A REMOTELY ADJUSTABLE MICROMACHINED CHECK-VALVE WITH A VARIABLE LENGTH CANTILEVER-BEAM STRUCTURE FOR IMPLANTABLE BIOMEDICAL MICROSYSTEMS

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ABSTRACT

In this paper, we present a remotely adjustable micromachined check-valve with a variable length cantilever-beam structure for implantable biomedical microsystems. The adjustable cantilever-beam structure is fabricated by gold thermo-compression bond of a thin silicon wafer over a glass substrate. The evaporated gold forms anchors on the silicon and strips on the glass substrate. Electrochemical dissolution of gold anchors and strips by a constant DC current (obtained via a telemetry link) is used to adjust the microfluidic characteristics of the valve (opening pressure and flow rate). The anchor pads are $10 \times 10 \mu\text{m}^2$ in size with $10 \mu\text{m}$ separation while strips are $50 \mu\text{m}$ wide and 1.5 mm long. A current density of 35 mA/cm^2 resulted in the release of gold anchors in 2 minutes with gold strips completely removed in 10 minutes.

Keywords: *Electrochemical Release, Implantable Biomedical Microsystem, Microfluidics, Microvalve, Thermo-Compression Bonding*

INTRODUCTION

Micromachined passive check-valves play an important role in a wide variety of implantable biomedical microsystems (e.g., drug delivery and therapeutic pressure relief devices) [1, 2]. In many applications, it is highly desirable to be able to remotely vary the opening pressure set-point over a period of time. For example, glaucoma drainage devices incorporate a passive check valve to reduce the intraocular pressure (IOP) [3]. The ability to remotely change the set-point after surgery provides surgeons more flexibility in regulating the IOP [2, 3]. We previously reported an adjustable SU-8 micromachined check-valve array based on electrochemical dissolution of gold membrane [2]. In this paper, we report on an alternative approach using a variable length cantilever-beam structure. Figure 1 shows a perspective view of the microvalve along with its fluidic schematic. It consists of a silicon cantilever beam bonded to a glass bottom plate containing an ultrasonically drilled inlet hole. The bonding is done using gold thermo-compression technique over selected areas defined by an array of gold anchor pads on the silicon substrate and thin gold strips on the glass substrate. The gold anchors and strips define various lengths and hence opening pressures for the valve. The

strips can be electrochemically removed in sequence, lowering the opening pressure accordingly.

Figure 1. A perspective view of the microvalve along with its fluidic schematic.

DESIGN AND FABRICATION

Figure 2 shows the fabrication sequence. The silicon substrate fabrication process starts with a thermal oxidation step of a thin silicon substrate ($200 \mu\text{m}$ thick). A $0.5 \mu\text{m}$ silicon dioxide is grown to prevent silicon diffusion during thermo-compression bond [4]; this also serves as an isolation layer during the electrochemical dissolution. The backside oxide layer is patterned for the final etching step. A Ti/Au ($10 \text{ nm}/1.0 \mu\text{m}$) bonding layer is then deposited by electron beam evaporation and patterned on the silicon wafer by lift-off process to create an array of gold bonding pads. The glass fabrication process starts with the deposition and patterning of the Ti/Au ($10 \text{ nm}/1.0 \mu\text{m}$) bonding strips on a Pyrex 7740 glass substrate. A flow channel of $600 \mu\text{m}$ in diameter is then ultrasonically drilled on the glass substrate. The bonding surfaces of the two substrates are exposed to oxygen plasma for 5 minutes to completely remove organic remains before wafer alignment. During the alignment process of two substrates, three spacers and three clamps on a bonder fixture are loaded consequentially to separate the wafers and hold them in position. The gold-to-gold thermo-compression is carried out in Karl Suss SB6 VAC bonder (Suss MicroTec, Germany) under atmospheric pressure. Following the withdrawal of the clamps, the temperature is ramped up and stabilized at 350°C for 30 minutes, 0.01 MPa pressure was applied over the wafer and then spacers are withdrawn.

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Then a bonding pressure of 0.1MPa is applied across the two substrates for 15 minutes (corresponding to 5MPa of pressure on the gold anchor pads). Afterwards, the temperature is ramped down slowly [4]. To machine the silicon substrate into a cantilever shape, we use deep reactive ion etch (DRIE) from the backside of the silicon wafer. Finally, the devices are diced and separated into individual 8x8mm² chips. The overall size is compatible with the currently implanted glaucoma drainage devices, *e.g.*, Ahmed glaucoma valve (which is the most widely used implantable devices) is 13mm wide, 16mm long, and 2mm thick.

Figure 2. Fabrication process includes: (1) thermal oxidation of silicon wafer; (2) e-beam evaporation and patterning of Ti/Au; (3) ultrasonic drilling of the flow channel; (4) thermo-compression bonding of Si-Glass; (5) DRIE Si from the backside followed by wafer dicing.

The release and activation mechanism is based on the electrochemical etching of gold anchors. Gold is a biocompatible metal and its ionic compounds have many medical applications (*e.g.*, Au^I salts used for treatment of rheumatoid arthritis). It has also been shown that one can dissolve gold in a saline solution by applying + (0.8~1.2) volts (with respect to a saturated calomel reference

electrode). Gold dissolution occurs through the formation of water-soluble chloro-gold III complexes (Au^{III}) [5, 6]. An array of 10x10μm² gold anchor pads with 10μm separation is used in our design. On the glass substrate, the gold pattern is a continuous narrow strip of 50μm wide connected to the outside pads. Figure 3 shows photographs of the front and the backside of the adjustable micromachined check-valve.

Figure 3. Photographs of (a) the front view and (b) the backside view of the adjustable micromachined check-valve.

EXPERIMENTAL SETUP AND RESULTS

As was mentioned previously, in our design, electrochemical dissolution of gold anchors is used to adjust the length of the cantilever beam, and therefore change the fluidic characterization of the microvalve. This is accomplished by a constant DC current obtained via a telemetry link. Figure 4 shows the test and measurement setup. For present experiments, we are using surface mount devices (SMD) to implement the receiver circuitry with a CMOS integrated version currently under design and fabrication. An AC source (4MHz, 20V_{p-p}) is used to inductively power a receiver. The received voltage is used to generate a constant current for the electrochemical dissolution. The receiver circuitry consists of a LC tank, a full wave rectifier (RH02 DICT-ND, Diodes Inc.) followed

by a capacitive filter and an adjustable current source (LM334M-ND, National Semiconductor). A current density of 35mA/cm² resulted in a release of gold anchors in 2 minutes and gold strips are completely removed in 10 minutes. Figure 5 shows microscopic photographs of the valve gold strips before and after complete release.

Figure 4. Schematic of the telemetry experimental setup and receiver circuitry.

Figure 5. Microscopic photographs of a gold strip on the valve (a) before and (b) after release.

Both gravity and syringe-pump driven flow were used to characterize the valve performance [7]. These can be easily achieved by using the following setup shown in Figure 6. The setup consists of a syringe pump (KDS Inc.) connected to a glass standpipe (1.1 mm ID) or a pressure sensor via a 3-way stopcock. The third outlet is connected to the valve through a silastic tubing. All tests are carried out at the atmospheric pressure and room temperature.

In the gravity-driven flow tests, the branch connected to the syringe pump is closed and the valve is subjected to the gravitational force of the fluid head, which forces the valve open and drains the fluid out until the gravitational head is reduced to the level at which the valve is closed. The flow rate through the device can be calculated as

Figure 6. Schematic of *in vitro* microfluidic test setup.

$$Q = A \frac{dh}{dt} \quad (1)$$

where A is the cross-section area of the standpipe and h is the height of the fluid in it. The flow resistance, defined by the pressure drop P over the flow rate Q , can be obtained from

$$R = \frac{P}{Q} = \frac{P}{A \frac{dh}{dt}} \quad (2)$$

This configuration allows the flow resistance to be measured over a wide range of flow rates. Figure 7 shows the measurement results of the pressure head (left) and flow resistance (right) vs. flow rate of a microvalve with the cantilever flap having a width of 1.5mm and a length of 4mm. As expected, the valve exhibits nonlinear fluidic behavior due to the pressure response of the cantilever beam structure, *i.e.*, the higher the pressure, the larger the displacement of valve flap, and therefore the much larger the flow rate and smaller the flow resistance. Figure 7 also indicates a closing pressure of 7mmHg.

Figure 7. The measurement results of the pressure head (left) and flow resistance (right) vs. flow rate of a microvalve with the cantilever flap having a width of 1.5mm and a length of 4mm.

In the syringe-pump driven tests, the syringe pump provides a constant flow rate and pressure variation across the valve is monitored using the standpipe or a pressure sensor. This configuration also simulates the operation of pressure activated passive check valves encountered in actual physiological settings (e.g., constant aqueous humor inflow in the eye) [8]. Following the dissolution of each anchor, deionized (DI) water at 10 μ L/min flow rate is injected into the microvalve. Figure 8 shows the pressure measurement results for the check-valve illustrating the basic forward operation of the device. Opening pressures of 178, 78, 26, 12mmHg and stabilized pressure of 145, 72, 22, 7mmHg show four-stage performance by releasing the gold strips sequentially.

Figure 8. The pressure measurement results for the check-valve characteristic tests illustrating the basic forward operation of the device.

CONCLUSIONS AND FUTURE WORK

In this paper, we report on a remotely adjustable micromachined check-valve with a variable length cantilever-beam structure for implantable biomedical microsystems. The adjustable cantilever-beam structure is fabricated by gold thermo-compression bond of a thin silicon wafer over a glass substrate. A current density of 35mA/cm² resulted in a release of gold anchors in 2 minutes and gold strips are completely removed in 10 minutes. Both gravity and syringe-pump driven flow were used to characterize the valve performance. Nonlinear valve behavior and four-stage valve behavior have been clearly

demonstrated. A CMOS integrated version of the receiver circuitry is currently under design and fabrication.

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