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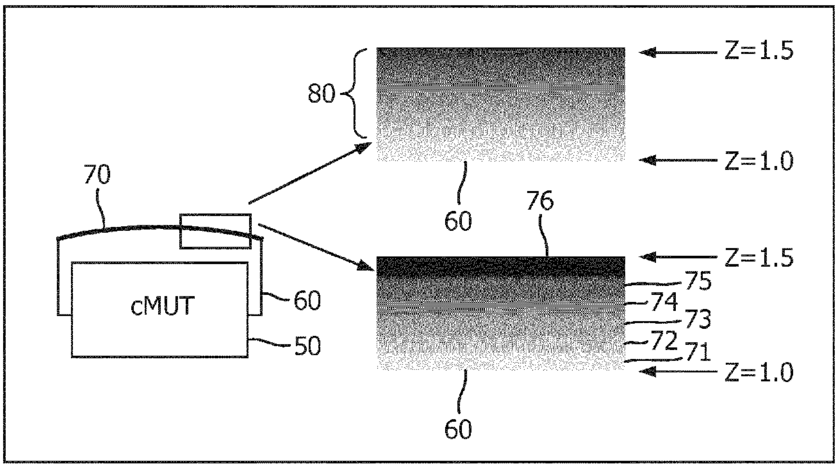

FIG. 7

(57) Abstract: A CMUT transducer array probe has an impedance matching lens with a low impedance silicone elastomer proximal to the CMUT array and providing a majority of the thickness of the lens. The low impedance elastomer is overlaid with a gradated impedance matching layer. The gradated impedance matching layer may provide a progressively increasing impedance from that of the low impedance elastomer to that of tissue, or a stepwise progression of layers of discrete impedances which range from the impedance of the low impedance elastomer to that of tissue.

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CMUT transducer array with impedance matching lens

FIELD OF THE INVENTION

This invention relates to medical diagnostic ultrasonic imaging and, in particular, to ultrasound probes which use capacitive micromachined ultrasonic transducers (CMUTs).

BACKGROUND OF THE INVENTION

The ultrasonic transducers used for medical imaging have numerous characteristics which lead to the production of high quality diagnostic images. Among these are broad bandwidth and high sensitivity to low level acoustic signals at ultrasonic frequencies. Conventionally the piezoelectric materials which possess these characteristics and thus have been used for ultrasonic transducers have been made of PZT, PMN-PT, and PVDF materials, and variants thereof, with PZT being the most preferred. However the ceramic PZT and the single crystal PMN-PT materials require manufacturing processes including dicing, matching layer bonding, fillers, electroplating and interconnections which are distinctly different and complex and require extensive handling, all of which can result in transducer stack unit yields which are less than desired. Accordingly it is desirable to be able to manufacture transducer arrays with improved yields and at lower cost to facilitate the need for low-cost ultrasound systems.

Recent developments have led to the prospect that medical ultrasound transducers can be manufactured by semiconductor processes. Desirably these processes should be the same ones used to produce the circuitry needed by an ultrasound probe such as a CMOS process. These developments have produced micromachined ultrasonic transducers or MUTs. The individual MUT cells can have round, rectangular, hexagonal, or other peripheral shapes. MUTs have been fabricated in two design approaches, one using a semiconductor layer with piezoelectric properties (PMUTs) and another using a diaphragm and substrate with electrode plates that exhibit a capacitive effect (CMUTs). The CMUT transducers are tiny diaphragm-like devices with electrodes that convert the sound vibration of a received ultrasound signal into a modulated capacitance. For transmission the capacitive charge applied to the electrodes is modulated to vibrate the diaphragm of the device and thereby transmit a sound wave. Since these devices are manufactured by semiconductor processes the devices generally have dimensions in the 10-200 micron range, but can range up to device diameters of 300-500 microns. Many such individual CMUTs can be connected together and operated in unison as a single transducer element. For example, four to sixteen CMUTs can be coupled together to function in unison as a single transducer element. A typical 2D transducer array currently will have 2000-10,000 piezoelectric transducer elements. When fabricated as a CMUT array, upwards of 50,000 CMUT cells will be used. Surprisingly, early results have indicated that the yields on semiconductor fab CMUT arrays of this size should be markedly improved over the yields for PZT arrays of several thousand transducer elements.

CMUTs were initially produced to operate in what is now known as an "uncollapsed" mode. Referring to FIGURE 1, a typical uncollapsed CMUT transducer cell 10 is shown in cross-section. The CMUT transducer cell 10 is fabricated along with a plurality of similar adjacent cells on a substrate 12 such as silicon. A diaphragm or membrane 14 which may be made of silicon nitride is supported above the substrate by an insulating support 16 which may be made of silicon oxide or silicon nitride. The cavity 18 between the membrane and the substrate may be air or gas-filled or wholly or partially evacuated. A conductive film or layer 20 such as gold forms an electrode on the diaphragm, and a similar film or layer 22 forms an electrode on the substrate. These two electrodes, separated by the dielectric cavity 18, form a capacitance. When an acoustic signal causes the membrane 14 to vibrate the variation in the capacitance can be detected, thereby transducing the acoustic wave into a corresponding electrical signal. Conversely, an a.c. signal applied to the electrodes 20,22 will modulate the capacitance, causing the membrane to move and thereby transmit an acoustic signal.

The CMUT is inherently a quadratic device so that the acoustic signal is normally the harmonic of the applied signal, that is, the acoustic signal will be at twice the frequency of the applied electrical signal frequency. To prevent this quadratic behavior a bias voltage is applied to the two electrodes which causes the diaphragm to be attracted to the substrate by the resulting coulombic force. This is shown schematically in FIGURE 2, where a DC bias voltage V_B is applied to the device as terminal 24 and is coupled to the membrane electrode 20 by a path which poses a high impedance Z to a.c. signals such as an inductive impedance. A.c. signals are capacitively coupled to and from the membrane electrode from a signal terminal 26. The positive charge on the membrane 14 causes the membrane to distend as it is attracted to the negative charge on the substrate 12. The CMUT cell only weakly exhibits the quadratic behavior when operated continuously in this biased state.

It has been found that the CMUT is most sensitive to received ultrasonic signal vibrations when the membrane is distended so that the two oppositely charged plates of the capacitive device are as close together as possible. A close proximity of the two plates will cause a greater coupling between acoustic and electrical signal energy by the CMUT. Thus it is desirable to increase the bias voltage V_B until the dielectric spacing 32 between the membrane 14 and substrate 12 is as small as can be maintained under operating signal conditions. In constructed embodiments this spacing has been on the order of one micron or less. If the applied bias voltage is too great, however, the membrane can contact the substrate, short-circuiting the device as the two plates of the device are stuck together by VanderWals forces. This sticking can occur when the CMUT cell is overdriven, and can vary from one device to another with the same bias voltage V_B due to manufacturing tolerance variations. While permanent sticking can be reduced by embedding the device electrodes in an electrical isolation layer (*e.g.,* silicon nitride), the nonlinearity of operation between collapsed and uncollapsed states is an inherent disadvantage when trying to operate an uncollapsed CMUT in a range of maximal sensitivity.

A solution to this coulombic sticking problem is to operate the CMUT transducers in a collapsed mode, in which each membrane is intentionally collapsed to the floor of its CMUT cell. With reference to FIGURE 3, a schematic cross-section of a CMUT element or cell 5 is depicted. CMUT cell 5 includes a substrate layer 12, an electrode 22, a membrane layer 14, and a membrane electrode ring 28. In this example, the CMUT cell has a circular shape and electrode 22 is circularly configured and embedded in the substrate layer 12. In addition, the membrane layer 14 is fixed relative to the top face of the substrate layer 12 and configured/dimensioned so as to define a spherical or cylindrical cavity 18 between the membrane layer 14 and the substrate layer 12. As previously mentioned, the cell and its cavity 18 may define alternative geometries. For example, cavity 18 could define a rectangular and/or square cross-section, a hexagonal cross-section, an elliptical cross-section, or an irregular cross-section.

The bottom electrode 22 is typically insulated on its cavity-facing surface with an additional layer (not pictured). A preferred insulating layer is an oxide-nitride-oxide (ONO) dielectric layer formed above the substrate electrode and below the membrane electrode. The ONO-dielectric layer advantageously reduces charge accumulation on the electrodes which leads to device instability and drift and reduction in acoustic output pressure. The fabrication of ONO-dielectric layers on a CMUT is discussed in detail in European patent application no. 08305553.3 by Klootwijk et al., filed September 16, 2008 and entitled "Capacitive micromachined ultrasound transducer." Use of the ONO-dielectric layer is desirable with collapsed CMUTs, which are more susceptible to charge retention than are uncollapsed devices. The disclosed components may be fabricated from CMOS compatible materials, *e.g.*, Al, Ti, nitrides (e.g., silicon nitride), oxides (various grades), tetraethyl orthosilicate (TEOS), poly-silicon and the like. In a CMOS fab, for example, the oxide and nitride layers may be formed by chemical vapor deposition and the metallization (electrode) layer put down by a sputtering process. Suitable CMOS processes are LPCVD, ALD and PECVD, the latter having a relatively low operating temperature of less than 400^{o}C.

Exemplary techniques for producing the disclosed cavity 18 involve defining the cavity in an initial portion of the membrane layer 14 before adding a top face of the membrane layer 14. Other fabrication details may be found in US Pat. 6,328,697 (Fraser). In the exemplary embodiment depicted in FIGURE 3, the diameter of the cylindrical cavity 18 is larger than the diameter of the circularly configured electrode plate 22. Electrode ring 28 may have the same outer diameter as the circularly configured electrode plate 22, although such conformance is not required. Thus, in an exemplary embodiment of the present invention, the electrode ring 28 is fixed relative to the top face of the membrane layer 14 so as to align with the electrode plate 22 below.

FIGURE 4 shows the CMUT cell of FIGURE 3 when biased to a collapsed state, in which the membrane 14 is in contact with the floor of the cavity 18. This is accomplished by applying a DC bias voltage to the two electrodes as indicated by voltage V_B applied to the electrode ring 28 and a reference potential (ground) applied to the substrate electrode 22. While the electrode ring 28 could also be formed as a continuous disk without the hole in the center, FIGURE 4 illustrates why this is not necessary. When the membrane 14 is biased to its collapsed state as shown in this drawing, the center of the membrane is in contact with the floor of the cavity 18. As such, the center of the membrane 14 does not move during operation of the CMUT. Rather, it is the peripheral area of the membrane 14 which moves, that which is above the remaining open void of the cavity 18 and below the ring electrode. By forming the membrane electrode 28 as a ring, the charge of the upper plate of the capacitance of the device is located above the area of the CMUT which exhibits the motion and capacitive variation when the CMUT is operating as a transducer. Thus, the coupling coefficient of the CMUT transducer element is improved.

The membrane 14 may be brought to its collapsed state in contact with the floor of the cavity 18 as indicated at 36 by applying the necessary bias voltage, which is typically in the range of 50-100 volts. As the voltage is increased, the capacitance of the CMUT cell is monitored with a capacitance meter. A sudden change in the capacitance indicates that the membrane has collapsed to the floor of the cavity. The membrane can be biased downward until it just touches the floor of the cavity as indicated at 36, or can be biased further downward to increased collapse beyond that of minimal contact.

Another way to bring the membrane 14 to its collapsed state is to apply pressure to the top of the membrane. When the cavity is formed in a partial or complete vacuum, it has been found that the application of atmospheric pressure of 1 Bar is sufficient to precollapse the membrane 14 to contact with the floor of the cavity 18. It is also possible to use a combination of pressure differential and bias voltage to controllably precollapse the membrane 14, which is effective with smaller devices that may have a high atmospheric collapse pressure (*e.g.*, 10 Bar.) Once the membrane has been collapsed, it can be maintained in that state during operation by the bias voltage V_B or by physical means such as by forming a retention member such as a lens cast on top of the collapsed membrane which retains the membrane in its collapsed state.

When a CMUT array is fabricated for use in a transducer probe, it is necessary to form a biocompatible nonconductive outer cover over the array for several reasons. One is to insulate the patient from contact with the high voltage bias V_B applied to the membrane electrodes on top of the array. Another is to protect the array from damage during use. Such a cover is generally referred to as a lens, regardless of its refractive acoustic properties. A typical cover material is RTV silicone with filler particles added to increase the impedance of the lens to approximate the impedance of tissue. However such a high impedance lens can detract from one of the major advantages of CMUT transducers, which is their broad bandwidth performance. The typical RTV lens will attenuate higher acoustic frequencies, limiting the probe bandwidth and its broadband performance. This characteristic is particularly disadvantageous when the CMUT array is designed to operate over a broad range of center frequencies and frequency bands by changing the bias voltage V_B as described in international patent publication WO 2015/028949 (Davidsen et al.) Accordingly it is desirable to provide a lens for a CMUT array probe which does not detract from the broadband performance of the array and enables a broad band of operation without significant attenuation of higher frequencies.

SUMMARY OF THE INVENTION

Accordingly, it is an object of the present invention to provide a lens for a CMUT transducer array which enables broadband performance.

It is a further object of the present invention to provide a lens for a CMUT transducer array which provides impedance matching between the CMUT array and tissue.

In accordance with the principles of the present invention, an impedance matching lens for an ultrasonic transducer CMUT cell array is provided with a low impedance silicone elastomer proximal to the array and one or more gradated impedance matching layers (gradient transition layer) distal to the array with transition from the low impedance of the silicone elastomer to approximately the impedance of tissue. The CMUT array is designed to operate at least at a first center frequency of a transmitted ultrasound wave and the silicone elastomer has a first thickness and forms a majority of the thickness of the lens. The first thickness layer closely matches the array impedance, while the impedance matching layer or layers having a second overall thickness provide the remaining lens thickness. The second thickness of the gradient transition layer is an irrational number of the wavelength of the first center frequency.

In contrast to commonly used in ultrasound imaging PZT transducers, whose acoustic impedance is considerably larger than the acoustic impedance of the tissue, the CMUT transducers exhibit acoustic impedance (about 1.0 MRayl) lower than the tissue's impedance (about 1.6 MRayl). The impedance matching lens of the present invention leverages this reduced impedance difference between the CMUT array and the tissue to its advantage. The lens of the present invention provides a reduced acoustic wave attenuation, due to the relatively thick low impedance layer of the elastomer having the first thickness. Since the CMUT transducer has a relatively low impedance, the impedance of the elastomer layer does not need to be further matched to the transducer and can be about 1.0 MRayl. A method of an acoustic impedance increase may comprise adding a filler, such as particles (insulating, for example), into elastomer material. The presence of the filler in the elastomer layer introduces additional attenuation of the acoustic wave generated by the array. Therefore, the low impedance elastomer layer, which does not require a filler for its impedance increase (or requires a reduced amount of the filler compared to the PZT application), proximal to the CMUT array provides an improved acoustic wave propagation, due to the efficient acoustic wave coupling to the tissue via the gradient layer(s).

Due to the improved acoustic wave propagation through the elastomer layer of the first thickness, the impedance gradient layer(s) are thus can be very thin compared to the thickness of the low impedance layer. In contrast to the prior art impedance matching lenses used for the PZT arrays, the thickness of each impedance gradient layer in the lens of the present invention does not need to be compared with a quarter (or fraction) of the acoustic wavelength of center frequency generated by the array. The lens of the present invention provides a reduced acoustic wave attenuation, due to the relatively thick low impedance layer, combined with improved acoustic wave propagation, due to the efficient acoustic wave coupling to the tissue via the gradient layer(s). In another embodiment the CMUT array is further arranged to operate at least at a second center frequency of the transmitted ultrasound wave, wherein the second thickness of the gradient layer is an irrational number of the wavelength of the second center frequency.

The impedance matching lens enables the CMUT array to operate over a broad range of center frequencies, which includes at least the first center and the second center frequency. Due to the innovative lens construction the gradient layer thickness does not need to be compared with the fraction of the acoustic wavelength of any center frequency generated by the array.

BRIEF DESCRIPTION OF THE DRAWINGS

In the drawings:

FIGURE 1 is a cross-sectional view of a typical CMUT transducer cell.

FIGURE 2 is a schematic illustration of the electrical properties of a typical CMUT cell.

FIGURE 3 is a cross-sectional view of a CMUT cell when operated in the uncollapsed state.

FIGURE 4 is a cross-sectional view of a CMUT cell when operated in the collapsed state.

FIGURE 5 illustrates a typical transducer probe acoustic stack with matching layers.

FIGURE 6 illustrates a CMUT array with an impedance matching lens constructed in accordance with the principles of the present invention.

FIGURE 7 illustrates the distal impedance gradient layer of the lens of FIGURE 6 in greater detail.

DETAILED DESCRIPTION OF THE EMBODIMENTS

Referring now to FIGURE 5, a transducer probe acoustic stack 100 is shown schematically. A piezoelectric layer 110 such as PZT and two matching layers 120, 130 bonded to the piezoelectric layer are diced by dicing cuts 75 to form an array 170 of individual transducer elements 175, four of which are seen in FIGURE 5. The transducer array 170 may comprise a single row of transducer elements (a 1-D array) or is a piezoelectric plate diced in two orthogonal directions to form a two-dimensional (2D) matrix array of transducer elements. The array 170 may also comprise a one or two dimensional array of micromachined ultrasound transducer (MUTs) formed on a semiconductor substrate by semiconductor processing. The matching layers match the acoustic impedance of the piezoelectric material or MUTs to that of human tissue, generally in steps of progressive matching layers. In this example the first matching layer 120 is formed as an electrically conductive graphite composite and the second matching layer 130 is formed of a polymer loaded with electrically conductive particles. A ground plane 180 is typically formed by gold sputtered polyester film (AKA PET or Polyester terephthalate)coupled to the second matching 130 layer and located in between the second matching layer and a third matching layer 150 made of polyolefin or Pebax resin. The third matching layer is a final matching layer of an acoustic window 140 of the stack. The ground plane is electrically coupled to the transducer elements through the electrically conductive matching layers and is connected to a ground conductor of flex circuit 185. Additionally, the array may have an outermost layer (not shown) made of an RTV or polyolefin, said layer provides the acoustic window 140 with additional mechanical resistance against an impact. The three matching layers transition in steps from the impedance of the transducer elements 175 to the impedance of tissue, about 1.3 to 2.0 MRayl preferably to 1.6 MRayl.

Below the transducer elements is an integrated circuit 160, an ASIC, which provides transmit signals for the transducer elements 175 and receives and processes signals from the elements. Conductive pads on the upper surface of the integrated circuit 160 are electrically coupled to conductive pads on the bottoms of the transducer elements by stud bumps 190, which may be formed of solder or conductive epoxy. Signals are provided to and from the integrated circuit 160 by connections to the flex circuit 185. Below the integrated circuit 160 is a backing block 165 which attenuates acoustic energy emanating from the bottom of the transducer stack. The backing block also conducts heat generated by the integrated circuit away from the integrated circuit and the transducer stack and away from the patient-contacting distal end of the transducer probe.

FIGURE 6 illustrates a CMUT array 50 with an impedance matching lens 60 constructed in accordance with the principles of the present invention. The CMUT array 50 is designed to operate over a broad range of center frequencies and frequency bands by changing the bias voltage V_B as described in international patent publication WO 2015/028949 (Davidsen et al.) This broad range of center frequencies includes at least a first center and a second center frequencies, said frequencies being different from each other. The lens comprises a relatively thick layer 60 of a low impedance elastomer having a first thickness, such as an unfilled silicone elastomer. Since CMUTs exhibit a different electro-acoustical transformation mechanism than conventional PZT, interactions between the CMUT membranes and the acoustic materials overlaying the CMUT array impose different requirements for an impedance matching lens than does PZT. The low impedance layer needs to mechanically adapt its inner surface to the displacement of the CMUT membranes. The relative low hardness of the elastomers (preferably below 50 ShoreA for soft or uncured elastomers) provides a desirable acoustic contact between the lens and the CMUT array. Suitable elastomers for most implementations include silicon based elastomers, polydimethylsiloxane (PDMS) and polybutadiene. The elastomer preferably exhibits a frequency characteristic which is relatively flat with increasing central frequency, meaning that the elastomer causes relatively little attenuation of higher frequencies relative to that of lower frequencies. An unfilled silicone elastomer will typically have an impedance of about 1.0 MRayl, closely matching the impedance of an array of CMUT cells. Overlaying this low impedance layer 60 is one or more transition gradient layers 70 having an overall thickness being a second thickness, said layer forms a transition from the relatively low impedance of the elastomer layer 60 to the impedance of tissue. The impedance value of the elastomer of about 1.0 MRayl is low compared to the tissue impedance value of about 1.6 MRayl, a difference which is accommodated by the transition gradient layer(s). The acoustic impedance of the gradient layer 70 gradually varies throughout its thickness from the low impedance value at its surface facing the low impedance layer 60 to an increased impedance value at its outer edge facing the tissue, wherein the increased impedance value is about the same as the tissue's impedance. Thus, the acoustic impedance value of the gradient layer 70 gradually increases with its thickness and can be described as a positive gradient. The impedance gradient transition layer of present invention exhibits the positive gradient of the acoustic impedance variation with respect to its thickness increase.

It is seen that the low impedance elastomer layer comprises most of the thickness of the lens in this example. In a constructed implementation the low impedance elastomer will comprise at least a majority of the thickness of the lens, and preferably will exhibit a thickness ratio to the thickness of the impedance gradient layer(s) of at least 5:1 and most preferably at least 9:1. The impedance gradient layers are thus very thin compared to the thickness of the low impedance layer. For example, the thickness of the low impedance layer can be about 900 micrometer and a combined thickness of the transition gradient layers can be about 100 micrometer. In contrast to the prior art impedance matching lenses used for the PZT arrays, the thickness of each impedance gradient layer in the lens of the present invention does not need to be compared with a quarter (or a fraction) of the acoustic wavelength. The overall thickness of the impedance gradient transition layer equals to the irrational number of the wavelength of the center frequency, at which the array is designed to operate. The lens of the present invention provides a reduced acoustic wave attenuation, due to the relatively thick low impedance layer, combined with improved acoustic wave propagation, due to the efficient acoustic wave coupling to the tissue via the gradient layer(s). The thickness ratio of the first thickness of the elastomer layer to the second thickness of the impedance gradient layer(s) being at least 5:1 and most preferably at least 9:1 provides a beneficial range of the ratios. In this embodiment potential losses in acoustic wave intensity (propagating through the impedance matching lens overlaying the CMUT array) due to its reflection at the interface(s) in between the layers are equilibrated by a minimized attenuation of the acoustic wave provided by the low impedance elastomer layer, wherein said low impedance layer forms the majority of the thickness of the impedance matching lens.

FIGURE 7 illustrates two types of impedance gradient layers, one providing a smooth gradient transition from the low impedance layer to tissue impedance and another providing a gradient transition in discrete steps. The upper enlarged view in the drawing illustrates a smooth gradient transition layer 80, transitioning from 1.0 MRayl at the bottom to 1.3-2.0 MRayl at the top. The lower enlarged view illustrates six layers 71-76, each of a progressively increasing impedance to step from 1.0 MRayl at the bottom to 1.5 MRayl at the top. A typical final impedance at the top will be 1.5 to 1.6 MRayl in most applications. The smoothly progressing gradient layer and the discrete impedance step layers can be formed by filling a silicone based resin with filler particles such as a metal oxide powder. Suitable particles include aluminum oxide, silicon dioxide, or iron oxide in the 1 to 5 micron range. The impedance of the resin is increased by using a greater concentration of filler particles. The hardness of the material increases with an addition of filler. By reducing the relative thickness of the gradient layer(s) compared to the thickness of the low impedance layer an implementation of the present invention provides an improved acoustic lens for the CMUT application. Typical construction of the step-wise impedance gradient layers 71-76 comprises three (or five) to ten layers with the thickness of each layer in the range of 20um to 50um. In a constructed embodiment the thickness of the low impedance layer was about 1 mm and the overall (combined) thickness of the impedance gradient layers was about 0.1 mm. If desired the impedance gradient layer 71-76 or 80 may be coated with a thin protective polyolefin or polymethylpentene (TPX) layer on the order of the same thickness as one of the gradient layers. This impedance variation provides an optimal impedance matching for the acoustic wave propagating from the CMUT array 50 to the patient’s body through the acoustic lens 60, 70.

It shall be noted that in all disclosed embodiment the first thickness of low impedance elastomer layer does not need to be compared with a quarter (or a fraction) of the acoustic wavelength. The first thickness can be equal to any irrational number of the wavelength of the center frequency generated by the array.

CLAIMS:

1. An ultrasonic CMUT array probe comprising:

an array of CMUT cells, said array designed to operate at a first center frequency of an ultrasound wave; and

an impedance matching lens overlaying the CMUT array, the lens having a thickness of which a majority comprises a low impedance elastomer having a relatively large first thickness and being in contact with the CMUT array, and a minority comprises an impedance gradient transition layer, having a relatively low second thickness, providing an impedance transition from that of the low impedance elastomer to that of tissue, wherein the second thickness is an irrational number of the wavelength of the first center frequency.

2. The ultrasonic CMUT array probe of Claim 2, wherein the array is further designed to operate at a second center frequency and the second thickness an irrational number of the wavelength of the second center frequency.

3. The ultrasonic CMUT array probe of Claim 1 or 2, wherein the ratio of the thickness of the low impedance elastomer to that of the impedance gradient transition layer is at least 5:1.

4. The ultrasonic CMUT array probe of Claim 3, wherein the ratio of the thickness of the low impedance elastomer to that of the impedance gradient transition layer is at least 9:1.

5. The ultrasonic CMUT array probe of Claim 4, wherein the thickness of the low impedance elastomer is about 1 mm and the thickness of the impedance gradient transition layer is about 0.1 mm.

6. The ultrasonic CMUT array probe of Claim 1, wherein the impedance gradient transition layer further comprises a plurality of layers, each exhibiting a discrete impedance and layered in order of progressively greater impedance.

7. The ultrasonic CMUT array probe of Claim 6, wherein the plurality of layers further comprises five to ten layers.

8. The ultrasonic CMUT array probe of Claim 6, wherein the lowest layer from the plurality of layers is in contact with the low impedance elastomer and exhibits an impedance of about 1.0 MRayl, and the uppermost layer exhibits an impedance of 1.5-1.6 MRayl.

9. The ultrasonic CMUT array probe of Claim 6, wherein each layer from the plurality exhibits a thickness which is in the range of 20 um to 50um.

10. The ultrasonic CMUT array probe of Claim 6, wherein each layer is loaded with filler particles.

11. The ultrasonic CMUT array probe of Claim 10, wherein the filler particles further comprise aluminum oxide, silicon dioxide, or iron oxide.

12. The ultrasonic CMUT array probe of Claim 11, wherein the filler particles exhibit a size ranging from 1 to 5 microns.

13. The ultrasonic CMUT array probe of Claim 1 or 2, wherein the impedance gradient transition layer further comprises a single layer exhibiting a gradient transition from that of the low impedance elastomer to that of tissue.

14. The ultrasonic CMUT array probe of Claim 13, wherein the single layer further exhibits a density of filler particles ranging from relatively low at a surface facing the low impedance layer to relatively high at its thickness most remote from the low impedance elastomer layer.

15. The ultrasonic CMUT array probe of Claim 13, wherein the single layer further exhibits an impedance of about 1.0 MRayl near the low impedance elastomer to about 1.5 MRayl at its thickness most remote from the low impedance elastomer.

16. The ultrasonic CMUT array probe of Claim 1 wherein the low impedance elastomer further comprises one of polydimethylsiloxane (PDMS) or polybutadiene.

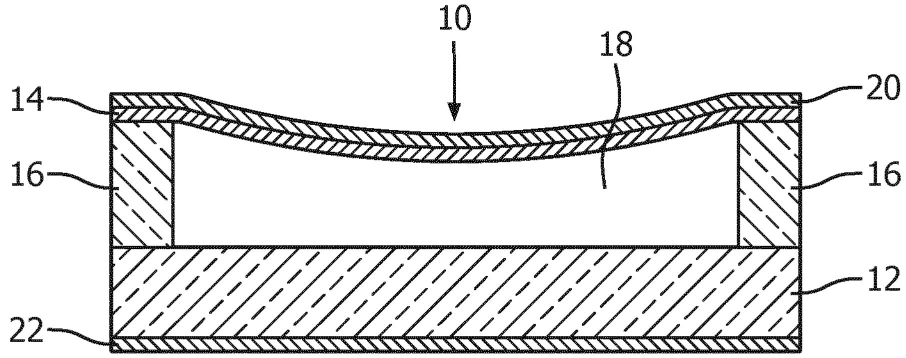

FIG. 1

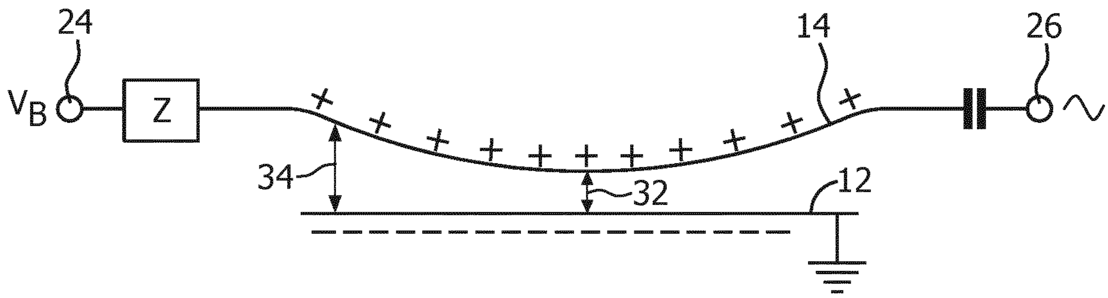

FIG. 2

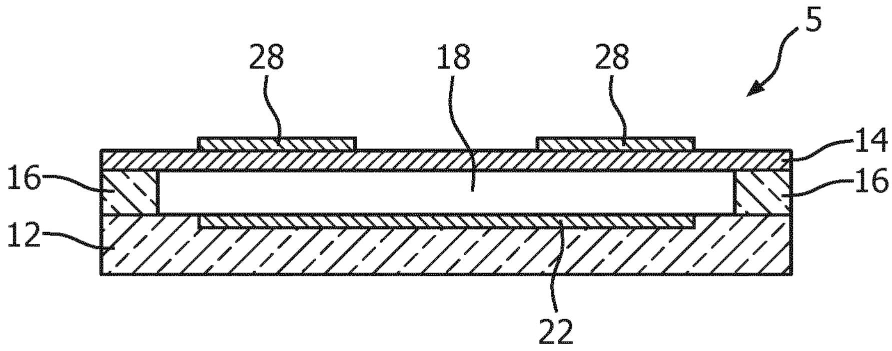

FIG. 3

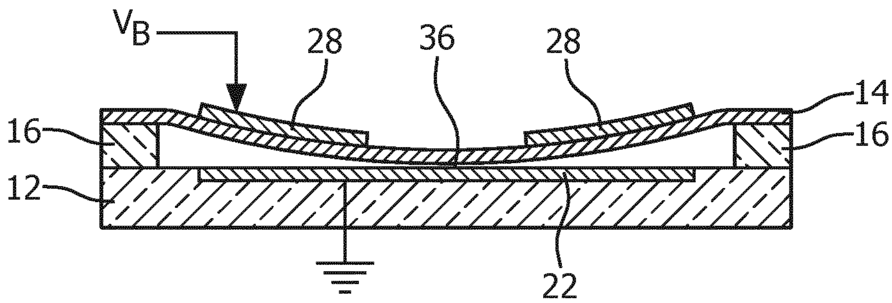

FIG. 4

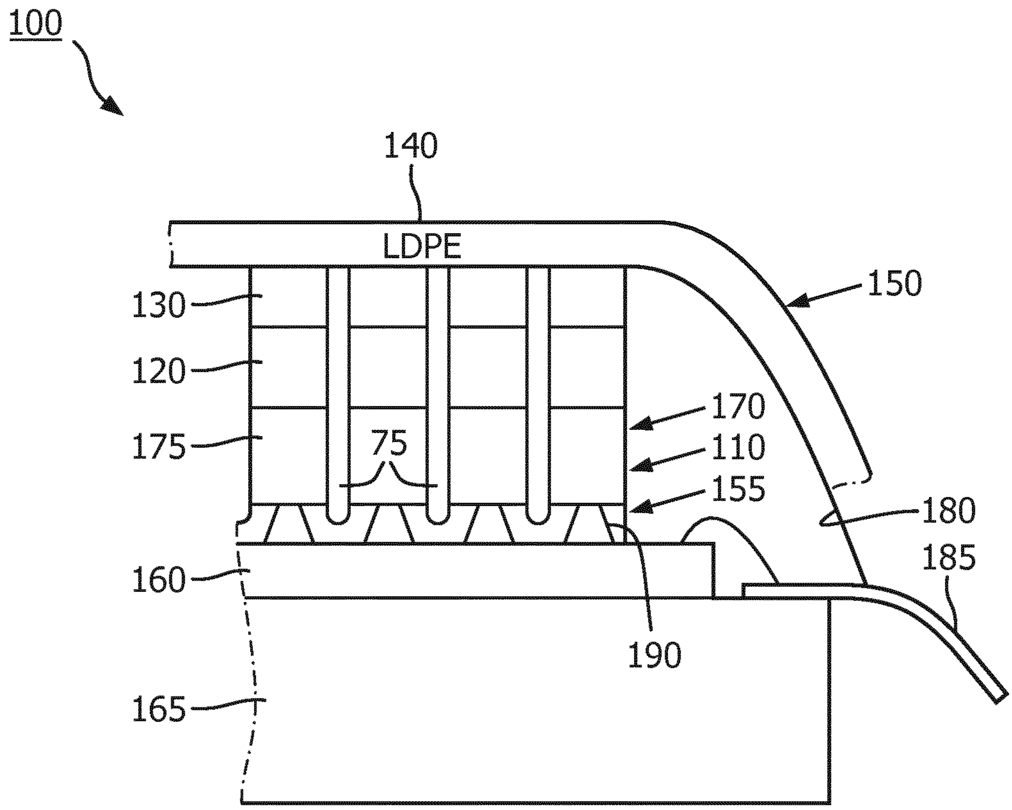

FIG. 5
Prior art

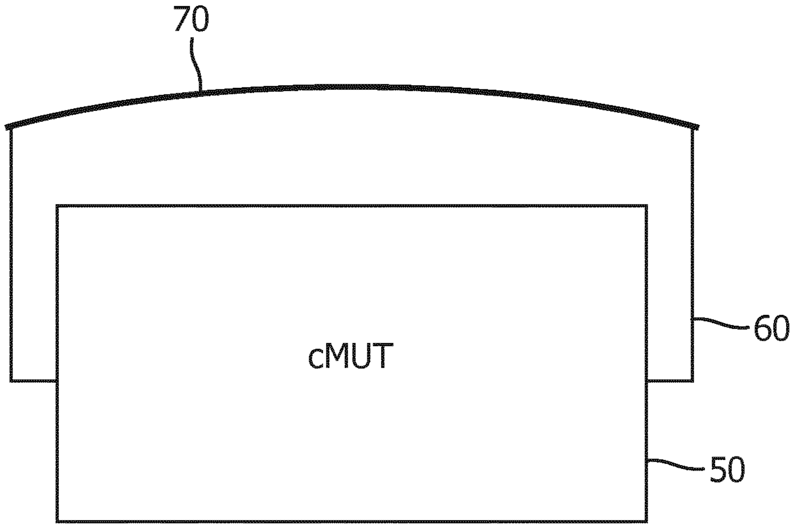
70
cMUT
60
50

FIG. 6

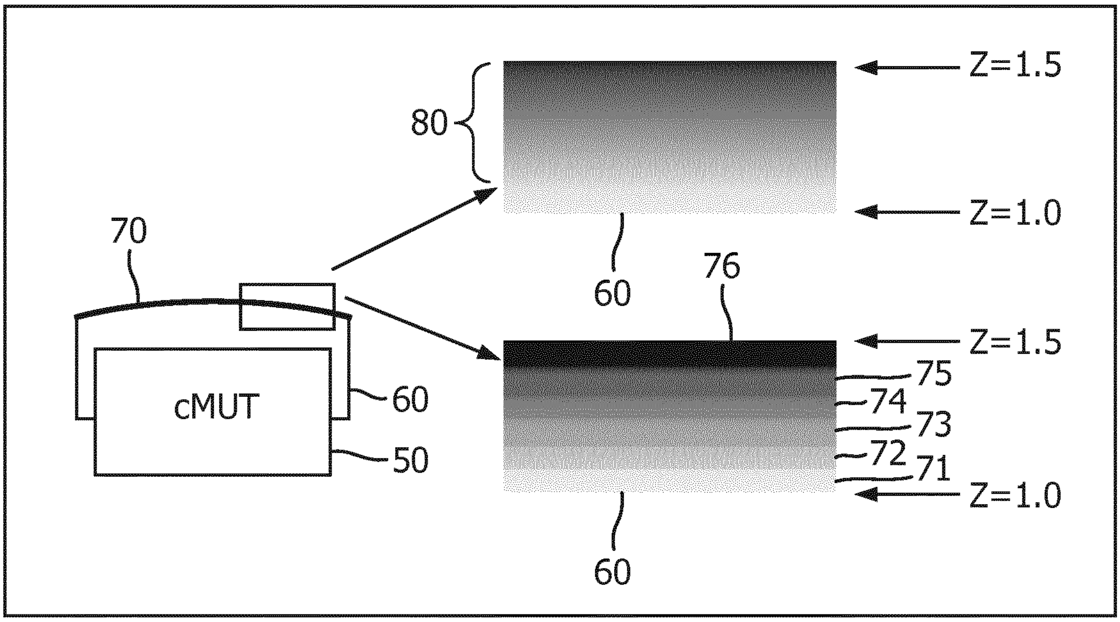
80
Z=1.5
Z=1.0
70
60
76
Z=1.5
75
74
73
cMUT
60
72
71
50
Z=1.0
60

FIG. 7

INTERNATIONAL SEARCH REPORT

International application No
PCT/EP2017/059935

A. CLASSIFICATION OF SUBJECT MATTER
INV. B06B1/02 G10K11/02 A61B8/00 G10K11/30
ADD.

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
B06B A61B G10K

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

EPO-Internal, WPI Data

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X Y	US 2013/301394 A1 (CHEN LI [US] ET AL) 14 November 2013 (2013-11-14) figure 18 paragraph [0056] paragraph [0051] - paragraph [0052] paragraph [0075] - paragraph [0076] paragraph [0099] - paragraph [0100] claims 20, 23 paragraph [0036] -----	1-6,8, 15,16 7,9-14
Y A	US 2014/265728 A1 (LI WEI [US] ET AL) 18 September 2014 (2014-09-18) paragraph [0024] paragraph [0029] paragraph [0020] -----	10-12,14 6
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[X] Further documents are listed in the continuation of Box C.

[X] See patent family annex.

* Special categories of cited documents :

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"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
8 June 2017	19/06/2017

Name and mailing address of the ISA/	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL - 2280 HV Rijswijk Tel. (+31-70) 340-2040, Fax: (+31-70) 340-3016	Pitzer, Hanna

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INTERNATIONAL SEARCH REPORT

International application No
PCT/EP2017/059935

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

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