CARDIAC SIGNAL RECORDING USING ULTRASONIC SILICON MICROPROBES

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ABSTRACT

For the first time we report on a silicon-based microprobe integrated with ultrasonic horn actuator that successfully penetrated isolated and perfused canine heart at low driving voltage ($6V_{pp}$) and recorded simultaneously the electrical signals on multiple sites within the heart. The penetration into tissues was not possible without ultrasound, and was made predictable and reliable using ultrasonic actuation. These recordings provide information for reconstruction of the physioelectrical wave propagation within heart tissue which is important in the study of the mechanism and development of cardiac arrhythmias.

INTRODUCTION

Sudden cardiac death—mainly caused by ventricular fibrillation (VF)—is the leading cause of mortality in the industrialized world. Despite the chaotic nature of the disturbed cardiac rhythm during VF, researchers have hypothesized that just prior to VF, rotating spiral waves with distinct spatial and temporal order are generated and travel around the heart, breaking into multiple wavelets upon collision and eventually lead the heart to the chaotic state incapable of pumping and sustaining life. Such kind of spiral pattern has been observed on surface of isolated canine heart by staining the heart with voltage sensitive dyes [1]. However these studies could not provide information of the electrical activities underneath the surface. Since heart is a three dimensional entity with electrical pathways expanding in all three directions, it is important to obtain three dimensional mapping of physioelectrical wave propagation within cardiac tissue.

Silicon based microprobes have been reported for electrical activity recording in neural tissues [2, 3]. They provide high spatial resolution, reduced tissue damage, and the ease to integrate with microelectronics. These probes therefore become a good candidate for recoding cardiac signals inside the heart. However, these thin probes do not have enough flexural rigidity to go into the much denser and harder cardiac muscle tissues, when encountered with resistance from the cardiac tissue, they easily buckle and break. This is in contrast to neural tissue with much lower stiffness. Thicker probes provide greater rigidity but the increased probe dimension causes more damage to the tissue being investigated and may affect their physioelectrical activities.

We have reported before on microfabricated silicon ultrasonic surgical tools that reduce cutting force on biological tissues by actuating the surgical tool at its ultrasonic resonance [4]. In this paper we integrate ultrasonic horn actuator with microprobes that have multiple electrodes for 2-dimensional recordings on one device. Therefore by inserting a number of probes at different locations on the heart, we will be able to record the 3-dimensional distribution of electrical activities within the heart, and the wave propagation can be reconstructed and studied. Here we demonstrate the feasibility of measuring cardiac potentials, opening the way for the more involved 3D studies.

DESIGN AND FABRICATION

The design and fabrication of the silicon ultrasonic probe is similar to the ultrasonic surgical tools reported before [4], except that two tips (5mm long x $100\mu\text{m}$ wide x $70\mu\text{m}$ thick) are defined at the high-velocity section of the silicon ultrasonic horn. The Au/Cr electrodes on the tips are $20\mu\text{m} \times 20\mu\text{m}$, there are 5 electrodes on each tip with $500\mu\text{m}$ spacing, forming a 2 by 5 electrode array. The catenary silicon horn is 6cm long, with end-to-tip area ratio 15:1, and a longitudinal $\lambda/2$ resonance of 75kHz. The tips are of much smaller dimensions, so their longitudinal resonant frequency is much higher and will move in longitudinal direction at the same ultrasonic speed as the small end of the horn. To reduce unwanted bending and flexural motion of the tips, another silicon horn with same shape and dimension but without the two small tips are bonded to the one with tips to achieve symmetry in thickness direction. Figure 1 shows the entire process flow.

Figure 1. Process flow for fabricating the silicon ultrasonic microprobe. **a:** Evaporate and pattern Au/Cr metallization on 4" wafer with LPCVD nitride layer; **b:** Deposit and pattern PECVD oxide passivation layer; **c:** Pattern photoresist defining the probe tips; **d:** DRIE with the PR mask for desired probe tip thickness; **e:** Backside-only KOH etching to release the device; **f:** Another silicon horn without tips is bonded to the device for symmetry and reducing bending mode of the tips.

Next PZT-4 plates (6mm $\times$ 20mm) are bonded to the probe at the displacement node. The probe is clamped to a customized PC board, also at the location of the displacement node. The Au/Cr pads on the probe are wire-bonded to the PC board and then connected to external circuit through ribbon cable. Figure 2 shows the assembled device.

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Figure 2. Silicon ultrasonic microprobe packaged with PZT plates and printed circuit board

EXPERIMENTAL RESULTS

The silicon ultrasonic microprobe was tested on chicken thigh sample first with ultrasonic driving voltage 0~10V_{pp}. The device was attached to and driven by a motorized micromanipulator at constant cutting speed 200μm/sec. The sample was fixed on a load cell to record the force applied by the probe. The penetration force is the maximum force the probe tip encounters during a penetration event. Figure 3 shows a typical penetration force vs. ultrasonic driving voltage curve. The penetration force decreases approximately linearly with the driving voltage after an initial flat region, which is probably due to a threshold energy necessary for the ultrasound to take effect, and is worth further investigation.

Figure 3. Penetration force measurement of the ultrasonic microprobe on chicken thigh sample

On canine heart experiment, the probes can be inserted into the cardiac tissue with $\geq 6V_{pp}$ driving, at lower voltage the tips often buckle significantly and break. To test cardiac signal recording, perfused canine heart was used and all measurements were conducted within three hours after the heart was isolated from the subject. Diacetylmoxime was added to the perfusion fluid to reduce the muscle contractions while still allow normal electrical activities. Figure 4 is a photo showing the probe tips penetrated and embedded into the heart tissue. The signals were amplified in unipolar configuration and sampled at 1kHz. Figure 5 is the recorded signal on four electrodes on one tip during ventricular fibrillation. The signals are at higher frequency (~500Hz) than that of normal rhythm and less regular. Figure 6 shows the five channel signals recorded when externally pacing the heart with a 300Hz square wave stimulator. Closer examination in time scale showed that the time delay between the furthest apart electrode pair (2mm away) is approximately 5ms. These results showed that our ultrasonic microprobe is suitable for penetrating the relatively tough cardiac tissue and can provide multi-site recordings with

high resolution (currently 500μm which can be easily improved by reducing electrode spacing) and good signal quality.

Figure 4. The in-vitro canine cardiac recording experiment

Figure 5. Signals during VF from four electrodes on one tip. This is probing during ventricular fibrillation resulting in the measurement of the high frequency content.

Figure 6. Signals from five electrodes on one tip during external stimulation. The temporally shifted signals indicate that we are indeed measuring electric potential wave propagation along the probe.

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