BioImplantable Bone Stress Sensor

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Abstract—The clinical management of skeletal trauma and
disease relies on radiographic imaging to infer bone quality.
However, bone strength does not necessarily correlate well with
image intensity. There is a need for a safe and convenient way
to measure bone strength in situ. This paper presents a new
technique to directly measure bone strength in situ at a micro-
level scale through a MicroElectroMechanical System (MEMS)
sensor.

The proposed MEMS stress sensor comprises an array of
piezoresistive sensor "pixels" to detect stress across the interfa-
cial area between the MEMS chip and bone with resolution to
100 Pa, in 1 sec averaging. The sensors are located within
a textured surface to accommodate sensor integration into
bone. From initial research, surface topography with 30-60
µm features was found to be conducive to guiding new cell growth.
Finite element analysis has led to a sensor design for normal
and shear stress detection.

I. INTRODUCTION

Clinical management of fractures, bone grafts, and im-
plants relies on x-ray imaging to infer the biomechanical
quality of bone regeneration. Recent developments use non-
invasive vibratory stimulus to indirectly assess bone stiffness
and mass [1]. NASA reported a new device, Osteosonic™ [2], to
detect the status of damaged tissues and to monitor
bone fractures using a sensor that analyzes the vibration
response of bone tissue. Nogata et al. 2003 [3], reported
estimation of bone strength with ultrasound inspection. Even
though some of these methods have good accuracy, they only
use the bone density as an indicator of bone strength.

To directly assess the biomechanical properties of bone,
extensive work has been done using implanted strain gauges
to measure bone strains in vivo [4]. While useful as research
tools, these sensors are not practical for broad clinical use
due to cost, size, ease of use, and power issues. Regarding
scale, there are other qualities of bone at a micro-level scale
that affect the bone strength, including microarchitecture and
damage accumulation [5]. If clinicians had a practical means
to directly measure and quantify biomechanical properties of
healing or diseased bone in situ, within bone, this capability
could provide improved and timely information for treatment
management options, including drugs, fixation adjustments,
rehabilitation regimens, or pre-emptive surgical intervention.

Fig. 1. Integrated implantable bone sensor concept with SEM of the stress
sensor prototype. The envisioned final device incorporates a coil antenna on
chip. The inset is a prototype of 24 pixels.

A. Surface texturing

Microfabrication has enabled surface features with peri-
odicity, depth, and shape to be precisely controlled. In in
vivo results from Chehroudi et al. [6] mineralized tissue was found more frequently on micromachined grooved or pitted surfaces. Guided by this prior work, our group explored several test surface topographic characteristics using microfabrication techniques in silicon prototype chips with a titanium coating. Three basic types of surfaces were chosen: dimple, pimple and pimple/dimple. The texturing exploited the aspect ratio depth etching in the DRIE process (narrow openings in the layout have lower etch rate than wider openings). Figure 2 depicts the textures chosen with simple layouts consisting of an array of square frames with different spacing. Three depths of the topography, 15 µm, 37.5 µm and 60 µm, were created for each texture type.

![Figure 2. (a) Layout for pimple, dimple and pimple/dimple textures. (b) 3D surface topography generated with MATLAB.](image)

B. Microfabrication techniques

The chips were processed using DRIE. This is a dry silicon etching technique which involves the use of SF$_6$/O$_2$ in a low-pressure inductively coupled plasma (ICP) [7], [8]. A known characteristic of DRIE is its dependence of the etch rates on the aspect ratio of the trench, known as lag or aspect ratio-dependent etching (ARDE). Surfaces can be textured by exploiting ARDE to obtain different levels of heights in an array of microholes of different diameters with the use of one mask [9].

The process starts with a photolithography step that transfers the pattern to a top silicon oxide mask layer for the texturing in a 4"<100> Si wafer that is 400 µm thick. The first DRIE step is a timed etch 340 µm deep to define through holes for additional in vivo bone integration. The hole etch was masked with a second photolithography step over the oxide mask layer. After stripping the photoresist, a second timed etch defines the desired surface texture. The intended depth for the texturing is 60 µm at most. The last steps is an isotropic etch, to remove the 2 µm silicon walls left after the prior texturing step. An SEM of a pimple/dimple chip after processing is shown in Figure 3. The entire process flow has been repeated successfully.

C. Cell growth experiments

After the first generation of prototype chips were fabricated, samples were sputter-coated with a thin layer of titanium to provide bio-compatibility for in vitro experiments. The textured surfaces were compared to control groups (non-textured surfaces) to confirm the existence of cell growth on the chips [10]. Cellular attachment was assessed using human Adult Mesenchymal Stem Cells (hAMSC), a cell population isolated from bone marrow that possess the ability to differentiate into multiple cell lineages including osteoblasts. MG-63 cells, a human osteoblast-like cell line, were also used for in vitro assessment of cellular interactions with the prototype sensors. Initial studies show that hAMSC and MG-63 cells readily attach, proliferate and remain viable on the surface with all textures. Cell attachment is assessed using live/dead fluorescent staining and scanning electron microscopy (Figure 4). This study has shown high directionality on groove topographies. Further studies after 14 days in a medium containing an osteogenic supplement induced the differentiation of hAMSC into osteoblasts, and resulted in calcium deposition on the prototype sensor.

III. STRESS SENSOR DESIGN

This section summarizes the results from a set of ANSYS simulations to verify the mechanics of the sensor inside the bone environment.

A. Finite Element Analysis

A reduced model of the sensor array consists of a silicon substrate, two silicon pillars and a silicon dioxide beam,
enclosed in a cubic cortical bone (Figure 5). The material properties are summarized in Table I. The bone is subjected to a set of axial or shear loads, i.e. a compressive stress applied to opposite sides of the bone.

### Table I

<table>
<thead>
<tr>
<th>Material</th>
<th>Young Modulus, $E$ [GPa]</th>
<th>Poisson ratio, $\nu$</th>
<th>Density, $\rho$ [kg/m$^3$]</th>
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<tr>
<td>Cortical Bone</td>
<td>18</td>
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<td>2000</td>
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<td>Silicon</td>
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1The materials are modeled as isotropic.

Dependent on the selection of the piezoresistor arrangement relative to crystal orientation, the sensitivity is maximized. FEA provides the compressive or shear stress values to calculate (1). The two sets of piezoresistive arrangements in Figure 7 were studied. Selecting the [100] direction for $n$-type piezoresistors and [110] for $p$-type, gives the maximum sensitivity according to FEA results. For piezoresistors oriented in the [100] it is necessary to use a coordinate transformation to find the corresponding stress components with respect to the primed coordinate frame ($1', 2', 3'$).

### B. Piezoresistive Stress Detection

Piezoresistive silicon is designed through layout of doped resistor areas. The piezoresistance change for a two terminal piezoresistor can be expressed as (1):

$$\frac{\Delta R}{R} = \pi^{'}_{11}\sigma_1 + \pi^{'}_{12}\sigma_2 + \pi^{'}_{13}\sigma_3 + \pi^{'}_{14}\tau_4 + \pi^{'}_{15}\tau_5 + \pi^{'}_{16}\tau_6 \tag{1}$$

where $\pi^{'}_{ij}$ are the piezoresistive coefficients.

![Fig. 6. (a) Sample data line for chip enclosed in cortical bone. (b) Stress components along sample line due to 10 kPa compressive stress.](image)

![Fig. 7. (a) Piezoresistors oriented in the [110] direction. (b) Piezoresistors oriented in the [100] direction.](image)

The resistance change for piezoresistors in [110] orientation (Figure 7(a)) is:

$$\frac{\Delta R_1}{R_1} = \frac{1}{2}(\pi_{11} + \pi_{12} + \pi_{44})\sigma_1 + \frac{1}{2}(\pi_{11} + \pi_{12} - \pi_{44})\sigma_2 + \pi_{12}\sigma_3 \tag{2}$$

and in [100] orientation (Figure 7(b)) is:

$$\frac{\Delta R_2}{R_2} = \left(\frac{\pi_{11} + \pi_{12}}{2}\right)\sigma_1 + \left(\frac{\pi_{11} + \pi_{12}}{2}\right)\sigma_2 + (\pi_{11} - \pi_{12})\sigma_3 + \pi_{12}\sigma_3 \tag{3}$$

### C. Stress Tensor Derivation

A Wheatstone bridge configuration is selected to measure the change in resistance in order to cancel common mode influences, primarily temperature. From the structural analysis in section III.A, it is possible to design piezoresistance elements within a bridge to isolate a single component of
the stress sensor. Since the entire chip consists of an array of posts, the bridge circuits can be distributed to maximize the fill factor in one cell. The change of piezoresistance given an input load (compressive or shear) for all n-type and p-type element combinations was calculated through simulations. For a post with p-type piezoresistors oriented in the [110] direction (Figure 7(a)) the output voltage is:

$$V_{o1} = \frac{1}{4} \left[ \frac{\Delta R_{1b} + \Delta R_{3d}}{R_{1b} + R_{1d}} - \frac{\Delta R_{1a} + \Delta R_{3c}}{R_{1a} + R_{1c}} \right] V_s$$  (4)

Table II shows the output voltage for different bridge circuits. In bridges 4, 5, 6 the compressive loads applied to the bone model do not have any effect on the output. On the other hand, the shear loads will produce an output shift.

The output voltage from each piezoresistive bridge needs to be amplified, and care must be taken to design the amplifier with sufficiently low thermal and flicker noise. For a minimum stress of 100Pa, and using the longitudinal piezoresistive coefficient, the output voltage is 0.359mV. The calculated thermal noise is 1.92nV/rtHz for a single 10 µm piezoresistor, at room temperature and a 1Hz bandwidth. Adding a typical noise for a simple amplifier the signal to noise ratio (SNR) is 34.1 dB.

### TABLE II

<table>
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<th></th>
<th>$\Delta V_1$</th>
<th>$\Delta V_2$</th>
<th>$\Delta V_3$</th>
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<th>$\Delta V_5$</th>
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<td>[100]</td>
<td>[110]</td>
<td>[110]</td>
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<tr>
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</tr>
<tr>
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<td>0</td>
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### REFERENCES


### V. Conclusion

This paper presents a new approach to directly measure bone stress in situ at a micro-level scale through an implantable MEMS stress sensor. The array of piezoresistive elements provides the capability to extract the stress tensor in a material with statistical analysis of stress. A model of the sensor is developed and will be verified. This instrument may be used to validate bone stress data from FEA models at a micro-level scale, where the analysis is complex and separate phenomena occur, such as crack, friction or decohesion.

ACKNOWLEDGMENT

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