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(54) **POINT SOURCE TRANSMISSION AND SPEED-OF-SOUND CORRECTION USING MULTI-APERATURE ULTRASOUND IMAGING**

(75) Inventors: **Donald F. Specht**, Los Altos, CA (US); **Kenneth D. Brewer**, Santa Clara, CA (US)

(73) Assignee: **MAUI IMAGING, INC.**, Sunnyvale, CA (US)

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Primary Examiner — Sanjay Cattungal
Assistant Examiner — Lawrence Laryea
(74) *Attorney, Agent, or Firm* — Shay Glenn LLP

(57) **ABSTRACT**

A Multiple Aperture Ultrasound Imaging system and methods of use are provided with any number of features. In some embodiments, a multi-aperture ultrasound imaging system is configured to transmit and receive ultrasound energy to and from separate physical ultrasound apertures. In some embodiments, a transmit aperture of a multi-aperture ultrasound imaging system is configured to transmit an omnidirectional unfocused ultrasound waveform approximating a first point source through a target region. In some embodiments, the ultrasound energy is received with a single receiving aperture. In other embodiments, the ultrasound energy is received with multiple receiving apertures. Algorithms are described that can combine echoes received by one or more receiving apertures to form high resolution ultrasound images. Additional algorithms can solve for variations in tissue speed of sound, thus allowing the ultrasound system to be used virtually anywhere in or on the body.

41 Claims, 7 Drawing Sheets

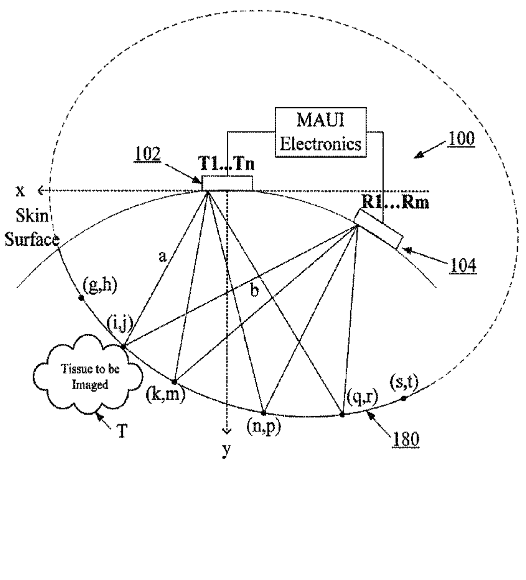

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FIG. 1A

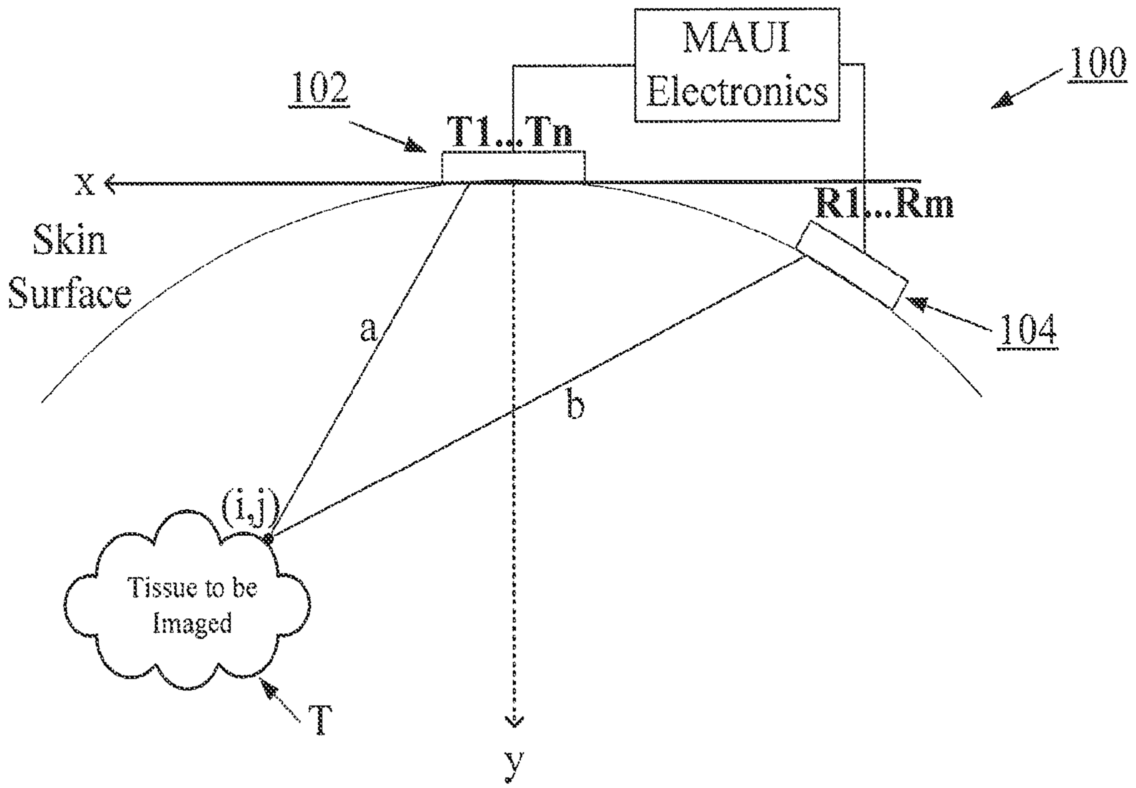

MAUI
Electronics
100
102
T1...Tn
x
R1...Rm
Skin
Surface
a
104
b
(i,j)
Tissue to be
Imaged
T
y

FIG. 1B

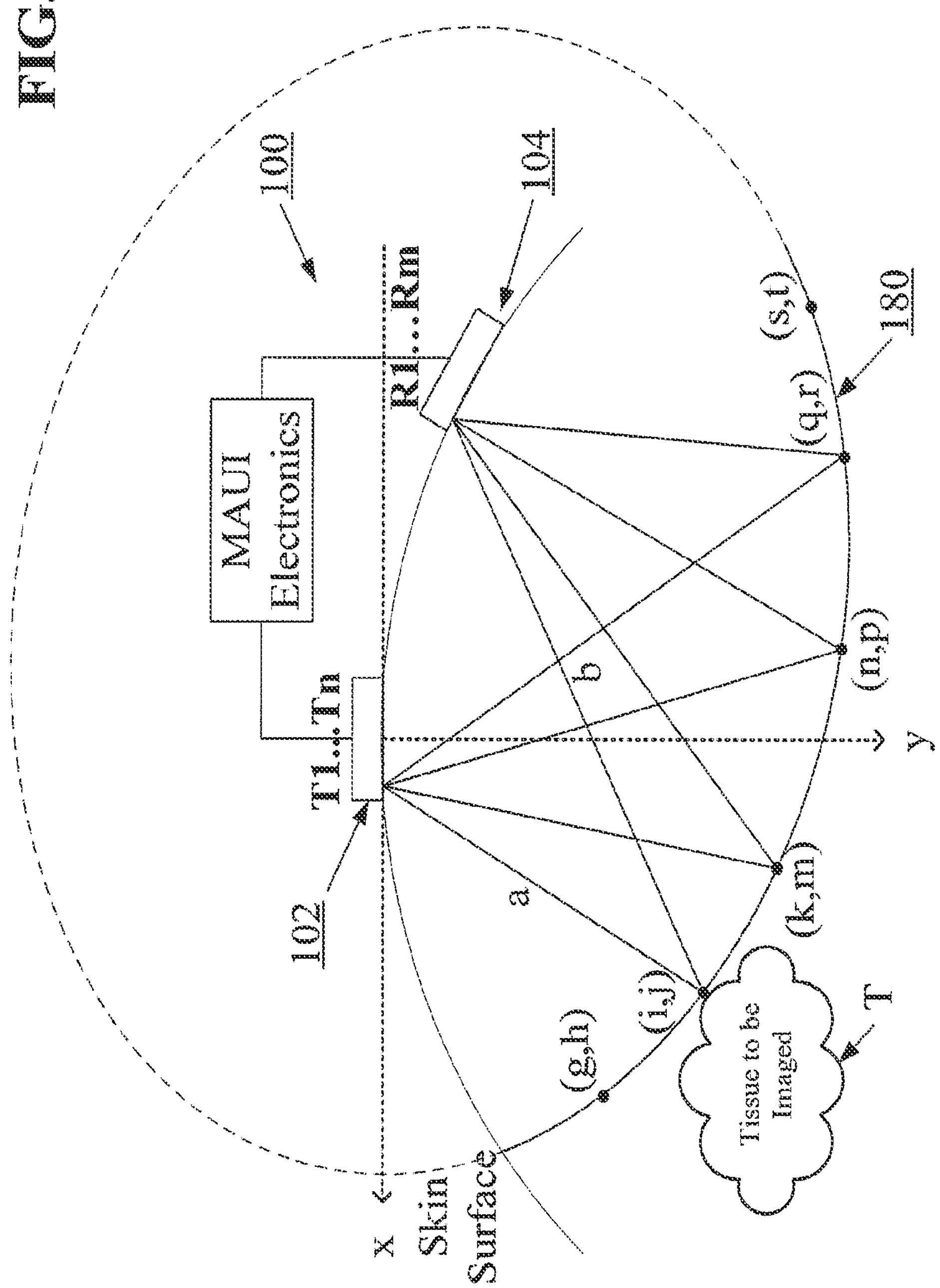

100
MAUI
Electronics
R1...Rm
104
T1...Tn
102
a
b
y
x
Skin
Surface
(g,h)
(i,j)
(k,m)
(n,p)
(q,r)
(s,t)
180
Tissue to be
Imaged
T

FIG. 1C

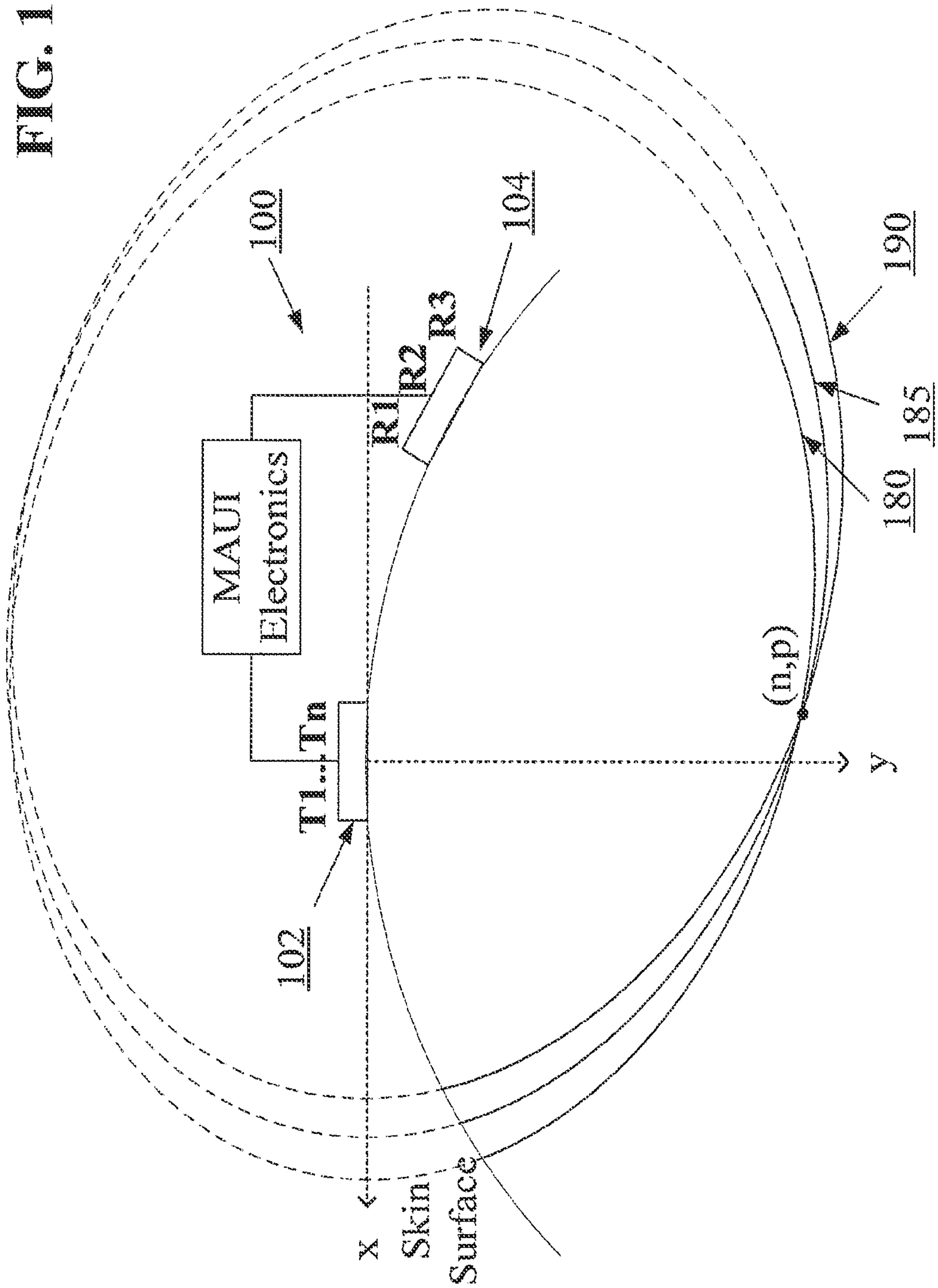

100
104
MAUI
Electronics
R1
R2
R3
T1 ... Tn
102
(n,p)
y
x
Skin
Surface
190
185
180

FIG. 2

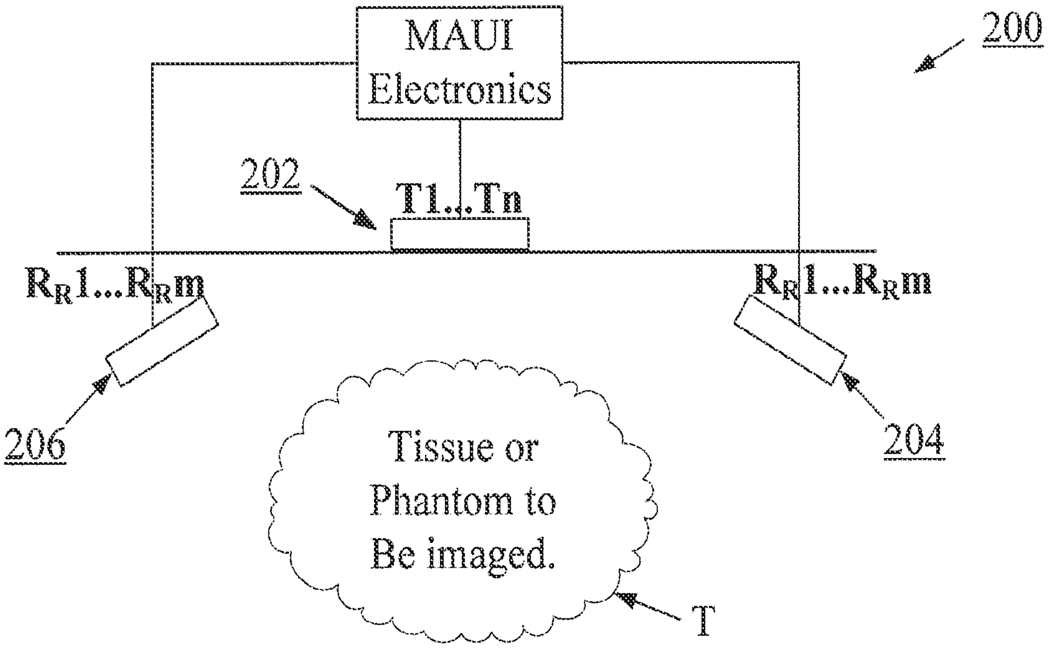

200
MAUI Electronics
202
T1...Tn
R_R1...R_Rm
R_R1...R_Rm
206
204
Tissue or Phantom to Be imaged.
T

FIG. 3

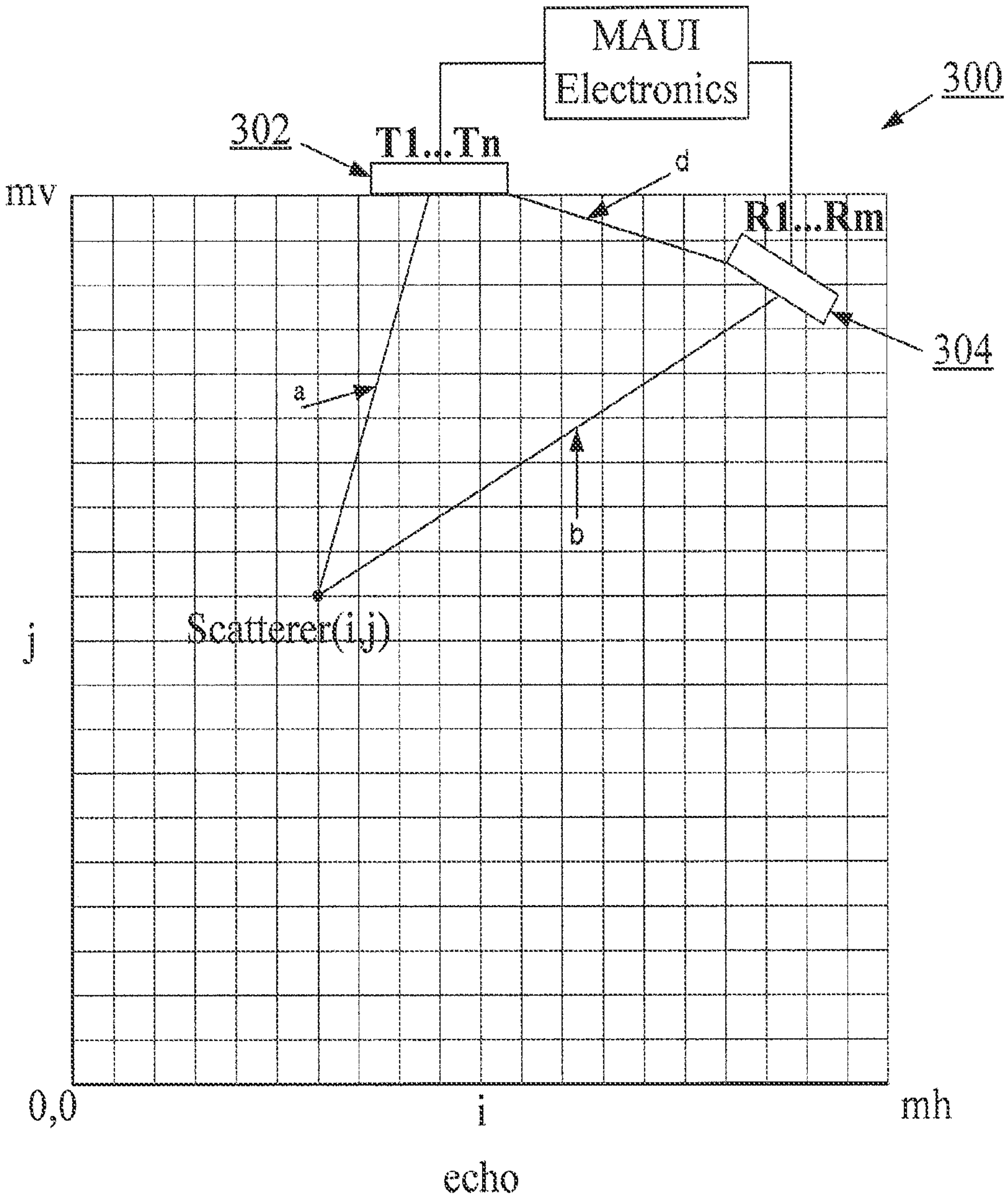

MAUI
Electronics
300
302
T1...Tn
d
R1...Rm
304
mv
a
b
Scatterer(i,j)
j
0,0
i
mh
echo

FIG. 4

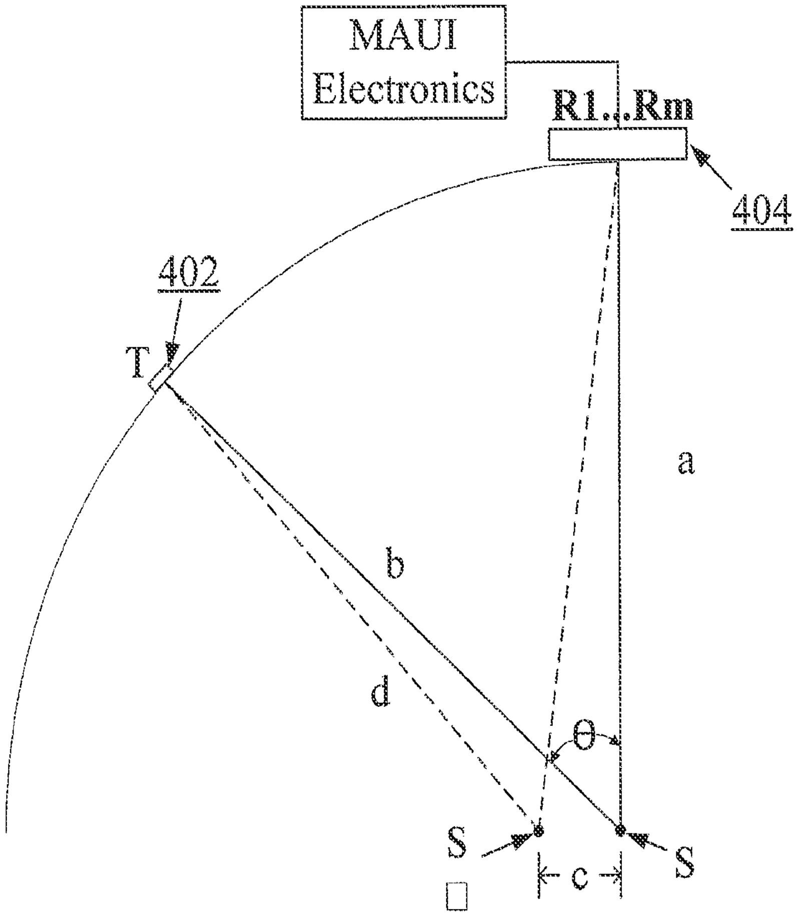
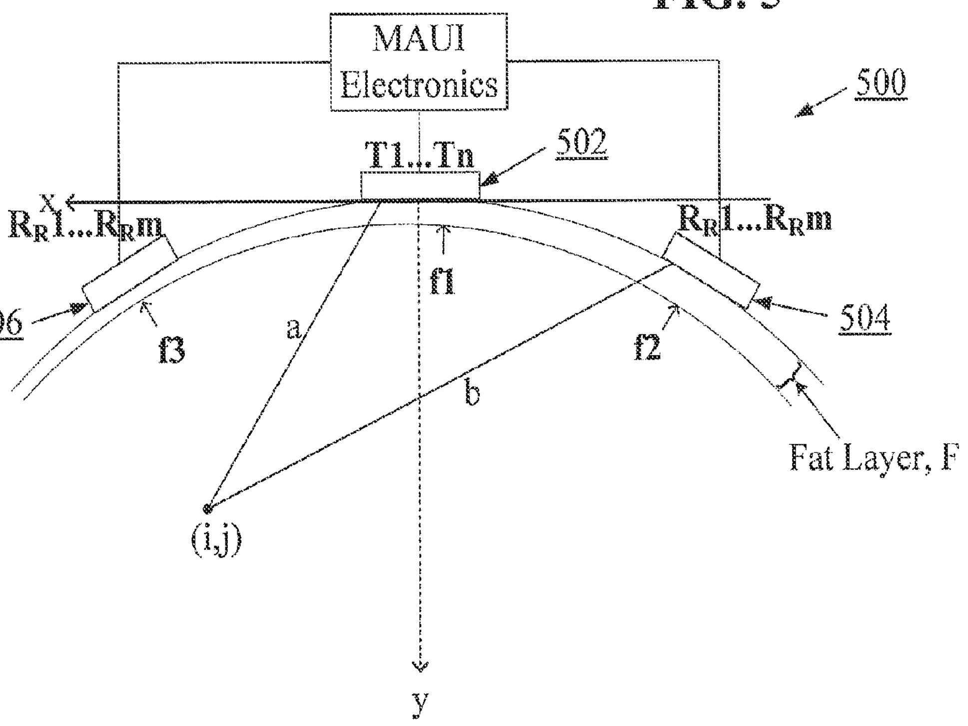


FIG. 5


MAUI
Electronics
500
T1...Tn
502
X
R_R1...R_Rm
R_R1...R_Rm
f1
506
f3
a
f2
504
b
Fat Layer, F
(i,j)
y

POINT SOURCE TRANSMISSION AND SPEED-OF-SOUND CORRECTION USING MULTI-APERATURE ULTRASOUND IMAGING

CROSS REFERENCE TO RELATED APPLICATIONS

This application claims the benefit of U.S. Provisional Patent Application No. 61/305,784, filed on Feb. 18, 2010, titled "Alternative Method for Medical Multi-Aperture Ultrasound Imaging".

This application is also related to U.S. patent application Ser. No. 11/865,501, filed Oct. 1, 2007, titled "Method and Apparatus to Produce Ultrasonic Images Using Multiple Apertures", and to U.S. patent application Ser. No. 11/532,013, filed Sep. 14, 2006, titled "Method and Apparatus to Visualize the Coronary Arteries Using Ultrasound"; all of which are herein incorporated by reference in their entirety.

INCORPORATION BY REFERENCE

All publications, including patents and patent applications, mentioned in this specification are herein incorporated by reference in their entirety to the same extent as if each individual publication was specifically and individually indicated to be incorporated by reference.

BACKGROUND OF THE INVENTION

In conventional ultrasonic imaging, a focused beam of ultrasound energy is transmitted into body tissues to be examined and the returned echoes are detected and plotted to form an image. The basic principles of conventional ultrasonic imaging are well described in the first chapter of "Echocardiography," by Harvey Feigenbaum (Lippincott Williams & Wilkins, 5th ed., Philadelphia, 1993).

In order to insonify body tissues, an ultrasound beam is typically formed and focused either by a phased array or a shaped transducer. Phased array ultrasound is a commonly used method of steering and focusing a narrow ultrasound beam for forming images in medical ultrasonography. A phased array probe has many small ultrasonic transducer elements, each of which can be pulsed individually. By varying the timing of ultrasound pulses (e.g. by pulsing elements one by one in sequence along a row), a pattern of constructive interference is set up that results in a beam directed at a chosen angle. This is known as beam steering. Such a steered ultrasound beam may then be swept through the tissue or object being examined. Data from multiple beams are then combined to make a visual image showing a slice through the object.

Traditionally, the same transducer or array used for transmitting an ultrasound beam is used to detect the returning echoes. This design configuration lies at the heart of one of the most significant limitations in the use of ultrasonic imaging for medical purposes: poor lateral resolution. Theoretically, the lateral resolution could be improved by increasing the width of the aperture of an ultrasonic probe, but practical problems involved with aperture size increase have kept apertures small. Unquestionably, ultrasonic imaging has been very useful even with this limitation, but it could be more effective with better resolution.

In the practice of cardiology, for example, the limitation on single aperture size is dictated by the space between the ribs (the intercostal spaces). Such intercostal apertures are typically limited to no more than about one to two centimeters.

For scanners intended for abdominal and other use, the limitation on aperture size is less a matter of physical constraints, and more a matter of difficulties in image processing. The problem is that it is difficult to keep the elements of a large aperture array in phase because the speed of ultrasound transmission varies with the type of tissue between the probe and the area of interest. According to the book by Wells (cited above), the speed varies up to plus or minus 10% within the soft tissues. When the aperture is kept small (e.g. less than about 2 cm), the intervening tissue is, to a first order of approximation, all the same and any variation is ignored. When the size of the aperture is increased to improve the lateral resolution, the additional elements of a phased array may be out of phase and may actually degrade the image rather than improving it.

US Patent Application Publication 2008/0103393 to Specht teaches embodiments of ultrasound imaging systems utilizing multiple apertures which may be separated by greater distances, thereby producing significant improvements in lateral resolution of ultrasound images.

SUMMARY OF THE INVENTION

One embodiment of a method describes a method of constructing an ultrasound image, comprising transmitting an omni-directional unfocused ultrasound waveform approximating a first point source within a transmit aperture on a first array through a target region, receiving ultrasound echoes from the target region with first and second receiving elements disposed on a first receive aperture on a second array, the first array being physically separated from the second array, determining a first time for the waveform to propagate from the first point source to a first pixel location in the target region to the first receiving element, and determining a second time for the waveform to propagate from the first point source to the first pixel location in the target region to the second receiving element, and forming a first ultrasound image of the first pixel by combining the echo received by the first receiving element at the first time with the echo received by the second receiving element at the second time.

In some embodiments, the method further comprises repeating the determining and forming steps for additional pixel locations in the target region. In one embodiment, additional pixel locations are located on a grid without scan-conversion.

In one embodiment, determining the first time and the second time comprises assuming a uniform speed of sound.

In another embodiment, the method further comprises transmitting a second omni-directional unfocused ultrasound waveform approximating a second point source within the transmit aperture through the target region, receiving ultrasound echoes from the target region with first and second receiving elements disposed on the first receive aperture, determining a third time for the second waveform to propagate from the second point source to the first pixel location in the target region to the first receiving element, and determining a fourth time for the second waveform to propagate from the second point source to the first pixel location in the target region to the second receiving element, and forming a second ultrasound image of the first pixel by combining the echo received by the first receiving element at the third time with the echo received by the second receiving element at the fourth time.

In some embodiments, the method further comprises combining the first ultrasound image with the second ultrasound image. The combining step can comprise coherent addition. In another embodiment, the combining step can comprise incoherent addition. In yet another embodiment, the combining step can comprise a combination of coherent addition and incoherent addition.

In some embodiments, the method can further comprise receiving ultrasound echoes from the target region with third and fourth receiving elements disposed on a second receive aperture on a third array, the third array being physically separated from the first and second arrays, determining a third time for the waveform to propagate from the first point source to the first pixel location in the target region to the third receiving element, and determining a fourth time for the waveform to propagate from the first point source to the first pixel location in the target region to the fourth receiving element, and forming a second ultrasound image of the first pixel by combining the echo received by the third receiving element at the third time with the echo received by the fourth receiving element at the fourth time.

In some embodiments, the method further comprises repeating the determining and forming steps for additional pixel locations in the target region. In some embodiments, the additional pixel locations are located on a grid without scan-conversion.

In one embodiment, the method further comprises transmitting a second omni-directional unfocused ultrasound waveform approximating a second point source within the transmit aperture through the target region, receiving ultrasound echoes from the target region with first and second receiving elements disposed on the first receive aperture and with the third and fourth receiving elements disposed on the second receive aperture, determining a fifth time for the second waveform to propagate from the second point source to the first pixel location in the target region to the first receiving element, determining a sixth time for the second waveform to propagate from the second point source to the first pixel location in the target region to the second receiving element, determining a seventh time for the second waveform to propagate from the second point source to the first pixel location in the target region to the third receiving element, determining an eighth time for the second waveform to propagate from the second point source to the first pixel location in the target region to the fourth receiving element, and forming a third ultrasound image of the first pixel by combining the echo received by the first receiving element at the fifth time with the echo received by the second receiving element at the sixth time, and forming a fourth ultrasound image of the first pixel by combining the echo received by the third receiving element at the seventh time with the echo received by the fourth receiving element at the eighth time.

In some embodiments, the method further comprises combining the first, second, third, and fourth ultrasound images. In some embodiments, the combining step comprises coherent addition. In other embodiments, the combining step comprises incoherent addition. In additional embodiments, the combining step comprises a combination of coherent addition and incoherent addition.

In some embodiments, the method comprises combining the first ultrasound image with the second ultrasound image. The combining step can comprise coherent addition. In another embodiment, the combining step can comprise incoherent addition. In yet another embodiment, the combining step can comprise a combination of coherent addition and incoherent addition.

In some embodiments, the method further comprises comparing the first ultrasound image to the second, third, and fourth ultrasound images to determine displacements of the second, third, and fourth ultrasound images relative to the first ultrasound image.

In another embodiment, the method further comprises correcting the displacements of the second, third, and fourth ultrasound images relative to the first ultrasound image and then combining the first, second, third and fourth ultrasound images.

In an additional embodiment, the method comprises adjusting the third, fourth, fifth, sixth, seventh, and eighth times to correct the displacements of the second, third, and fourth ultrasound images relative to the first ultrasound image.

In some embodiments, the method further comprises comparing the first ultrasound image to the second ultrasound image to determine a displacement of the second ultrasound image relative to the first ultrasound image.

The method can further comprise correcting the displacement of the second ultrasound image relative to the first ultrasound image and then combining the first and second ultrasound images.

In another embodiment, the method comprises adjusting the third time and the fourth time to correct the displacement of the second ultrasound image relative to the first ultrasound image.

In some embodiments, the first pixel is disposed outside a plane defined by the point source, the first receiving element, and the second receiving element. In other embodiments, the first pixel is disposed inside a plane defined by the point source, the first receiving element, and the second receiving element.

Various embodiments of a multi-aperture ultrasound imaging system are also provided, comprising a transmit aperture on a first array configured to transmit an omni-directional unfocused ultrasound waveform approximating a first point source through a target region, a first receive aperture on a second array having first and second receiving elements, the second array being physically separated from the first array, wherein the first and second receiving elements are configured to receive ultrasound echoes from the target region, and a control system coupled to the transmit aperture and the first receive aperture, the control system configured to determine a first time for the waveform to propagate from the first point source to a first pixel location in the target region to the first receiving element, and is configured to determine a second time for the waveform to propagate from the first point source to the first pixel location in the target region to the second receiving element, the control system also being configured to form a first ultrasound image of the first pixel by combining the echo received by the first receiving element at the first time with the echo received by the second receiving element at the second time.

In some embodiments of the system, there are no transducer elements disposed between the physical separation of the transmit aperture and the first receive aperture.

In one embodiment of the system, the transmit aperture and the first receive aperture are separated by at least twice a minimum wavelength of transmission from the transmit aperture. In another embodiment, the transmit aperture and the receive aperture comprise a total aperture ranging from 2 cm to 10 cm.

In some embodiments, the ultrasound system further comprises a second receive aperture on a third array having third and fourth receiving elements, the third array being physically separated from the first and second arrays, wherein the third and fourth receiving elements are configured to receive ultrasound echoes from the target region.

In another embodiment of the multi-aperture ultrasound imaging system, the control system can be coupled to the transmit aperture and the first and second receive apertures, wherein the control system is configured to determine a third time for the waveform to propagate from the first point source to a first pixel location in the target region to the third receiving element, and is configured to determine a fourth time for the waveform to propagate from the first point source to the first pixel location in the target region to the fourth receiving element, the control system also being configured to form a second ultrasound image of the first pixel by combining the echo received by the third receiving element at the third time with the echo received by the fourth receiving element at the fourth time.

In some embodiments, the control system is configured to correct a displacement of the second ultrasound image relative to the first ultrasound image due to speed of sound variation.

In other embodiments of the multi-aperture ultrasound imaging system, the transmit aperture, the first receive aperture, and the second receive aperture are not all in a single scan plane.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. **1**A. A two-aperture system.

FIG. **1**B. Equidistant time delay points forming an ellipse around a transmit transducer element and receive transducer element.

FIG. **1**C. Loci of points relative to equidistant time delays for different receive transducer elements.

FIG. **2**. A three-aperture system.

FIG. **3**. Grid for display and coordinate system.

FIG. **4**. Fat layer model with a three-aperture system.

FIG. **5**. Construction for estimation of point spread function.

DETAILED DESCRIPTION OF THE INVENTION

Greatly improved lateral resolution in ultrasound imaging can be achieved by using multiple separate apertures for transmit and receive functions. Systems and methods herein may provide for both transmit functions from point sources and for compensation for variations in the speed-of-sound of ultrasound pulses traveling through potentially diverse tissue types along a path between a transmit aperture and one or more receive apertures. Such speed-of-sound compensation may be performed by a combination of image comparison techniques (e.g., cross-correlation), and the coherent and/or incoherent averaging of a plurality of received image frames.

As used herein the terms "ultrasound transducer" and "transducer" may carry their ordinary meanings as understood by those skilled in the art of ultrasound imaging technologies, and may refer without limitation to any single component capable of converting an electrical signal into an ultrasonic signal and/or vice versa. For example, in some embodiments, an ultrasound transducer may comprise a piezoelectric device. In some alternative embodiments, ultrasound transducers may comprise capacitive micromachined ultrasound transducers (CMUT). Transducers are often configured in arrays of multiple elements. An element of a transducer array may be the smallest discrete component of an array. For example, in the case of an array of piezoelectric transducer elements, each element may be a single piezoelectric crystal.

As used herein, the terms "transmit element" and "receive element" may carry their ordinary meanings as understood by those skilled in the art of ultrasound imaging technologies. The term "transmit element" may refer without limitation to an ultrasound transducer element which at least momentarily performs a transmit function in which an electrical signal is converted into an ultrasound signal. Similarly, the term "receive element" may refer without limitation to an ultrasound transducer element which at least momentarily performs a receive function in which an ultrasound signal impinging on the element is converted into an electrical signal. Transmission of ultrasound into a medium may also be referred to herein as "insonifying." An object or structure which reflects ultrasound waves may be referred to as a "reflector" or a "scatterer."

As used herein the term "aperture" refers without limitation to one or more ultrasound transducer elements collectively performing a common function at a given instant of time. For example, in some embodiments, the term aperture may refer to a group of transducer elements performing a transmit function. In alternative embodiments, the term aperture may refer to a plurality of transducer elements performing a receive function. In some embodiments, group of transducer elements forming an aperture may be redefined at different points in time. FIG. **3** demonstrates multiple apertures used in a multiple aperture ultrasound probe. An aperture of the probe has up to three distinct features. First, it is often physically separated from other transducers located in other apertures. In FIG. **3**, a distance 'd' physically separates aperture **302** from aperture **304**. Distance 'd' can be the minimum distance between transducer elements on aperture **302** and transducer elements on aperture **304**. In some embodiments, no transducer elements are disposed along the distance 'd' between the physical separation of apertures **302** and **304**. In some embodiments, the distance can be equal to at least twice the minimum wavelength of transmission from the transmit aperture. Second, the transducer elements of an aperture need not be in the same rectangular or horizontal plane. In FIG. **3**, all the elements of aperture **304** have a different vertical position 'j' from any element of aperture **302**. Third, apertures do not share a common line of sight to the region of interest. In FIG. **3**, aperture **302** has a line of sight 'a' for point (i,j), while aperture **304** has a line of sight 'b'. An aperture may include any number of individual ultrasound elements. Ultrasound elements defining an aperture are often, but not necessarily adjacent to one another within an array. During operation of a multi-aperture ultrasound imaging system, the size of an aperture (e.g. the number and/or size and/or position of ultrasound elements) may be dynamically changed by re-assigning elements.

As used herein the term "point source transmission" may refer to an introduction of transmitted ultrasound energy into a medium from single spatial location. This may be accomplished using a single ultrasound transducer element or combination of adjacent transducer elements transmitting together. A single transmission from said element(s) approximates a uniform spherical wave front, or in the case of imaging a 2D slice it creates a uniform circular wave front within the 2D slice. This point source transmission differs in its spatial characteristics from a "phased array transmission" which focuses energy in a particular direction from the transducer element array. Phased array transmission manipulates the phase of a group of transducer elements in sequence so as to strengthen or steer an insonifying wave to a specific region of interest. A short duration point source transmission is referred to herein as a "point source pulse." Likewise, a short duration phased array transmission is referred to herein as a "phased array pulse."

As used herein, the terms "receive aperture," "insonifying aperture," and/or "transmit aperture" can carry their ordinary meanings as understood by those skilled in the art of ultrasound imaging, and may refer to an individual element, a group of elements within an array, or even entire arrays within a common housing, that perform the desired transmit or receive function from a desired physical viewpoint or aperture at a given time. In some embodiments, these various apertures may be created as physically separate components with dedicated functionality. In alternative embodiments, the functionality may be electronically designated and changed as needed. In still further embodiments, aperture functionality may involve a combination of both fixed and variable elements.

In some embodiments, an aperture is an array of ultrasound transducers which is separated from other transducer arrays. Such multiple aperture ultrasound imaging systems provide greatly increased lateral resolution. According to some embodiments, a multi-aperture imaging method comprises the steps of insonifying a target object with an ultrasound pulse from a first aperture, detecting returned echoes with a second aperture positioned at a distance from the first aperture, determining the relative positions of the second aperture with respect to the first aperture, and processing returned echo data to combine images white correcting for variations in speed-of-sound through the target object.

In some embodiments, a distance and orientation between adjacent apertures may be fixed relative to one another, such as by use of a rigid housing. In alternative embodiments, distances and orientations of apertures relative to one another may be variable, such as with a movable linkage. In further alternative embodiments, apertures may be defined as groups of elements on a single large transducer array where the groups are separated by at least a specified distance. For example, some embodiments of such a system are shown and described in U.S. Provisional Patent Application No. 61/392,896, filed Oct. 13, 2010, titled "Multiple Aperture Medical Ultrasound Transducers". In some embodiments of a multi-aperture ultrasound imaging system, a distance between adjacent apertures may be at least a width of one transducer element. In alternative embodiments, a distance between apertures may be as large as possible within the constraints of a particular application and probe design.

A multi-aperture ultrasound imaging system with a large effective aperture (the total aperture of the several sub apertures) can be made viable by compensation for the variation of speed-of-sound in the target tissue. This may be accomplished in one of several ways to enable the increased aperture to be effective rather than destructive, as described below.

FIG. **1**A illustrates one embodiment of a simplified multi-aperture ultrasound imaging system **100** comprising two apertures, aperture **102** and aperture **104**. Each of apertures **102** and **104** can comprise a plurality of transducer elements. In the two-aperture system shown in FIG. **1**A, aperture **102** can comprise transmit elements T**1** . . . Tn to be used entirely for transmit functions, and aperture **104** can comprise receive elements R**1** . . . . Rm to be used entirely for receive functions. In alternative embodiments, transmit elements may be interspersed with receive elements, or some elements may be used both for transmit and receive functions. The multi-aperture ultrasound imaging system **100** of FIG. **1**A can be configured to be placed on a skin surface of a patient to image target object or internal tissue T with ultrasound energy. As shown in FIG. **1**A, aperture **102** is positioned a distance "a" from tissue T, and aperture **104** is positioned a distance "b" from tissue T. Also shown in FIG. **1**A, MAUI electronics may be coupled to the transmit and receive apertures **102** and **104**. In some embodiments, the MAUI electronics can comprise a processor, control system, or computing system, including hardware and software configured to control the multi-aperture imaging system **100**. In some embodiments, the MAUI electronics can be configured to control the system to transmit an omni-directional unfocused ultrasound waveform from an aperture, receive echoes on an aperture, and form images from the transmitted waveform and the received echoes. As will be described in further detail below, the MAUI electronics can be configured to control and achieve any of the methods described herein.

Ultrasound elements and arrays described herein may also be multi-function. That is, the designation of transducer elements or arrays as transmitters in one instance does not preclude their immediate re-designation as receivers in the next instance. Moreover, embodiments of the control system described herein include the capabilities for making such designations electronically based on user inputs or pre-set scan or resolution criteria.

Another embodiment of a multi-aperture ultrasound imaging system **200** is shown in FIG. **2** and includes transducer elements arranged to form three apertures **202**, **204**, and **206**. In one embodiment, transmit elements T**1** . . . Tn in aperture **202** may be used for transmit, and receive elements R_R**1** . . . R_Rm in apertures **204** and **206** may be used for receive. In alternative embodiments, elements in all the apertures may be used for both transmit and receive. The multi-aperture ultrasound imaging system **200** of FIG. **2** can be configured to image tissue T with ultrasound energy. Also shown in FIG. **2**, MAUI electronics may be coupled to the transmit and receive apertures **202** and **204**. In some embodiments, the MAUI electronics can comprise a processor, control system, or computing system, including hardware and software configured to control the multi-aperture imaging system **200**. In some embodiments, the MAUI electronics can be configured to control the system to transmit an omni-directional unfocused ultrasound waveform from an aperture, receive echoes on an aperture, and form images from the transmitted waveform and the received echoes. As will be described in further detail below, the MAUI electronics can be configured to control and achieve any of the methods described herein.

Multi-aperture ultrasound imaging systems described herein may be configured to utilize transducers of any desired construction. For example, 1D, 1.5D, 2D, CMUT or any other transducer arrays may be utilized in multi-aperture configurations to improve overall resolution and field of view.

Point Source Transmission

In some embodiments, acoustic energy may be transmitted to as wide a two-dimensional slice as possible by using point source transmission. For example, in some embodiments, a transmit aperture, such as transmit apertures **102** or **202** in FIGS. **1**A and **2**, respectively, may transmit acoustic energy in the form of a point source pulse from a single substantially omni-directional transducer element in an array. In alternative embodiments, a plurality of transducer elements may be provisioned to transmit a point source pulse that is relatively wide in three dimensions to insonify objects in a three dimensional space. In such embodiments, all of the beam formation may be achieved by the software or firmware associated with the transducer arrays acting as receivers. There are several advantages to using a multi-aperture ultrasound imaging technique by transmitting with a point source pulse rather than a phased array pulse. For example when using a phased array pulse, focusing tightly on transmit is problematic because the transmit pulse would have to be focused at a particular depth and would be somewhat out of focus at all other depths. Whereas, with a point source transmission an entire two-dimensional slice or three-dimensional volume can be insonified with a single point source transmit pulse.

Each echo detected at a receive aperture, such as receive apertures **104** or **204**/**206** in FIGS. **1**A and **2**, respectively, may be stored separately. If the echoes detected with elements in a receive aperture are stored separately for every point source pulse from an insonifying or transmit aperture, an entire two-dimensional image can be formed from the information received by as few as just one element. Additional copies of the image may be formed by additional receive apertures collecting data from the same set of insonifying point source pulses. Ultimately, multiple images can be created simultaneously from one or more apertures and combined to achieve a comprehensive 2D or 3D image.

Although several point source pulses are typically used in order to produce a high-quality image, fewer point source pulses are required than if each pulse were focused on a particular scan line. Since the number of pulses that can be transmitted in a given time is strictly limited by the speed of ultrasound in tissue, this yields the practical advantage that more frames can be produced per second by utilizing a point source pulse. This is very important when imaging moving organs, and in particular, the heart.

In some embodiments, a spread spectrum waveform may be imposed on a transmit aperture made up of one or more ultrasound transducer elements. A spread spectrum waveform may be a sequence of frequencies such as a chirp (e.g., frequencies progressing from low to high, or vice versa), random frequency sequence (also referred to as frequency hop), or a signal generated by a pseudo random waveform (PN sequence). These techniques can be collectively referred to as pulse compression. Pulse compression provides longer pulses for greater depth penetration without loss of depth resolution. In fact, the depth resolution may be greatly improved in the process. Spread spectrum processing typically involves much more signal processing in the form of matched filtering of each of the received signals before the delay and summation steps. The above examples of transmit pulse forms are provided for illustration only. The techniques taught herein may apply regardless of the form of the transmit pulse.

Basic Image Rendering

FIG. **1**A illustrates one embodiment of a multi-aperture ultrasound imaging system **100** containing a first aperture **102** with ultrasound transmitting elements T**1**, T**2**, . . . Tn and a second aperture **104** with ultrasound receive elements R**1**, R**2**, . . . Rm. This multi-aperture ultrasound imaging system **100** is configured to be placed on the surface of an object or body to be examined (such as a human body). In some embodiments, both apertures may be sensitive to the same plane of scan. In other embodiments, one of the apertures may be in a different plane of scan. The mechanical and acoustic position of each transducer element of each aperture must be known precisely relative to a common reference point or to each other.

In one embodiment, an ultrasound image may be produced by insonifying the entire region to be imaged, such as internal tissue or target object T, (e.g., a plane through the heart, organ, tumor, or other portion of the body) with a transmitting element (e.g., transmit element T**1** of aperture **102**), and then receiving echoes from the entire imaged plane on a receive element (e.g. receive element R**1** of aperture **104**). In some embodiments, receive functions may be performed by all elements in the receive probe (e.g., R**1** through Rm). In alternative embodiments, echoes are received on only one or a select few elements of the receive aperture. The method proceeds by using each of the elements on the transmitting aperture **102** (e.g., T**2**, . . . Tn) and insonifying the entire region to be imaged with each of the transmitting elements in turn, and receiving echoes on the receive aperture after each insonifying pulse. Transmit elements may be operated in any desired sequential order, and need not follow a prescribed pattern. Individually, the images obtained after insonification by each transmitting element may not be sufficient to provide a high resolution image, but the combination of all the images may provide a high resolution image of the entire region to be imaged. For a scanning point represented by coordinates (i,j) as shown in FIG. **1**A, it is a simple matter to calculate the total distance "a" from a particular transmit element Tx to an element of internal tissue or target object at (i,j), and the distance "b" from that point to a particular receive element. These calculations may be performed using basic trigonometry. The sum of these distances is the total distance traveled by one ultrasound wave.

When the speed of ultrasound in tissue is assumed to be uniform throughout the tissue, it is possible to calculate the time delay from the onset of the transmit pulse to the time that an echo is received at the receive element. (Non uniform speed-of-sound in tissue is discussed below.) This one fact means that a scatterer (i.e., a reflective point within the target object) is a point in the medium for which a+b=the given time delay. The same method can be used to calculate delays for any point in the desired tissue to be imaged, creating a locus of points. FIG. **1**B demonstrates that points (g,h), (i,j), (k,m), (n,p) (q,r), (s,t) all have the same time delay for transmit element T_1 and receive element R_1. A map of scatter positions and amplitudes can be rendered by tracing the echo amplitude to all of the points for the locus of equal-time-delay points. This locus takes the form of an ellipse **180** with foci at the transmit and receive elements. FIG. **1**B also illustrates MAUI electronics, which can comprise the MAUI electronics described above with reference to FIGS. **1**A and **2**.

The fact that all points on the ellipse **180** are returned with the same time delay presents a display challenge, since distinguishing points along the ellipse from one another within a single image is not possible. However, by combining images obtained from multiple receive points, the points may be more easily distinguished, since the equal-time-delay ellipses defined by the multiple receive apertures will be slightly different.

FIG. **1**C shows that with a transmit pulse from element T**1**, echoes from a single scatterer (n,p) are received by different receive elements such as R**1**, R**2**, and R**3** at different times. The loci of the same scatterer can be represented by ellipses **180**, **185** and **190** of FIG. **1**C. The location at which these ellipses intersect (point n,p) represents the true location of the scatterer. Beam forming hardware, firmware, or software can combine the echoes from each receive element to generate an image, effectively reinforcing the image at the intersection of the ellipses. In some embodiments, many more receiver elements than the three shown may be used in order to obtain a desirable signal-to-noise ratio for the image. FIG. **1**C also illustrates MAUI electronics, which can comprise the MAUI electronics described above with reference to FIGS. **1**A and **2**.

A method of rendering the location of all of the scatterers in the target object, and thus forming a two dimensional cross section of the target object, will now be described with reference to multi-aperture ultrasound imaging system **300** of FIG. **3**. FIG. **3** illustrates a grid of points to be imaged by apertures **302** and **304**. A point on the grid is given the rectangular coordinates (i,j). The complete image will be a two dimensional array called "echo." In the grid of FIG. **3**, mh is the maximum horizontal dimension of the array and mv is the maximum vertical dimension. FIG. **3** also illustrates MAUI electronics, which can comprise the MAUI electronics described above with reference to FIGS. **1**A and **2**.

In one embodiment, the following pseudo code may be used to accumulate all of the information to be gathered from a transmit pulse from one transmit element (e.g., one element of T1 . . . Tn from aperture **302**), and the consequent echoes received by one receive element (e.g., one element of R1 . . . Rm from aperture **304**) in the arrangement of FIG. **3**.

```
for (i = 0; i < mh; i++){
    for (j = 0;j < mv; j++){
        compute distance a
        compute distance b
        compute time equivalent of a+b
        echo[ i ][ j ] = echo[i ][ j]+stored received echo at the computed
        time delay.
    }
}
```

The fixed delay is primarily the time from the transmit pulse until the first echoes are received. As will be discussed later, an increment can be added or subtracted to compensate for varying fat layers.

A complete two dimensional image may be formed by repeating this process for every receive element in aperture **304** (e.g., R1 . . . Rm). In some embodiments, it is possible to implement this code in parallel hardware resulting in real time image formation.

Combining similar images resulting from pulses from other transmit elements will improve the quality (e.g., in terms of signal-to-noise ratio) of the image. In some embodiments, the combination of images may be performed by a simple summation of the single point source pulse images (e.g., coherent addition). Alternatively, the combination may involve taking the absolute value of each element of the single point source pulse images first before summation (e.g., incoherent addition). In some embodiments, the first technique (coherent addition) may be best used for improving lateral resolution, and the second technique (incoherent addition) may be best applied for the reduction of speckle noise. In addition, the incoherent technique may be used with less precision required in the measurement of the relative positions of the transmit and receive apertures. A combination of both techniques may be used to provide an optimum balance of improved lateral resolution and reduced speckle noise. Finally, in the case of coherent addition, the final sum should be replaced by the absolute value of each element, and in both cases, some form of compression of the dynamic range may be used so that both prominent features and more-subtle features appear on the same display. In some embodiments, additional pixel locations are located on a grid without scan-conversion.

In some embodiments, compression schemes may include taking the logarithm (e.g., 20 $\log_{10}$ or "dB") of each element before display, or taking the nth root (e.g., 4^{th} root) of each element before display. Other compression schemes may also be employed.

Referring still to FIG. **3**, any number of receive probes and transmit probes may be combined to enhance the image of scatterer (i,j) as long as the relative positions of the transducer elements are known to a designed degree of precision, and all of the elements are in the same scan plane and are focused to either transmit energy into the scan plane or receive energy propagated in the scan plane. Any element in any probe may be used for either transmit or receive or both.

The speed-of-sound in various soft tissues throughout the body can vary by +/−10%. Using typical ultrasound techniques, it is commonly assumed that the speed-of-sound is constant in the path between the transducer and the organ of interest. This assumption is valid for narrow transducer arrays in systems using one transducer array for both transmit and receive. However, the constant speed-of-sound assumption breaks down as the transducer's aperture becomes wider because the ultrasound pulses pass through more tissue and possibly diverse types of tissue, such as fat, muscle, blood vessels, etc. Tissue diversity under the width of the transducer array affects both the transmit and the receive functions.

When a scatterer is insonified by a point source pulse from a single transmit element, it reflects back an echo to all of the elements of the receiver group. Coherent addition of images collected by elements in this receive aperture can be effective if the speed-of-sound variations in the paths from scatterer (i,j) to each of the receiver elements do not exceed +−180 degrees phase shift relative to one path chosen as reference. Referring to FIG. **3**, the maximum size of the receive aperture for which coherent addition can be effective is dependent on tissue variation within the patient and cannot be computed in advance. However, a practical maximum for a particular transmit frequency can be determined from experience.

When insonifying with unfocused point source pulses, the aperture size of the transmit group is not highly critical since variation in the path time from transmitter elements to a scatterer such as scatterer (i,j) will change only the displayed position of the point. For example, a variation resulting in a phase shift of 180 degrees in the receive paths results in complete phase cancellation when using coherent addition, whereas the same variation on the transmit paths results in a displayed position error of only a half wavelength (typically about 0.2 mm), a distortion that would not be noticed.

Thus, in a multi-aperture imaging system with one aperture used only for transmit and the other used only for receive during a single transmit/receive cycle, as is illustrated in FIG. **1**A, very little additional compensation for the speed-of-sound variation is needed. Although the aperture has been increased from element T1 to Rm which can be many times the width of a conventional sector scanner probe, the concern of destructive interference of the signals from scatterer (i,j) is independent of the width of the transmit aperture or the separation of the apertures, and is dependent only on the width of the receive aperture (element R1 to Rm). The standard width for which speed-of-sound variation presents a minimal problem in practice is about 16-20 mm for 3.5 MHz systems (and smaller for higher frequencies). Therefore, no explicit compensation for speed-of-sound variation is necessary if the receive aperture has the same or smaller width than standard apertures.

Substantial improvement in lateral resolution is achieved with a receive aperture of the same width as a conventional single array 1D, 1.5D or 2D ultrasound probe used for both transmit and receive, because received energy when imaging adjacent cells (i.e., regions of the target object) to that which represents a scatterer is dependent on the time difference between when an echo is expected to arrive and the time that it actually arrives. When the transmit pulse originates from the same array used for receive, the time difference is small. However, when the transmit pulse originates from a second array at some distance from the receive array, the time difference is larger and therefore more out of phase with the signal for the correct cell. The result is that fewer adjacent cells will have signals sufficiently in phase to falsely represent the true scatterer.

Referring to FIG. **4**, consider the signal received at a single element (e.g., one of receive elements R1 . . . Rm) of a receive aperture **404** from a scatterer at "S". If both the transmit and receive functions are performed on the same element, the time for the ultrasound to propagate to "S" and be returned would be 2a/C (where C is the speed-of-sound in tissue). When the reconstruction algorithm is evaluating the signal received for a possible scatterer in an adjacent cell "S" separated "c" distance from the true scatterer, "S", the expected time of arrival is $2(\mathrm{sqrt}(a^2+c^2)/C)$. When "c" is small, this time is almost the same and so the signal from "S" will be degraded only slightly when estimating the magnitude of the scatterer "S'" in the adjacent cell. FIG. **4** also illustrates MAUI electronics, which can comprise the MAUI electronics described above.

Now consider moving the transmitting aperture **402** away from the receive aperture **404** by an angle theta ("θ"). For convenience in comparison, let the distance "b" from aperture **402** to scatterer "S" be equal to the distance "a" from aperture **404** to scatterer "S". The time for the ultrasound to propagate from the transmit aperture **402** to "S" and be returned to the receive aperture **404** would still be $(a+b)/C=2a/C$ (with a=b), but the expected time for the signal to propagate to the adjacent cell "S" would be $(d+\mathrm{sqrt}(a^2+c^2)/C=(\mathrm{sqrt}((a\sin\theta-c)^2+(a\cos\theta)^2)+\mathrm{sqrt}(a^2+c^2))/C$. The difference between the expected time of arrival and actual would then be $\mathrm{Diff}=(\mathrm{sqrt}((a\sin\theta-c)^2+(a\cos\theta)^2)+\mathrm{sqrt}(a^2+c^2)-2a)/C$.

To put some numbers in this equation, suppose that the separation of aperture **402** and aperture **404** is only 5 degrees, distance a=400 cells, and distance c=1 cell. Then the ratio of the difference in time-of-arrival for θ=5 degrees to that for θ=0 degrees is 33.8. That is, the drop off of display amplitude to adjacent cells is 33 times faster with θ=5 degrees. The larger difference in time-of-arrival greatly simplifies the ability to uniquely distinguish echo information from adjacent cells. Therefore, with high theta angles, the display of a point will be less visible as noise in adjacent cells and the result will be higher resolution of the real image. With multiple aperture transmitters and receivers, we can make the angle as high as needed to improve resolution.

Simulation for a realistic ultrasound system with multiple reflectors in multiple cells shows that the effect is still significant, but not as dramatic as above. For a system comprising a receive aperture of 63 elements, a θ of 10 degrees, and a transmit pulse from a point-source transmit aperture that extends for 5 cycles with cosine modulation, the lateral spread of the point spread function was improved by a factor of 2.3.

Explicit Compensation for Speed-of-Sound Variation

A single image may be formed by coherent averaging of all of the signals arriving at the receiver elements as a result of a single point source pulse for insonification. Summation of these images resulting from multiple point source pulses can be accomplished either by coherent addition, incoherent addition, or a combination of coherent addition by groups and incoherent addition of the images from the groups. Coherent addition (retaining the phase information before addition) maximizes resolution whereas incoherent addition (using the magnitude of the signals and not the phase) minimizes the effects of registration errors and averages out speckle noise. Some combination of the two modes may be preferred. Coherent addition can be used to average point source pulse images resulting from transmit elements that are close together and therefore producing pulses transmitted through very similar tissue layers. Incoherent addition can then be used where phase cancellation would be a problem. In the extreme case of transmission time variation due to speed-of-sound variations, 2D image correlation can be used to align images prior to addition.

When an ultrasound imaging system in a second aperture, using the second aperture for receiving as well as transmitting produces much better resolution. In combining the images from two or more receive arrays; it is possible and beneficial to use explicit compensation for the speed-of-sound variation.

Consider the tissue layer model for the three-aperture ultrasound imaging system **500** as shown in FIG. **5**, which illustrates the effects of varying thicknesses of different types of tissue, such as fat or muscle. A fat layer "F" is shown in FIG. **5**, and the thickness of the tissue layers f**1**, f**2**, and f**3** under each aperture **502**, **504**, and **506**, respectively, is different and unknown. It is not reasonable to assume that the tissue layer at aperture **506** will be the same as at aperture **504**, and so coherent addition of the signals from all of the receive elements together is not usually possible. In one example, if the tissue layer at aperture **504** were as much as 3 cm larger than that at aperture **506**, this corresponds to about 3 wavelengths (at 3.5 MHz) displacement of the signals, but this is only 1.3 mm displacement of the representation of the deep tissues. For such small displacements, only a tiny amount of geometric distortion of the image would be observed. Therefore, although coherent addition is not possible, incoherent addition with displacement of one image relative to the other is possible.

Image comparison techniques may be used to determine the amount of displacement needed to align image frames from left and right apertures (e.g., apertures **506** and **504**, respectively). In one embodiment, the image comparison technique can be cross-correlation. Cross-correlation involves evaluating the similarity of images or image sections to identify areas with a high degree of similarity. Areas with at least a threshold value of similarity may be assumed to be the same. Thus, by identifying areas within images with high degrees of similarity, one image (or a section thereof) may be shifted such that areas with substantial similarity overlap and enhance overall image quality. FIG. **5** also illustrates MAUI electronics, which can comprise the MAUI electronics described above.

Further, these image comparison techniques can also be used by applying sub-image analysis, which can be used to determine displacement of sub-images and accommodate for localized variation in speed-of-sound in the underlying tissue. In other words, by breaking down the images into smaller segments (e.g. halves, thirds, quarters, etc), small portions of a first image may be compared to the corresponding small portion of a second image. The two images may then be combined by warping to assure alignment. Warping is a technique understood by those skilled in the art, and is described, for example in U.S. Pat. No. 7,269,299 to Schroeder.

The same technique of incoherent addition of images from multiple receive transducer arrays may be applied to any number of apertures. The same idea may be applied even to a single element array which is too wide to be used for coherent addition all at once. An ultrasound imaging system with a single wide array of elements may be divided into sections (apertures) each of which is small enough for coherent addition, and then the images resulting from these sections may be combined incoherently (with displacement if necessary).

Even a slight distortion of the image may be compensated for with sufficient computational power. Image renderings may be computed for one receive array using varying amounts of delay in the rendering algorithm (echo[i][j]=echo[i][j]+stored receive echo at the computed time+delay). Then the best matched of these (by cross-correlation or some other measure of acuity) may be incoherently added to the image from the other receive array(s). A faster technique includes calculating the cross correlation network for the uncorrected pair of images, and feeding this into a neural network trained to pick the correction delay.

Because multiple aperture ultrasound systems that can correct for speed of sound incongruences allow for significantly larger apertures, some embodiments of the multi-aperture ultrasound systems described herein can have apertures located 10 cm apart from one another. Since resolution is proportional to 2λ/D, this larger aperture leads to higher resolution of tissues located well below the surface of the skin. For instance, the renal arteries are frequently located 10 cm to 15 cm below the skin and are 4 mm to 6 mm in size near the abdominal aorta. Phased array, linear array and synthetic aperture ultrasound systems usually cannot detect this physiology in most patients; specifically because the aperture size is not large enough to have adequate lateral resolution. Typically, phased array systems have aperture sizes of approximately 2 cm. Increasing the aperture size from larger than 2 cm to approximately 10 cm in a multi-aperture ultrasound system can increase the resolution by up to 5×.

3D Imaging

In some embodiments, three-dimensional information may be obtained by moving a two-dimensional imaging system and acquiring 2D slices at a number of positions or angles. From this information and using interpolation techniques, a 3D image at any position or angle may be reconstructed. Alternatively, a 2D projection of all of the data in the 3D volume may be produced. A third alternative is to use the information in a direct 3D display.

Because multi-aperture ultrasound imaging systems may result in wider probe devices, the easiest way to use them to obtain 3D data is to not move them on the patient's skin but merely rock them so that the 2D slices span the 3D volume to be imaged. In some embodiments, a mechanical rotator mechanism which records position data may be used to assist in the collection the 2D slices. In other embodiments, a freely operated ultrasound probe with precision position sensors (such as gyroscopic sensors) located in the head of the probe may be used instead. Such an apparatus allows for complete freedom of movement while collecting 2D slices. Finally, intravenous and intracavity probes may also be manufactured to accommodate wide apertures. Such probes may be manipulated in similar ways in order to collect 2D slices.

This combination is particularly desirable for 3D cardiac imaging using a multi-aperture ultrasound imaging system. Most patients have good acoustic windows in two intercostal spaces next to the sternum. A multi-aperture imaging system is ideal in this case since the intervening rib would render a flat probe useless, while a probe with at least two widely spaced apertures can be positioned such that a send aperture and a receive aperture align with separate intercostal spaces. Once a probe with multiple apertures is in place, it cannot be rotated, but it can be rocked to obtain the 3D information, A multi-aperture probe may also be used in the same intercostal space but across the sternum.

3D information may also be obtained directly with multi-aperture imaging systems having apertures that are not all in the same scan plane. In this case the elements making up the transmit aperture preferably propagate spherical waveforms (rather than circular waveforms confined to one plane of scan). The elements making up the receive apertures may likewise be sensitive to energy arriving from all directions (rather than being sensitive only to ultrasonic energy in a single plane of scan). The reconstruction pseudo code provided above may then be extended to three dimensions.

As for additional details pertinent to the present invention, materials and manufacturing techniques may be employed as within the level of those with skill in the relevant art. The same may hold true with respect to method-based aspects of the invention in terms of additional acts commonly or logically employed. Also, it is contemplated that any optional feature of the inventive variations described may be set forth and claimed independently, or in combination with any one or more of the features described herein. Likewise, reference to a singular item, includes the possibility that there are plural of the same items present. More specifically, as used herein and in the appended claims, the singular forms "a," "and," "said," and the include plural referents unless the context clearly dictates otherwise. It is further noted that the claims may be drafted to exclude any optional element. As such, this statement is intended to serve as antecedent basis for use of such exclusive terminology as "solely," "only" and the like in connection with the recitation of claim elements, or use of a "negative" limitation. Unless defined otherwise herein, all technical and scientific terms used herein have the same meaning as commonly understood by one of ordinary skill in the art to which this invention belongs. The breadth of the present invention is not to be limited by the subject specification, but rather only by the plain meaning of the claim terms employed.

What is claimed is:

1. A method of constructing an ultrasound image, comprising:

transmitting an omni-directional unfocused ultrasound waveform from a transmit aperture comprising at least one transducer element and approximating a first point source within the transmit aperture on a first array through a target region;

receiving ultrasound echoes from the target region with first and second receiving elements disposed on a first receive aperture on a second array, the first array being physically separated from the second array;

retrieving position data describing a mechanical and acoustic position of each of the first receiving element, the second receiving element, and the at least one transducer element of the transmit aperture relative to a common reference point;

determining, using the position data, a first time for the waveform to propagate from the first point source to a first pixel location in the target region to the first receiving element, and determining a second time for the waveform to propagate from the first point source to the first pixel location in the target region to the second receiving element; and

forming a first ultrasound image of the first pixel location by combining a first echo received by the first receiving element at the first time with a second echo received by the second receiving element at the second time.

2. The method of claim **1** further comprising repeating the determining and forming steps for additional pixel locations in the target region.

3. The method of claim **2** further comprising transmitting a second omni-directional unfocused ultrasound waveform approximating a second point source within the transmit aperture through the target region;

receiving ultrasound echoes from the target region with first and second receiving elements disposed on the first receive aperture;

determining a third time for the second waveform to propagate from the second point source to the first pixel location in the target region to the first receiving element, and determining a fourth time for the second waveform to propagate from the second point source to the first pixel location in the target region to the second receiving element; and

forming a second ultrasound image of the first pixel location by combining a third echo received by the first receiving element at the third time with a fourth echo received by the second receiving element at the fourth time.

4. The method of claim **3** further comprising combining the first ultrasound image with the second ultrasound image.

5. The method of claim **4** wherein the combining step comprises coherent addition.

6. The method of claim **4** wherein the combining step comprises incoherent addition.

7. The method of claim **4** wherein the combining step comprises a combination of coherent addition and incoherent addition.

8. The method of claim **1** wherein the additional pixel locations are located on a grid without scan-conversion.

9. The method of claim **1** wherein determining the first time and the second time comprises assuming a uniform speed of sound.

10. The method of claim **1** further comprising:

receiving ultrasound echoes from the target region with third and fourth receiving elements disposed on a second receive aperture on a third array, the third array being physically separated from the first and second arrays;

determining a third time for the waveform to propagate from the first point source to the first pixel location in the target region to the third receiving element, and determining a fourth time for the waveform to propagate from the first point source to the first pixel location in the target region to the fourth receiving element; and

forming a second ultrasound image of the first pixel location by combining a third echo received by the third receiving element at the third time with a fourth echo received by the fourth receiving element at the fourth time.

11. The method of claim **10** further comprising repeating the determining and forming steps for additional pixel locations in the target region.

12. The method of claim **10** wherein the additional pixel locations are located on a grid without scan-conversion.

13. The method of claim **11** further comprising transmitting a second omni-directional unfocused ultrasound waveform approximating a second point source within the transmit aperture through the target region;

receiving ultrasound echoes from the target region with first and second receiving elements disposed on the first receive aperture and with the third and fourth receiving elements disposed on the second receive aperture;

determining a fifth time for the second waveform to propagate from the second point source to the first pixel location in the target region to the first receiving element, determining a sixth time for the second waveform to propagate from the second point source to the first pixel location in the target region to the second receiving element, determining a seventh time for the second waveform to propagate from the second point source to the first pixel location in the target region to the third receiving element, determining an eighth time for the second waveform to propagate from the second point source to the first pixel location in the target region to the fourth receiving element; and

forming a third ultrasound image of the first pixel location by combining a fifth echo received by the first receiving element at the fifth time with a sixth echo received by the second receiving element at the sixth time, and forming a fourth ultrasound image of the first pixel location by combining a seventh echo received by the third receiving element at the seventh time with an eighth echo received by the fourth receiving element at the eighth time.

14. The method of claim **13** further comprising combining the first, second, third, and fourth ultrasound images.

15. The method of claim **14** wherein the combining step comprises coherent addition.

16. The method of claim **14** wherein the combining step comprises incoherent addition.

17. The method of claim **14** wherein the combining step comprises a combination of coherent addition and incoherent addition.

18. The method of claim **10** further comprising combining the first ultrasound image with the second ultrasound image.

19. The method of claim **18** wherein the combining step comprises coherent addition.

20. The method of claim **18** wherein the combining step comprises incoherent addition.

21. The method of claim **18** wherein the combining step comprises a combination of coherent addition and incoherent addition.

22. The method of claim **13** further comprising comparing the first ultrasound image to the second, third, and fourth ultrasound images to determine displacements of the second, third, and fourth ultrasound images relative to the first ultrasound image.

23. The method of claim **22** further comprising correcting the displacements of the second, third, and fourth ultrasound images relative to the first ultrasound image and then combining the first, second, third and fourth ultrasound images.

24. The method of claim **22** further comprising adjusting the third, fourth, fifth, sixth, seventh, and eighth times to correct the displacements of the second, third, and fourth ultrasound images relative to the first ultrasound image.

25. The method of claim **10** further comprising comparing the first ultrasound image to the second ultrasound image to determine a displacement of the second ultrasound image relative to the first ultrasound image.

26. The method of claim **25** further comprising correcting the displacement of the second ultrasound image relative to the first ultrasound image and then combining the first and second ultrasound images.

27. The method of claim **25** further comprising adjusting the third time and the fourth time to correct the displacement of the second ultrasound image relative to the first ultrasound image.

28. The method of claim **10** wherein the first pixel location is disposed outside a plane defined by the point source, the first receiving element, and the second receiving element.

29. The method of claim **10** wherein the first pixel location is disposed inside a plane defined by the point source, the first receiving element, and the second receiving element.

30. The method of claim **10** wherein the first pixel location is disposed outside a plane defined by the point source, the third receiving element, and the fourth receiving element.

31. The method of claim **10** wherein the first pixel location is disposed inside a plane defined by the point source, the third receiving element, and the fourth receiving element.

32. The method of claim **1** wherein the first pixel location is disposed outside a plane defined by the point source, the first receiving element, and the second receiving element.

33. The method of claim **1** wherein the first pixel location is disposed inside a plane defined by the point source, the first receiving element, and the second receiving element.

34. A multi-aperture ultrasound imaging system, comprising:

a transmit aperture comprising at least one transducer element on a first array configured to transmit an omni-directional unfocused ultrasound waveform approximating a first point source through a target region;

a first receive aperture on a second array having first and second receiving elements, the second array being physically separated from the first array, wherein the first and second receiving elements are configured to receive ultrasound echoes from the target region;

a position memory containing position data describing a mechanical and acoustic position of each of the first receiving element, the second receiving element, and the at least one transducer element of the transmit aperture relative to a common reference point;

a control system coupled to the transmit aperture and the first receive aperture, the control system configured to retrieve the position data from the position memory, determine, using the position data, a first time for the waveform to propagate from the first point source to a first pixel location in the target region to the first receiving element, and is configured to determine a second time for the waveform to propagate from the first point source to the first pixel location in the target region to the second receiving element, the control system also being configured to form a first ultrasound image of the first pixel location by combining a first echo received by the first receiving element at the first time with a second echo received by the second receiving element at the second time.

35. The multi-aperture ultrasound imaging system of claim **34** wherein there are no transducer elements disposed between the physical separation of the transmit aperture and the first receive aperture.

36. The multi-aperture ultrasound imaging system of claim **34** wherein the transmit aperture and the first receive aperture are separated by at least twice a minimum wavelength of transmission from the transmit aperture.

37. The multi-aperture ultrasound imaging system of claim **34** wherein the transmit aperture and the receive aperture comprise a total aperture ranging from 2 cm to 10 cm.

38. The multi-aperture ultrasound imaging system of claim **34** further comprising a second receive aperture on a third array having third and fourth receiving elements, the third array being physically separated from the first and second arrays, wherein the third and fourth receiving elements are configured to receive ultrasound echoes from the target region.

39. The multi-aperture ultrasound imaging system of claim **38** wherein the control system is coupled to the transmit aperture and the first and second receive apertures, wherein the control system is configured to determine a third time for the waveform to propagate from the first point source to a first pixel location in the target region to the third receiving element, and is configured to determine a fourth time for the waveform to propagate from the first point source to the first pixel location in the target region to the fourth receiving element, the control system also being configured to form a second ultrasound image of the first pixel location by combining a third echo received by the third receiving element at the third time with a fourth echo received by the fourth receiving element at the fourth time.

40. The multi-aperture ultrasound imaging system of claim **39** wherein the control system is configured to correct a displacement of the second ultrasound image relative to the first ultrasound image due to speed of sound variation.

41. The multi-aperture ultrasound imaging system of claim **38** wherein the transmit aperture, the first receive aperture, and the second receive aperture are not all in a single scan plane.

* * * * *

专利名称(译)	使用多孔径超声成像的点源传输和声速校正		
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申请号	US13/029907	申请日	2011-02-17
[标]申请(专利权)人(译)	SPECHT DONALD˚F BREWER KENNETHĐ		
申请(专利权)人(译)	SPECHT DONALD˚F BREWER KENNETHĐ		
当前申请(专利权)人(译)	MAUI IMAGING，INC.		
[标]发明人	SPECHT DONALD F BREWER KENNETH D		
发明人	SPECHT, DONALD F. BREWER, KENNETH D.		
IPC分类号	A61B8/00 G01S15/89 G01S7/52 A61B8/08 A61B8/14		
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摘要(译)

多孔径超声成像系统和使用方法具有许多特征。在一些实施例中，多孔径超声成像系统被配置为向单独的物理超声孔径发送超声能量和从单独的物理超声孔径接收超声能量在一些实施例中，多孔径超声成像系统的发射孔径被配置为发射通过目标区域近似第一点源的全方向未聚焦超声波形。在一些实施例中，利用单个接收孔接收超声能量。在其他实施例中，利用多个接收孔接收超声能量。描述了可以组合由一个或多个接收孔接收的回波以形成高分辨率超声图像的算法。额外的算法可以解决声音组织速度的变化，从而允许超声系统几乎可以在身体内或身体上的任何地方使用。