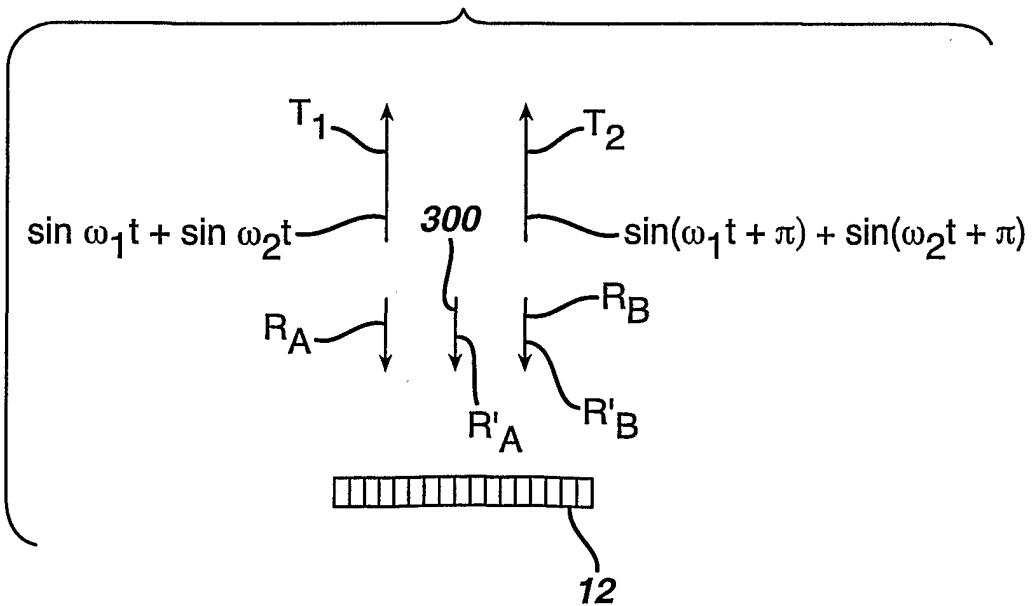
专利名称(译)	非线性互调和谐波频率分量的超声诊断成像		
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申请号	EP2001982368	申请日	2001-09-26
[标]申请(专利权)人(译)	皇家飞利浦电子股份有限公司		
申请(专利权)人(译)	皇家飞利浦电子N.V.		
当前申请(专利权)人(译)	皇家飞利浦电子N.V.		
[标]发明人	AVERKIOU MICHALAKIS		
发明人	AVERKIOU, MICHALAKIS		
IPC分类号	A61B8/00 G01S7/52 G01S15/89		
CPC分类号	G01S15/8963 G01S7/52026 G01S7/52038 G01S7/52095 G01S15/8952 G01S15/8954 G01S15/8959		
代理机构(译)	LOTTIN，CLAUDINE		
优先权	09/923682 2001-08-06 US 60/237407 2000-10-02 US		
外部链接	Espacenet		

摘要(译)

描述了一种超声诊断成像系统和方法，用于利用谐波和互调产物（和频或差频）分量执行非线性回声信号成像。谐波和互调产物都是由组织或造影剂的非线性效应产生的，并且两者都有利地通过脉冲反转处理与回波信号的基本发射分量分离。使用两个非线性分量可以改善超声图像的信噪比，并且可以在图像的不同区域中混合或使用两种类型的分量以抵消深度相关衰减的影响。