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Improvement of Integrated Ultrasonic Transducer Sensitivity

JIAN-HUA MO, ANDREW L ROBINSON and FRED L TERRY, JR

Solid-State Electronics Laboratory, Department of Electrical Engineering and Computer Science, University of Michigan, Ann Arbor, MI 48109-2122 (U S A)

DALE W FITTING* and PAUL L CARSON

Radiology Physics and Engineering Division, Department of Radiology, University of Michigan Medical Center, Ann Arbor, MI 48104 (U S A)

Abstract

In this paper, we present a micromachined diaphragm structure for integrated ultrasound transducers This structure greatly reduces the parasitic capacitance between the lower electrode and the conductive Si substrate in a nonmicromachined structure The micromachining improves both sensitivity and minimum detectable signal It also reduces crosstalk between transducer elements

Introduction

The concept of Si-based integrated ultrasound transducers was first realized with a PVDF-MOSFET (POSFET) structure (Fig 1) by Swartz and Plummer in 1979 [1] The POSFET structure combines a PVDF transducer with a MOSFET input amplifier fabricated with conventional integrated circuit (IC) technology With the help of well-developed silicon IC technology in design and fabrication, a large number of small-size transducers can be made on an Si substrate with the potential for further integration of on-chip signal-processing circuitry Possible applications include medical imaging and non-destructive evaluation since, as recent studies show, large arrays of small-size transducers improve image quality [2]

However, as indicated by Swartz and Plummer, some problems exist in the POSFET structure [1] (1) a large parasitic capacitance between the extended lower electrode and the conductive silicon substrate, which shunts the input to the preamplifier and therefore causes sensitivity loss, and (2) lateral propagation of acoustic waves (possibly through the Si substrate), which causes acoustic crosstalk between the elements in a transducer array As the size of the transducer is scaled down with integration, any sensitivity loss from already small signals may be costly, and close proximity of neighboring transducer elements in an integrated environment may produce more severe crosstalk Furthermore, the relatively high propagation velocity of acoustic waves in Si may seriously limit the acceptance angle of a transducer array through crosstalk

A POSFET structure based on a dielectric substrate has been suggested [1] to deal with the problem of parasitic capacitance In that version, the problem of lateral propagation through the substrate will still remain To our knowledge, no work has been done to deal with these problems

In this paper, we propose a diaphragm structure for integrated ultrasound transducers based on micromachining of Si This structure greatly reduces the aforementioned parasitic capacitance and reduces the lateral propagation of acoustic waves through Si substrate As compared to a non-micromachined structure, we see improvement in both sensitivity and minimum detectable signals, we have also found a narrower effective width for each micromachined transducer, which is an indirect indication of reduced acoustic crosstalk

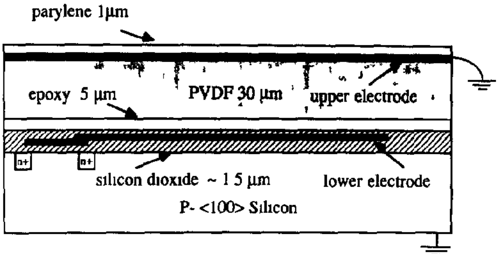

Fig 1 POSFET structure

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*D W Fitting is now with the National Institute of Standards and Technology (formerly the National Bureau of Standards), Boulder, CO 80303, U S A

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The Micromachined Diaphragm Structure

The proposed diaphragm structure is schematically illustrated in Fig 2 This structure is quite similar to the POSFET structure, except that the Si substrate underneath the extended lower electrode and the thin dielectric layer is removed using silicon micromachining technologies [3] and reactive ion etching Because the conductive Si substrate is physically removed, the parasitic capacitance can be basically eliminated except for a component due to peripheral areas For the same reason, the acoustic crosstalk between neighboring transducer elements can also be reduced by the removal of the Si substrate propagation medium for lateral travelling of acoustic waves Therefore the micromachined structure can provide an improvement in transducer sensitivity and a reduction in crosstalk between elements in a transducer array With improved transducer sensitivity, smaller minimum detectable signals may also be expected

Three sets of transducers have been fabricated (Fig 3) type I—devices with solid substrate and with SiO_2 separating the lower electrode from the Si substrate, type II—devices with a micromachined substrate and with a 1-μm-thick stress-balanced composite $SiO_2/Si_3N_4/SiO_2$ diaphragm under the lower electrode, and type III—micromachined devices with the same composite diaphragm and a 5-μm-thick p^+ silicon layer Type III devices are intermediate products in the fabrication of type II devices and are included here to help distinguish the electrical effects of micromachining from the acoustical effects

The fabrication of type I devices starts from a lightly doped p-type ⟨100⟩ silicon substrate A wet oxidation is performed at 1100 °C for three hours to grow a SiO_2 layer of 1 μm thick Cr and Au are then evaporated on the SiO_2 with thicknesses of 400 Å and 2000 Å, respectively, and are patterned into 1 mm squares with extended bonding pads to serve as lower electrodes Next, a 40-μm-thick PVDF sheet, with Au on its top as the upper electrode, is bonded onto the Si substrate, using non-conductive epoxy The upper electrode and the Si substrate are both grounded, and the signal comes out from the lower electrode

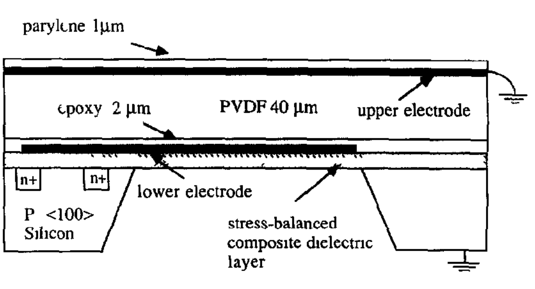


Fig 2 Micromachined diaphragm structure

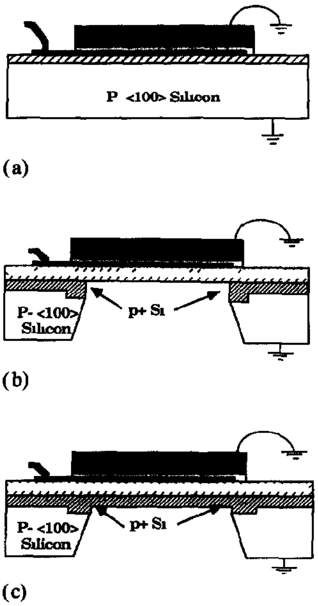


Fig 3 Three types of transducers with (a) type I—solid substrate, (b) type II—diaphragm w/o5/micron p^+ Si layer, and (c) type III—diaphragm with 5/micron p^+ Si layer

There are several different processing steps in the fabrication of types II and III devices as compared to the fabrication sequence of type I devices First, before the formation of the thin dielectric layer, deep and shallow boron diffusions are performed to form a p^+ rim and a p^+ silicon layer as mechanical support for diaphragms formed later [4] These diffusions are formed with solid boron sources at 1175 °C for 3 5 hours and 16 hours, respectively Second, instead of a single SiO_2 layer in type I devices, a short wet oxidation at 1100 °C is conducted to form a 2000 Å SiO_2 layer, and then CVD Si_3N_4 and SiO_2 layers with thicknesses of 1500 Å and 6500 Å are deposited at 820 °C and 920 °C, respectively These conditions provide a flat composite dielectric layer Third, before bonding the PVDF sheet onto the chip, the Si-etch window on the backside of the wafer is defined with the lower electrode at its center The Si substrate in the backside etch window is anisotropically etched with a water and EDP solution at 110 °C The Si-etch will stop at p^+ rim and p^+ layer with etch stop depth at about 10 μm and 5 μm respectively, from the front side For type II devices, an additional step is taken after bonding the PVDF sheet, reactive ion etching (RIE) is performed to remove the 5-μm p^+ silicon layer underneath the lower electrode

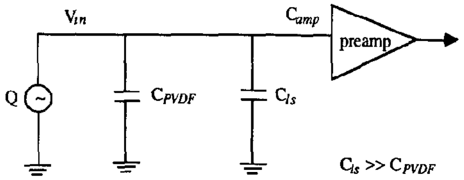
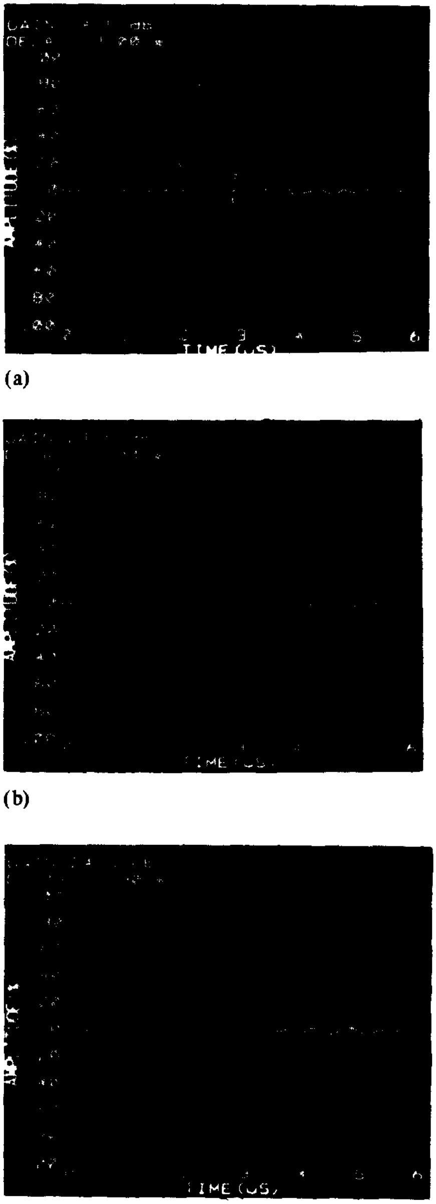


Fig 4 Simplified equivalent circuit



(a)

(b)

(c)

Fig 5 Typical response waveforms for transducers (a) Type II DG = 14 5 dB, DP = 80%, (b) type I DG = 24 5 dB, DP = 90%, (c) type III DG = 24 5 dB, DP = 80%

Simple Modeling and Experimental Results

Sensitivity and Minimum Detectable Signals

Measurements of transducer response to acoustical input were performed through a water path on all transducers with an air backing The device under test drives a simple buffer

Figure 4 shows a simplified equivalent circuit for POSFET structure Based on this simple model, the sensitivity ratio *SR* for two types of transducers with the same acoustical backing condition is

$$SR_{a/b} \equiv \frac{S^a}{S^b} = \frac{C^b_{PVDF} + C^b_{amp} + C^b_{ls}}{C^a_{PVDF} + C^a_{amp} + C^a_{ls}}$$

where S^a and S^b are the sensitivities of the transducers being compared, C_{PVDF} is the clamped capacitance associated with PVDF sheet with its area defined by the lower electrode, C_{ls} is the previously mentioned parasitic capacitance between the lower electrode and the conductive Si substrate, and C_{amp} is the input capacitance of the pre-amplifier

Figure 5 illustrates typical waveforms for each type of transducer Measured $SR_{II/III}$ = 8 9 dB and $SR_{II/I}$ = 9 6 dB, in good agreement with predicted values (Table 1) The minimum detectable signals are about 4000 Pa, 1600 Pa, and 7000 Pa (Fig 6) for types I, II, and III devices, respectively We believe that the minimum detectable signals are mostly limited by the noise level of the test set-up With the same air backing for all transducers, the

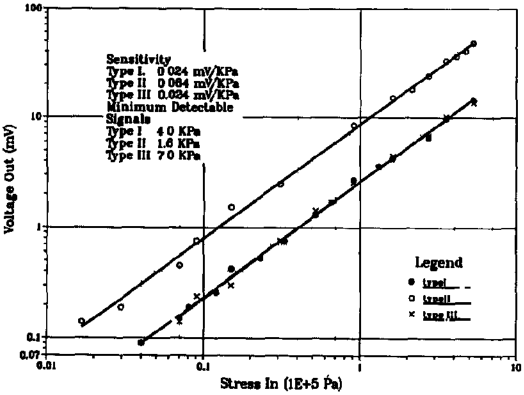


Fig 6 Transducer sensitivity and minimum detectable signals

TABLE 1 Predicted and measured sensitivity ratio (*SR*) (dB)

$SR_{a/b}$	Predicated	Actual
$SR_{II/I}$	10 9	8 9
$SR_{II/III}$	11 3	9 6

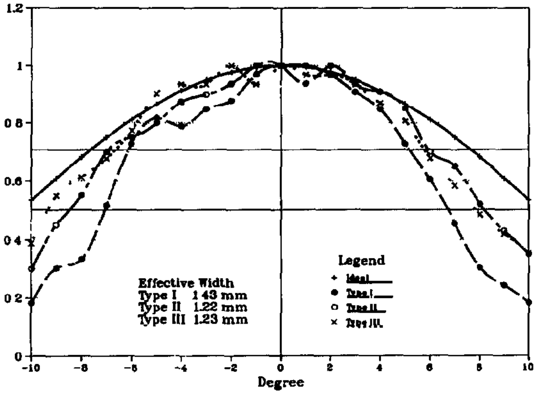


Fig 7 Transducer directivity

sensitivity improvement and smaller minimum detectable signal are electrical effects of the micromachining of Si substrate rather than acoustical in nature

Crosstalk

In general, crosstalk between transducer elements makes a transducer acoustically wider than it is physically The effective element width can be determined through directivity measurement as an indirect indication of acoustic crosstalk in devices The directivity of each type of transducer was measured by changing angles between the incoming acoustic beam and the transducer under test Figure 7 shows the directivities for the ideal case and for each type of transducers The average of −3 dB and −6 dB effective widths is about 1 22 mm for types II and III devices and 1 43 mm for type I devices, evidence of reduced acoustical crosstalk in micromachined transducers

Conclusions

We have presented a micromachined diaphragm structure for integrated ultrasound transducers This structure greatly reduces a large parasitic capacitance in a non-micromachined structure and thus improves transducer sensitivity and allows a smaller minimum detectable signal It also reduces acoustic crosstalk between transducer elements because the Si substrate underneath the lower electrode, a lateral acoustic wave propagation medium, is removed The improved device performance through micromachining, coupled with other advances in material quality and process compatibility, may permit development of two-dimensional transducers with better image quality

The above experimental results are also applicable to transducers based on P(VDF-TrFE) copolymer, which is under study The process compatibility provided by P(VDF-TrFE) copolymer makes it an attractive alternative to PVDF in terms of further integration of ultrasound transducer with the signal-processing circuitry on the Si wafer [5]

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