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A Tactile Sensor for Ultrasound Imaging Systems

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Abstract

Medical ultrasound systems are capable of monitoring a variety of health conditions while avoiding invasive procedures. However this function is complicated by ultrasound contrast of the tissue varying with contact pressure exerted by the probe. The knowledge of the contact pressure is beneficial for a variety of screening and diagnostic procedures involving ultrasound. This paper introduces a solid-state sensor array which measures the contact pressure distribution between the probe and the tissue marginally affecting the ultrasound imaging capabilities. The probe design utilizes the dielectrostriction mechanism which relates the change in dielectric properties of the sensing layer to deformation. The concept, structure, fabrication, and performance of this sensor array are discussed. The prototype device is highly tolerant to overloads (>1 MPa tested) and provides stress measurements in the range of 0.14 to 10 kPa. Its loss of ultrasound transmissivity is less 3dB at 9 MHz ultrasound frequency. This performance is satisfactory for clinical and biomedical research in ultrasound image formation and interpretation, however for commercial product, a higher ultrasound transmissivity is desired. Directions for improving the sensor ultrasound transparency and electrical performance are discussed. The sensor array described in this paper has been developed specifically for ultrasound diagnosis during breast cancer screening. However, the same sensing mechanism, similar configuration and sensor array structure can be applied to other applications involving ultrasound tools for medical diagnostics.

Keywords

Tactile array; Ultrasound diagnosis; Dielectrostriction; Stress/strain-dielectric response; Planar capacitor sensor

I. INTRODUCTION

Ultrasound waves are emitted by a (typically piezoelectric) transducer and propagate into the tissue. Inhomogeneities in acoustic impedance within tissue cause the ultrasound wave to scatter and/or reflect. The echo signals are detected and displayed as an image with varying

shades of gray relative to their amplitude, where higher scattering amplitudes usually appear as lighter shade of gray on the image [1].

During an ultrasound scan, an ultrasound transducer is in contact with the patient body. In many imaging tasks there are irregular layers of tissue and structures, such as ligaments, that interfere with wave propagation. Transmission of the acoustic wave through these layers improves when contact pressure with the transducer is increased to the point that the layers become (reasonably) planar and perpendicular to the direction of wave propagation. Consequently, ultrasound image brightness and contrast is strongly affected by the pressure between the ultrasound probe and the tissue. Knowledge of the contact pressure between the probe and the tissue is beneficial for learning ultrasound imaging techniques and performing repeatable screening and diagnostic tasks. Contact pressure information is useful in ultrasound elastography for breast cancer screening [2, 3] to aid in reproducibility and quantitative accuracy: this is the application which motivates the development reported in this paper. Tissues exhibit nonlinear elastic behavior, therefore the shear modulus varies with deformation beyond a linear elastic region. The (quasi-static) elastography testing procedure involves comparison of ultrasound echo signals obtained before and after applying a compressive force to the tissue by the ultrasound probe. Pre-and post-loading echo signals are compared to estimate the displacement fields in tissue and, potentially, for images of relative tissue strain. The distribution of contact pressure between the probe and the breast tissue can be combined with the deformation data to provide a distribution of the elastic modulus in the tissue. Performing this data acquisition and modulus reconstruction over relatively large deformations (20% strain) allows estimating the elastic nonlinearity of tissues. Malignant lesions tend to be stiffer than benign lesions, and malignant lesions have higher elastic nonlinearity than normal and benign tissues [4-6]. The reconstruction of tissue modulus and elastic nonlinearity can help classify the lesion as benign or malignant and, therefore, reduce the unnecessary biopsies. Besides benefiting modulus reconstruction, the knowledge of contact pressure is also helpful for comparison of repeat (visit-to-visit) images, telemedicine, sonographer assistance, and monitoring workload of sonographers for reducing risk of musculoskeletal injury.

To provide data on contact pressure between an ultrasound probe and tissue surface, an array of sensors operates between the patient body and the ultrasound probe. This dictates technical requirements for the contact pressure sensor array:

- (a) good ultrasound transparency,
- (b) high pressure sensitivity,
- (c) recording time-dependent loads with good temporal resolution and stability (0.1 to 50 sec),
- (d) large dynamic range, and tolerance to overload.

In addition, the materials of the sensor should be biocompatible, and the excitation voltage should be low. Further, minimal bending of the pressure sensor array is desirable to avoid stray sources of pressure-dependent signals.

Flexible and semi-flexible pressure arrays are currently produced by a number of companies. These products can be divided into several groups based on the physical approaches, dynamic ranges, tolerance to overloads, and energy efficiency in wearable implementations. The following is a brief summary of some typical sensing solutions.

Pressure indicating films change color under the applied pressure: Typical products in this category are Fuji Prescale® film (Tekscan [7], Sensor Products [8]) and Pressurex-micro® (Sensor Products [8]). However, such films detect pressures and deformations in a narrow load range and are not suitable for time dependent and periodic testing applications.

Piezoresistive pressure arrays use a layer of semiconductor material or a composite of conductive inclusions at the percolation limit. The resistance between two electrodes deposited on opposite sides of such layer is proportional to the applied pressure. This approach benefits from a simple electronic interface and the ability to have long wires between the sensor and a data acquisition (DAQ) unit. However, such sensors have low dynamic range of the detected pressure, high hysteresis for cycling loads, significant temperature drift and electric noise. Piezoresistive sensors dissipate electric energy which limits service life in wearable and portable implementations and requires a correction for temperature drift due to self-heating. There is also a very limited selection of available sensing materials. Typical products of this type are sensors manufactured by Tekscan [7].

Resistive strain gages arrays utilize a thin film gage deposited on a deformable substrate. In-plane deformations of the substrate are detected through a change of the sensor resistance. Resistive strain gages are the industrial standard for detection of deformations and provide good sensitivity and temperature stability with proper selection of the gage material [9]. There are many well established and readily available off-the-shelf data acquisition (DAQ) solutions for sensing these signals. However, conversion of a pressure load to an in-plane deformation may be challenging – a typical approach involves a set of micro cells where the sensor substrate works as a membrane, and this would not be compatible with ultrasound imaging. Another critical limitation is the low detectable range of deformation which should stay within the elastic limits of the gage material, sensor substrate, adhesive layer and the membrane assembly. The energy efficiency of the resistive strain gages is also low. Typical products of this type are manufactured by Strain Measurement Devices [10] and Sensor Products [8].

Fiber optic pressure arrays detect deformations of optical fibers [9, 11]. The advantage of this approach is their high resistance to electromagnetic noise, high temperature stability, and the capability of monitoring several data nodes with a single fiber. There are many established DAQ methods for such pressure arrays, though they are quite expensive. Limitations of this sensing approach are a narrow range of deformation sensitivity, high cost of manufacturing and assembly, and low pressure and spatial resolutions. Optical fiber arrays would scatter the ultrasound beam and cannot be used with ultrasound probes. Typical products of this type are manufactured by Sensing Tex [12].

Capacitive pressure arrays have electrodes deposited on opposite sides of a sensing layer [13-15]. The advantage of this approach is its high sensitivity to the pressure signals and

high energy efficiency in portable implementations. A typical implementation of this sensor requires an air-gap between the elastic spacers. This implementation has low dynamic range of the detected signal and sensing elements have multiple cavities which would scatter the ultrasound beam. Typical products of this type are manufactured by Pressure Profile Systems [16].

Dielectrostriction pressure array is introduced in this paper. It employs solid-state technology and therefore has no features which scatter the ultrasound beam, provides large dynamic range and can withstand large overloads. The sensing mechanism is based on the dielectrostriction effect, that is, the contact pressure is determined by a change in the dielectric constant of the sensing layer [17-19]. The sensor implementation presented here employs an interdigitated configuration of electrodes deposited on a single plane. Compared to traditional parallel-plate sensors, the planar configuration of interdigitated electrodes simplifies sensor manufacturing, eliminates errors arising from the potential misalignment of two plates, and enables the sensor to be easily accessed at any desired location. Electronic and DAQ techniques for the dielectrostriction sensing approach are similar to capacitive sensors and are readily available.

In the following sections the dielectrostriction phenomenon and stress-dielectric relation are introduced. The concept of a dielectrostrictive sensor is discussed. Then the configuration and structure of sensor array are described, followed by the sensor fabrication and performance. Although in this paper an electrode pattern and the sensor structure are designed specifically for ultrasound diagnosis during breast cancer screening, the same sensing mechanism can be applied, and the configuration and structure of the sensor can be easily modified, to other applications involving ultrasound tools for screening and diagnosis.

II. Background

A. Dielectrostriction and stress-dielectric relation

Dielectrostriction is defined as the variation of dielectric properties of a material with deformation, and is a fundamental property of any dielectric material [17, 21, 22]. Deformation affects the relative positions of the dipoles and the local electric field of the material which leads to a change in dielectric properties.

The dielectric properties of a deformed isotropic material are described by a second order tensor, ε_{ij} , which can be approximated as a linear function of the strain tensor, u_{ij} , [22, 23],

$$\Delta\varepsilon_{ij} = \varepsilon_{ij} - \varepsilon_0 \delta_{ij} = \alpha_1 u_{ij} + \alpha_2 u_{ii} \delta_{ij} \quad (1)$$

where α_1 and α_2 are strain-dielectric coefficients, $u_{ii} = u_{11} + u_{22} + u_{33}$ and δ_{ij} is the Kronecker delta function. Small variations of dielectric properties with deformation are required to justify the assumption of linearity in Eq.(1) namely,

$$\|\alpha_1 u_{ik}\| \ll \varepsilon, \quad \text{and} \quad |\alpha_2 u_{ii}| \ll \varepsilon, \quad (2)$$

where ε is the dielectric constant of the initially isotropic material. By using a local field model, Shkel and Klingenberg [21] derived strain-dielectric coefficients, α_1 and α_2 , for a mixture with randomly distributed rigid inclusions under affine deformation,

$$\alpha_1 = -\frac{2}{5} \frac{(\varepsilon_{mix} - \varepsilon_c)^2}{\varepsilon_c}, \quad \text{and} \quad \alpha_2 = -\frac{1}{3} \frac{(\varepsilon_{mix} - \varepsilon_c)(\varepsilon_{mix} + 2\varepsilon_c)}{\varepsilon_c} + \frac{2}{15} \frac{(\varepsilon_{mix} - \varepsilon_c)^2}{\varepsilon_c}, \quad (3)$$

where ε_{mix} and ε_c are the relative dielectric constants of the mixture and the continuum matrix medium, respectively. A pure elastic solid, which can be treated as a mixture of numerous polarizable molecules dispersed in a vacuum ($\varepsilon_c = 1$), is a common example of the above-mentioned material systems.

In the stress-dielectric study of polymers, it has been found that a linear relationship exists between the change of dielectric constant and the stress under small deformations for both solid and liquid polymers [17, 19, 24], as both the polarizability tensor and stress tensor are strongly related to the distribution of the end-to-end vectors of a material system. A stress-dielectric relation can be formulated similar to the strain-dielectric relation as,

$$\Delta\varepsilon_{ij} = \lambda_1 \sigma_{ij} + \lambda_2 \sigma_{ll} \delta_{ij}, \quad (4)$$

where λ_1 and λ_2 are stress-dielectric coefficients, σ_{ij} is a stress component, $\sigma_{ll} = \sigma_{11} + \sigma_{22} + \sigma_{33}$ and δ_{ij} is the Kronecker delta function.

Strain-based and stress-based formulations of dielectrostriction are equivalent for materials in the linear elastic region: strain-dielectric coefficients, α_1 , α_2 , and stress-dielectric coefficients, λ_1 , λ_2 can be easily converted to each other, since Hooke's law provides a linear relationship between strains and stresses:

$$\lambda_1 = \frac{1+\nu}{E_Y} \alpha_1, \quad \text{and} \quad \lambda_2 = -\frac{\nu}{E_Y} \alpha_1 + \left(\frac{1-2\nu}{E_Y} \right) \alpha_2, \quad (5)$$

where E_Y is the Young's modulus and ν is the Poisson's ratio. In the non-linear stress-strain region, the strain-based and the stress-based descriptions of dielectrostrictive response cannot easily be converted to each other. They therefore should be treated independently and may be related to each other only for some given deformation modes.

B. Planar capacitive sensor

A planar sensor with interdigitated electrodes deposited on a non-conductive substrate is described below. The sensing layer is a dielectric material located in close proximity to the electrodes. Deformation changes the dielectric constant of the sensing layer, which is characterized by the change in the sensor capacitance. A detailed examination of the rheological and electronic aspects of dielectrostriction measurements using planar capacitive sensor is discussed in Refs. [18, 19, 24-26]. The following gives a brief description of the concept and dielectrostrictive response of this sensor under a compressive loading.

A planar capacitive sensor is formed by interdigitated electrodes (having equal width, $W=2w$, and being separated by a distance, $A=2a$) which is attached to a dielectric material,

as shown in Fig. 1. Both the thickness of the dielectric material, h , in Fig. 1, and the length of the electrode, l , are much larger than the electrode width and separation ($h, l \gg W, A$). The electrodes in Fig. 1 are located in the xy -plane and form an angle, θ , with respect to the y -axis. The capacitance, C_θ , of such a planar sensor with an isotropic material on top is

$$C_\theta = \frac{1}{2} C_0 (\varepsilon + \varepsilon_s) \quad (6)$$

where ε and ε_s are the dielectric constants of the dielectric material and the sensor substrate, respectively. The value of C_0 can be obtained experimentally or estimated as $C_0 = 2\varepsilon_0 L \ln(1 + w/a)/\pi$, where parameter L is the combined length of all electrodes, and C_0 represents the capacitance of the electrodes in free space. In this study, the width $2w$ and the spacing $2a$ are the same, so $C_0 = 2\varepsilon_0 L \ln 2/\pi$.

Variation of dielectric constant of a dielectric layer subjected to a compressive strain, u_{zz} , depends on constraints on the layer surfaces. If the material is free to expand, then the other non-zero strain components are lateral expansion strains in x and y direction, $u_{xx} = u_{yy} = -\nu u_{zz}$. The capacitance of the planar sensor under small compressive deformations becomes,

$$C_\theta = C_0 \left[\varepsilon + \varepsilon_s + \frac{(1-\nu)\alpha_1 + 2(1-2\nu)\alpha_2}{2} u_{zz} \right] \quad \text{in strains,} \quad (7)$$

and

$$C_\theta = C_0 \left[\varepsilon + \varepsilon_s + \frac{(1-\nu)\alpha_1 + 2(1-2\nu)\alpha_2}{2E} \sigma_{zz} \right] \quad \text{in stresses.}$$

If the material is fully constrained, then $u_{xx} = u_{yy} = 0$. a result, the sensor capacitance becomes,

$$C_\theta = C_0 \left[\varepsilon + \varepsilon_s + \frac{\alpha_1 + 2\alpha_2}{2} u_{zz} \right] \quad \text{in strains,} \quad (8)$$

and

$$C_\theta = C_0 \left[\varepsilon + \varepsilon_s + \frac{(1-\nu)(1-2\nu)(\alpha_1 + 2\alpha_2)}{2(1-\nu)E} \sigma_{zz} \right] \quad \text{in stresses.}$$

Eqs. (7) and (8) indicate that the capacitance of the planar capacitive sensor in both situations is a linear function of strain and independent of sensor orientation.

C. Sensitivity of the interdigitated design

The interdigitated design of the pressure sensor has numerous advantages for application in ultrasound imaging systems. Compared with the parallel plate design, electrodes are located on a single substrate which reduces number of layers in the sensor and increases its ultrasound transparency. In addition, the thickness of the sensing layer has little influence on the performance of this pressure sensor once it exceeds 3 times the distance between the

centers of two adjacent electrodes. However, it could be instructional to compare sensitivity to deformation of interdigitated and parallel plate sensors.

Consider the simple case of a parallel plate sensor where two rigid plates are separated by elastic spacers. This design is similar to the one which has been commercialized by Pressure Profile Inc. [16]. The sensor response to pressure-induced strains is

$$\frac{\Delta C}{C} = -\frac{\Delta h}{h} = -u_{zz},$$

where h is the plate separation and u_{zz} is the pressure-induced strain in the elastic spacers.

Similarly, the response of the planar sensor with interdigitated electrodes is given by Eq. (8) which simplifies to

$$\frac{\Delta C}{C} = \frac{\alpha_1 + 2\alpha_2}{2(\varepsilon + \varepsilon_s)} u_{zz}.$$

Eq. (3) provides an estimate for dielectrostriction coefficients which is

$$\alpha_1 + 2\alpha_2 = -\frac{1}{3} \left(\frac{1}{5} (\varepsilon - 1)^2 + (\varepsilon - 1)(\varepsilon + 2) \right).$$

The sensor substrate used in this study was polyimide (PI) with $\varepsilon_s = 3.4$ and the sensing layer was polydimethylsiloxane (PDMS) with $\varepsilon = 2.8$. In this case, sensitivity of interdigitated design is 0.5 times the sensitivity of parallel plate design. For a sensing layer with $\varepsilon = 4.8$, the sensing response of both designs would be equivalent, and with $\varepsilon = 9.7$ the sensing response of interdigitated electrodes would be twice that of an air-gap parallel plate design. For future implementation, we considered a silicone rubber as the sensing material since it has well-established biocompatible properties and has dielectric constants in range of $\varepsilon = 3.2$ -9.8 [27]. The sensitivity of the resulting interdigitated design would range from 0.65 to 2 times of the parallel plate design.

III. Sensor Array: Design, manufacturing, performance

A. Electrode pattern and sensor structure

This sensor array is designed to be mounted on the front surface of an ultrasound transducer. During ultrasound imaging, the contact pressure between the integrated device and the tissue would deform the sensing layer, change its dielectric properties, and therefore lead to a change in the sensor capacitance. The sensor array layout is shown in Fig. 2(a) and the cross-layer structure of a single sensor element is presented in Fig. 2(b). Fig. 2(a) depicts the whole sensing array having an area of 38.5 mm × 31.1 mm and the interdigitated electrode layout. These design dimensions were selected to accommodate the acoustic aperture of a 2D capacitive micromachined handheld ultrasound transducer such as the prototype transducer from Siemens Ultrasound reported in [28]. This design also allows the same array to be repeated to cover the scanning area of an Automated Breast Volume Scanner (ABVS),

such as the system based on the Siemens Acuson S2000. The pressure sensor array contains 16 elements with each element covering an area of $6\text{ mm} \times 6\text{ mm}$. The width of each electrode is $20\text{ }\mu\text{m}$, and the distance between the centers of two adjacent electrodes is $40\text{ }\mu\text{m}$. An electrode pattern with $40\text{ }\mu\text{m}$ electrode pitch can be readily manufactured on a flexible substrate and the sensing element capacitance which is proportional to total length of leads per sensor area is sufficient for reliable operation. In addition we assume an array of 6 mm by 6 mm sensing elements provides sufficient resolution to accurately detect a pressure distribution over contact area of the ultrasound probe if it is produced by elastic deformation of lesions located 18 mm or deeper in the tissue.

To determine the thickness of the sensing layer before the fabrication of the sensor array, the impedance measurements were conducted on the sensor arrays with PDMS layers at thicknesses of 40 , 60 , 80 , 100 , and $120\text{ }\mu\text{m}$. The results showed that the impedance of the sensing element varied little when the sensing layer's thickness was above $80\text{ }\mu\text{m}$. At a distance of $80\text{ }\mu\text{m}$ away from the electrodes, the electric field strength drops to around 6% of its maximum value. Therefore, $90\text{ }\mu\text{m}$ has been finally chosen as the thickness of the PDMS sensing layer in this study.

B. Fabrication

In the device part, as shown in Fig. 2 (b), an adhesive polyimide (PI) film has an interdigitated electrode pattern made of titanium (Ti) and gold (Au) on top, which is covered by polydimethylsiloxane (PDMS) as a sensing layer. The shielding part consists of a non-adhesive polyimide (PI) film covered by Ti and Au. The whole sensor array is bonded to polymethylpentene (PMP; is also called TPX) substrate using silicone adhesive on the back surface of the adhesive PI film. TPX has a high acoustic transparency and is commonly used in ultrasound devices.

The fabrication process involves four main steps as shown in Fig. 3. A $25\text{ }\mu\text{m}$ thick adhesive PI film (3M Kapton HN-1) was attached to a $3''$ silicon wafer (Fig. 3 (a)). AZ5214E photoresist (Microchemicals GmbH) was spin-coated on the PI film, went through baking, exposing and developing processes. A 20 nm thick Ti layer followed by a 200 nm thick Au layer were deposited by e-beam evaporation. During the lift-off process, the patterned photoresist was removed by ultrasonication in acetone (Fig. 3 (b)). A liquid PDMS prepolymer (Dow Corning Sylgard 184) was mixed with its curing agent at a ratio of 10:1 by weight, degassed in vacuum, spun onto the PI film to form a $90\text{ }\mu\text{m}$ layer, and then heat cured at 60°C for 14 hours (Fig. 3 (c)). During transferring process, the device part with silicon substrate was heated on a hotplate at 150°C to melt the adhesive on the back surface of the PI film, detached from the substrate, and then bonded, at room temperature, to a $127\text{ }\mu\text{m}$ thick TPX film.

To make a shielding part, an additional PDMS layer was spun on a silicon wafer and heat cured following the same procedure as described for the sensing device. A non-adhesive PI film was temporarily attached to the PDMS layer, a 20 nm thick Ti layer and a 200 nm thick Au layer were deposited on this PI film. Finally, the PI film covered by the metal layer was detached from PDMS layer. A strong bond between the device part and the shielding part is achieved by following the procedure described in Ref. [29]: A PI film in the shielding part

was treated with a 50 W oxygen plasma for 1 min, and then placed in an aqueous solution of 1% v/v (3-Aminopropyl) triethoxysilane (APTES) for 20 minutes. Plasma and chemical treatments were employed to activate and modify the molecules at the bonding surfaces. After the chemical treatment, the PI film was dried with a stream of air. Meanwhile, the PDMS surface was treated with a 50 W oxygen plasma for 1 min. Then the surfaces of the PI film and the PDMS layer were kept in conformal contact. After a few minutes, a strong bond was formed between the device and shielding parts (Fig. 3 (d)).

C. Performance

1) Electrical impedance measurements—A perfect pressure sensor is modeled as an ideal variable-capacitor. In the real world all sensor leads have resistance, there is a leakage current between the capacitor plates, and there are dielectric losses in the material between the capacitor plates. All these effects can be modelled by three element schematics shown in Fig 4. In this model C_p represents the sensor capacitance, R_p describes the effects of current leakage and dielectric losses and R_s represents the combined resistance of all electric leads. A good impedance analyzer can readily measure all these parameters. The ideal relationship between the equivalent circuit parameters would be $R_s \ll 1/\omega C_p (=Z_c) \ll R_p$. If R_s is too high and/or R_p too low it would reduce the sensing signal. We are prepared to accept up to 10% reduction of the output in the real circuits, and therefore, as the rule of thumb, we would accept the relation $R_s < 10 Z_c < R_p$.

The electrical impedance, Z , phase, θ , capacitance, C_p , and resistance, R_p (in parallel mode) of the sensing elements on three sensor arrays were measured using an HP 4194A impedance analyzer. Table 2 lists the mean values and standard deviations of these three electrical properties. From the measured impedance, Z , and phase shift, θ , in Table 2, the impedances of the capacitor, C_p , and the resistors, R_p , and R_s in Fig. 4 can be estimated as, $Z_c \approx 600k\Omega$, $R_p \approx 80M\Omega$, and $R_p \approx 6k\Omega$ ($R_s < 100Z_c < R_p$ which corresponds to a 1% signal reduction and is much better than we expected). Since and $R_s \ll Z \ll R_p$, the sensor design is optimal.

2) Acoustic transmission measurements—An experimental setup for acoustic transmittance measurements is shown in Fig. 5. A function generator (Tektronix AFG3251) produces electronic pulses (sinusoidal bursts with an amplitude of 20 mV, pulse repetition frequency of 50 Hz, and duration of 20 cycles) whose amplitude are magnified by a power amplifier (EN 240L). The electronic pulses excite an ultrasound transducer (on the left side of the in Figure 5), which converts the electronic pulses to acoustic pulses. The ultrasound transducers and pressure sensor array are submerged in degassed distilled water. The acoustic waves emitted from the transmitting transducer are propagated through the water only or the water and the pressure sensor array in two separate measurements (a typical insertion loss experiment), and then received by the other ultrasound transducer (on the right side of the sample in Fig. 5). The received acoustic pluses are converted back to electronic signals, which are displayed on an oscilloscope (Agilent Technologies DSO3152A). The acoustic transmissivity of the pressure sensor array can be determined by dividing the detected electric signals with and without the pressure sensor array between the transmitting and receiving transducers [30]. The acoustic transmissivity measured at three frequencies of

7.5, 10 and 15 MHz are given in Table 3. The mean and standard deviation of each value were calculated based on testing of five pressure sensor arrays.

Transmissivity data in Table 3 suggest suitability of the current design of the pressure array for clinical and biomedical research. For commercial applications a higher transmissivity is desired (a loss in transmissivity equates to a loss in acoustic penetration into the tissue). Higher transmissivity can be achieved by better selection of matching materials in the sensor and/or using a thinner layer for the polyimide substrate.

Acoustic transmissivity in Table 3 has been verified in an experiment mimicking clinical use where images of a tissue-mimicking phantom composed of micron-size acoustic-scattering glass beads embedded in agar were acquired with a Siemens Acuson S2000 system using an 18L6 linear array transducer operated at 9MHz shown in Fig. 6.

Figure 7 shows the resulting B mode ultrasound images from the tissue mimicking phantom. The image on the left was acquired without the sensor array between the transducer and the surface of the phantom and at a gain of 0dB. The image on the right was acquired with the sensor array between the transducer and the phantom and the gain was increased, until at 3dB the image looked similar to that of the phantom without the sensor, which indicating an approximately 3dB loss of power caused by the presence of the sensor.

3) Dielectrostriction measurements—Figure 8 shows the schematic of the experimental setup for dielectrostriction measurement on the sensor array. An AR-1000 rheometer (TA Instruments, New Castle, DE), as shown in Fig. 8(a), exerts normal forces on the sensor array by an acrylic plate with a surface area of 40 mm × 35 mm. A 2 mm thick silicone rubber sheet is attached to the acrylic plate. Compared to acrylic plate, a soft silicone rubber sheet can provide a better surface contact, and therefore a better stress transfer between the indenter and the sensor array. The impedance bridge circuit in Fig. 8(b) registers variations in the capacitance of the sensing elements caused by the deformation of the PDMS sensing layer. All tests were run in the squeeze/pull off mode of the rheometer with controllable variables the travel distance and loading speed of the indenter. During the tests, the starting travel point of the indenter was set to be 200 μm above the shielding layer; the travel distance was varied from 60 to 800 μm with an increment of 20 μm, and the loading speed was 20 μm/s. The corresponding normal forces were recorded by the rheometer.

The excitation voltage and frequency of the impedance bridge circuit were 2.5 V and 10 kHz, respectively. The output signal was measured by an Elvis II data acquisition system (National Instruments Corporation, Austin, TX) at a sampling rate of 250 kS/sec, and processed by Fast Fourier Transform of 1,000 data point blocks of data.

A time-dependent viscoelastic behavior of the sensing layer can be an issue for implementing the dielectrostriction stress-sensing approach [17, 19, 24]. A preliminary test has been run to evaluate transition effects. Figure 9 shows the voltage response to incremental loading. Detected voltage and several representative stresses are shown in Fig. 9(a) as the function of time. Same measurements are presented in Fig. 9(b) as the voltage

response to applied stress. Obtained results show that time-transition effects in the sensing layer can be neglected on timescales 1 to 10 sec.

Figure 10(a) presents a typical normal stress vs. time profile. The corresponding dielectrostrictive responses measured by a representative sensing element is shown in Fig. 10(b), a scatter of the data points is within 1 to 3%. A short interval time between two adjacent steps in preprogramed rheometer loading subroutine was observed during the test. An exact duration of this intervals between squeeze to pull off, or pull off to squeeze cannot be set by the user and was not recorded by the rheometer. Voltage data in Fig. 10(b) show 10 sec-plateaus at zero and maximum stresses with the data scattering less than 3% (not seen at the selected scale).

Based on the data in Fig. 10, the sensor output signal is replotted in terms of stress in Fig. 11. From Fig. 11, it has been found that the sensor signal is not linear with stress over the entire stress range. Instead, the signal vs. stress curve can be split into three regions. In the two regions with stresses less than about 0.7 kPa and greater than about 2.2 kPa, the sensor response increases linearly with stress but with different slopes. In the transition region between these two stress values, the slope of sensor output vs. stress curve decreases with an increase in stress. The mean and standard deviation of the smallest stress detected by sensing element, and the slopes of signal vs. stress curve in the two linear regions are listed in Table 4 for 12 sensing elements on 3 sensor arrays.

IV. Discussion

The dielectric response-stress profile of PDMS shows piecewise linear behavior under compressive loadings, as shown in Fig. 11. This piecewise linearity of stress-dielectric relation in PDMS might be attributed to the different mechanical constraints on PDMS sensing layer under small and large deformations. The PDMS sensing layer is fixed on a PI film at its bottom surface. Under a small compressive load, σ_{zz} , the PDMS layer is free to expand at its top and lateral side surfaces. Under a large compressive load the PDMS layer between two rigid plates is hard to expand, that is, the Poisson effect is not free to occur. PDMS has a Poisson's ratio close to 0.5. Therefore, its constrained stiffness under large deformations would be much larger than its stiffness in the free expanding situation under small deformations. The differences in the stiffness of the sensing layers and the sensor capacitances (as expressed in Eqs. 7 and 8) under small and large deformations explain the piecewise linearity observed in the stress-dielectric relation.

Experimental data from Table 4 can be verified by the theory developed in Section II. Slopes of experimental curves in Fig.10 are linked to the sensor capacitance as

$$\frac{1}{V} \frac{dV}{d\sigma_{zz}} = \frac{1}{C} \frac{dC}{d\sigma_{zz}} \quad (9)$$

This expression follows from balancing conditions of the bridge circuit in Fig 8(b). Expressions 7 and 8 provide a capacitance of interdigitated planar sensor, C . In addition, the sensor design in Fig. 2 has a shielding layer which acts as a parallel plate capacitor, C_{shield} .

which 25% increases the total capacitance of the sensor. A capacitance of the shield also varies with deformations.

For unconstrained layer

$$\frac{1}{C} \frac{dC}{d\sigma_{zz}} = \frac{(1-v)\alpha_1 + 2(1-2v)\alpha_2}{2E} \quad (10)$$

$$\frac{1}{C_{shield}} \frac{dC_{shield}}{d\sigma_{zz}} = \frac{\alpha_1 + (1-2v)\alpha_2 - 1}{2E}$$

For fully constrained layer:

$$\frac{1}{C} \frac{dC}{d\sigma_{zz}} = \frac{(1+v)(1-2v)(\alpha_1 + 2\alpha_2)}{2(1-v)E} \quad (11)$$

$$\frac{1}{C_{shield}} \frac{dC_{shield}}{d\sigma_{zz}} = \frac{(1+v)(1-2v)(\alpha_1 + 2\alpha_2 - 1)}{2(1-v)E}$$

Ref [31] provides material data for Dow Corning Sylgard 184 PDMS which has been utilized for sensing layer in the developed sensor array. We utilized curing in our manufacturing process. According to Ref. [31] PDMS cured under temperature 60C should yield Young' modulus $E_y=1.8\text{MPa}$ and compressive modulus $E_c= 170\text{MPa}$. These modules in turn are in agreement with commonly accepted approximation of $\nu \approx 0.499$ for Poison's ratio.

Substituting material data from Ref. [31] in the expressions above provides comparisons of the sensor sensitivity with experimental values in Table 4. Predicted sensitivity to the pressure is highly influenced by the Poison's ratio and Young' modulus of the sensing layer. Table 5 shows predicted slops for range of Poison's ratio approaching value 0.5 which corresponds to incompressible material.

The testing results for a single sensing element based on 4 measurements show that the standard deviation of the dielectrostriction signal is about 30% of the mean value at small deformations (stress less than 0.7 kPa), and about 10% at large deformations (stress greater than 2.2 kPa). For 12 sensing elements on 3 different sensor arrays, the variation of the dielectrostriction is about 75% at small deformations, and about 38% under large deformations as shown in Table 3. From Table 2 it can be seen that the standard variation of the impedance is about 5% of the mean value for all sensing elements on different arrays. The large variation of the dielectric responses among the sensing elements is believed to be related to the contact surface between the indenter and the sensing layer. Since the thickness of the sensing layer is only about 90 μm , a tiny bump on the surface of the indenter or the sensing layer, a slight deviation of the indenter's axis from the normal loading direction, or a small variation in the thickness of the sensor array would cause the contact surface between the indenter and the sensing layer to be much less than the area seen by one sensing element.

This phenomenon is more critical under small deformations than that under large deformations, and has been seen less significant for a soft (silicone rubber) indenter than a rigid (acrylic) indenter as the former can deform to some extent to increase the contact surface.

V. Conclusion

In this study, a microsensor array based on the dielectrostriction sensing mechanism has been developed to measure the contact pressure between an ultrasound probe and breast tissue to assist ultrasound diagnosis during breast cancer screening. The same sensing mechanism, and similar structure and configuration of this sensor array, can be applied for other screening and diagnostic ultrasound tasks. This sensor array adopts solid-state technology, therefore is able to work over a large dynamic range and withstand large overload. The planar electrode configuration and simple structure of this sensor array allows for the reduction in the thickness of the sensor array and the insertion of an acoustic matching layer which would improve the acoustic transmittance of the sensor array. The piecewise linearity observed in the stress-dielectric responses of the current sensor array is believed to be attributed to the different mechanical constraints on the sensing layer at small and large deformations, which might be resolved by replacing the current bulk structure with a meta-structure having a low effective Poisson's ratio. The demonstrated performance is satisfactory for clinical biomedical research in ultrasound image formation and interpretation, however for commercial applications a higher ultrasound transmissivity is desired. To increase the acoustic transmissivity of the sensor array, the width of the electrode and the distance between two neighboring electrodes can be reduced to allow for the thickness reduction of the sensing layer. The other options to improve the acoustic transmittance of the sensor array include reducing the thicknesses of the other layers in the sensor array, and inserting acoustic matching layers to minimize the discrepancy of the acoustic impedances between the integrated device (pressure sensor and ultrasound probe) and the tissue.

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REFERENCES

1. Prince, J.L.; Links, J. *Medical Imaging Signals and Systems*. Prentice Hall; 2005.
2. Sarvazyan A, et al. An overview of elastography - an emerging branch of medical imaging. *Current Medical Imaging Reviews*. 2011; 7(4):255–282. [PubMed: 22308105]
3. Leong LCH, et al. A prospective study to compare the diagnostic performance of breast elastography versus conventional breast ultrasound. *Clinical Radiology*. 2010; 65:887–894. [PubMed: 20933643]
4. O'Hagan JJ, Samani A. Measurement of the hyperelastic properties of 44 pathological ex vivo breast tissue samples. *Phys. Med. Biol.* 2009; 54:2557–2569. [PubMed: 19349660]
5. Hall TJ, et al. Recent results in nonlinear strain and modulus imaging. *Current Medical Imaging Reviews*. 2011; 7(4):313–327. [PubMed: 22754425]

6. Goenezen S, et al. Linear and Nonlinear Elastic Modulus Imaging: An Application to Breast Cancer Diagnosis. *IEEE Trans Med Imaging*. 2012; 31(8):1628–1637. [PubMed: 22665504]
7. Tekscan, Inc.. [cited; Available from: <http://www.tekscan.com>]
8. Sensor Products Inc.. [cited; Available from: <http://www.sensorprod.com/>]
9. Dally, JW. Experimental stress analysis. 3rd ed.. McGraw-Hill; New York: 1991.
10. Strain Measurement Devices Inc.. [cited; Available from: <http://www.smdsensors.com>]
11. Roriz P, et al. Review of fiber-optic pressure sensors for biomedical and biomechanical applications. *Journal of Biomedical Optics*. 2013; 18(5):050903–(1-18).
12. SensingTex.. [cited; Available from: <http://www.sensingtex.com/>]
13. Maiolino P, et al. A Flexible and Robust Large Scale Capacitive Tactile System for Robots. *Ieee Sensors Journal*. 2013; 13(10):3910–3917.
14. Pritchard, E., et al. Sensors. 2008 IEEE; 2008. Flexible capacitive sensors for high resolution pressure measurement..
15. Baxter, LK. Capacitive sensors: design and applications. IEEE Press; New York: 1997.
16. Pressure Profile Systems Inc.. [cited; Available from: <http://www.pressureprofile.com/>]
17. Shkel, YM. Electrostriction: material parameters and stress/strain constitutive relations. Vol. 87. *Philosophical Magazine*; 2007. p. 1743-1767.
18. Lee HY, Peng Y, Shkel YM. Strain-dielectric Response of Dielectrics as Foundation for Electrostriction Stresses. *J. Appl. Phys*. 2005; 98(7):074104.
19. Peng Y, Shkel YM, Kim GH. Stress Dielectric Response in Liquid Polymers. *J. Rheol*. 2005; 49(1):297–311.
20. Shkel YM, Klingenberg DJ. Material parameters for electrostriction. *J. Appl. Phys*. 1996; 80(8): 4566–4572.
21. Shkel YM, Klingenberg DJ. Electrostriction of polarizable materials: Comparison of models with experimental data. *J. Appl. Phys*. 1998; 83(12):7834–7843.
22. Stratton, JA. Electromagnetic Theory. McGraw-Hill; New York: 1941.
23. Landau, LD.; Lifshitz, EM.; Pitaevskii, LP. Course of Theoretical Physics. 2nd ed.. Vol. 8. Butterworth-Heinenann; Oxford: 1984. Electrodynamics of Continuous Media..
24. Peng, YY.; Shkel, YM. Rheo-Dielectric Effect in Liquid Suspensions. Vol. 89. *Philosophical Magazine*; 2009. p. 1473-1487.
25. Peng, YY. Mechanical Engineering Department. University of Wisconsin; Madison: 2008. Rheo-dielectric Studies in Polymeric Systems; p. 176
26. Lee HY, Shkel YM. Dielectric Response of Solids for Contactless Detection of Stresses and Strains. *Sensors and Actuators A*. 2007; 137:287–295.
27. Clipper controls. Dielectric Constant Values. 2014. [cited; Available from: <http://www.clippercontrols.com/pages/Dielectric-Constant-Values.html#S>]
28. Fisher TG, et al. Volumetric Elasticity Imaging with a 2-D CMUT Array. *Ultrasound Med Biol*. 2010; 36(6):978–990. [PubMed: 20510188]
29. Sunkara V, et al. Simple room temperature bonding of thermoplastics and poly(dimethylsiloxane). *Lab on a chip*. 2010; 11:962–965. [PubMed: 21152492]
30. Madsen EL, et al. Inter-laboratory comparison of ultrasonic attenuation and speed measurements. *J. Ultrasound Med*. 1986; 5:569–576. [PubMed: 3534290]
31. Johnston ID, D.K.M. Tan CKL, Tracey MC. Mechanical characterization of bulk Sylgard 184 for microfluidics and microengineering. *Journal of Micromechanics and Microengineering*. 2014; 24:035017, 7.

Figure 1.

(a) Interdigitated electrodes are located in the xy -plane and form an angle, θ , with respect to the y -axis. A dielectric layer is on the top of the electrodes. (b) Each electrode has width, $2w$, length, l , and is separated from each other by distance $2a$.

Figure 2.
The schematic of (a) the electrode pattern, and (b) the structure of the sensor array.

Figure 3.

Fabrication steps of the array: (a) adhesive PI film is attached to Si-wafer and (b) electrode pattern is deposited; (c) PDMS sensing layer is spun-coated and (d) metalized PI film is chemically bonded to the sensing layer.

Figure 4.
A 3-component circuit to model each sensing element in the array

Figure 5.

The schematic of the experimental setup for acoustic transmission measurements. Acoustic waves are emitted by the transducer on the left side, propagate through the media, and then detected by the transducer on the right side.

Figure 6.

Layout of ultrasound imaging system and tactile sensor array from top to bottom: Siemens Acuson S2000 system using an 18L6 linear array transducer, tactile pressure array, and tissue-mimicking phantom.

Figure 7.

Images of a tissue-mimicking phantom were acquired using a Siemens Acuson S2000 system with an 18L6 linear array transducer operated at 9MHz. The image on the left was acquired without the sensor array between the transducer and the surface of the phantom and at a gain of 0dB. The image on the right was acquired with the sensor array between the transducer and the phantom and the gain of 3dB.

Figure 8.

(a) The rheometer exerts normal forces on the sensor array by an acrylic indenter with a surface area of $40 \text{ mm} \times 35 \text{ mm}$ and a 2 mm thick silicone rubber tip. (b) The impedance bridge circuit detects the dielectrostrictive response of the sensing elements which are excited by a 2.5 V AC, 10 KHz sine wave.

Figure 9.

Sensor response to incremental step loading: (a) Voltage response and several representative stresses are shown as the function of time. (b) Voltage response as a function of applied stress.

Figure 10.

(a) Exerted normal stresses, and (b) voltage response versus time for one representative sensing element.

Figure 11.

A typical dielectrostrictive signals vs. stresses curve obtained with one sensing element.

Table 1

Sensing technologies which were evaluated for ultrasound application.

Sensing Technology	Resolution (mm)	Dynamic range	Tolerance to overloads	Ultrasound application
Chemical [8]	0.3	1:5	Bad	Yes
Piezoresistive [7]	0.6	1:20	Good	Yes
Optical [12]	5	1:200	Fair	No
Air-gap capacitive [16]	2	1:200	Fair	No
Dielectrostriction [17, 20]	6	1:10 ⁴	Good	Yes

Table 2

The electrical impedance, phase, capacitance, and resistance of the sensing elements on three arrays measured at 10 kHz.

No.	Z (k Ω)	θ (°)	C_p (pF)	R_p (M Ω)
1	636.69±17.41	89.42±0.03	25.02±0.69	62.76±1.94
2	651.42±32.01	89.56±0.05	24.60±1.26	84.76±9.06
3	651.34±21.94	89.62±0.06	24.43±0.85	99.88±15.43

Table 3

Acoustic transmissivity (in amplitude) of the sensor arrays measured at various frequencies.

Frequency (MHz)	7.5	10	15
Transmissivity(%)	71.14±2.86	67.90±1.45	57.87±1.45

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Table 4

The smallest detectable stress, and the slopes of dielectrostrictive signal vs. stress in two linear regions.

Min detectable σ	Slope $\times 10^{-6}$ @ $\sigma < 0.7 \text{ kPa}$	Slope $\times 10^{-6}$ @ $\sigma > 2.2 \text{ kPa}$
$0.14 \pm 0.09 \text{ kPa}$	$176.65 \pm 132.11 \text{ kPa}^{-1}$	$9.69 \pm 3.70 \text{ kPa}^{-1}$

Table 5

Slopes for unconstrained constrained sensing layers calculated for the range reported in Ref. [31] material data, Poisson's ratio $\nu = 0.495 \div 0.498$ and Young's modulus $E = 1.8MPa/1.25MP$.

ν	$\frac{1}{V} \frac{dV}{d\sigma} \times 10^{-6} \frac{V}{VkPa}$	
	$\sigma < 0.7 \text{ kPa}$	$\sigma > 2.2 \text{ kPa}$
0.495	65.7/94.7	12.4/17.9
0.496	65.3/93.9	9.95/14.3
0.497	64.8/92.6	7.49/10.7
0.498	64.3/91.9	5.0/7.2