Modeling and simulation of a monolithic self-actuated microsystem for fluid sampling and drug delivery

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ABSTRACT

A novel MEMS-based microsystem, including a microneedle array with a self-actuated structure for fluid sampling and drug delivery, is modeled, designed and simulated. The self-actuating mechanism and the microneedle array can be fabricated on a monolithic chip with a very reliable fabrication procedure. The microsystem is composed of a microneedle array at the center of the microsystem and an actuating mechanisms of symmetrically arranged Z-shaped PZT unimorph benders. The characteristics of the microsystem are simulated. The simulations show that a large displacement can be accomplished with a relative low actuating voltage. This monolithic microsystem opens up a wide application area for the commercialization of microsystems for fluid sampling and drug delivery.

Keywords: microneedle array, actuator, BioMEMS

1 INTRODUCTION

Hypodermic needles have played an important role in biological fluid extraction and drug delivery systems since they don not have the orally administered drug disadvantages of degradation in the gastrointestinal tract and/or elimination by the liver [1]. But, the disadvantages are also significant such as tissue trauma, insertion pain, difficulty in providing sustained drug release, and the need for expertise to perform an injection. In addition, bolus injections caused by hypodermic needles can lead to high concentrations of drugs being injected into the body and blood stream with the potential of toxic side effects. The need for new drug delivery routes and microbiological sampling has been recognized for some time, especially for new biotechnology drugs that cannot be administered using conventional approaches [2].

Z-shape benders are perfectly suited to actuation of microneedle arrays because they can enlarge and advance the distance of the array. In this paper we discuss the simulation of such an actuator and also include the modal resonance of the system in these simulations. The actuation issue is a required feature and also a considerable challenge for microneedle commercialization. Very few publications on microneedles with actuating mechanisms appear in peer reviewed publications, let alone both microneedle and actuation structures on a monolithic chip.

Microneedles based on MEMS technologies, which are a byproduct of the advances in microelectronics and integrated circuit technologies, can overcome the disadvantages above and provide pathways for drug delivery and fluid extraction across the skin, and offer further potential functionality.

The earliest microneedles were fabricated using solid silicon as microprobes for neural electrical activity recording [3]. Following that work, a variety of micromachined needle designs were reported [4]. Many different approaches have been employed, including surface micromachining [5], bulk micromachining [6], and LIGA techniques, and many materials have been used including silicon dioxide, silicon nitride, metal, plasma, plastic and others.

Actuation of microneedles is an important issue in real applications, but there are few publications in the literature relating to actuation. Microactuators represent major components for MEMS-based microsystems and can be driven by various forces suitable in the micro domain. The common actuation principles are based on: electrostatic [7], piezoelectric [8], electro-thermal [9][10], magnetic [11][12], electrochemical [13] effects at the device level. When choosing a driving mechanism, system level requirements for performance should be considered such as displacement, force, size, power, and energy consumption.

Magnetic actuators can produce both high force and large displacement, but it is more commonly used in macro actuators owing to scaling consideration and difficulties with integration into batch-fabricated micro assemblies. Most microacutuators exploit the other driving principles such as electrostatic forces.

Electrostatic actuation has been widely employed in various microstructures [14][15] where there are no requirements for special materials except for silicon and its oxides, and these are well established micromachining techniques that can be integrated with a variety of silicon-based sensors and circuits to form complete microsystems. However, electrostatic actuators are susceptible to particulates and

moisture, requiring high voltages, and very narrow gaps for large forces.

Polysilicon electrothermal microactuators have also shown considerable promise in MEMS applications because of their large displacement and force output capabilities combined with their ability to be driven at CMOS compatible voltages and currents. However, electrothermal actuators consume considerably more power than comparable electrostatic or piezoelectric actuation strategies.

Electrochemical actuation, which is based on the electrolysis of an aqueous electrolyte solution, is a relatively new principle. The reversible chemical reactions lead to gas evolution and gas pressure can be used to change the deflection of a membrane. Integration and package represent the major challenges for this method.

Piezoelectric actuation is another option for microactuators. The conversion of electrical energy into mechanical motion by piezoelectric thin films is a promising technique for microactuation applications. The multimorph piezoelectric actuator has several advantages including low energy losses, fast response, high induced forces, and easy geometrical adaptation in comparison with the electrostatic, electromagnetic, and electrothermic conversion mechanisms, respectively.

Silicon is an attractive choice for microactuators due to well established micromachining techniques available and the ability to be integrated with the variety of silicon-based sensors and circuits to form complete microsystems. Since silicon lacks magnetic, piezoelectric and other such properties, that are often exploited in mechanical actuators, electrostatic and electrothermal actuation are the primary options available.

In this paper, a microneedle array with a self-actuating structure on a monolithic chip is presented. The proposed microneedle array is fabricated by employing a bi-mask technique to facilitate very sharp tips with keen edges, a cylindrical body and side ports. The microneedle array uses a piezoelectric mechanism for actuation, and we present simulation results for the microneedle array.

2 DESIGN AND MODELING

The needle structure mainly defines the needle's properties. Due to the small dimensions of the microneedles, the fluid flow is quite small; thus, the high needle density of a microneedle array is needed. An out of plane microneedle array design can yield high needle density to to provide high fluid flow rate. In addition, the properties of the fabrication material should be also taken into account. Briefly: silicon dioxide is fragile; the strength of metal is good, but thin films of metal formed by sputtering or depositing are soft. The mechanics of microneedle insertion are also critically important for practical applications. Sharper needle tips can be expected to require less force for insertion, but the reduced penetration force comes at the expense of reduced strength near the tip. Only microneedles with the correct geometry and physical properties are able to penetrate skin without breaking or bending during insertion.

The geometries of the needle tip have a great effect on the forces required for insertion and fracture. Insertion force can be shown to be independent of wall thickness; thin-walled hollow needles and solid needles with the same outer tip radii require the similar insertion force [16]. Fracture force increases strongly with increasing wall thickness and increases weakly with increasing wall angle, but is independent of tip radius [16]. Therefore needles with small tip radius and large wall thickness are considered the best choice. Also, the blockage (clogging) problem upon insertion must be taken into consideration when designing microneedle structures.

We present a new monolithic self-actuated MEMS-based microneedle array. The basic structure of the microsystem is shown in Fig. 1

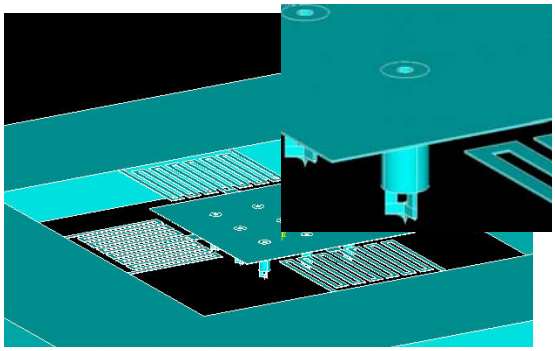

Figure 1 Structure of the microsystem

In order to realize a reliable side port design, and to be able to place the tip on the top of a cylindrical needle base, rather than directly on the wafer surface, the fabrication process employs a bi-mask technique [17][18].

3 FBRICATION

The process flow is shown in Fig. 2.

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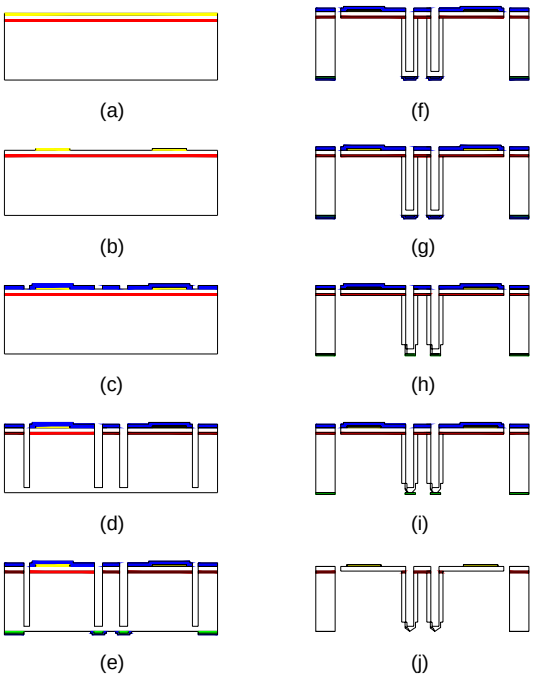

Figure 2 Process flow

In this work, a SOI silicon wafer is used. The PZT film is prepared by sol-gel method and the metal films are deposited as electrodes (Fig. 2a). After the PZT benders and electrodes are formed by etching, a thin film of SiO2 is patterned and the SOI wafer is etched to form the silicon benders (Fig. 2b). Following that, a thin film of SiO2 is deposited and etched as the mask to form the microneedle channels (Fig. 2c). The channels are anisotropically etched by anisotropic ICP, as shown in Fig. 2d. The next step is patterning the bi-mask (Fig. 2e), which is aligned to the center of the hole on the front side of the wafer. An anisotropic ICP step follows, forming the body of the needle (Fig. 2f). Then, the wafer is given another thermal growth of oxide (Fig. 2g), which forms a thin film of SiO2 on the surface of both needle bodies and channels and is used to protect both the needle bodies and channels for the subsequent steps, thereby ensuring the needle thickness. After the upper mask from the bi-mask is removed, the wafer is isotropically etched using ICP. Following that, a second anisotropic ICP is carried out. During this step, the side ports are formed (Fig. 2h). In order to guarantee a sharp needle tip, a third isotropic ICP is employed (Fig. 2i). Finally, the SiO2 is etched (Fig. 2j), and the processing is complete. Fig. 3 shows an SEM micrograph of one of the fabricated microneedle arrays.

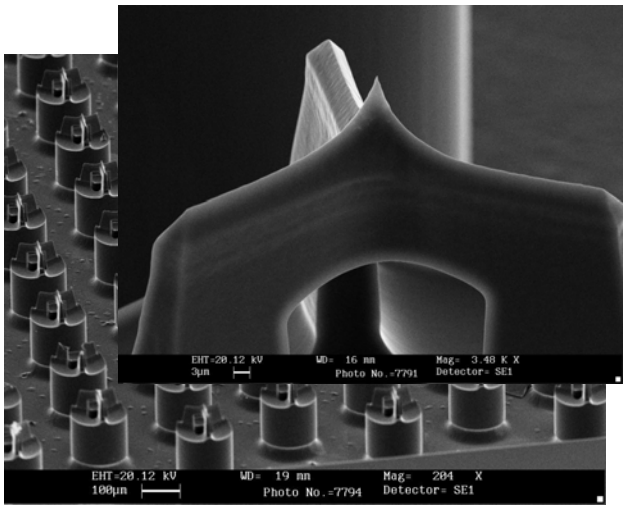

Figure 3 microneedle array

4 SIMULATION AND DISCUSSION

A series of FEM simulations were performed in order to characterize the microsystem in order to prove the adequacy of the actuating mechanisms and the efficiency of the system during actuation. The needle array plate can be actuated up to 700 m and more. Von Mises stress distribution and deformation of microsystem with five-levels are shown in fig.4.

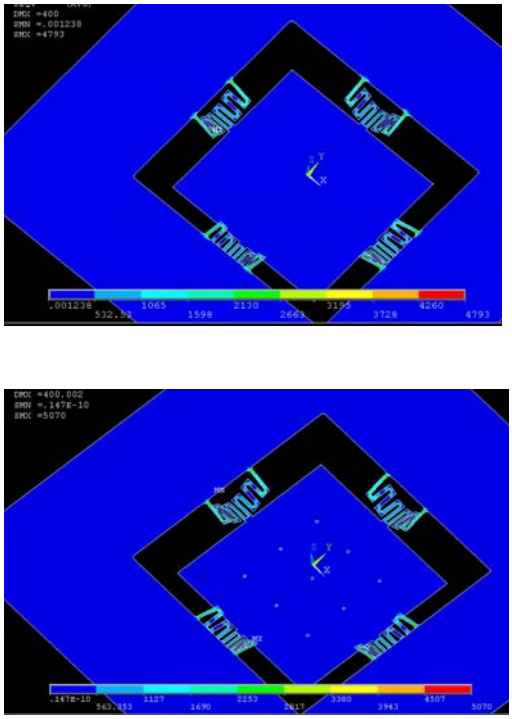

Fig 4 Von mises stress distribution and deformation of microsystem with five-level

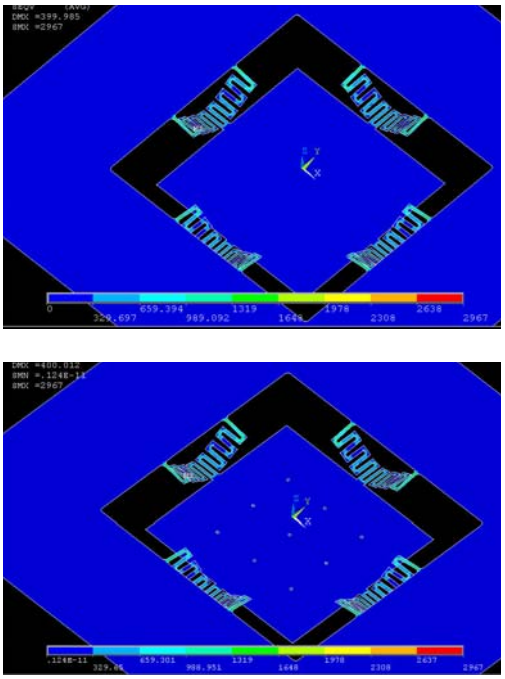

Fig 5 Von Mises stress distribution and deformation of a microsystem with seven-level

The stress distribution during actuation shows that the highest stress is concentrated where the levels cross. Low magnitude stress is found in the area of the needle array plate except at the linkage parts (Fig. 4). The deformation force of self weight is less than 1 nm and can be neglected. At the first mode resonance of the system, the microneedle array moves parallel up and down. The length of levels has a large effect on the stress.

Fig.5 shows the shape and stress distribution of the system, with seven-levels, during operation. The simulation clearly predicts that the needle array plate can maintain its initial shape during actuation and can advance the microneedle array with sufficient length (Fig. 5). The first five modal resonance frequencies of system with five-level are 2792, 4088, 4088, 6337, 10394 separately.

5 CONCLUSION

A self-actuating microsystem for fluid sampling and drug delivery is presented and simulated. The actuating mechanisms and microneedle array are on a monolithic chip to avoid having to use a bonding process. The large displacement can be accomplished through employing Z-shaped PZT unimorph benders. The properties of a target microsystem have also been simulated. This monolithic microsystem appears to provide a secure path towards commercialization.

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