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Fabrication and characterization of a smart epitaxial piezoelectric micromachined ultrasonic transducer

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ABSTRACT

A novel piezoelectric micromachined ultrasonic transducer (pMUT) array was designed and fabricated using epitaxially grown functional $Pb(Zr_{0.52}Ti_{0.48})O_3$ (PZT) thin film on Si(1 1 1)/γ-Al_2O_3(1 1 1)/$SrRuO_3$(1 1 1) substrate for biomedical applications. The crystallographic orientation of PZT film was controlled by the incorporation of epitaxial γ-Al_2O_3 film on Si substrate. Modal shape of pMUT was analyzed employing advanced 3D finite element modeling taking the crystallographic anisotropy of materials and the properties of immersed medium (air or water) into account. Eigenfrequency with mode shapes has shown to have significant influence on transmitting-receiving characteristics of pMUT. Modal shapes of pMUT were also quantitatively determined using Laser Doppler Vibratometry (LDV). An excellent correlation was obtained between computational and experimental results. A significantly high sensitivity of 3.9 μV/kPa was obtained in an under-water ultrasonic wave transmission experiment conducted using fabricated pMUT as wave transmitter and a commercial transducer as receiver at a fundamental frequency of 1.20 MHz. Advanced FE computation thus serves as a tool to a priori optimize device structure for the successful transmission of ultrasonic waves with sufficient power to generate high resolution 3D imaging.

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1. Introduction

Sensing architect is confronted with a big challenge to realize ultrasonic transducers with high sensitivity maintaining device integrity. The ultrasonic transducers transform electrical energy into mechanical energy (an acoustic pulse is transmitted) and transform mechanical energy into electrical energy (an echo is received). The vibrating material commonly used for ultrasonic transducer in the probe of medical ultrasonic diagnostic equipment is piezoelectric ceramics.

The ultrasonic wave transmitted towards the object spreads in a medium and the reflected ultrasonic wave returns towards the transducer. Since the received echo differs in intensity depending on the structure of the living organisms and the reflected medium, thus the relative intensity of the received signal can be used to construct two-dimensional image with the variation of brightness. The ultrasonographic image enables the outline of internal organs and the boundary of a liquid-living body is detected thereby recognizing the structure inside a living body.

A significant improvement in sensitivity can be obtained using functional thin films based epitaxial piezoelectric micromachined ultrasonic transducer (pMUT) arrays to fully compete with capacitive micromachined ultrasonic transducer (cMUT) arrays [1–15] particularly in medical diagnostic applications where high depth of penetration is required [16–30]. pMUT does not require high voltage, responds linearly with drive voltage, and its deflection and micro fabrication are not confined by gap size.

In addition, in order to improve resolution in depth direction, a higher frequency transducer is required. Moreover, due to miniaturization demand is increasing for applications of the ultrasonic transducer at the tip of ultrasound endoscopy or catheter for intravascular ultrasonic diagnosis. However, with the conventional piezo-electric bulk ceramics, it is difficult to realize miniaturized elements using ultra-fine processing technology. For example, if an ultrasonic transducer array is required to attach at the tip of a catheter, a chip size would be about 2 mm × 2 mm or less.

However, a class of ferroelectric materials namely $Pb(Zr_xTi_{1-x})O_3$ (PZT) as thin films is promising for piezo-electric

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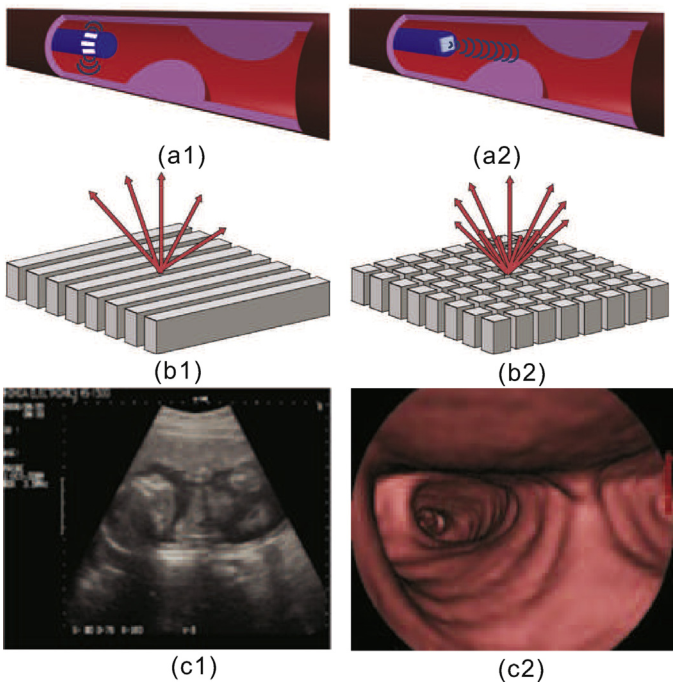

Fig. 1. Ultrasonic transducer array: 1-D (a1–c1) and 2-D (a2–c2) [31].

ceramics of pMUT. PZT attracts considerable attention due to its excellent piezoelectricity [31]. Fig. 1(a1) and (a2) shows arteriosclerosis in a blood vessel with positioned pMUT on the tip of an endoscope in 1-D and 2-D transduction modes, respectively.

Generally, pMUTs are arranged in 1-D array (Fig. 1(b1)). Since the information in the depth direction is acquired by the ultrasonic wave transmitted and received with one transducer and the information on a transverse direction is acquired by changing the transducer which transmits and receives an ultrasonic wave enabling to acquire a 2-D image (Fig. 1(c1)). However, in recent years, there is tremendous demand for 3-D imaging in medical diagnostics (Fig. 1(a2)). To address this demand, it is required to fabricate ultrasonic transducer in 2-D array (Fig. 1(b2)). By acquiring the information in the depth direction by the ultrasonic wave transmitted and received with one transducer, and changing the transducer which transmits and receives ultrasonic wave in both lengthwise and transverse directions allow to capture a 3-D image (Fig. 1(c2)).

Moreover, broad bandwidth technology allows medical transducers to be used with more than one operating frequency. Optimal imaging is thus a balance between acoustic beam attenuation and high resolution. High resolution imaging of superficial targets is captured with a short signal wavelength at higher frequencies (10–15 MHz) but these high intensity signals are quickly attenuated in tissue [32]. In contrast, imaging of large or deep organs can be realized at lower frequencies (<5 MHz) which allows a large number of prospective ultrasonic applications, so the first mode resonant frequency between 1 and 5 MHz can be considered as a practical range in the optimal designing of pMUTs.

However, no research has been conducted yet to quantify mode shapes of diaphragm both experimentally and numerically and correlate it with resonance frequencies thereby extracting operating frequency for maximum energy transfer keeping device integrity intact for medical diagnostic applications. In this context, a pMUT structure consisting of Si(1 1 1)/γ-Al_2O_3(1 1 1)/Pt(1 1 1)/$SrRuO_3$(1 1 1)/PZT(1 1 1)/$SrRuO_3$(1 1 1) was designed and fabricated combining miniaturized ultrasonic transducer and a signal-processing circuit. We considered epitaxial $Pb(Zr_xTi_{1-x})O_3$ ($x = 0.52$) thin films in smart pMUTs design, which was grown epitaxially on γ-Al_2O_3 buffer insulating layer using Si substrate. Such epitaxial PZT thin films have higher sensitivity than crystalline bulk PZT counterpart.

2. Designing of pMUT array

We propose a new design (see Fig. 2(a)) of an array of 8×8 channel circular shaped (as high density is attained with circular shape compared to square one) pMUT based on a structure consisting of Si(1 1 1)/γ-Al_2O_3(1 1 1)/Pt(1 1 1)/$SrRuO_3$(1 1 1)/PZT(1 1 1)/$SrRuO_3$(1 1 1). In this structure, PZT thin film was epitaxially grown on epitaxial γ-Al_2O_3/Si substrate. Such epitaxial PZT thin film has shown to have higher sensitivity than crystalline PZT thin films on SiO_2. A detail view of a single channel pMUT is presented in Fig. 2(b) where each channel consisting of 42 circular transducer elements with a diameter of 100 μm in each element.

The pMUT array thus designed was integrated with Si Large-Scale-Integration (LSI) technology and each element of pMUT possesses a semispherical cavity underneath the functional piezoelectric PZT thin film created by surface bulk micromachining technique with etching in XeF_2 gas through a hole of 10 μm in diameter. This technique ensured miniaturization resulting high density integration compared to that obtained with conventional back-side etching technique.

2.1. Fabrication

The pMUT array was fabricated using our LSI facility as follows. At first, γ-Al_2O_3 thin film of 10 nm was epitaxially grown on a Si (1 1 1) substrate by molecular beam epitaxy i.e. MBE (Al flux: 1 nm/min, O_2:3.0×10^{-2} Pa, time: 120 min) at 750 °C. Next, a Pt thin film of 100 nm was deposited on γ-Al_2O_3 at 600 °C using RF sputtering method (RF 75 W, Ar: 66 sccm, 5.0×10^{-1} Pa, time: 10 min). Then, to obtain an interface with sufficient strength and strain compatibility a $SrRuO_3$ (SRO) thin film (sputtering condition: RF 15 W, Ar:O_2 = 12:3 sccm, 1.0 Pa, time: 60 min) of 10 nm was deposited on Pt film at 700 °C using RF sputtering technique too (see Fig. 3(a)).

Subsequently, a PZT film was deposited employing a chemical solution deposition (CSD) technique namely sol–gel method using a PZT precursor solution of Pb:Zr:Ti = 1.15:0.52:0.48 by 15 wt.% (Mitsubishi Material Corporation). Spin coating was conducted first at 1000 rpm for 10 s and next 3000 rpm for 20 s. The deposited film was dried at 150 °C for 5 min followed by pyrolysis at 250 °C for 5 min. To form PZT with perovskite phase (crystallization), deposited film was sintered at 650 °C for 90 s under O_2 environment (Pr: 0.5 slm) employing rapid thermal annealing (RTA). The above procedure was repeated for 5 times to obtain a film thickness of 500 nm (100 nm film per layer). Then, an SRO thin film of 100 nm was deposited for upper electrode using RF sputtering technique (condition: RF 100 W, Ar: 12 sccm, 1 Pa, time: 3 min) at room temperature. The thickness of deposited thin film was determined using optical profilometry or spectroscopic ellipsometry depending on the type of film.

Next, the top SRO and PZT thin films were patterned using conventional lithography technique followed by inductively coupled plasma reactive ion etching (ICP-RIE) using CHF_3 gas. The bottom SRO/Pt and γ-Al_2O_3 layers were patterned using ICP-RIE in Ar gas. A SiN film (insulator) of 1 μm was deposited by plasma enhanced chemical vapor deposition (PECVD) at 300 °C. Deposited SiN film was patterned by ICP-RIE in SiF_6 gas. To make interconnect, 'Al' thin film of 1.2 μm was deposited by electron beam (EB) evaporator. Then, 'Al' interconnect was pattered by RIE technique employing BCl_3, Cl_2, and N_2 gases.

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