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(54) **Title**: METHODS AND APPARATUS FOR ULTRASOUND IMAGING

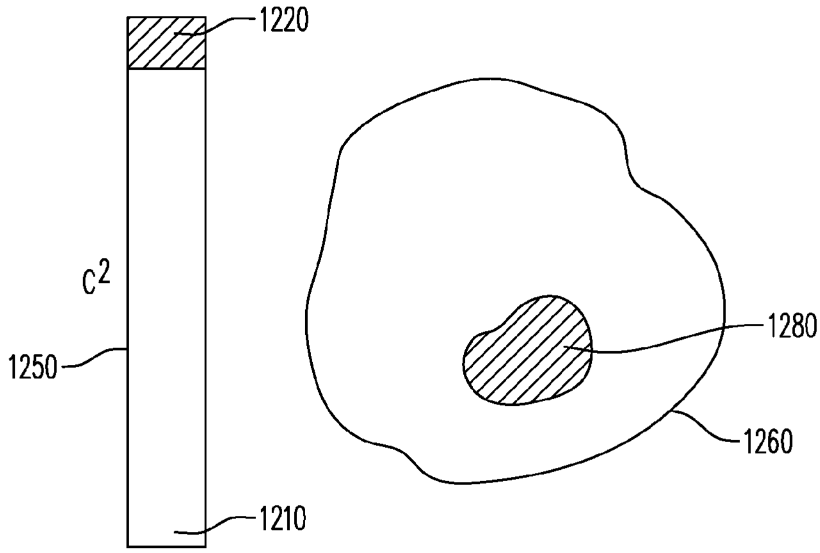

Figure 12

(57) Abstract: A first ultrasound pulse is applied to biological tissue to create shear waves in the biological tissue, a focused ultrasound pulse is transmitted into the biological tissue, one or more ultrasound signals is received from the biological tissue, and shear waves are detected in the biological tissue based on the received one or more ultrasound signals. At least one shear wave propagation property associated with the detected shear waves is determined, and the determined at least one propagation property is displayed. Ultrasound beam steering is used to improve measurement accuracy.

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Methods and Apparatus for Ultrasound Imaging

Cross-reference to related application

The present application claims priority to U.S. Provisional Patent Application Serial No. 61/469,295, filed on March 30, 2011 and entitled "Method and Apparatus for Ultrasound Imaging", the contents of which are incorporated herein by reference for all purposes.

Background

Systems and methods described herein generally relate to the field of ultrasound imaging. More specifically, embodiments described below relate to methods and systems for measuring shear wave velocity in tissue.

Pathological conditions may result in soft tissue which is stiffer than would be present under physiological conditions. Physicians therefore use palpation to locate stiff tissue within a body and thereby identify pathological conditions. For example, breast cancers are known to be generally harder than healthy breast tissue and may be detected as a hard lump through palpation.

The propagation velocity of shear waves in tissue is related to the stiffness (Young's modulus or shear modulus) of tissue by the following equation,

$$E = 3\rho \cdot c^2 \quad (1)$$

where

c is the propagation velocity of shear wave, E is Young's modulus, and ρ is the tissue density. Therefore, cancers or other pathological conditions may be detected in tissue by measuring the propagation velocity of shear waves passing through the tissue.

A shear wave may be created within tissue by applying a strong ultrasound pulse to the tissue. The ultrasound pulse may exhibit a high amplitude and a long duration (e.g., on the order of 100 microseconds). The ultrasound pulse generates an acoustic radiation force which pushes the tissue, thereby causing layers of tissue to slide along the direction of the ultrasound pulse. These sliding (shear) movements of tissue may be considered shear waves, which are of low frequencies (e.g., from 10 to 500 Hz) and may propagate in a direction perpendicular to the direction of the ultrasound pulse. The ultrasound pulse may propagate at a speed of 1540 m/s in tissue. However, the shear wave propagates much more slowly in tissue, approximately on the order of 1-10 m/s.

Since the tissue motion is generally in the axial direction (i.e., the ultrasound pulse direction) the shear waves may be detected using conventional ultrasound Doppler techniques. In this regard, the ultrasound Doppler technique is best suited to detect velocity in the axial direction. Alternately, shear waves may be detected by measuring a tissue displacement caused by the acoustic radiation force.

In order to accurately measure the propagation velocity of the shear wave, the shear wave needs to be tracked at a fast rate or a fast frame rate of several thousands frames per second. An image in a frame may consist of a few hundred ultrasound lines. A typical frame rate of regular ultrasound imaging is about 50 frames/s, which is too slow to track the shear wave propagation. Therefore, there exists a need to increase the frame rate while maintaining a good signal to noise ratio and good spatial resolution. Also, there exists a need to efficiently provide an indication of tissue stiffness.

Brief Description Of The Drawings

Figure 1. A diagram of shear wave generation resulting from an acoustic radiation

force.

Figure 2A. A diagram of an ultrasound imaging system of some embodiments.

Figure 2B. A diagram of a composite image processor according to some embodiments.

Figure 3. A diagram of a conventional ultrasound imaging system.

Figure 4. A diagram of multiple ultrasound transmitted/received beams.

Figure 5. A diagram of an ultrasound transmitted beam and multiple ultrasound received beams.

Figure 6. Color coding of shear wave propagation velocity squared.

Figure 7. Color coding of shear wave propagation velocity squared.

Figure 8. A diagram illustrating generation of shear waves by acoustic radiation forces and the propagation of shear waves.

Figure 9. A diagram illustrating sliding movements of shear waves.

Figure 10. A diagram illustrating the propagation of shear waves.

Figure 11. A diagram illustrating the propagation of shear waves.

Figure 12. An example of a color-coded image of shear wave propagation velocity squared in tissue.

Figure 13. A diagram to illustrate tissue displacement caused by an acoustic radiation force.

Figure 14. Scale of shear wave velocity squared c^2 by color coding bar composed of RGB representation.

Figure 15. A diagram to show the ultrasound coordinate system with respect to an ultrasound transducer.

Figure 16. Steered acoustic radiation force.

Figure 17. Steered ultrasound beam.

Figure 18. Steered ultrasound beams.

Figure 19. Image depicting a shear wave property at a first ultrasound beam steering angle.

Figure 20. Image depicting a shear wave property at a second ultrasound beam steering angle.

Figure 21. Image depicting a shear wave property at a third ultrasound beam steering angle.

Figure 22. Image depicting a shear wave property according to some embodiments.

Detailed Description

Embodiments will be described with reference to the accompanying drawing figures wherein like numbers represent like elements throughout. Before embodiments of the invention are explained in detail, it is to be understood that embodiments are not limited in their application to the details of the examples set forth in the following description or illustrated in the figures. Other embodiments may be practiced or carried out in a variety of applications and in various ways. Also, it is to be understood that the phraseology and terminology used herein is for the purpose of description and should not be regarded as limiting. The use of “including,” “comprising,” or “having,” and variations thereof herein is meant to encompass the items listed thereafter and equivalents thereof as well as additional items. The terms “mounted,” “connected,” and “coupled,” are used broadly and encompass both direct and indirect mounting, connecting, and coupling. Further, “connected,” and “coupled” are not restricted to physical or mechanical connections or couplings.

Acoustic radiation force is created by a strong ultrasound pulse 120 as shown in Figure 1. The ultrasound pulse 120 exhibits a high amplitude as well as a long duration, (e.g., on the order of 100 microseconds). The ultrasound pulse 120 is transmitted from an ultrasound transducer array 110. The ultrasound pulse 120 is focused at a focal point 130 in biological tissue 160, resulting in an acoustic radiation force which pushes the tissue 160 at the focal point 130. The ultrasound pulse 120 may be transmitted multiple times and may be focused at a different focal point for each of multiple transmitted ultrasound pulses.

The tissue 160 is pushed mostly in the axial direction of the ultrasound pulse 120, creating shear waves 140, 150 which may propagate in the lateral direction or directions other than the axial direction (i.e., vertical direction). The propagation velocity of the shear waves 140, 150 depends on the stiffness (Young's modulus or the shear modulus) of the tissue 160. Greater tissue stiffness results in greater shear wave propagation velocity as shown in equation 1. Pathological conditions such as cancer may increase tissue stiffness thus these conditions may be diagnosed by determining the propagation velocity. For example, the shear wave propagation velocity may vary from 1 m/s to 10 m/s, depending on tissue conditions.

Since the shear wave may be characterized by tissue movement (or motion), the shear wave may be detected by the ultrasound Doppler technique (e.g., see US4573477, US4622977, US4641668, US4651742, US4651745, US4759375, US4766905, US4768515, US4771789, US4780837, US4799490, and US4961427). To detect this tissue movement (motion), the ultrasound pulse is transmitted multiple times to the tissue, and the ultrasound is scattered by scatterers in tissue and received by an ultrasound transducer as received ultrasound signals. The received ultrasound signals from the ultrasound array transducers are filtered, amplified, digitized, apotized, and beamformed (i.e. summed) after applying delays and/or phase-rotations for focusing and steering. The order of these processing steps may be interchanged. Received beamformed RF ultrasound signals undergo quadrature demodulation, resulting in complex, Doppler I-Q signals. In a color Doppler technique, the ultrasound is transmitted at a pulse repetition frequency (PRF) and the velocity is detected as the shift in frequency (Doppler shift frequency) in the received ultrasound signal. The received ultrasound is mixed with in-phase (0 degrees) and quadrature (90 degrees) reference signals of the same frequency as the transmitted ultrasound frequency, resulting in complex I-Q Doppler signals.

Generally, the complex I-Q signal is used to derive the Doppler shift frequency because the Doppler shift frequency and the blood velocity have the following relationship

$$\Delta f = \frac{2 f_t v \cos\theta}{c_S} \quad (2),$$

where Δf is the Doppler shift frequency, f_t is the transmitted frequency, v is the blood velocity, θ is the angle between the ultrasound beam direction and the velocity vector, and c_S is the speed of sound. The Doppler shift frequency is thus dependent on the angle between the velocity direction and the ultrasound beam direction and is a measurement that an ultrasound color Doppler system may obtain.

In the case of color Doppler, the number of the sampled signals may be limited to several. Therefore, an auto-correlation technique is usually used to determine the phase differences between the I-Q signals and then to determine the Doppler shift frequency and the velocity as follows. The color Doppler's I-Q signals $z(m) = x(m) + jy(m)$ are used to calculate "auto-correlation" r as shown in the following equation, where $z(m)$ is the complex I-Q Doppler signal, $x(m)$ is the in-phase (real) signal, $y(m)$ is the quadrature phase (imaginary) signal, m indicates the signal number, j is the imaginary unit and * indicates the complex conjugate.

$$r = \sum z(m) \cdot z^*(m-1) \quad (3)$$

The real ($\mathrm{Re}\,al(r)$) and imaginary ($\mathrm{Im}\,ag(r)$) parts of r are used to obtain the phase φ as shown in the following equation.

$$\varphi = \tan^{-1} \frac{\mathrm{Im}\,ag(r)}{\mathrm{Re}\,al(r)} \quad (4)$$

Since $\tan^{-1}$ usually provides only -0.5π to 0.5π, the position of complex value r in the complex coordinate may be also used to derive φ in the range of $-\pi$ to π. The phase (i.e., color Doppler phase) φ is then related to the Doppler shift frequency as shown in the following equation.

$$\Delta f = \frac{\varphi f_{PRF}}{2\pi} \quad (5)$$

Autocorrelation r between the received complex baseband ultrasound signals is thus obtained to detect tissue velocity or movement.

Tissue movement is detected at multiple lateral points in a field of tissue region by multiple ultrasound beams (for example, 540, 545, 550 in Figure 5) in order to monitor movement. This movement reflects action of the shear wave at those multiple lateral points (or multiple ultrasound beams). Consequently, the lateral propagation velocity of the shear wave may be determined from the detected tissue movement.

Alternately, the shear wave may be detected by measuring tissue displacement caused by acoustic radiation force which is in turn caused by a strong ultrasound pulse as shown in Figure 13. Tissue 1310 is positioned at a position 1320 before the acoustic radiation is applied and then is moved to a position 1330 after the acoustic radiation force was applied. To measure tissue displacement caused by the strong ultrasound pulse, ultrasound pulses are transmitted to tissue from an ultrasound transducer 1305 and then the ultrasound pulses are scattered from scatterers in tissue and returned to the transducer 1305 and received by the transducer 1305 as received ultrasound signals. The ultrasound pulses are focused at a depth in order to increase a signal-to-noise ratio of the resulting received ultrasound signals in comparison to unfocused ultrasound pulses. Using correlation of the received ultrasound signals from tissue the displacement 1340 (from the position 1320 to the position 1330) of the tissue 1310 due to the acoustic radiation force may be obtained and the tissue 1310 may be tracked thereafter. The ultrasound pulses may thereby track shear waves after shear waves are created by acoustic radiation force.

Ultrasound signals resulting from the first ultrasound pulse and received from the tissue 1310 before acoustic radiation force is applied are cross-correlated with received ultrasound signals resulting from the second ultrasound pulse after the acoustic radiation force is applied in order to find the best match between the received ultrasound signals. The best match may be found by finding a maximum correlation value to track the tissue and its displacement due to the acoustic radiation force. Therefore, when tissue displacement is observed or measured, a shear wave is detected. The displacement and tissue velocity may be related in that the displacement is a time integral $\int v_s dt$ of tissue velocity v_s. Therefore, the tissue displacement may be obtained by calculating the time integral of color Doppler velocity. Received ultrasound signals may be RF (Radio Frequency), IF (Intermediate Frequency) or baseband signals after demodulation. Alternately, the displacement may be further differentiated to obtain tissue strain, which may be then used to detect the shear wave propagation velocity.

Cross correlation $CC(t,\tau)$ of signals in the previous paragraphs may be mathematically expressed as follows,

$$CC(t,\tau) = \int_t^{t+W} S_1(t')S_2(t'-\tau)dt' \quad (6)$$

where $CC(t,\tau)$: cross correlation; $S_1(t')$: received signal from the first ultrasound transmission; $S_2(t'-\tau)$: received ultrasound signal from the second ultrasound transmission; W: window length; t: time, t’: time; τ : time displacement. Time displacement value τ, which makes the maximum cross correlation (or the best match), determines the tissue displacement. Interpolation of signals using an interpolation function (e.g. cubic-spline) may be performed before cross correlation to increase spatial resolution.

The cross correlation may be replaced by the sum of absolute differences (SAD), the sum of square differences (SSD), the sum of absolute cubic differences (SCD), or the sum of absolute power differences (SPD) as follows.

$$SAD[l,k] = \sum_{n=0}^{N} \left| S_1[l+n] - S_2[l+n-k] \right| \quad (7)$$

$$SSD[l,k] = \sum_{n=0}^{N} (S_1[l+n] - S_2[l+n-k])^2 \quad (8)$$

$$SCD[l,k] = \sum_{n=0}^{N} \left| S_1[l+n] - S_2[l+n-k] \right|^3 \quad (9)$$

$$SPD[l,k] = \sum_{n=0}^{N} \left| S_1[l+n] - S_2[l+n-k] \right|^p \quad (10)$$

S_1 is the received ultrasound signal from the first ultrasound transmission before displacement, S_2 is the received ultrasound signal from the second ultrasound transmission after displacement. N: the number of signals in the signal window. k: window displacement by the number of signals and equivalent of τ. l: the position of the window. p is a real number. For SAD, SSD, SCD and SPD, the tissue displacement is determined based on the value of k that makes the minimum (or best match) of each of the SAD, SSD, SCD and SPD.

Figures 8 and 9 are used to illustrate shear wave generation and detection in detail. A strong ultrasound pulse 820 is applied to tissue 860, 960 from an ultrasound transducer 810, 910 once or more times to increase the amplitude of shear waves which are caused by acoustic radiation forces resulting from the ultrasound pulse. Shear waves attenuate very quickly in tissue and thus a greater amplitude results in a greater propagation distance. One or multiple ultrasound pulses may be focused at one focal point or different focal points. The ultrasound pulse creates acoustic radiation forces which push a layer of tissue, resulting in tissue movement 830, 910 mostly in the axial (vertical) direction as illustrated in Figure 9. The tissue layer movement 910 causes adjacent tissue layer movements 920, 925 mostly in the axial direction. The tissue layer movements 920, 925 then in turn cause next tissue layer movements 930, 935 which then cause adjacent tissue layer movements 940, 945. This succession of tissue movements represents a propagation of shear waves 840, 850 in the lateral (horizontal) direction as shown in Figure 8. Since the tissue movements (or motions) caused by acoustic radiation forces are mostly in the axial direction, the motion may be detected by the color Doppler technique, which is sensitive to motions in the axial direction.

For example, the color Doppler technique transmits and receives several ultrasound pulses, determines phase differences between the received ultrasound signals, and calculates a velocity of tissue or blood using the autocorrelation technique as previously discussed and known in the art. Variance and power of color Doppler signals may be also calculated in addition to the velocity. As in the conventional display of moving tissue or blood, one of these parameters may be used to display shear waves as shown in Figures 10, 11. It will be assumed that shear waves 1040 (1140), 1050 (1150) are determined in a color Doppler frame representing a certain time and shear waves 1060 (1160), 1070 (1170) are determined at a next moment or in a next frame. More image frames of shear waves may be obtained to track the shear waves and to create a movie of shear wave propagation. In alternate embodiments, tissue displacement due to acoustic radiation forces may be detected.

Figures 10 and 11 depict shear wave propagation at two points in time. Local shear wave propagation velocities, as illustrated by arrows 1080, 1090, may be derived by correlating two images of shear waves at two points in time. More image frames of shear waves may be used to track the propagation of shear waves in more image areas in order to present local shear wave propagation velocities or shear wave propagation velocity squared in a two-dimensional image as described below.

Correlation coefficient (CCV) between a first frame signal S^1 and the second frame signal S^2 may be obtained as speckle tracking as follows,

$$CCV(S^1,S^2)=\frac{\sum_{x=1}^{m}\sum_{z=1}^{n}(S^1{}_{x,z}-\overline{S^1})(S^2{}_{x+X,z+Z}-\overline{S^2})}{\sqrt{\sum_{x=1}^{m}\sum_{z=1}^{n}(S^1{}_{x,z}-\overline{S^1})^2\cdot\sum_{x=1}^{m}\sum_{z=1}^{n}(S^2{}_{x+X,z+Z}-\overline{S^2})^2}} \quad (11)$$

where $S^1{}_{x,z}$ is the ultrasound signal at x, z of the first frame, $S^2{}_{x+X,z+Z}$ is the ultrasound signal at x+X, z+Z of the second frame, $\overline{S^1}$ is mean signal value in the window of the first frame signal, $\overline{S^2}$ is mean signal value in the window of the second frame signal. The coordinate system (x,y,z) is shown with respect to an ultrasound transducer 1510 in Figure 15. The elevational axis y is perpendicular to the paper of Figure 15 although it is shown slightly different for illustration purposes.

The displacement X, Z, that yields the maximum correlation coefficient determines the correct speckle tracking and the distance, and thus the velocity (i.e., the distance per time).

Similar to the 1D case, the correlation coefficient may be replaced by the sum of absolute differences (SAD), the sum of square differences (SSD), the sum of absolute cubic differences (SCD) and the sum of absolute power differences (SPD) as follows.

$$SAD(S^1, S^2, X, Z) = \sum_{x=1}^{m}\sum_{z=1}^{n}\left|S^1{}_{x,z} - S^2{}_{x+X,z+Z}\right| \quad (12)$$

$$SSD(S^1, S^2, X, Z) = \sum_{x=1}^{m}\sum_{z=1}^{n}\ (S^1{}_{x,z} - S^2{}_{x+X,z+Z})^2 \quad (13)$$

$$SCD(S^1, S^2, X, Z) = \sum_{x=1}^{m}\sum_{z=1}^{n}\left|S^1{}_{x,z} - S^2{}_{x+X,z+Z}\right|^3 \quad (14)$$

$$SPD(S^1, S^2, X, Z) = \sum_{x=1}^{m}\sum_{z=1}^{n}\left|S^1{}_{x,z} - S^2{}_{x+X,z+Z}\right|^p \ (15)$$

p is a real number; m and n are integers. The 2D speckle tracking may be approximated by a 1D speckle tracking to obtain the shear wave propagation velocity and the shear wave propagation velocity squared. The mathematical expression will be similar to that used in the displacement measurement.

Alternately, a shear wave equation (16) may be used to derive the shear wave propagation velocity as follows,

$$\rho\frac{\partial^2 u_i}{\partial t^2}=\mu(\frac{\partial^2 u_i}{\partial x^2}+\frac{\partial^2 u_i}{\partial y^2}+\frac{\partial^2 u_i}{\partial z^2}) \quad (16)$$

where $i=x,y,z$, ρ is tissue density, μ is the shear modulus, u_i is the displacement vector, x is lateral coordinate, y is elevational coordinate and z is axial coordinate as shown in Figure 15. For incompressible materials, the Young's modulus E and the shear modulus μ have the following relationship.

$$E=3\mu \quad (17)$$

Therefore, the shear wave propagation velocity squared may be obtained as a ratio of the shear modulus to the density as the following equation.

$$c^2=\frac{\mu}{\rho} \quad (18)$$

One of the displacement components u_z in equation 16 may be determined by cross-correlation as previously discussed. By combining z component of equation 16 and equation 18, the shear wave propagation velocity squared and velocity are obtained as follows,

$$c^2=\frac{\frac{\partial^2 u_z}{\partial t^2}}{\frac{\partial^2 u_z}{\partial x^2}+\frac{\partial^2 u_z}{\partial y^2}+\frac{\partial^2 u_z}{\partial z^2}} \quad (19)$$

and

$$c=\sqrt{\frac{\frac{\partial^2 u_z}{\partial t^2}}{\frac{\partial^2 u_z}{\partial x^2}+\frac{\partial^2 u_z}{\partial y^2}+\frac{\partial^2 u_z}{\partial z^2}}} \quad (20).$$

Therefore, the shear wave propagation velocity is obtained as the square root of the ratio between the temporal second-order derivative of the displacement and the spatial second-order derivatives of the displacement. Likewise, the shear wave propagation velocity squared is obtained as the ratio between the temporal second-order derivative of the displacement and the spatial second-order derivatives of the displacement. Since the spatial derivative of the displacement in elevational direction $\frac{\partial^2 u_z}{\partial y^2}$ may be considered negligible compared with the other spatial derivatives, the shear wave propagation velocity squared and velocity may be obtained from the other measurement values.

It is desirable to monitor and to track the shear wave frequently, meaning at a fast rate or frame rate. To speed up the frame rate, a wide, focused ultrasound pulse 520 may be transmitted and multiple ultrasound signals 540, 545, 550 may be simultaneously received as shown in Figure 5. The received ultrasound beams are used as described previously to detect shear waves and to derive shear wave propagation properties (i.e., velocity and velocity squared) therefrom. The focused transmit ultrasound beam 520 may be particularly suitable for maintaining a good signal-to-noise ratio of resulting received ultrasound beams during the detection of shear waves.

In some embodiments, multiple ultrasound beams (pulses) are simultaneously applied and transmitted to the tissue field and multiple ultrasound beams (pulses) per transmitted ultrasound pulse are received to increase the frame rate, as shown in Figure 4. In Figure 4, ultrasound pulses 420, 430 are simultaneously transmitted to biological tissue 480 from an ultrasound transducer array 410. For each transmitted ultrasound pulse 420, 430, multiple ultrasound receive signals 440, 445, 465, 460, 465, 470 are simultaneously received. The multiple ultrasound pulses may be transmitted simultaneously or at substantially simultaneous times. The multiple ultrasound pulses may be simultaneously transmitted. Or a second ultrasound pulse may be transmitted after a first ultrasound pulse is transmitted and before the first ultrasound pulse returns to the ultrasound transducer from a deepest depth of an ultrasound field. This transmission method increases the frame rate.

Figure 4 shows an example of two simultaneous transmitted ultrasound pulses but more than two transmitted ultrasound pulses may be also used. In some embodiments, coded ultrasound waveforms may be transmitted for better separation of simultaneous multiple ultrasound signals. For example, chirp codes, Barker codes, Golay codes or Hadamard codes may be used for better separation of ultrasound pulses. Again, the received signals are analyzed using the methods previously described to determine tissue movement at multiple points, and shear wave propagation properties are derived therefrom.

An image of a shear wave can be created based on the motion (or velocity) detected at multiple points in the imaging field. Subsequent transmit/receive sequences of ultrasound may create multiple images of the shear wave at multiple points in time. Correlation between the images of the shear wave is then calculated to obtain the shear wave propagation velocity and velocity squared as previously discussed. Alternately, tissue displacement caused by acoustic radiation force is determined and the shear wave propagation velocity is calculated as the square root of the ratio between the temporal second-order derivative of the displacement and the spatial second-order derivatives of the displacement. Likewise, the shear wave propagation velocity squared is calculated as the ratio between the temporal second-order derivative of the displacement and the spatial second-order derivatives of the displacement.

In some embodiments, the propagation velocity of a detected shear wave (c) may be displayed. In some embodiments, the propagation velocity squared (c^2) of the detected shear wave may be displayed. Advantageously, the propagation velocity squared (c^2) may be more closely related than the propagation velocity (c) to the Young's modulus or the shear modulus as shown in equation 1. Therefore the propagation velocity squared (c^2) may provide an efficient proxy for the actual stiffness. In some embodiments, the propagation velocity squared (c^2) may be multiplied by three and then displayed. If tissue density is close to 1 g/cm^3, this number (i.e., $3c^2$) may be close to the actual Young's modulus. In some embodiments, a product (bc^2) of any real number (b) and the propagation velocity squared (c^2) may be displayed. Determinations of actual stiffness are difficult and error-prone because the density of the tissue is unknown and must be estimated.

A color coding technique, a grayscale technique, or a graphical coding technique may be employed to present a shear wave propagation property (i.e., velocity c or velocity squared c^2) to a user. In some embodiments, a propagation velocity squared (c^2) of shear waves within tissue is displayed in a two-dimensional color image. Graphical-coding and/or two-dimensional images may also be used to represent the propagation velocity c or velocity squared c^2 in some embodiments.

A low value of shear wave propagation velocity squared c^2 may be coded using a red color while a high value of c^2 may be coded using a blue color. For example, Figure 6 illustrates a legend indicating that a red-colored tissue area includes shear waves associated with low c^2 values (e.g., 1 m^2/s^2) and that a blue-colored tissue area includes shear waves associated with high c^2 values (e.g., 100 m^2/s^2). Embodiments are not limited to color-based coding. Images of shear wave propagation properties within tissue may be coded using grayscale or any combination of graphical patterns (e.g., vertical lines, horizontal lines, cross-hatching, dot patterns of different densities, etc.) and colors.

After determining the propagation velocity squared (c^2), c^2 may be coded linearly with respect to the color wavelength as shown in Figure 6. For example, if c^2 within a tissue area is determined to be 50 m^2/s^2, the tissue area may be displayed using a yellow color 630.

Alternately, color-coding of the shear wave propagation velocity squared (c^2) may be defined as shown in Figure 7. Tissue areas associated with low values of the shear wave propagation velocity squared may be displayed as blue 710 while areas associated with high values of the velocity squared may be displayed as red 720. Different color-coding methods may be also used to represent the propagation velocity squared (c^2) or velocity c of shear waves. For example, color coding may be based on hue, brightness, and other color characteristics. The color-coded scale may represent different maximums and minimums of the shear wave propagation velocity squared or velocity than shown in Figure 6, 7. In this regard, the velocity squared maximum of 100 m^2/s^2 and velocity squared minimum of 1 m^2/s^2 in Figures 6 and 7 are only for the illustration purposes and do not limit the scope of the claims. Other values may represent the maximum or minimum values of the coding scale.

Color coding based on Red, Green and Blue (RGB) values may be used to represent the propagation velocity c or velocity squared (c^2) of shear waves as shown in Figure 14. In this example (Figure 14), the propagation velocity squared (c^2) of a shear wave within tissue is represented according to a color coding bar 1410 which is based on RGB values 1420, 1430 and 1440. The shear wave propagation velocity squared has 256 possible values in this example, as represented 256 colors in the color coding bar 1410. The smallest velocity squared $c^2(0)$ 1412 is represented by a color composed of a combination of R(0) 1422, G(0) 1432 and B(0) 1442. The middle velocity squared $c^2(127)$ 1415 is represented by a color composed of a combination of R(127) 1425, G(127) 1435 and B(127) 1445. The highest velocity squared $c^2(255)$ 1418 is represented by a color composed of a combination of R(255) 1428, G(255) 1438 and B(255) 1448. In this example, R(255) only indicates a Red color associated with the red index 255 and does not necessarily indicate a Red color value of 255, which is the brightest Red color. Likewise, G(255) indicates a Green color associated with the green index 255 and B(255) indicates a Blue color associated with the blue index 255.

Alternately, Red, Green, Blue and Yellow may be used to define a color coding bar. Alternately, a Hue-based color coding bar may be used.

Figure 12 represents an example of a color-coded image 1260 displaying a shear wave propagation velocity squared c^2 within human soft tissue (e.g. breast). A color coding scale 1250 is illustrated, in which a color code 1210 (i.e., representing a red color although displayed as white in this black/white document) represents a low shear wave propagation velocity squared value and a color code 1220 (i.e., representing a blue color although displayed as hatched in this black/white document) represents a higher shear wave propagation velocity squared value.

Based on the coding scale 1250, it can be seen that the color coded image 1260 includes an area 1280 of high propagation velocity squared c^2. Since the shear wave propagation velocity squared c^2 is proportional to the Young's modulus, the tissue area corresponding to area 1280 is likely to be hard. Since a tumor is generally hard, image 1260 may indicate pathological conditions.

The color-coding method provides efficient distinction between an area including shear waves having a high propagation velocity squared value and other areas including shear waves having a low propagation velocity squared value. The color coding method therefore allows efficient identification of hard tissue areas within soft tissue areas. An image displaying shear wave propagation velocity or velocity squared may be combined (e.g., superimposed) with a regular image of ultrasound, e.g. B-mode image, or a combined B-mode image and color Doppler image and/or spectral Doppler image. Alternately, the shear wave propagation velocity squared or velocity may be displayed numerically. In some embodiments, the shear wave propagation velocity squared may be displayed in gray scale or based on other graphic coding methods such as using patterns rather than colors. For example, low values of shear wave propagation velocity or square of the shear wave propagation velocity may be displayed in black or dark gray while high values of shear wave propagation velocity or shear wave propagation velocity squared may be displayed in light gray or white using a grayscale coding method.

Figure 3 shows a diagram of a conventional ultrasound diagnostic imaging system with B-mode imaging, Doppler spectrum and color Doppler imaging. The system may include other imaging modes, e.g. elasticity imaging, 3D imaging, real-time 3D imaging, tissue Doppler imaging, tissue harmonic imaging, contrast imaging and others. An ultrasound signal is transmitted from an ultrasound probe 330 driven by a transmitter/ transmit beamformer 310 through a transmit/receive switch 320. The probe 320 may consist of an array of ultrasound transducer elements which are separately driven by the transmitter/transmit beamformer 310 with different time-delays so that a transmit ultrasound beam is focused and steered. A receive beamformer 340 receives the received ultrasound signals from the probe 330 through the switch 320 and processes the signals 325. The receive beamformer 340 applies delays and/or phases to the signals and the resultant signals are summed for focusing and steering a received ultrasound beam. The receive beamformer 340 may also apply apodization, amplification and filtering.

In an alternate embodiment, an ultrasound beam 1620 for acoustic radiation force may be steered by applying appropriate delays for ultrasound beam angle steering as shown in Figure 16. As an example, the ultrasound beam 1620 is steered to right in Figure 16. Shear waves may be also detected by using steered transmit ultrasound beams 1720, 1820, 1830 as shown in Figures 17 and 18.

Shear wave propagation velocity and velocity squared may be determined at every image point as previously discussed, using ultrasound beams transmitted at two or more steered angles. Then, a shear wave propagation velocity or square of shear wave propagation velocity for a given image point may be determined based on (e.g., by averaging) each of the two or more velocities or squared velocities determined for the given image point. This process may improve the accuracy of the resulting image.

For example, as described above, a first ultrasound pulse may be applied to biological tissue to create shear waves in the biological tissue in a first direction. Next, a focused ultrasound pulse is transmitted into the biological tissue in a second direction. A first one or more ultrasound signals generated in response to the focused ultrasound pulse is then received from the biological tissue, and shear waves are detected in the biological tissue based on the received first one or more ultrasound signals. A first set of at least one shear wave propagation property associated with the detected shear waves (e.g., shear wave propagation velocity and/or velocity squared) for each image pixel in the field of view is then determined.

Figure 19 illustrates image 1950 of a first set of at least one shear wave propagation property determined as described above. According to Figure 19, the focused ultrasound pulse has been transmitted into the biological tissue at a 0 degree beam steering angle. The first set consists of a value of the shear wave propagation property for each point in image 1950. In other words, the value of the shear wave propagation property determined for a given point in image 1950 determines the value assigned to the image pixel which represents that point.

Next, a second ultrasound pulse may be applied to the biological tissue to create second shear waves in the biological tissue in a third direction, and a second focused ultrasound pulse is transmitted into the biological tissue in a fourth direction. A second one or more ultrasound signals generated in response to the second focused ultrasound pulse is then received from the biological tissue, and the second shear waves are detected in the biological tissue based on the received second one or more ultrasound signals. A second set of at least one shear wave propagation property associated with the detected second shear waves (e.g., shear wave propagation velocity and/or velocity squared) for each image pixel in the field of view is then determined.

Figure 20 illustrates image 2050 of a second set of at least one shear wave propagation property determined as described above. The focused ultrasound pulse of the Figure 20 example has been transmitted into the biological tissue at a beam steering angle of 10 degrees to the left. The second set consists of a value of the shear wave propagation property for each point in image 2050, where the value of the shear wave propagation property determined for a given point in image 2050 determines the value assigned to the image pixel which represents that point.

Additionally, Figure 21 illustrates image 2150 of a third set of at least one shear wave propagation property determined as described above, using a focused ultrasound pulse transmitted into the biological tissue at a beam steering angle of 10 degrees to the right (i.e., -10 degrees). Again, the third set consists of a value of the shear wave propagation property for each point in image 2150, where the value of the shear wave propagation property determined for a given point in image 2150 determines the value assigned to the image pixel which represents that point.

Next, a fourth set of shear wave propagation properties are determined based on the determined sets of shear wave propagation properties. According to the present example, the shear wave propagation property values determined for a given point are averaged to determine a composite shear wave propagation property value for the given point. Then, an image is generated in which the composite value of each given point is used to determine the value assigned to the image pixel which represents the given point.

The shear waves which are created in the biological tissue as described with respect to Figures 19 through 21 may travel in any direction, depending on the direction of the applied acoustic radiation forces, and one or more of these shear waves may travel in a same direction.

Figure 22 depicts image 2250 generated based on composite values as described above. For example, shear wave velocities determined at areas (i.e., pixels) 1970, 2070, 2170 (i.e., C_{1970}, C_{2070} and C_{2170}) are averaged to yield mean shear wave velocity $C_{2270} =(C_{1970}+C_{2070}+C_{2170})/3$, which is used to determine image pixel value at area (i.e., pixel) 2270. Alternatively, the image pixel value at area (i.e., pixel) 2210 may be determined based on the average shear velocity squared $(C_{2270})^2=((C_{1970})^2+(C_{2070})^2+(C_{2170})^2)/3$.

Region 2210 of image 2250 is therefore composed of image pixels whose values are based on shear wave propagation property values represented in images 1950, 2050 and 2150. However, due to the different fields of view of images 1950, 2050 and 2150, some regions of image 2250 are determined based on only two or one of images 1950, 2050 and 2150. For example, region 2220 is composed of image pixels whose values are based on shear wave propagation property values represented in images 1950 and 2050, region 2230 is composed of image pixels whose values are based on shear wave propagation property values represented in images 1950 and 2150, region 2240 is composed of image pixels whose values are based on shear wave propagation property values of image 2050, and region 2260 is composed of image pixels whose values are based on shear wave propagation property values of image 2150.

Different ultrasound speckle signals result from different steering angles, thus the above-described averaging improves the accuracy of the determination of shear wave propagation velocity and velocity squared more effectively. Different ultrasound beam steering angles create less correlated ultrasound signals or uncorrelated ultrasound signals. Averaging uncorrelated signals will result in reduction of uncorrelated noise in the signals and thus better improvement of measurement accuracy than averaging correlated signals. Therefore, the beam steering technique described above will improve measurement accuracy of shear wave propagation velocity or velocity squared.

Although averaging is discussed above, any mathematical function may be applied to multiple propagation property values for a given point to determine a composite value for the given point. The above discussion also contemplates the use of three beam steering angles in order to improve measurement accuracy. However, the number of beam steering angles may be two or more than three. Also, a beam steering angle may be other than 0, 10, and/or -10 degrees. The beam steering angle of an ultrasound pulse to create shear waves may be different from the beam steering angle of a focused ultrasound pulse used to detect such shear waves.

The processed signal 345 is coupled to a Doppler spectrum processor 350, a color Doppler processor 360, and a B-mode image processor 370. The Doppler spectrum processor 350 includes a Doppler signal processor and a spectrum analyzer, and processes Doppler flow velocity signals and calculates and outputs a Doppler spectrum 355. The color Doppler processor 360 processes the received signal 345 and calculates and outputs velocity, power and variance signals 365. The B-mode image processor 370 processes the received signal 345 and calculates and outputs a B-mode image 375 or the amplitude of the signal by an amplitude detection.

The Doppler spectrum signals 355, color Doppler processor signals (velocity, power, and variance) 365 and B-mode processor signals 375 are coupled to a scan converter 380 that converts the signals to scan-converted signals. The output of scan converter 380 is coupled to a display monitor 390 for displaying ultrasound images.

Figure 2A shows a diagram of elements of an ultrasound imaging system including a shear wave processor 295 according to some embodiments. The ultrasound system in Figure 2A transmits strong ultrasound pulses to biological tissue to create acoustic radiation forces which push the biological tissue. Shear waves are created and propagate in the tissue after the biological tissue is pushed. The ultrasound system then transmits and receives ultrasound pulses to track the shear waves as the shear waves propagate in the biological tissue. Multiple received ultrasound beams may be simultaneously formed by the receive beamformer 240. Likewise, multiple transmitted ultrasound beams may be simultaneously formed by the transmitter/transmit beamformer 210. Received ultrasound signals from the receive beamformer 240 are processed to obtain tissue displacement, Doppler velocity, correlation, shear wave propagation velocity and/or shear wave propagation velocity squared as previously described. The shear wave processor 295 may perform the shear wave processing methods described previously. The shear wave processor 295 receives output 245 from the receive beamformer 240. Output 297 comprises shear wave velocity data or other shear wave properties. For example, the shear wave processor 295 outputs the propagation velocity or the square of the propagation velocity of the shear wave to a scan converter 280 and a representation of the shear wave propagation velocity or the square of the shear wave propagation velocity is output to the display monitor 290 along with the B-mode, color Doppler or spectral Doppler images via a composite image processor 285.

For the B-mode signals, the data from the B-mode image processor 275 are line data which consist of processed beam signals for each receive ultrasound beam and may not have signals for all image pixels with the correct vertical-to-horizontal distance relationship for the display. Line data may be also vector data in the direction of the ultrasound beam and not necessary in the direction of (x, z) display. The scan converter 280 interpolates the line data in two dimensions (x, z) and fills in all image pixels with ultrasound image data. Also, color Doppler data 265 are line data which consist of processed beam signals for each receive color Doppler beam and may not include signals for all image pixels having the correct vertical-to-horizontal distance relationship for the display. The scan converter 280 interpolates the line data in two dimensions (x, z) and fills in all color Doppler image pixels with scan-converted color Doppler image data. Likewise, shear wave data 297 may also be line data thus may require scan-conversion. The scan converter 280 interpolates the line data in two dimensions (x, z) and fills in all shear wave image pixels with scan-converted shear wave image data.

The composite image processor 285 receives multiple images of shear wave properties (e.g., shear wave velocity, shear wave velocity squared) obtained at multiple beam steering angles and calculates a composite image, e.g., an averaged image or an image calculated based on multiple images. For averaging of image signals at 2 beam steering angles, a composite image signal $I_{x,z}$ at an image position (x, z) may be obtained from an image signal $I_{1,x,z}$ at the same image position (x, z) obtained at the first beam steering angle and an image signal $I_{2,x,z}$ at the same image position (x, z) at the second beam steering angle. The image signal $I_{x,z}$ may be either the shear wave velocity or the shear wave velocity squared.

$$I_{x,z} = \frac{I_{1,x,z}+I_{2,x,z}}{2} \quad (21)$$

For averaging of images at 3 beam steering angles, a first image $I_{1,x,z}$, a second image $I_{2,x,z}$ and a third image $I_{3,x,z}$ may be averaged at each image position (x, z) as follows.

$$I_{x,z} = \frac{I_{1,x,z}+I_{2,x,z}+I_{3,x,z}}{3} \quad (22)$$

Alternately, a composite image may be calculated as a function f of multiple images $I_{1,x,z}, I_{2,x,z}$, ... at multiple beam steering angles at each image pixel position (x, z) as follows.

$$I_{x,z} = f(I_{1,x,z}, I_{2,x,z}, ...) \quad (23)$$

The composite image processor 285 may comprise of image processor 284 and multiple memories 281, 282, 283 which store multiple images. The multiple images are used to calculate a composite image of shear wave properties (e.g., shear wave velocity or shear wave velocity squared) as shown in Figure 2B.

The foregoing discussion relates to two-dimensional images. However, the averaging or mathematical image function f may be performed in three-dimensional images (or volumes) of shear wave propagation property (e.g., shear wave velocity or shear wave velocity squared).

Transmitter 210 may contain a transmit beamformer which may apply time delays to signals for transducer elements for focusing and beam steering. For example, a first set of transmit time delays are either generated or read from memory and loaded to a transmit delay table, and a first set of receive time delays/phases are generated or read from memory and loaded to a receive delay table. A first shear wave image (i.e. shear wave velocity or shear wave velocity squared) is then acquired at a first beam steering angle. Next, a second set of transmit time delays are either generated or read from memory and loaded to the transmit delay table and a second set of receive time delays/phases are generated or read from memory and loaded to the receive delay table. A second shear wave image is then acquired at a second beam steering angle. This process continues multiple times as the transmit beamformer and the receive beamformer update each of the delay tables and multiple shear wave images are acquired at multiple beam steering angles.

The shear wave processor 295 may comprise of general purpose central processing units (CPUs), digital signal processors (DSPs), field programmable Arrays (FPGAs), graphic processing units (GPUs) and/or discreet electronics devices.

Figure 2A represents a logical architecture according to some embodiments, and actual implementations may include more or different elements arranged in other manners. Other topologies may be used in conjunction with other embodiments. Moreover, each element of the Figure 2A system may be implemented by any number of computing devices in communication with one another via any number of other public and/or private networks. Two or more of such computing devices may be located remote from one another and may communicate with one another via any known manner of network(s) and/or a dedicated connection. The system may comprise any number of hardware and/or software elements suitable to provide the functions described herein as well as any other functions. For example, any computing device used in an implementation of the Figure 2A system may include a processor to execute program code such that the computing device operates as described herein.

All systems and processes discussed herein may be embodied in program code stored on one or more non-transitory computer-readable media. Such media may include, for example, a floppy disk, a CD-ROM, a DVD-ROM, a Blu-ray disk, a Flash drive, magnetic tape, and solid state Random Access Memory (RAM) or Read Only Memory (ROM) storage units. Embodiments are therefore not limited to any specific combination of hardware and software.

One or more embodiments have been described. Nevertheless, various modifications will be apparent to those in the art.

Claims

1. A method comprising:

applying a first ultrasound pulse to biological tissue to create shear waves in the biological tissue in a first direction

transmitting a focused ultrasound pulse into the biological tissue in a second direction;

receiving a first one or more ultrasound signals from the biological tissue generated in response to the focused ultrasound pulse;

detecting the shear waves in the biological tissue based on the received first one or more ultrasound signals;

determining a first set of at least one shear wave propagation property associated with the detected shear waves;

applying a second ultrasound pulse to biological tissue to create second shear waves in the biological tissue in a third direction;

transmitting a second focused ultrasound pulse into the biological tissue in a fourth direction;

receiving a second one or more ultrasound signals from the biological tissue generated in response to the second focused ultrasound pulse;

detecting the second shear waves in the biological tissue based on the second received one or more ultrasound signals;

determining a second set of at least one shear wave propagation property associated with the second detected shear waves;

determining a third set of at least one shear wave propagation property based on the first and second sets of at least one propagation property; and displaying the third set of at least one shear wave propagation property.

2. A method according to claim 1, wherein the first, second or third sets of the at least one shear wave propagation property comprises one or more of;

a propagation velocity associated with one or more of the detected shear waves; and

a product (bc^2) of a real number (b) and the square of the shear wave propagation velocity (c^2).

3. A method according to claim 1, wherein detecting the shear waves comprises calculating a correlation, the sum of absolute differences (SAD), the sum of square differences (SSD), the sum of absolute cubic differences (SCD) or the sum of absolute power differences (SPD) between the received ultrasound signals at one or multiple positions in time.

4. A method according to claim 1, wherein determining the first or second sets of the at least one shear wave propagation property comprises calculating a correlation, the sum of absolute differences (SAD), the sum of square differences (SSD), the sum of absolute cubic differences (SCD) or the sum of absolute power differences (SPD) between the detected shear waves at one or multiple instances.

5. A method according to claim 1, further comprising transmitting a third focused ultrasound pulse from a transducer in the second direction after transmitting the focused ultrasound pulse into the biological tissue in the second direction and before the focused

ultrasound pulse returns to the transducer from a deepest position in an ultrasound field.

6. A method according to claim 1, wherein the transmitted focused ultrasound pulses comprise coded waveform signals.

7. A method according to claim 6, wherein the coded waveform signals comprise one of Chirp codes, Barker codes, Golay codes or Hadamard codes.

8. A method according to claim 1, wherein displaying the third set of at least one shear wave propagation property comprises:

displaying a graphical representation of the third set of at least one shear wave propagation property using color coding, grayscale coding or numerals.

9. A method according to claim 8, wherein the color coding is based on RGB (Red, Green, Blue) values, RGBY (Red, Green, Blue, Yellow) values, Hue, luminance, the wavelength or a color chart.

10. A method according to claim 1, wherein detecting the shear waves comprises determining a displacement of the biological tissue.

11. A method according to claim 1, wherein detecting the shear waves comprises determining a velocity of the biological tissue using color Doppler technique.

12. A method according to claim 2, wherein the shear wave propagation velocity

is calculated based on the square root of the ratio between a temporal second-order derivative of the displacement of the biological tissue and a spatial second-order derivative of the displacement of the biological tissue.

13. A method according to claim 2, wherein the square of the shear wave propagation velocity is calculated based on the ratio between a temporal second-order derivative of the displacement of the biological tissue and a spatial second-order derivative of the displacement of the biological tissue.

14. A method according to claim 10, wherein determining a displacement of the biological tissue comprises calculating a time integral of tissue color Doppler velocity.

15. A method according to claim 1, wherein applying the first ultrasound pulse comprises applying a plurality of ultrasound pulses to the biological tissue to create shear waves in the biological tissue,

wherein each of the plurality of ultrasound pulses is focused at a different focal point.

16. A method according to claim 1, wherein transmitting the focused ultrasound pulse comprises transmitting a plurality of focused ultrasound pulses into the biological tissue more than one time in a same direction, and

wherein the one or more ultrasound signals are received from the biological tissue at one or more instances.

17. A non-transitory medium storing processor-executable program code, the program code executable by a device to:

apply a first ultrasound pulse to biological tissue to create shear waves in the biological tissue in a first direction;

transmit a focused ultrasound pulse into the biological tissue in a second direction;

receive a first one or more ultrasound signals from the biological tissue generated in response to the focused ultrasound pulse;

detect the shear waves in the biological tissue based on the received one or more ultrasound signals;

determine a first set of at least one shear wave propagation property associated with the detected shear waves;

apply a second ultrasound pulse to biological tissue to create shear waves in the biological tissue in a third direction;

transmit a second focused ultrasound pulse into the biological tissue in a fourth direction;

receive a second one or more ultrasound signals from the biological tissue generated in response to the second focused ultrasound pulse;

detect the second shear waves in the biological tissue based on the second received one or more ultrasound signals;

determine a second set of at least one shear wave propagation property associated with the second detected shear waves;

determine a third set of at least one shear wave propagation property based on the first and second sets of at least one shear wave propagation property; and display the third set of at least one shear wave propagation property.

18. A medium according to claim 17, wherein the first, second or third sets of the at least one shear wave propagation property comprises one or more of;

a propagation velocity associated with one or more of the detected shear waves; and

a product (bc^2) of a real number (b) and the square of the shear wave propagation velocity (c^2).

19. A medium according to claim 17, wherein detection of the shear waves comprises calculation of a correlation, the sum of absolute differences (SAD), the sum of square differences (SSD), the sum of absolute cubic differences (SCD) or the sum of absolute power differences (SPD) between the received ultrasound signals at one or multiple positions in time.

20. A medium according to claim 17, wherein determination of the first or second sets of the at least one propagation property comprises calculation of a correlation, the sum of absolute differences (SAD), the sum of square differences (SSD), the sum of absolute cubic differences (SCD) or the sum of absolute power differences (SPD) between the detected shear waves at one or multiple instances.

21. A medium according to claim 17, further comprising transmission of a third focused ultrasound pulse from a transducer in the second direction after transmitting the focused ultrasound pulse into the biological tissue in the second direction and before the

focused ultrasound pulse returns to the transducer from a deepest position in an ultrasound field.

22. A medium according to claim 17, wherein the transmitted focused ultrasound pulses comprise coded waveform signals.

23. A medium according to claim 22, wherein the coded waveform signals comprise one of Chirp codes, Barker codes, Golay codes or Hadamard codes.

24. A medium according to claim 17, wherein display of the third set of at least one shear wave propagation property comprises display of a graphical representation of the third set of at least one shear wave propagation property using color coding, grayscale coding or numerals.

25. A medium according to claim 24, wherein the color coding is based on RGB (Red, Green, Blue) values, RGBY (Red, Green, Blue, Yellow) values, Hue, luminance, the wavelength or a color chart.

26. A medium according to claim 17, wherein detection of the shear waves comprises determination of a displacement of the biological tissue.

27. A medium according to claim 17, wherein detection of the shear waves comprises determination of a velocity of the biological tissue using color Doppler technique.

28. A medium according to claim 18, wherein the shear wave propagation velocity is calculated based on the square root of the ratio between a temporal second-order derivative of the displacement of the biological tissue and a spatial second-order derivative of the displacement of the biological tissue.

29. A medium according to claim 18, wherein the square of the shear wave propagation velocity is calculated based on the ratio between a temporal second-order derivative of the displacement of the biological tissue and a spatial second-order derivative of the displacement of the biological tissue.

30. A medium according to claim 26, wherein determination of a displacement of the biological tissue comprises calculation of a time integral of tissue color Doppler velocity.

31. A medium according to claim 17, wherein application of the first ultrasound pulse comprises application of a plurality of ultrasound pulses to the biological tissue to create shear waves in the biological tissue,

wherein each of the plurality of ultrasound pulses is focused at a different focal point.

32. A medium according to claim 17, wherein transmission of the focused ultrasound pulse comprises transmission of a plurality of focused ultrasound pulses into the biological tissue more than one time in a same direction, and

wherein the one or more ultrasound signals are received from the biological tissue at one or more instances.

33. A system comprising:

a memory storing processor-executable program code; and

a processor to execute the processor-executable program code in order to cause the system to:

apply a first ultrasound pulse to biological tissue to create shear waves in the biological tissue in a first direction;

transmit a focused ultrasound pulse into the biological tissue in a second direction;

receive a first one or more ultrasound signals from the biological tissue generated in response to the focused ultrasound pulse;

detect the shear waves in the biological tissue based on the received first one or more ultrasound signals;

determine a first set of at least one shear wave propagation property associated with the detected shear waves;

apply a second ultrasound pulse to biological tissue to create shear waves in the biological tissue in a third direction;

transmit a second focused ultrasound pulse into the biological tissue in a fourth direction;

receive a second one or more ultrasound signals from the biological tissue generated in response to the second focused ultrasound pulse;

detect the second shear waves in the biological tissue based on the second received one or more ultrasound signals;

determine a second set of at least one shear wave propagation property associated with the second detected shear waves;

determine a third set of at least one shear wave propagation property based on the first and second sets of at least one propagation property; and

display the third set of at least one shear wave propagation property .

34. A system according to claim 33, wherein the first, second or third sets of the at least one shear wave propagation property comprises one or more of;

a propagation velocity associated with one or more of the detected shear waves; and

a product (bc^2) of a real number (b) and the square of the shear wave propagation velocity (c^2).

35. A system according to claim 33, wherein detection of the shear waves comprises calculation of a correlation, the sum of absolute differences (SAD), the sum of square differences (SSD), the sum of absolute cubic differences (SCD) or the sum of absolute power differences (SPD) between the received ultrasound signals at one or multiple positions in time.

36. A system according to claim 33, wherein determination of the first or second sets of the at least one propagation property comprises calculation of a correlation, the sum of absolute differences (SAD), the sum of square differences (SSD), the sum of

absolute cubic differences (SCD) or the sum of absolute power differences (SPD) between the detected shear waves at one or multiple instances.

37. A system according to claim 33, the processor further to execute the processor-executable program code in order to cause the system to:

transmit a third focused ultrasound pulse from a transducer in the second direction after transmitting the focused ultrasound pulse into the biological tissue in the second direction and before the focused ultrasound pulse returns to the transducer from a deepest position in an ultrasound field.

38. A system according to claim 33, wherein the transmitted focused ultrasound pulses comprise coded waveform signals.

39. A system according to claim 38, wherein the coded waveform signals comprise one of Chirp codes, Barker codes, Golay codes or Hadamard codes.

40. A system according to claim 33, wherein display of the third set of at least one shear wave propagation property comprises display of a graphical representation of the third set of at least one shear wave propagation property using color coding, grayscale coding or numerals.

41. A system according to claim 40, wherein the color coding is based on RGB (Red, Green, Blue) values, RGBY (Red, Green, Blue, Yellow) values, Hue, luminance, the wavelength or a color chart.

42. A system according to claim 33, wherein detection of the shear waves comprises determination of a displacement of the biological tissue.

43. A system according to claim 33, wherein detection of the shear waves comprises determination of a velocity of the biological tissue using color Doppler technique.

44. A system according to claim 34, wherein the shear wave propagation velocity is calculated based on the square root of the ratio between a temporal second-order derivative of the displacement of the biological tissue and a spatial second-order derivative of the displacement of the biological tissue.

45. A system according to claim 34, wherein the square of the shear wave propagation velocity is calculated based on the ratio between a temporal second-order derivative of the displacement of the biological tissue and a spatial second-order derivative of the displacement of the biological tissue.

46. A system according to claim 42, wherein determination of a displacement of the biological tissue comprises calculation of a time integral of tissue color Doppler velocity.

47. A system according to claim 33, wherein application of the first ultrasound pulse comprises application of a plurality of ultrasound pulses to the biological tissue in a

same direction to create shear waves in the biological tissue,

wherein each of the plurality of ultrasound pulses is focused at a different focal point.

48. A system according to claim 33, wherein transmission of the focused ultrasound pulse comprises transmission of a plurality of focused ultrasound pulses into the biological tissue more than one time in a same direction, and

wherein the one or more ultrasound signals are received from the biological tissue at one or more instances.

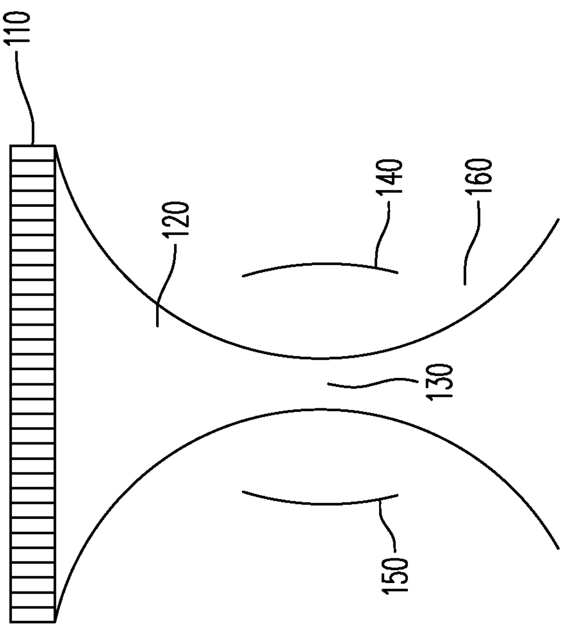


Figure 1

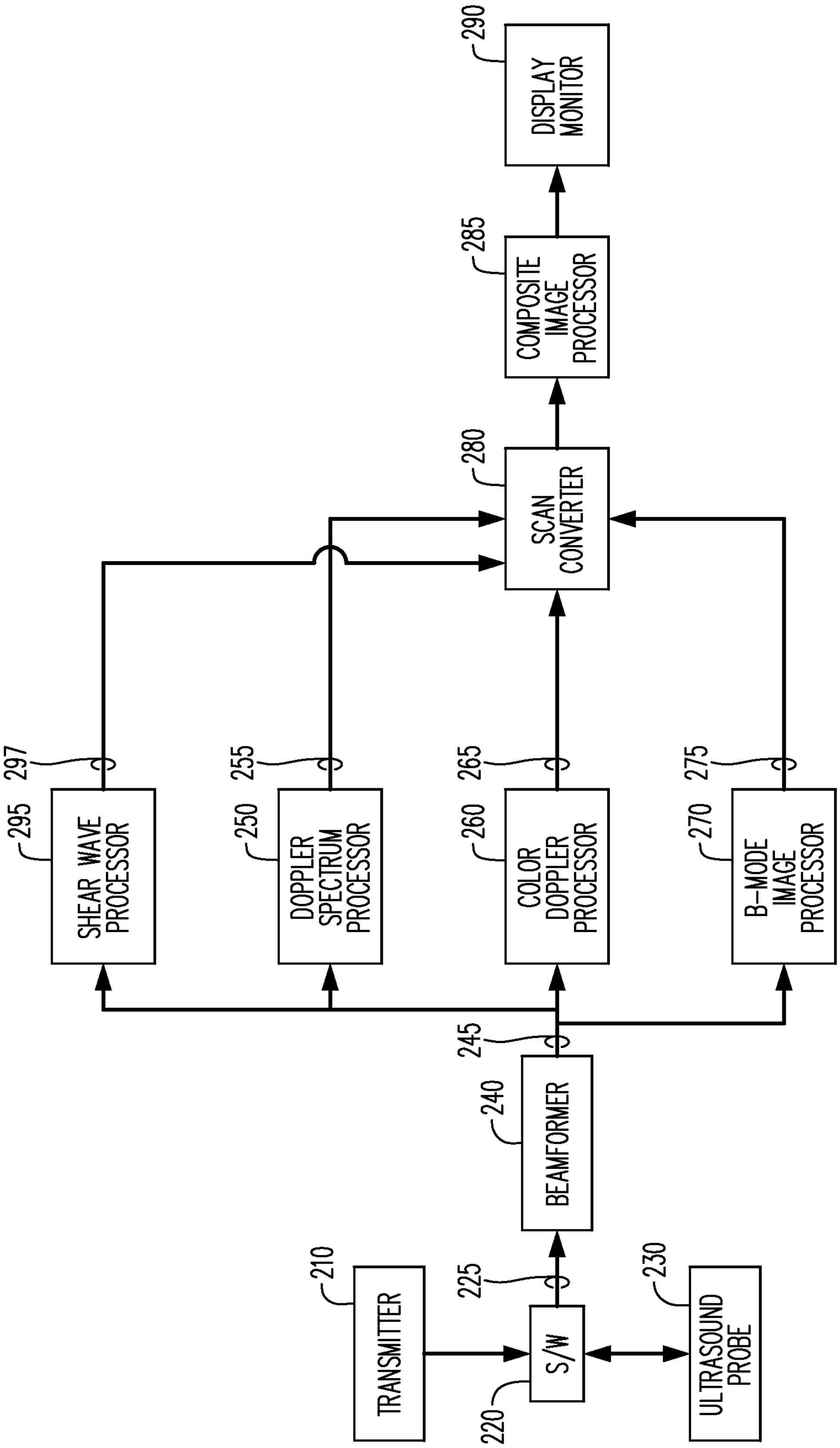

210
TRANSMITTER
220
S/W
225
230
ULTRASOUND PROBE
240
BEAMFORMER
245
295
SHEAR WAVE PROCESSOR
297
250
DOPPLER SPECTRUM PROCESSOR
255
260
COLOR DOPPLER PROCESSOR
265
270
B-MODE IMAGE PROCESSOR
275
280
SCAN CONVERTER
285
COMPOSITE IMAGE PROCESSOR
290
DISPLAY MONITOR

Figure 2A

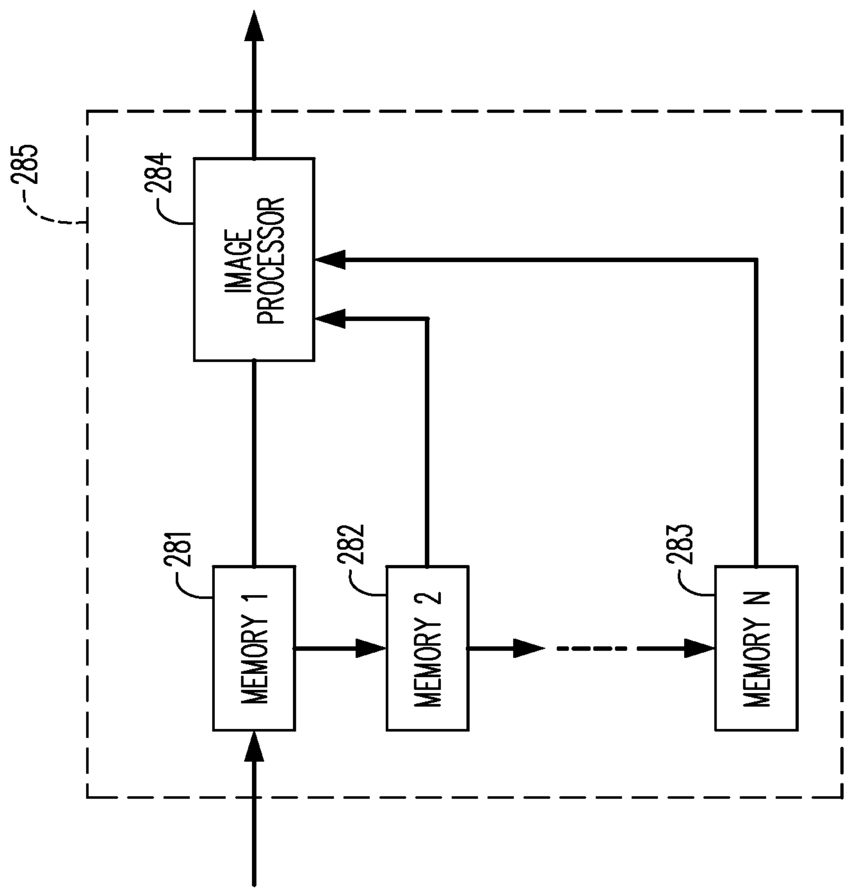

285
284
IMAGE PROCESSOR
281
MEMORY 1
282
MEMORY 2
283
MEMORY N

Figure 2B

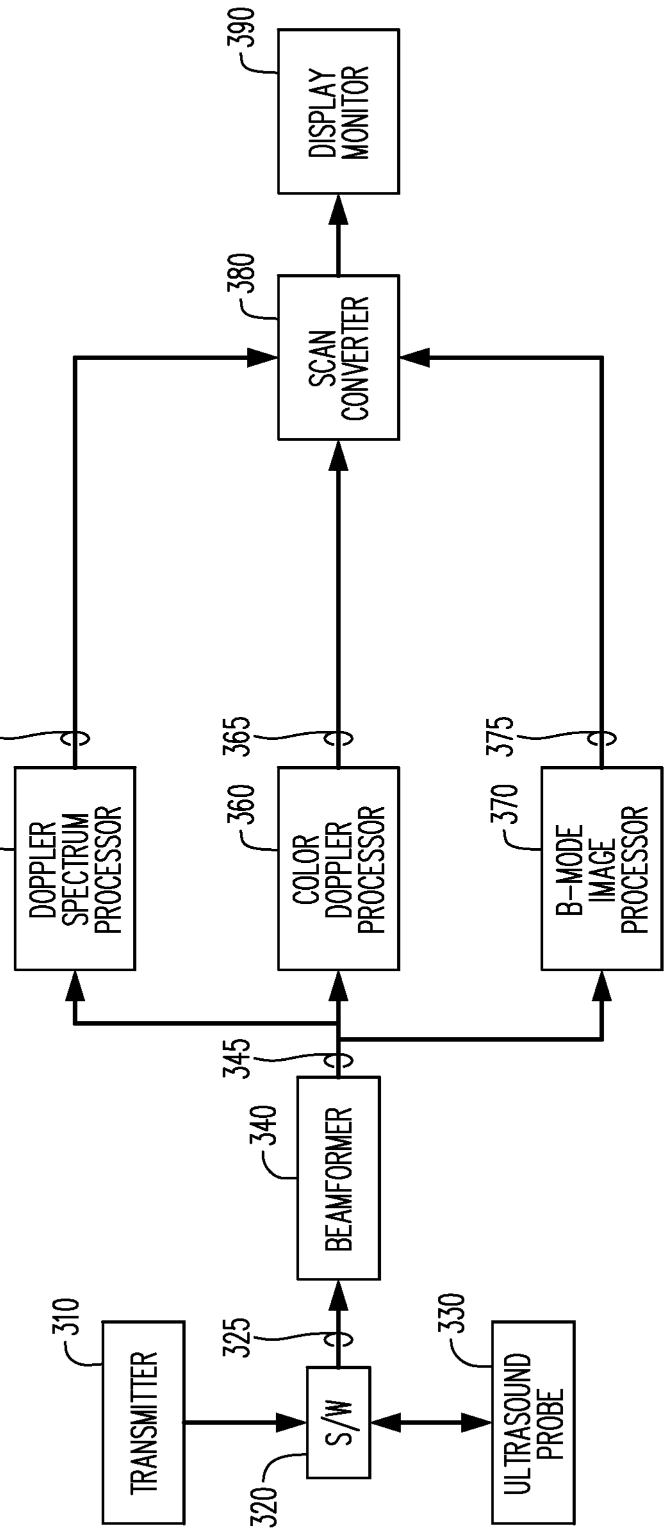


Figure 3
(PRIOR ART)

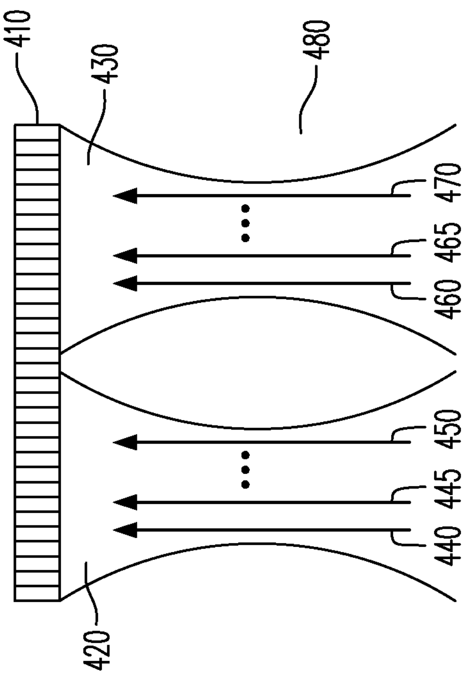


Figure 4

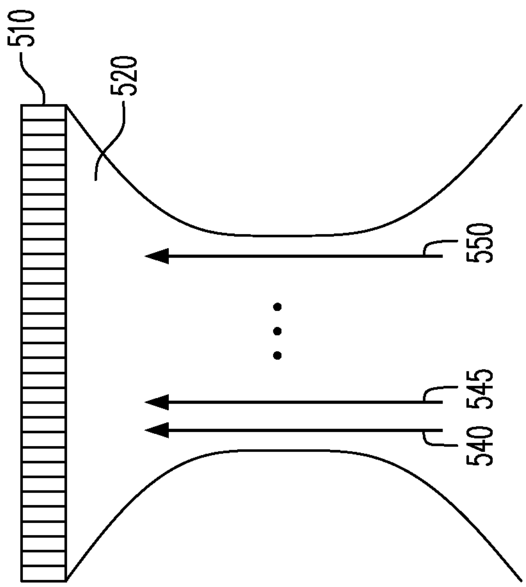


Figure 5

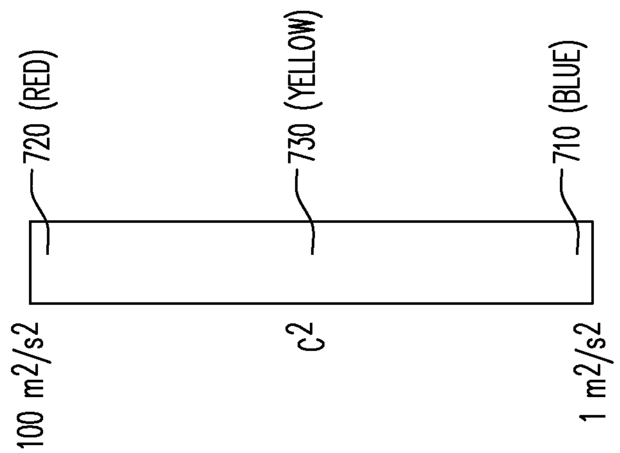


Figure 7

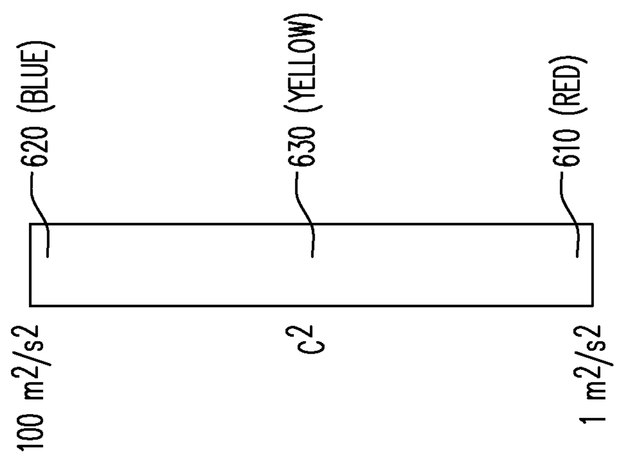
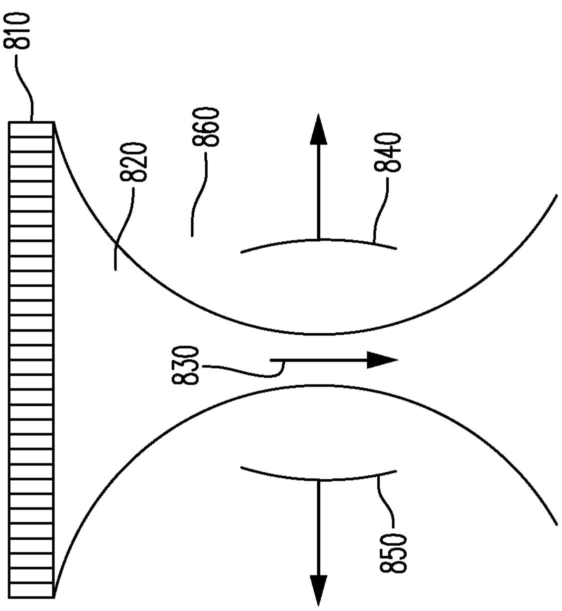


Figure 6



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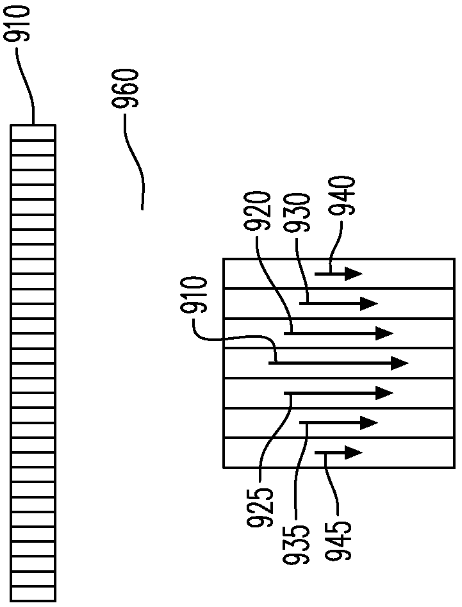


Figure 9

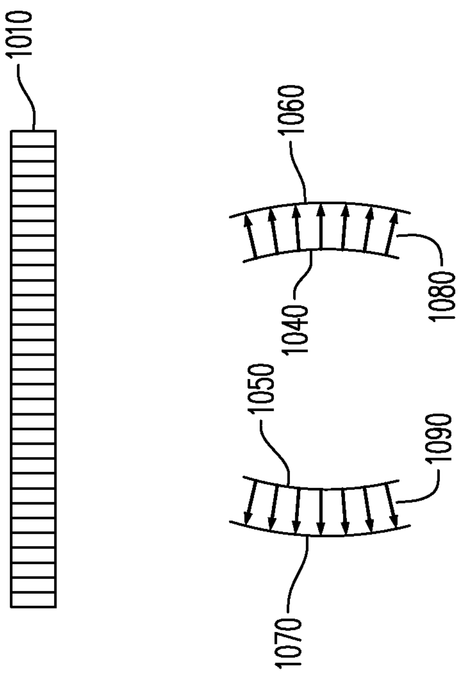


Figure 10

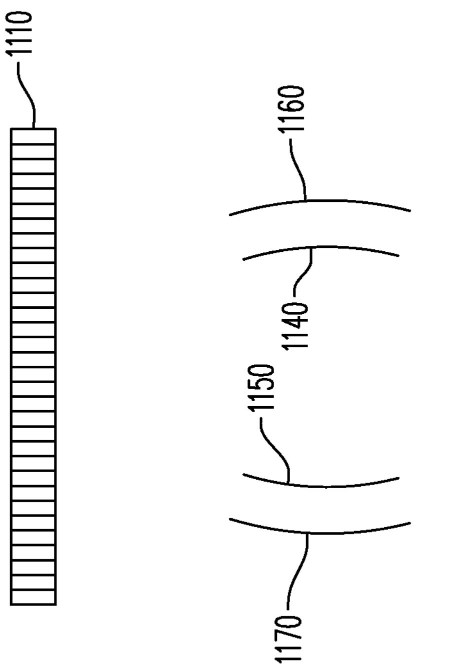


Figure 11

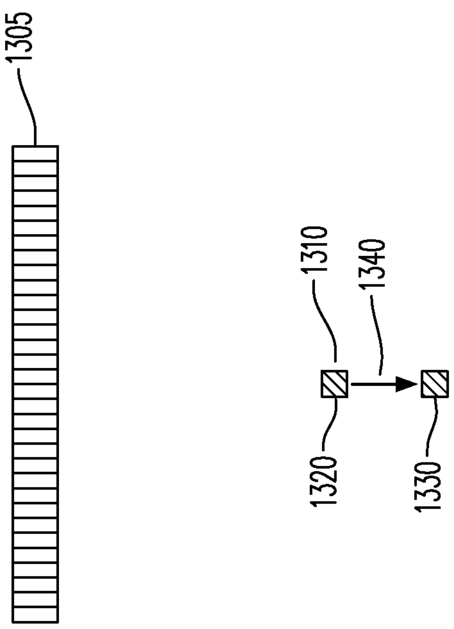
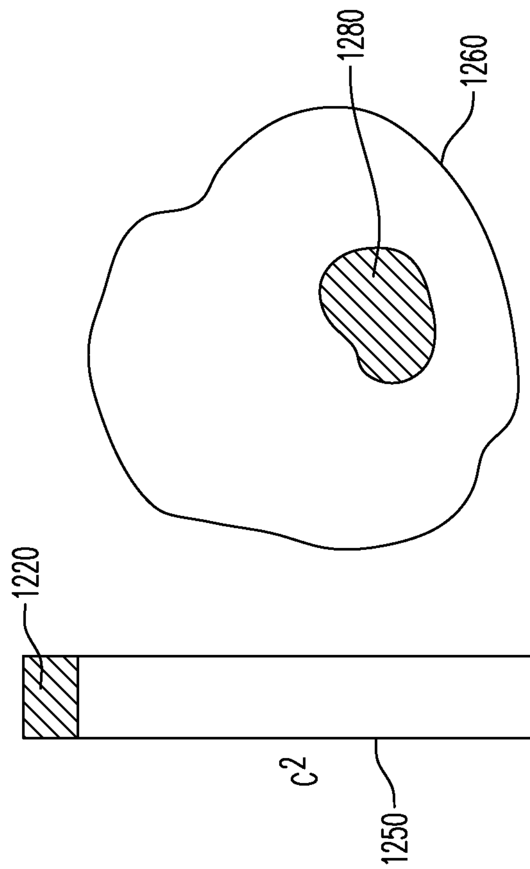


Figure 13



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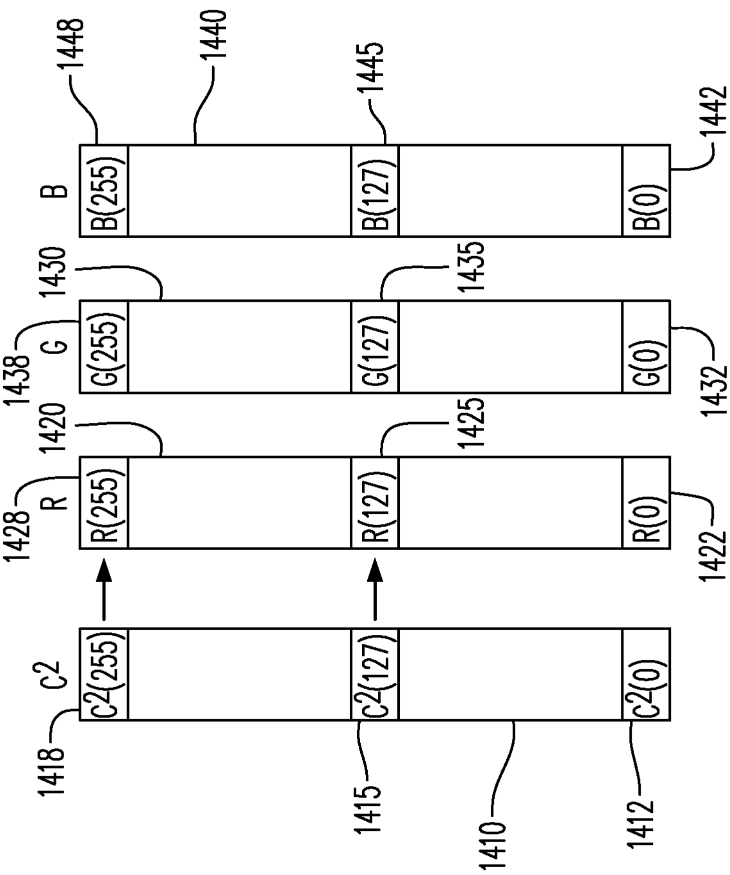

Figure 14

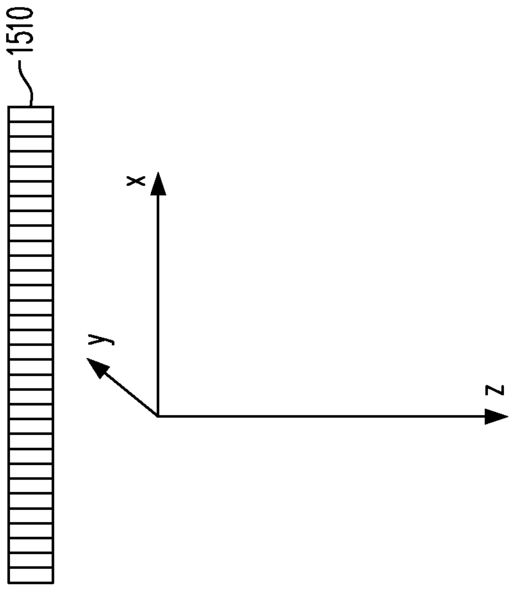

Figure 15

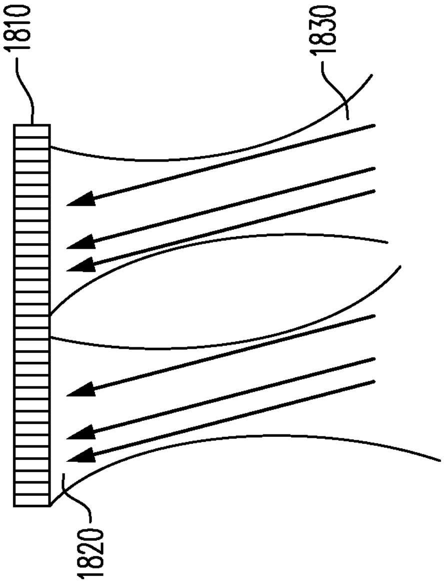

Figure 18

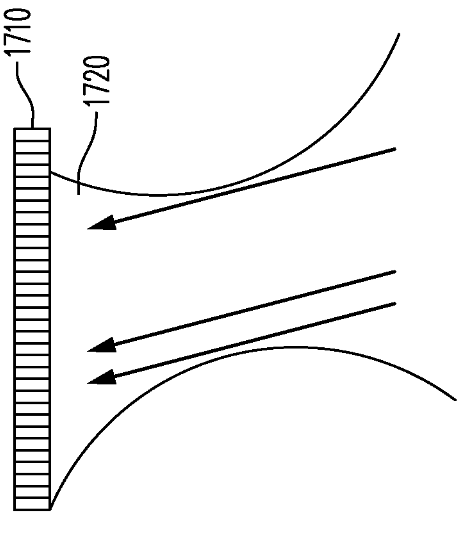

Figure 17

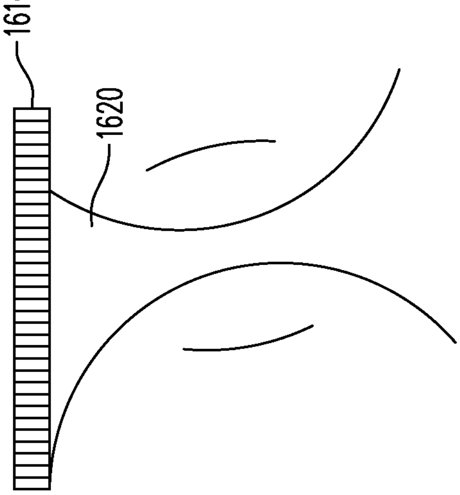

Figure 16

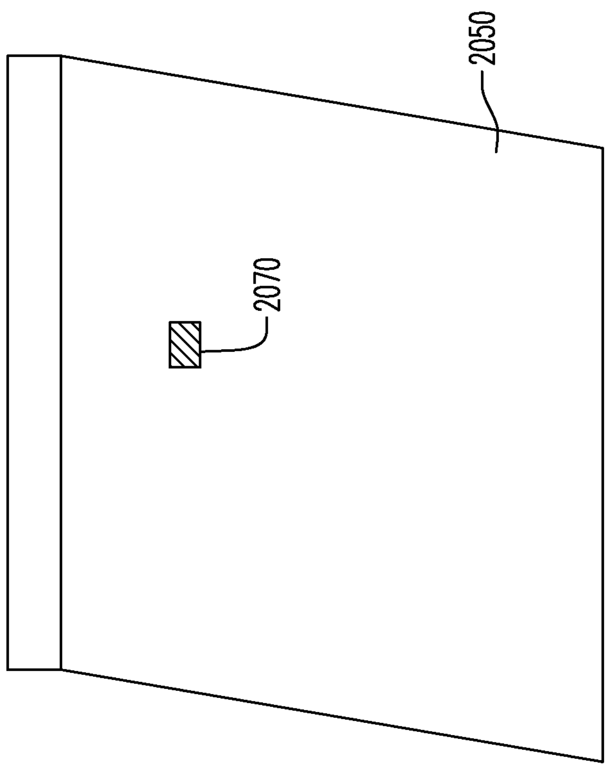

Figure 20

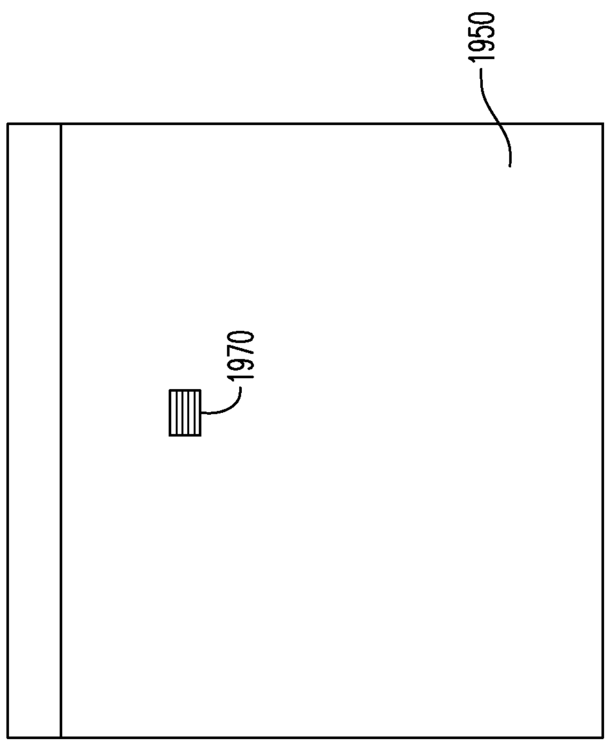

Figure 19

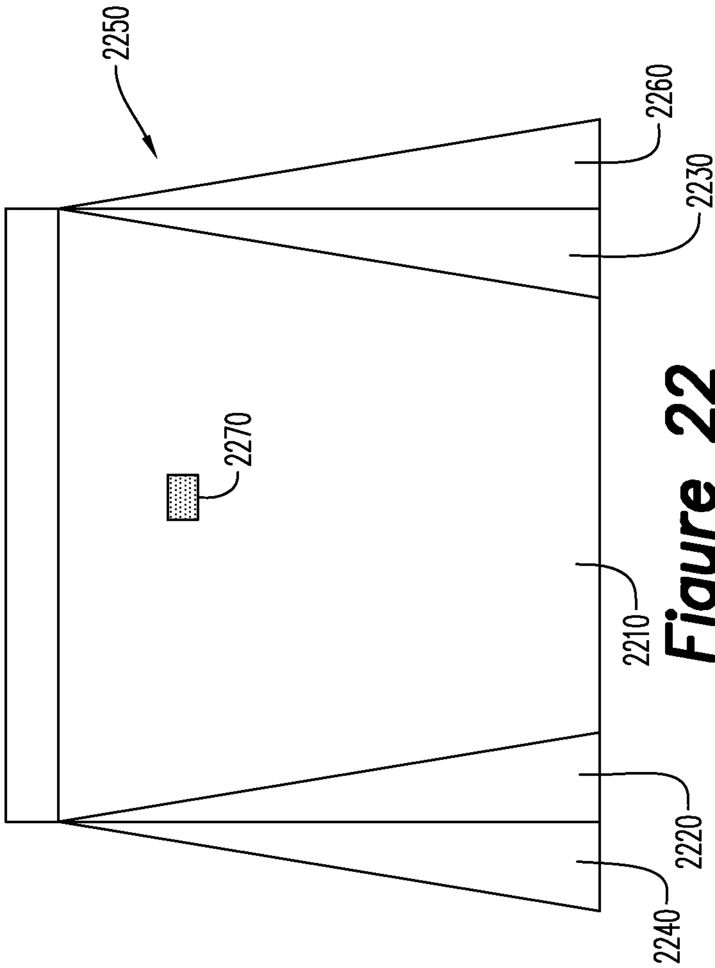

Figure 22

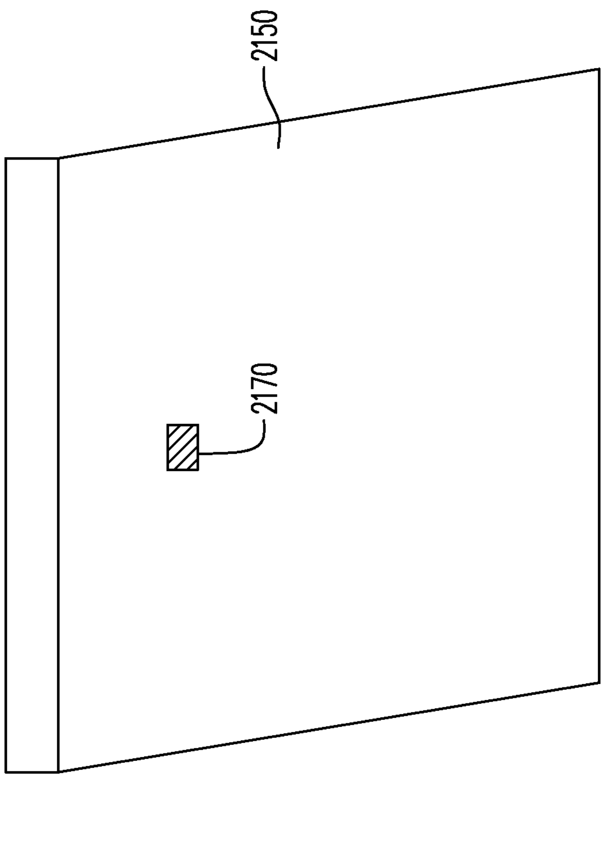

Figure 21

patsnap

专利名称(译)	用于超声成像的方法和设备		
公开(公告)号	EP2691026A2	公开(公告)日	2014-02-05
申请号	EP2012763027	申请日	2012-03-30
[标]申请(专利权)人(译)	日立阿洛卡医疗株式会社		
申请(专利权)人(译)	日立ALOKA MEDICAL.，LTD.		
当前申请(专利权)人(译)	HITACHI，LTD.		
[标]发明人	TAMURA TADASHI		
发明人	TAMURA, TADASHI		
IPC分类号	A61B8/08 G06F19/00 G01S15/89		
CPC分类号	A61B8/485 A61B8/0825 A61B8/463 A61B8/466 A61B8/483 A61B8/488 G01S7/52038 G01S7/52071 G01S7/5209 G01S7/52095 G01S15/8959		
优先权	61/469295 2011-03-30 US 13/314736 2011-12-08 US		
其他公开文献	EP2691026A4		
外部链接	Espacenet		

摘要(译)

将第一超声脉冲施加到生物组织以在生物组织中产生剪切波，将聚焦的超声脉冲发射到生物组织中，从生物组织接收一个或多个超声信号，并且在生物组织中检测剪切波。基于所接收的一个或多个超声信号。确定与检测到的剪切波相关联的至少一个剪切波传播特性，并且显示所确定的至少一个传播特性。超声波束转向用于提高测量精度。