



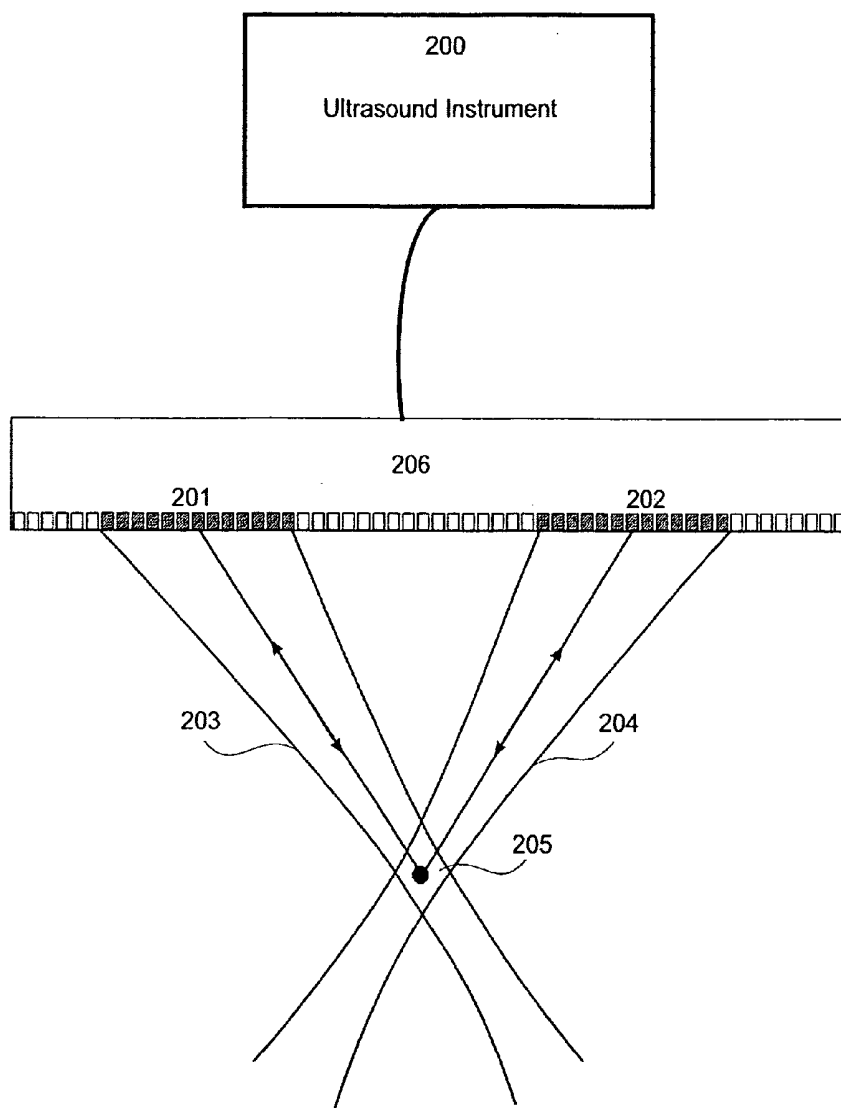
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(19) **United States**(12) **Patent Application Publication** (10) **Pub. No.: US 2005/0182324 A1**
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Angelsen et al.(54) **METHODS OF ESTIMATING ULTRASOUND
SCATTERER PARAMETERS IN SOFT
TISSUES****Publication Classification**(51) **Int. Cl.⁷** **A61B 8/00**(52) **U.S. Cl.** **600/437**(76) **Inventors: Bjorn A.J. Angelsen, Trondheim (NO);
Tonni F. Johansen, Trondheim (NO)**(57) **ABSTRACT**

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A method for characterizing ultrasound scatterers in soft tissue, wherein scattered signals are used to eliminate the absorption attenuation of an ultrasound wave from estimated parameters and to obtain directional scattering information. The scattered signals are obtained from either multiple regions where one region is used as a reference or multiple scattering directions, creating known reference signals from the same scattering region. The method provides new ultrasound parameters for the characterization and contrast enhancement of tissue structures in ultrasound imaging, such as tumor structures, ischemia of a myocardial wall and/or plaque compositions in vascular atheroma. The method of the invention can be used with various arrangements of ultrasound transducers, particularly switched linear arrays.

(21) **Appl. No.: 11/030,385**(22) **Filed: Jan. 6, 2005****Related U.S. Application Data**(60) **Provisional application No. 60/534,417, filed on Jan. 6, 2004.**

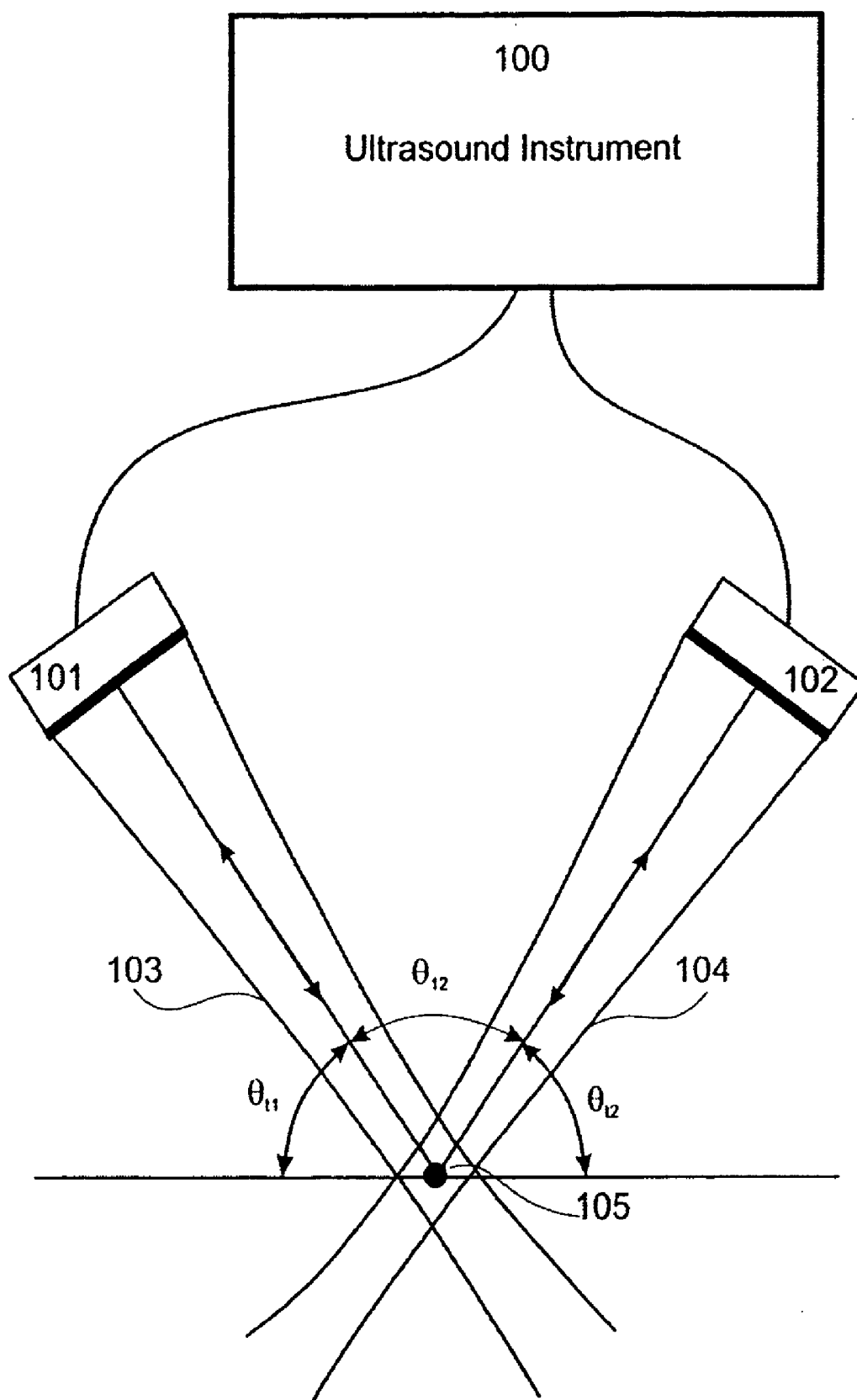


Figure 1

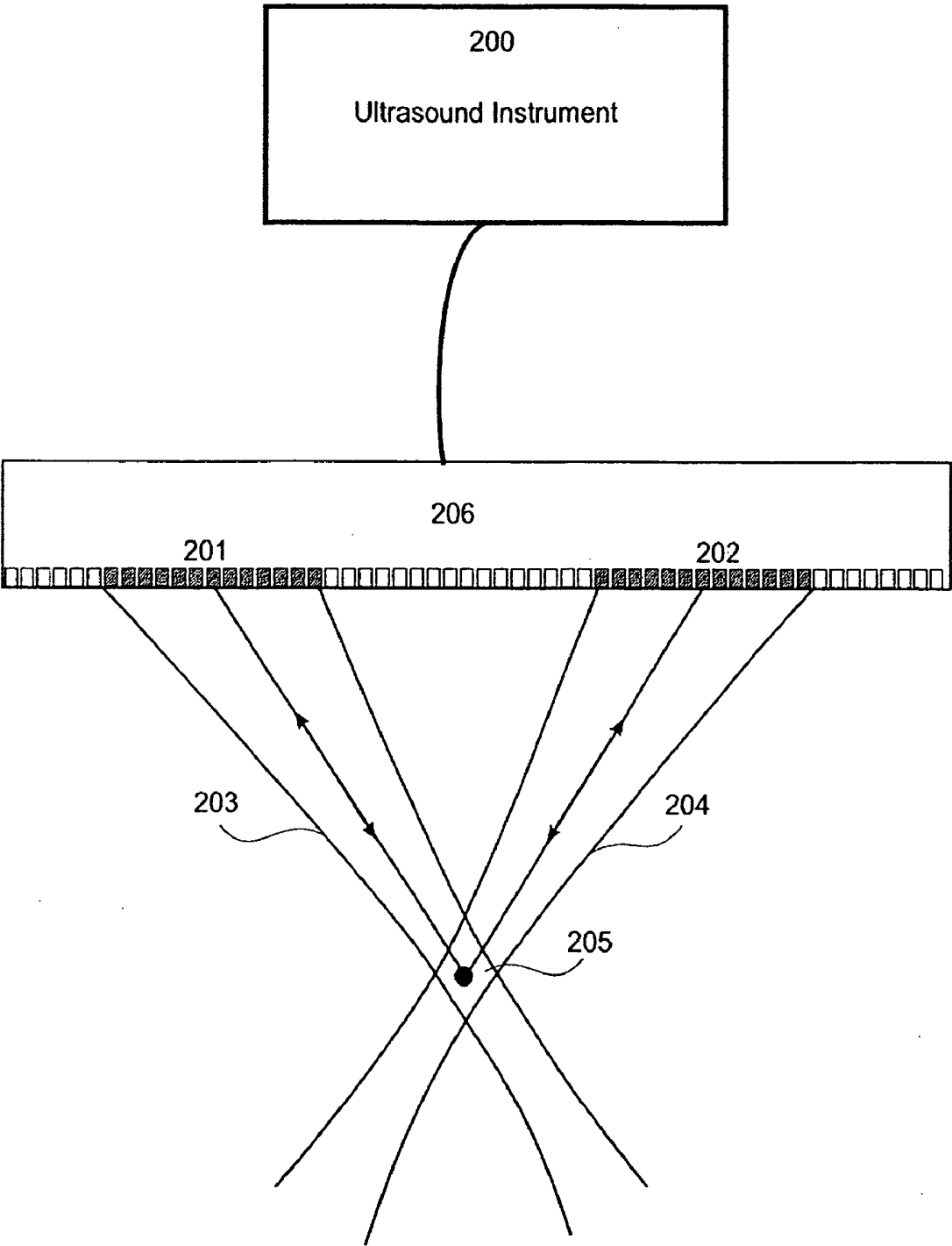


Figure 2

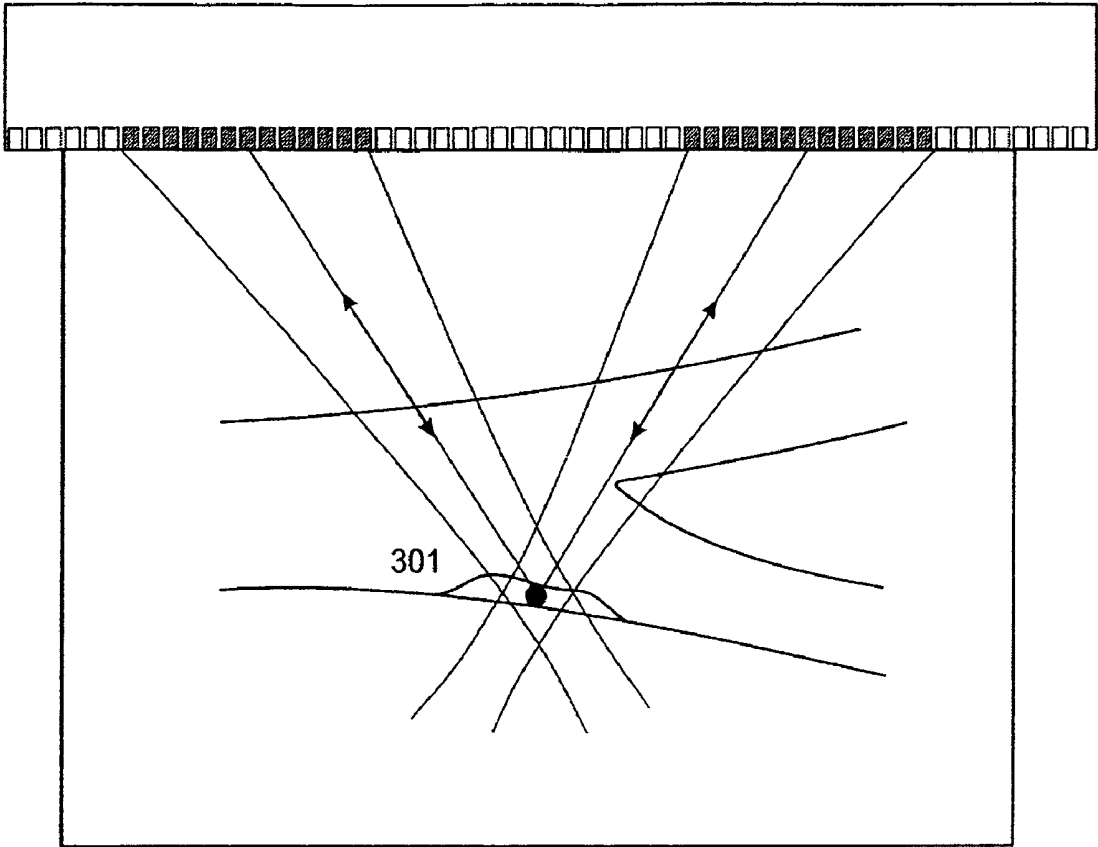


Figure 3

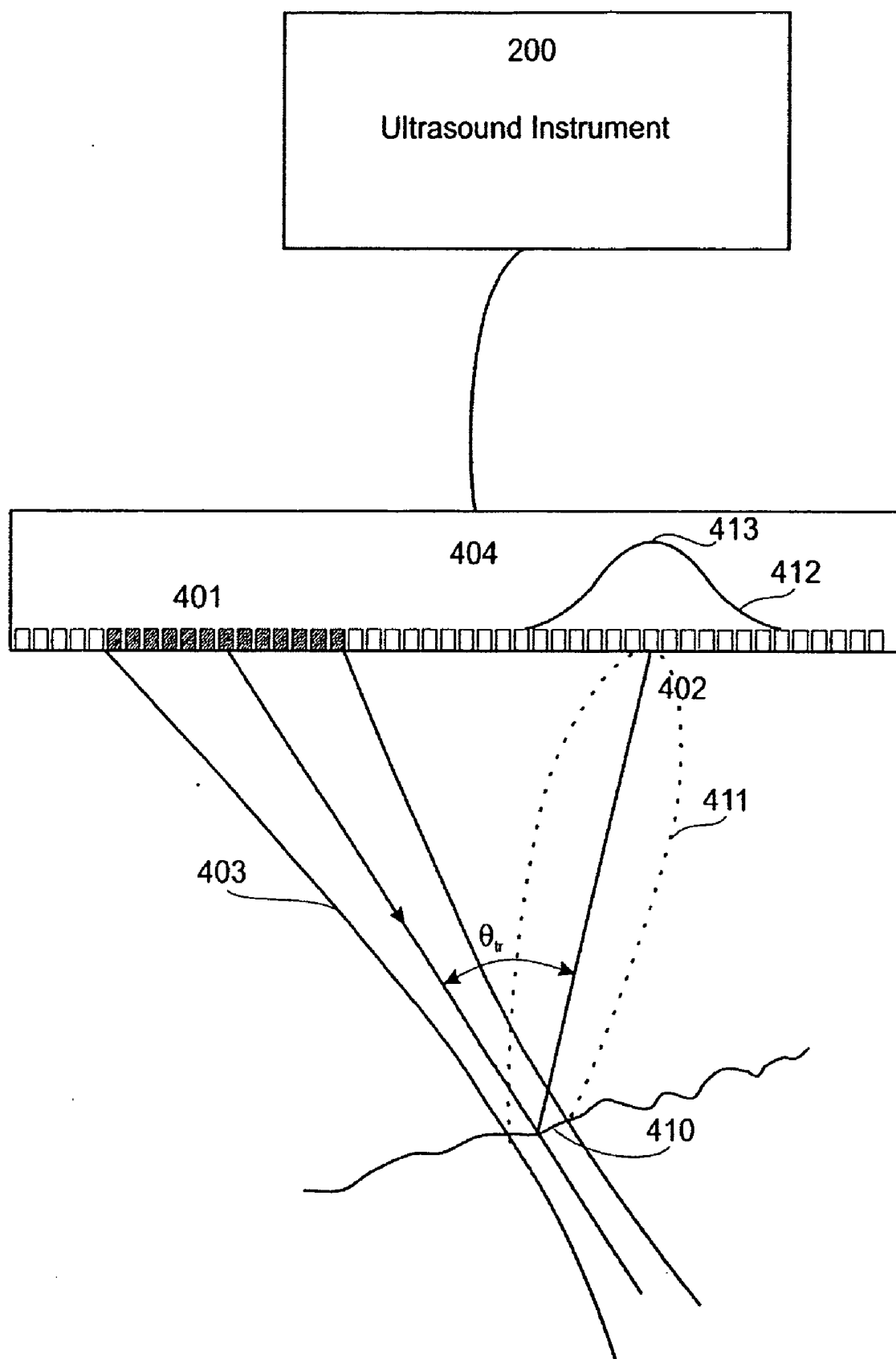


Figure 4

METHODS OF ESTIMATING ULTRASOUND SCATTERER PARAMETERS IN SOFT TISSUES

RELATED APPLICATIONS

[0001] This application claims priority from U.S. Provisional Patent Application Ser. No. 60/534,417 which was filed on Jan. 6, 2004.

BACKGROUND OF THE INVENTION

[0002] 1. Field of the Invention

[0003] The present invention, relates to characterization of ultrasound scatterers in soft tissue. The method eliminates the effect of unknown ultrasound absorption between the transducer and the scatterers, and produces local scattering parameters of the tissue and their frequency dependence.

[0004] 2. Description of the Related Art

[0005] Ultrasound back scatter imaging is a widely used imaging modality for diagnosis of many diseases of humans and animals. However, at present the method is used mainly to visualize anatomic structures, blood velocities, movements, and shear elasticity of tissue, and there is a further need for improved characterization of the scattering tissue structures with dimensions below the resolution in the anatomic image.

[0006] Transmitting on one transducer with the direction of the beam incident on the scatterers defined by the angle θ_i and receiving on the same or another transducer so that the angle between the transmit and the receive beams is θ_{tr} , one can write the frequency dependency of the received power of the scattered signal from the cross over region of the transmit and receive beams (105 of FIG. 1)

$$S_{tr} = K(\Delta\beta - \Delta\gamma \cos \theta_{tr})^2 V_s H(f, \theta_{tr}; \theta_i) f^4 e^{-2a(s)f} \quad (1)$$

[0007] where K is an amplitude parameter, f is the ultrasound frequency, and $e^{-2a(s)f}$ represents acoustic absorption where $a(s)$ is the integral of the absorption coefficient for the whole propagation distance s of the signal along transmit and the receive beams. $\Delta\beta$ is the relative deviation in the bulk compressibility and $\Delta\gamma$ is the relative deviation in the mass density between the scatterer and the surrounding medium, V_s is the scatterer volume, and $H(f, \theta_{tr}; \theta_i)$ is a function that represents that scatterer size and shape and the dependency of the scattered intensity on the direction θ_i of the incident beam, and the angle θ_{tr} between the incident and receiving beams, for example $\theta_{tr} = \theta_{12}$ as illustrated in FIG. 1. For scatterers that are much smaller than the wavelength $\lambda = c/f$, where $c \sim 1.54$ mm/ μ sec is the ultrasound propagation velocity in the tissue, we have $H(f, \theta_{tr}; \theta_i) = 1$. The scatterers are then referred to as point scatterers.

[0008] As the scatterer dimensions approaches or becomes larger than λ (for example as λ is reduced with higher frequencies), one gets interference from waves scattered at different regions of the scatterer. Depending on the angles θ_i and θ_{tr} , one can get destructive interference that reduces the scattered power in certain directions, or constructive interference that increases the scattered power in other directions. The invention provides methods on how one can use this angular variation to characterize the scatterers.

[0009] We hence notice that information about the acoustic parameters of the scatterers, $\Delta\beta$ and $\Delta\gamma$, and the scatter size and shape are found in the magnitude of the scattered

intensity, i.e. $(\Delta\beta - \Delta\gamma \cos \theta_{tr})^2 V_s H(f, \theta_{tr}; \theta_i)$, the frequency variation of the intensity, $H(f, \theta_{tr}; \theta_i) f^4 e^{-2a(s)f}$, and the variation of the scattered intensity with the directions of the incident and receive beams $(\Delta\beta - \Delta\gamma \cos \theta_{tr})^2 H(f, \theta_{tr}; \theta_i)$.

[0010] With direct back-scattering, i.e. the transmit and receive transducers are the same so that $\theta_{tr} = 0$, we can often approximate the absorption integral as

$$a(s) = 2az \quad (2)$$

[0011] where z is the distance between the transducer and the scatterer so that $s = 2z$, and a is an absorption parameter with typical values of 0.035-0.058 Neper/cmMHz. These values correspond to an absorption attenuation of 0.3-0.5 dB/cmMHz.

[0012] From Eq.(1) we see that absorption plays an important role in the frequency dependency of the scattered intensity, except when f is so low that the frequency variation of $e^{-2a(s)f}$ is negligible compared to the other terms. To get some more insight into this, we do as an initial exercise assume that we have point scatterers ($H(f, \theta_{tr}; \theta_i) = 1$), which gives the back scattered (i.e. $\theta_{tr} = 0$)

$$S_{back} = K(\Delta\beta - \Delta\gamma)^2 V_s f^4 e^{-2a(s)f} \quad (3)$$

[0013] Differentiation with respect to f gives a maximum of this function, and also a maximum in the frequency variation of $f^4 e^{-4azf}$, for

$$f_0 = \frac{1}{4az} \quad (4)$$

[0014] The absorption term will then have negligible effect on the frequency variation of the scattered intensity when $f \ll f_0$, say $f < 0.2 f_0$. From Eq.(4) we see that f_0 is approximately inversely proportional to the depth, where we shall analyze three examples.

[0015] For intravascular imaging (IVUS) of coronary plaque we have typically $z \sim 0.2$ cm, for noninvasive imaging of carotid plaque we have typically $z \sim 2.5$ cm, and for noninvasive imaging of liver disorders we have typically $z \sim 7$ cm. With the values of a given above, we get the following frequencies for peak of Eq.(3):

$$\begin{aligned} \text{IVUS Coronary Plaque: } f_0 &= 85\text{-}145 \text{ MHz} \\ \text{Noninvasive Carotid Plaque: } f_0 &= 7\text{-}12 \text{ MHz} \\ \text{Noninvasive Liver scatterers: } f_0 &= 2.5\text{-}4 \text{ MHz} \end{aligned} \quad (5)$$

[0016] For the carotid and liver imaging, $f_0 = 7\text{-}12$ MHz covers the actual frequencies used for imaging, so that in these situations the absorption has a dominating effect on the frequency variation of the scattering. Hence, in these situations one should find methods for scatterer characterization, where the effect of frequency variation of absorption on the scattered intensity is reduced.

[0017] For IVUS imaging at 20-30 MHz, the influence of absorption on the frequency variation of the scattered intensity can be neglected when the scatterer dimension approaches or gets larger than the wavelength, which is 50-80 μ m for these frequencies. Hence, the frequency variation of the scattered intensity can contain some information for scatterers with dimensions approaching ~ 50 μ m.

[0018] As the absorption is roughly proportional to the frequency, the image depth is inversely proportional to the frequency. For imaging of the carotid vessel and similar down to 40 mm depth, one generally uses 10 MHz ultrasound. Hence, IVUS imaging of coronary artery wall down to 4 mm is hence very attractive at ~100 MHz. According to Eq.(5) the ultrasound absorption will then play an important role in the frequency dependency of the back scattered intensity. The wave length at 100 MHz is ~15 μ m, and by reducing the effect of absorption on the frequency variation of the scattered intensity, one is able to extract information on scatterers down to ~2 μ m dimension.

SUMMARY OF THE INVENTION

[0019] The present invention provides methods for characterizing the scatterers in ultrasound imaging that strongly reduces the effect of absorption attenuation of the waves in the characterization, and makes it possible to eliminate the effect of frequency dependent absorption in the transmit path of the ultrasound pulse, and obtain frequency dependent scattering parameters from local regions in the tissue. It is furthermore possible to obtain information of scattering anisotropy in such regions, that can provide information about fiber direction in fibrous and muscular tissue, as well as degree of fibrosity. Moreover, by comparing ultrasound angular scattering with back scattering, one is able to derive anisotropic properties of the scattering cross section from a local region, that can give information of fibrous structures in the tissue.

[0020] The method provides new ultrasound parameters for characterization and contrast enhancement of tissue structures in ultrasound imaging, like tumor structures, ischemia of a myocardial wall, and plaque composition in vascular atheroma. It can be used with many arrangements of ultrasound transducers, particularly switched linear arrays.

[0021] The essence of the invention, is to use scattered signals either from multiple regions where one region is used as a reference, or multiple scattering directions, creating own reference signals from the same scattering region, so that the absorption attenuation of the ultrasound wave is eliminated from the estimated parameters, and directional scattering information can be achieved.

[0022] Other objects and features of the present invention will become apparent from the following detailed description considered in conjunction with the accompanying drawings. It is to be understood, however, that the drawings are designed solely for purposes of illustration and not as a definition of the limits of the invention, for which reference should be made to the appended claims. It should be further understood that the drawings are not necessarily drawn to scale and that, unless otherwise indicated, they are merely intended to conceptually illustrate the structures and procedures described herein.

BRIEF DESCRIPTION OF THE DRAWINGS

[0023] In the drawings:

[0024] FIG. 1 shows an example embodiment according to the invention of two transducers arranged so that the scattering parameters of the tissue in the overlap region of the transducers beams can be investigated;

[0025] FIG. 2 shows example embodiments according to the invention where a switched linear array is used to realize two transducer apertures, which are arranged so that the scattering parameters' of the tissue in the overlap region of the apertures' beams can be investigated;

[0026] FIG. 3 illustrates how the scattering from tissue consisting of scatterers comparable or larger than the wavelength can be investigated using a linear array; and

[0027] FIG. 4 shows an application of the method described, imaging of anisotropic structures of a carotid plaque combined with two dimensional backscatter imaging.

DETAILED DESCRIPTION OF THE PRESENTLY PREFERRED EMBODIMENTS

[0028] In the following we describe example embodiments of the invention with reference to the drawings. FIG. 1 shows two ultrasound transducers 101 and 102 that are connected to an ultrasound instrument 100 that allows selectable transmission of ultrasound pulses on each transducer, with selectable reception of the scattered signal on each transducer, independent of the transducer selected for transmission. The transducers can be directed at an angle towards each other, where θ_t represents the direction of the transmit beam and θ_r is the angle between the transmit beam and the receive beam. In the Figure we have $\theta_t=\theta_{t1}$ when transmitting on Transducer 1 (101) and $\theta_t=\theta_{t2}$ when transmitting on Transducer 2 (102). We denote the beam 103 from Transducer 1 as Beam 1, and the beam 104 from Transducer 2 as Beam 2. The beams intercept in a region labeled 105 in FIG. 1. The angle between Beam 1 and Beam 2 is indicated as θ_{12} in FIG. 1. When different beams are used to transmit and receive one gets $\theta_r=\theta_{12}$, and when the same beam is used to transmit and receive one gets $\theta_r=0$.

[0029] A first method for reduction of the effect of frequency absorption according to the invention, is to use the scattered signal from reference scatterers with known frequency variation of the scattering cross section, and located in neighboring region with close to the same absorption as scatterers to be characterised. Such reference scatterers can be the erythrocytes in blood, where for characterization of arterial wall plaque one would use the signal scattered from the blood close to the plaque as reference. For other situations (like the liver) one would find the blood vessels close to the area of interest as reference, or use other reference scatterers in the neighborhood of the region of interest. In such situations one can do back scatter imaging and use the same beam directions for the transmit and receive beams. This will create a reference signal for back scatter imaging obtained from Eq.(3) as

$$S_{ref} \sim \sigma_{ref} V_{ref} H_{ref}(f, 0; \theta_t) f^4 e^{-2a(2z)f} \quad (6)$$

[0030] where $\sigma_{ref}=(\Delta\beta-\Delta\gamma)^2$ for the parameters of the reference scatterers with back scatter imaging, and H_{ref} represents the frequency variation of the back scatterers for the reference scatterers. For point scatterers, like erythrocytes up to f~10 MHz, $H_{ref}=1$.

[0031] The backscattered signal from blood is often masked in close to stationary reverberation noise. As the blood is moving, the scatterers also move and the backscattered signal from blood can be retrieved from the stationary noise by collecting back scattered signal from several pulses in substantially the same beam direction and perform high

pass filtering of the signal along the pulse number coordinate. A small variation of the beam direction between the pulses can be accepted, as for example with continuous, mechanical scanning of the beam direction.

[0032] Ultrasound contrast agent micro bubbles can also conveniently be used as reference scatterers, both in visible blood vessels, and within the capillary vessels. The reference signal from the contrast agent bubbles can be separated from the signal from surrounding tissue by several known methods, such as harmonic imaging, pulse inversion imaging, and manipulation of the scattering properties by a low frequency pulse. Other reference scatterers can be identifiable cells, like normal liver cells, or fat cells in atheroma. The back scattered intensity from the reference scatterers can be approximated as

$$S_{\text{scat}} \sim \sigma_{\text{scat}} V_{\text{scat}} H_{\text{scat}}(f, 0; \theta_i) f^4 e^{-2\alpha(2z)f} \quad (7)$$

[0033] $\sigma_{\text{scat}} = (\Delta\beta - \Delta\gamma)^2$ for the scatterers in the actual region of the plaque, and V_{scat} is the volume of these scatterers. $H(f, 0; \theta_i)$ then includes the frequency variation of the scattered signal due to the scatterer size and shape. The following ratio will then be independent of the absorption attenuation of the ultrasound wave along the beams

$$\frac{S_{\text{scat}}}{S_{\text{ref}}} \sim \frac{\sigma_{\text{scat}} V_{\text{scat}}}{\sigma_{\text{ref}} V_{\text{ref}}} \frac{H_{\text{scat}}(f, 0; \theta_i)}{H_{\text{ref}}(f, 0; \theta_i)} \quad (8)$$

[0034] A second method according to the invention, is to use angular scattering with two beams as illustrated in FIG. 1-4. First transmitting on Transducer 1 (101) and receiving the back scattered signal at a depth range along the beam corresponding to the cross over region 105 between the two beams, one receives a back scattered signal power on Transducer 1 and Transducer 2 (102) as

$$\begin{aligned} S_{11} &= A_1^2 \sigma_{11} H(f, 0; \theta_{11}) \text{ Transmit Transd 1 and Receive Transd 1} \\ S_{12} &= A_1 A_2 \sigma_{12} H(f, \theta_{12}; \theta_{11}) \text{ Transmit Transd 1 and Receive Transd 2} \\ \sigma_{11} &= (\Delta\beta - \Delta\gamma)^2 \quad \sigma_{12} = (\Delta\beta - \Delta\gamma \cos \theta_{12})^2 \end{aligned} \quad (9)$$

[0035] where A_1 contains the one-way power attenuation along Beam 1 (103) from Transducer 1 (101) to the scattering region 105, A_2 contains the one-way power attenuation along the Beam 2 (104) from Transducer 2 (102) to the scattering region 105. Transmitting at Transducer 2 (102) one gets scattered signal power from the overlap region (105) as

$$\begin{aligned} S_{22} &= A_2^2 \sigma_{22} H(f, 0; \theta_{12}) \text{ Transmit Transd 2 and Receive Transd 2} \\ S_{21} &= A_2 A_1 \sigma_{21} H(f, \theta_{12}; \theta_{21}) \text{ Transmit Transd 2 and Receive Transd 1} \\ \sigma_{22} &= (\Delta\beta - \Delta\gamma)^2 \quad \sigma_{21} = \sigma_{12} = (\Delta\beta - \Delta\gamma \cos \theta_{12})^2 \end{aligned} \quad (10)$$

[0036] Direct calculation shows that the ratio

$$\sigma_a = \frac{S_{12} S_{21}}{S_{11} S_{22}} = \frac{(\Delta\beta - \Delta\gamma \cos \theta_{12})^4}{(\Delta\beta - \Delta\gamma)^4} \frac{H(f, \theta_{12}; \theta_{11}) H(f, \theta_{12}; \theta_{21})}{H(f, 0; \theta_{11}) H(f, 0; \theta_{21})} \quad (11)$$

[0037] is independent of the beam form and cumulative power absorption along Beam 1 (103) and Beam 2 (104). The frequency variation of the ratio σ_a contains informa-

tion on the scatterer size, and σ_a will also contain information on the degree of anisotropy of the scatterers in the overlap region 103.

[0038] For scatterers that are much smaller than the wavelength, we have $H=1$. For $\Theta_{12}=\pi/2$ we get

$$\sigma_a = \frac{\Delta\beta^4}{(\Delta\beta - \Delta\gamma)^4} = \frac{1}{(1 - \Delta\gamma/\Delta\beta)^4} \quad (12)$$

[0039] Typical values are $|\Delta\beta| \sim 0.3$ and $|\Delta\gamma| \sim 0.1$, where $\Delta\beta$ and $\Delta\gamma$ have opposite signs. This gives gives $\sigma_a \sim 0.3$. In the above situation, we can calculate

$$\Delta\gamma/\Delta\beta = 1 - \sigma_a^{-1/4} \quad (13)$$

[0040] As the scatterer dimensions become comparable to or larger than the wave length, the shape of the scatterers influences the scattering cross section so that the H 's in Eq.(11) are different from unity. This influences the frequency variation of σ_a , and σ_a becomes dependent on the direction of the incident and the receive beams. For unidirectional fibrous scatterers, like collagen or muscle fibers, one can get large σ_a when the angle of Beam 1 to the fiber direction is similar to the angle of Beam 2 to the fiber direction. The value of σ_a then also increases above 0.3.

[0041] When Beam 1 (103) and Beam 2 (104) have the same shape and crosses through tissue with similar absorption and scattering, we get $A_2 \sim A_1$ and we can calculate S_{12}/S_{11} or S_{12}/S_{22} as a measure of the tissue scattering anisotropy.

[0042] The basic method is conveniently implemented with a linear array as illustrated in FIG. 2, where 206 shows the linear array that is connected to the ultrasound instrument 200. The instrument is designed for free selection of the transmit apertures, where the Figure by way of example illustrates two transmit/receive apertures, Aperture 1 as 201 and Aperture 2 as 202, where one by time delay of the element signals steers the direction of the Beam 1 (203) from Aperture 1 and the direction of Beam 2 (204) from Aperture 2, so that we get an overlap region 205 between the beams. By transmitting and receiving from the two apertures in the same way as for the two transducers in FIG. 1, one can calculate the anisotropy scattering coefficient σ_a as defined in Eq.(3).

[0043] The linear array allows common lateral scanning of Beam 1 and Beam 2 that enables spatial imaging of σ_a , as illustrated in FIG. 3 for noninvasive imaging of a carotid plaque 301. Maintaining the distance between Aperture 1 and Aperture 2 and scanning the beams laterally with a fixed direction angle of the beams, one obtains a spatial image of σ_a at a fixed beam overlap depth. This overlap depth can be varied by varying the distance between Aperture 1 and Aperture 2, with constant direction angles of the beams. The angles between the beams can also be varied for more details of the anisotropic scattering. The two-dimensional back scatter image will indicate the location of the plaque, and one can hence use the back scatter image to limit the spatial region that is actual for interrogation or observation of anisotropic scattering structures by the methods presented above, hence reducing the time of interrogation and the possibility of using multiple directions of the transmit beam with adequate frame rate of the anisotropic scattering imaging.

[0044] Varying the direction angles of the beams, will in principle also vary all the scattering coefficients. FIG. 4 illustrates a method for at least semi-automatic observation of anisotropic scattering structures of unknown direction in relation to the transducer array. The Figure shows a linear array 404, and an anisotropic scattering structure 410. A pulsed beam 403 is emitted from the aperture 401 of the array, and the scattered wave is picked up by an array element or group of array elements 402, for several positions along the array. This gives an angle θ_r between the transmit and the receive beam directions, that for the given transmit beam varies with the position of the receive elements along the array.

[0045] The received signal power as a function of receive element position is monitored at a time interval after the pulse transmission that selects a particular depth along the transmitted beam. An anisotropic scattering structure like 410 will back scatter the main energy in a particular direction indicated by the scattering diagram 411. This scattering diagram produces a variation of the power in the received signal along the element position, indicated as 412 in FIG. 4. Hence, a peaked variation of the scattering power along the element position like at 413 indicates an anisotropic scattering structure at the particular depth, and also the anisotropy direction of the scattering structures. By calculating the power of the received element signals for several time intervals, one can interrogate several depths along the transmitting beam for imaging possible anisotropic scatterers along the transmitted beam. Lateral scanning of the transmit beam then provides a two-dimensional image of anisotropic scattering structures in the tissue. Using several direction angles of the transmitted beam provides possibility of imaging a larger spread of the anisotropy directions.

[0046] Thus, while there have shown and described and pointed out fundamental novel features of the invention as applied to a preferred embodiment thereof, it will be understood that various omissions and substitutions and changes in the form and details of the devices illustrated, and in their operation, may be made by those skilled in the art without departing from the spirit of the invention. For example, it is expressly intended that all combinations of those elements and/or method steps which perform substantially the same function in substantially the same way to achieve the same results, are within the scope of the invention. Moreover, it should be recognized that structures and/or elements and/or method steps shown and/or described in connection with any disclosed form or embodiment of the invention, may be incorporated in any other disclosed or described or suggested form or embodiment as a general matter of design choice. It is the intention, therefore, to be limited only as indicated by the scope of the claims appended hereto.

We claim:

1. A method for measuring at least one of the magnitude and the frequency dependency of ultrasound scatterer parameters in an observation region of soft tissue, which is insensitive to ultrasound absorption, where

one or more pulsed ultrasound beams are transmitted towards the tissue under investigation and the scattered signal from at least one observation region and at least one reference region is received, where

said observation regions consist of scatterers to be characterized, and said reference regions consist of scatter-

ers with adequately known scattering characteristics so that the signal from said reference regions can be used as reference,

said scatterer parameters in said observation region are obtained as a combination of the received signal from said observation regions and said reference regions.

2. A method for measuring at least one of the magnitude and the frequency dependency of ultrasound scatterer parameters in an observation region of soft tissue according to claim 1, where the scatterers in said reference regions are erythrocytes in blood.

3. A method for measuring at least one of the magnitude and the frequency dependency of ultrasound scatterer parameters in an observation region of soft tissue according to claim 2, where the signal from the scatterers in said reference regions is retrieved from stationary noise by transmitting multiple pulses along essentially the same beam direction and high pass filtering the received signal from said reference regions along the pulse number coordinate, to attenuate the close to stationary noise before the signal from said reference regions is used for said combination with the signal from said observation regions.

4. A method for measuring at least one of the magnitude and the frequency dependency of ultrasound scatterer parameters in an observation region of soft tissue according to claim 1, where the scatterers in said reference regions are parenchymatic cells like in the liver or a gland.

5. A method for measuring at least one of the magnitude and the frequency dependency of ultrasound scatterer parameters in an observation region of soft tissue according to claim 1, where the scatterers in said reference regions are fat cells in atheroma.

6. A method for measuring at least one of the magnitude and the frequency dependency of ultrasound scatterer parameters in an observation region of soft tissue according to claim 1, where the scatterers in said reference regions are contrast agent micro bubbles injected in the blood or other body fluids.

7. A method for measuring at least one of the magnitude and the frequency dependency of ultrasound scatterer parameters in an observation region of soft tissue according to claim 6, where the signal from said contrast agent bubbles is separated from surrounding tissue signal through one of harmonic imaging, and pulse inversion imaging, and manipulation of the scattering properties by a low frequency pulse.

8. A method for measuring at least one of the magnitude, the frequency dependency, and the spatial anisotropy of scatterer parameters in an observation region of soft tissue, where

a 1st and a 2nd ultrasound transducer area is used to generate transmit and receive beams along a 1st and a 2nd beam directions, and where said 1st and a 2nd beam directions intersects to define said observation region, and where

a 1st ultrasound pulse is transmitted from said 1st transducer area along said 1st beam direction, and a 1st receive signal power scattered from said 1st pulse from said observation region along said 1st beam direction is measured at said 1st transducer area, and a 2nd receive signal power scattered from said 1st pulse from said observation region along said 2nd beam direction is measured at said 2nd transducer area, and

a 2nd ultrasound pulse is transmitted from said 2nd transducer area along said 2nd beam direction, and a 3rd receive signal power scattered from said 2nd pulse from said observation region along said 2nd beam direction is measured at said 2nd transducer area, and a 4th receive signal power scattered from said 2nd pulse from said observation region along said 1st beam direction is measured at said 1st transducer area, and where

scatterer parameters in said observation region that is not influenced by the absorption in the tissue, is obtained through a combination of said 1st, 2nd, 3rd, and 4th received signal powers.

9. A method for measuring at least one of the magnitude, the frequency dependency, and the spatial anisotropy of scatterer parameters in an observation region of soft tissue according to claim 8, where said transducer areas are composed of array elements, so that selection of said observation region electronically controlled in an ultrasound instrument through delay direction steering of the said 1st and 2nd beam directions.

10. A method for measuring at least one of the magnitude, the frequency dependency, and the spatial anisotropy of scatterer parameters in an observation region of soft tissue according to claim 8, where said 2 ultrasound transducer areas are 2 separate array transducers.

11. A method for measuring at least one of the magnitude, the frequency dependency, and the spatial anisotropy of scatterer parameters in an observation region of soft tissue according to claim 8, where said 2 ultrasound transducer areas are formed as sub groups of the elements of a linear array and intersection of said beams is electronically controlled in an ultrasound instrument.

12. A method for measuring at least one of the magnitude, the frequency dependency, and the spatial anisotropy of scatterer parameters in an observation region of soft tissue according to claim 8, where said transducer areas are moved along the array surface in order to estimate angular variation of scattering parameters.

13. A method for imaging of spatial variation of at least one of the magnitude, the frequency dependency, and the spatial anisotropy of scatterer parameters in an image region of soft tissue, where

the scattering parameters in selected observation regions are measured according to claim 1 and

the image of the spatial variation of said scattering parameters are obtained by moving said observation region around in said image region.

14. A method for imaging of spatial variation of at least one of the magnitude, the frequency dependency, and the spatial anisotropy of scatterer parameters in an image region of soft tissue, where

the scattering parameters in selected observation regions are measured according to claim 8 and

the image of the spatial variation of said scattering parameters are obtained by moving said observation region around in said image region.

15. A method for characterizing different types of tissue based on measuring at least one of the magnitude, the frequency dependency, and the spatial anisotropy of scatterer parameters obtained according to claim 1.

16. A method for characterizing different types of tissue based on measuring at least one of the magnitude, the frequency dependency, and the spatial anisotropy of scatterer parameters obtained according to claim 8.

17. A method for measuring at least one of the magnitude, the frequency dependency, and the spatial anisotropy of scatterer parameters in an observation region of soft tissue according to claim 11, where said transducer areas are moved along the array surface in order to estimate angular variation of scattering parameters.

* * * * *

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摘要(译)

一种用于表征软组织中的超声散射体的方法，其中散射信号用于从估计的参数中消除超声波的吸收衰减并获得定向散射信息。散射信号从多个区域获得，其中一个区域用作参考或多个散射方向，从相同的散射区域产生已知的参考信号。该方法为超声成像中的组织结构的表征和对比度增强提供了新的超声参数，例如肿瘤结构，心肌壁缺血和/或血管粥样斑块中的斑块组合物。本发明的方法可以与超声换能器的各种布置一起使用，特别是切换线性阵列。

