A Multifunctional Silicon-based Microscale Surgical System

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ABSTRACT

In this paper we report our progress on the development of silicon-based surgical tools. We report on the integration of piezoresistive sensors in noise insensitive Wheatstone configuration, silicon nitride/platinum composite overhanging blades, fluid channels, and electrical packaging features to make the device highly multifunctional. Multiple polysilicon piezoresistors were integrated onto bulk micromachined ultrasonic transducers, that were characterized for measuring the ultrasonic strain and the force applied to the cutting tip. The use of thin film silicon nitride/metal composites for cutting swine retinas at the cellular level is demonstrated.

INTRODUCTION

Microscale smart surgical tools can benefit many surgical operations where microscale or cellular level tissue manipulation is required [1]. For example, retinal surgery requires the manipulation of few layers of retina cells in an often-performed retinal-detachment surgical procedure. Other examples of cellular-scale removal of cells are epithelial cell transplants, plastic surgery, corneal incisions, and skin grafts. Although the glass pipette patch probes have existed for several decades to measure cellular potentials and manipulate tissue at the microscale, they requires high optical magnification to view the area being cut with expensive micromanipulation technology without integrated sensors. Sharp microscale blades made of diamond or other ceramics have been used but result the technology precludes the integration of solid-state sensors. The alternative of bonding sensors onto microscale cutting tools requires expensive packaging and leads to reliability problems.

Micromachined silicon probe technology allows one to fabricate cutting tools that have microscale cutting edges suitable for cells ranging in size from 10 to 100 microns. Micromachined neural probes have been designed to measure cellular potentials for at least a decade [5] and continued advances in that field have resulted in highly engineered systems. Force measurement probes with micromachined probes have also been demonstrated using piezoresistors [6].

We have previously demonstrated silicon-based high-intensity ultrasonic actuators for tissue cutting [2-4]. We have previously presented the concept of high-intensity ultrasonically driven silicon-based surgical tools with

integrated sensors. Silicon's material properties of high yield strain and high speed of sound make it suitable for high-intensity ultrasound applications. The ability to micromachine and integrate sensors makes silicon ultrasonic actuators more versatile and allow integration of several functions on the transducer.

In this paper we present results on integration of piezoresistive strain gauges, microscale cutting edges, and packaging advances to result in a multifunctional tool that might allow tissue characterization while cutting the tissue. The fabricated device is a bulk-micromachined surgical tools that allow both mm³ and μm^3 volume of tissue removal using piezoelectric actuators that can be driven at ultrasonic frequencies. Integrated piezoresistors are implemented on the silicon micromachined structure that can be used for sensing forces and closed-loop control of the ultrasonic amplitude. Spatial distribution of piezoresistors enables the measurement of spatial variations in strain enabling novel tissue characterization. With microfluidic channels for fluid sampling and delivery, this system has the possibility of integrating several surgical functions of cutting tissue at mm and μm scale, pumping cut tissue, sensing tissue stiffness and closed loop control of ultrasonic energy.

FABRICATION and ASSEMBLY

A schematic and photograph of the device is shown in Figure 1. The half needles are etched using the TMAH (Tetra-Methyl Ammonium Hydroxide) anisotropic etchant and bonded together to form the needle-like structure using silk-screened adhesive. Each half needle consists of integrated three strain-gauges using four polysilicon piezoresistors formed by the process shown in Figure 4. A 3" <100> wafer is coated with 1.0 μm of low-stress silicon nitride, which is later used as the mask layer for the TMAH etch. LPCVD polysilicon thin film (1.3 μm) is deposited on the silicon nitride and will form the piezoresistors. The polysilicon is doped heavily p-type by Boron implantation using 60 keV with a dose of $4.0 \times 10^{15} / \text{cm}^2$. After patterning the polysilicon layer in the shape of resistive elements, the layer is thermally oxidized to form oxide passivation. Contacts are etched into the polysilicon resistors followed by sputter deposition of 0.4 μm thick aluminum film. Aluminum interconnects to form the Wheatstone bridge and wires to the external electronics are defined. The exposed silicon nitride layer is now patterned in a front-back side alignment system to create the horn shape and the embedded

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channels inside the needle structure. The exposed silicon is etched in 5.0 % TMAH solution with silicic acid as an aluminum passivation additive. The TMAH etch was chosen to ensure aluminum passivation and future compatibility with integration of active electronic devices on the cutting tool. The piezoresistors are arranged in a Wheatstone bridge configuration such that the strain sensitivity in the axial direction is amplified over the other possible direction as shown in Figure 5.

The Wheatstone bridge also eliminates the RF coupling from the PZT electrode driven at large voltages. The piezoresistors were also placed at three locations along the length of the needle structure as shown in Figure 3. The distribution of the strain gauges allows the measurement of the ultrasonic strain and strain due to forces during tissue cutting.

Figure 1. The schematic and photograph of the silicon needle structure mounted on a PC board

A piezoelectric PZT (lead-zirconate-titanate) ceramic plate is adhesively attached at the center of the needle structure. The placement at the node and the length of the PZT are chosen to optimize the electromechanical coupling to the silicon structure [8].

The structure is designed to operate in its half-wavelength resonance. Hence every point along the structure vibrates except at the node. The piezoresistors are connected

to the external electronics via metal pins placed at the node. Physical wires placed anywhere else on the vibrating surface would effect the modal shape by mass and spring loading. Aluminum pads are shaped around a square anisotropically etched orifice as shown in Figure 1. The pins of a DIP package are fed through the orifice and silver-epoxied or wire-bonded to the aluminum interconnect. In addition to providing the electrical connections the pins also provide a rigid anchor for the vibrating structure. The device mounted on the DIP package can easily be placed on a PC board with electronics.

A silicon nitride blade is formed at the edge of the silicon structure by using a double-side exposure technique (Figure 2). The nitride blades were shadow coated with sputtered titanium/platinum thin film sandwich of 0.3 μm thickness. The metal coating substantially decreases the brittleness of the cantilever and makes possible the ability to penetrate tissue without breaking.

Figure 2. The cutting tip of the ultrasonic device can have silicon nitride blades for microscale cutting.

DEVICE CHARACTERIZATION

The ultrasonic response of the device can be characterized by measuring the electrical impedance versus frequency of the device. For example, the electromechanical coupling to the different bimorph and longitudinal modes can be obtained from the measured electrical impedance versus drive frequency at the PZT terminals. The $\frac{1}{2}$ -wavelength longitudinal mode was found at 112 kHz as predicted from an analytical model. The parameters of the equivalent circuits can be extracted from the impedance data. The quality factor of the resonance was measured to be 306 and K_{eff}^2 was .025. The product of the two quantities is the figure of merit of an ultrasonic transducer and is 7.7 for the transducer. The displacement versus applied drive voltage was measured optically, and found to be linearly dependent on drive voltage as 50 nm/ V_{pp} drive at the resonance frequency of 112 kHz.

The piezoresistor value obtained after the fabrication and annealing of the contacts was 20 k Ω . A variation in resistance value of approximately 4 % was measured across the wafer resulting from polysilicon thickness, doping, and oxidation variations. These small differences lead to DC offsets for the resulting in large DC voltages from the differential amplifiers. In future, the resistors can be laser-trimmed to eliminate the DC offsets.

The strain gauges were characterized by measuring the differential signal across the Wheatstone bridge amplified by an Analog Devices AD620 instrumentation amplifier (set at a gain of 30) as a function of applied forces. A force transducer (Sensym FS-03) was mounted against the needle structure, as the needle was pressed against the transducer, the calibrated readings from the Sensym device and the needle piezoresistors were measured. Careful attention was given to eliminate bending as that also produces a strain signal. In particular the tip piezoresistor has negligible bending even if bending moments are applied. Figure 6 shows the measured voltage from the instrumentation amplifier as a function of applied force. Using this data and assuming that the compression was uniaxial from the needle tip to the anchor pins, one can derive an expression for the gauge factor of the device to be

$$G = \frac{\Delta V}{2V} \frac{1}{F} * E_{Si} * A,$$

where V is the DC voltage across the Wheatstone bridge, F is the applied force, E_{Si} is the silicon Young's modulus, and A is the silicon needle cross section at the piezoresistor. Using this equation and the measured data, we obtain a gauge factor of 25, which is in very good agreement with value of 27 for similarly doped polysilicon piezoresistor published earlier [7].

ULTRASONIC STRAIN MEASUREMENT

The ultrasonic strain as a function of applied drive voltage was also measured. The differential strain value as a function of applied voltage to the PZT is shown in Figure 7. The strain value was calculated using the gauge factor of 25 and the measured ultrasonic differential signal. The strain values obtained using the piezoresistors agree with the value obtained from knowing the tip displacement and the mode shape of the $\frac{1}{2}$ -wavelength resonance. This data clearly demonstrates that the piezoresistors can be used to measure ultrasonic strain for feedback control of the device.

THIN FILM SILICON NITRIDE/METAL BLADES

In order to investigate the ability of the silicon nitride/metal composite blades to withstand cutting forces, the devices, mounted on a micrometer, were pushed into a swine retina. The silicon nitride cutting edge was able to penetrate the vitreous and the retina layer several times without any blade fracture (Figure 3). Furthermore, the cantilever was able to release retinal cells, much more so than when cut without the silicon nitride blade. Cutting results on other tissues such as onion resulted in precision cellular cutting (Figure 3). In both cases, applied ultrasound energy resulted in acoustic streaming-related fluid motion

and release of materials in the tissue. Notably blades made only of silicon nitride sometimes broke during cutting indicating the importance of metal films to increase blade toughness.

Figure 3. Top shows the penetration of the silicon nitride cutting edge into swine retina and releasing highly reflective cells. Bottom shows the silicon nitride blade cutting an onion cell along a controllable cut line.

CONCLUSIONS

We have presented a multifunctional ultrasonic transducer suitable for surgery that includes integrated strain gauges, silicon nitride blades, microfluidic channels, and through holes for easy electrical interconnects to external circuitry. The integrated sensors have been characterized for use in feedback control of the ultrasonic amplitude and measure forces during cutting tissue. One of the exciting possibilities of the integrated sensors is to measure the applied force to the silicon micromachine the surgical tool operator. Such a capability might allow the doctor to a highly sensitive version of the traditional percussive testing of tissue properties. The silicon nitride blades coated with Ti/Pt coatings were strong enough to penetrate soft tissues repeatedly. The device allows cutting tissue ultrasonically or statically, while sensing the tissue forces being applied. In the near future we plan to study the efficacy of the device in performing ophthalmic surgery in animal eyes.

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Figure 4. The fabrication process for the silicon cutter. (a) Low- stress silicon nitride deposition, (b) polysilicon deposition and ion implantation, (c) oxidation of polysilicon after patterning, (d) contact opening to polysilicon, (e) metal deposition, (f) after patterning aluminum, patterning silicon nitride film, and bulk silicon etching in TMAH

Figure 5. SEM of the polysilicon piezoresistors arranged in a Wheatstone configuration to measure strain in the axial direction of the transducer

Figure 6. Amplified differential output from the tip strain gauge as a function of applied compressive force on needle yields linear response and a gauge factor of 25 for the polysilicon piezoresistors

Figure 7. Measured strain at the middle strain gauge as a function of drive sinusoidal voltage across the piezoelectric PZT ceramic