ME381 Introduction to MEMS

Term Project

Dynamic Wettability Switching by Surface Roughness Effect

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DYNAMIC WETTABILTY SWITCHING BY SURFACE ROUGHNESS EFFECT

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ABSTRACT

This paper presents a novel wettability switching mechanism for micro-droplet transportation by surface roughness effect. A roughness switchable device based on softlithography with Polydimethylsiloxane (PDMS) was fabricated and tested. It was found that the surface wettability can be dynamically switched from medium hydrophobic to super hydrophobic states by deflecting the membrane with a pneumatic method. Electrostatic actuation method will be utilized in our future device to achieve addressable micro-droplet manipulation. Finite Element Analysis software packages, ANSYS and ABAQUS, were used to simulate the membrane deflection under air pressure and electric/stress coupled field.

INTRODUCTION

Much attention has been paid to the surface-tension-driven actuation mechanism for its applications in micro-fluid handling techniques. A liquid-based micro optical switch was operated by thermally controlled surface tension known as thermocapillary [1]. A liquid handling system has been fabricated to move droplet by the electric control of surface wettability known as electrowetting [2]. However, there are apparent disadvantages of the above mechanisms in many applications because heat and electric potential may affect particles such as biomolecules in the solution.

This paper demonstrates a new type of surface-tension-actuation mechanism that uses purely mechanical means to control wettability to achieve liquid motion. The proposed method is more suitable to many applications including biological ones since it has no thermal or electrical effects on the content of the solution. Our device specifically targets an on-chip addressable micro-droplet manipulation for biomolecular mixing and amplification.

THEORY

The effect of surface roughness on wettability was studied more than a half century ago [3, 4]. Recently, more interest has been stimulated by experimental results showing that wettability can be tuned by surface geometry [5, 6].

Fig. 1 shows the wettability difference due to microfabricated surface roughness. Gold was e-beam evaporated on the surface, followed by vapor phase coating of a self-assembled monolayer.
According to Laplace-Young equation [7], there exists a pressure difference inside a droplet when it is deposited at a boundary between flat and rough surfaces due to the difference of radii of curvatures by contact angle difference. This pressure difference can cause the movement of microscale liquid, as shown in Fig. 2, where a micro droplet is moving across the boundary between flat (hydrophobic) and roughened (superhydrophobic) areas of Fig. 1.

**Fig. 1** Wettability shift due to roughness.

![Wettability shift due to roughness](image1.jpg)

(a) Flat surface  (b) Roughened surface

Fig. 2 Droplet motion across different wettability regions.

**REVIEW OF SOFT LITHOGRAPHY**

Our device is fabricated based on soft lithography technique. It is a non-photolithographic strategy based on self-assembly and replica molding for carrying out micro/nano fabrication. It provides a convenient, effective, and low-cost method for the formation and manufacturing of micro/nano structures.

PDMS is most widely used as "stamps" with patterned relief on the surface to generate features to generate patterns and structures with feature size ranging form 30 nm to 100 µm. The following techniques have been demonstrated [8]:

![Droplet motion across different wettability regions](image2.jpg)
**Microcontact Printing** (µCP) (as shown in Fig. 3). An "ink" of alkanethiols is spread on a patterned PDMS stamp. The stamp is then brought into contact with the substrate, which can range from coinage metals to oxide layers. The thiol ink is transferred to the substrate where it forms a self-assembled monolayer that can act as a resist against etching. Features as small as 300 nm have been made in this way.

**Fig. 3** Schematics of Microcontact Printing (µCP) process.

**Fig. 4** (a) Schematics of Replica Molding (REM) process, (b) final product.
**Replica Molding** (as shown in Fig. 4). A PDMS stamp is cast against a conventionally patterned master. Polyurethane is then molded against the secondary PDMS master. In this way, multiple copies can be made without damaging the original master. The technique can replicate features as small as 30 nm.

**Micromolding in Capillaries** (MIMIC, as shown in Fig. 5). Continuous channels are formed when a PDMS stamp is brought into conformal contact with a solid substrate. Capillary action fills the channels with a polymer precursor. The polymer is cured and the stamp is removed. MIMIC is able to generate features down to 1 µm in size.

![Fig. 5 Schematics of Micromolding in Capillaries (MIMIC).](image)

**Microtransfer Molding** (μTM). A PDMS stamp is filled with a prepolymer or ceramic precursor and placed on a substrate. The material is cured and the stamp is removed. The technique generates features as small as 250 nm and is able to generate multilayer systems.

Applications of soft lithography in the near future could include simple optical devices, such as polarizers, filters, wire grids, and surface acoustic wave (SAW) devices. Long-term goals include working toward optical data storage systems, flat panel displays, and quantum devices. The following example (Fig. 6) about Microcontact printing (μCP) reveals its application with micro fluidic networks (μFN) to pattern substrates with proteins.
Fig. 6 Microcontact printing (μCP) and microfluidic networks (μFN) are powerful techniques to pattern substrates with proteins. Examples of applications of these techniques:

(a) fluorescence from a patterned immunoglobulin G monolayer on a glass slide created by μCP;
(b) AFM image of a small stamped feature of antibodies on a silicon wafer;
(c) a neuron and its axonal outgrowth on affinity-stamped axonin-1;
(d) repetitive stamping of different proteins onto the same plastic substrate;
(e) water condensation pattern on micropatterned albumin forming droplets of ~2 μm in diameter;
(f) fluorescence micrograph of different proteins patterned by μFN.

There are, however, important unresolved problems (1) PDMS shrinks upon curing and swells in a number of non-polar solvents, which makes it difficult for high resolution molding, (2) the elasticity and thermal expansion of PDMS limit the accuracy in registration across a large area and application in multilayer fabrication, (3) The softness of an elastomer limits the aspect ratio of microstructures in PDMS.

**FABRICATION PROCESS**

Major fabrication steps are illustrated in Fig. 7. First, a hydrophobic silicone elastomer, PDMS is molded on a silicon wafer with SU8 pattern (a) (b). Then another PDMS mixture is diluted with hexane and spin-coated on a silicon wafer with photoresist pre-coated on it (c). Next, the PDMS rough substrate and thin PDMS membrane are treated by oxygen plasma for one minute, and brought into contact to form an irreversible bond (d) [9]. Finally, the photoresist is dissolved. The resulting device is a thin PDMS membrane bonded on top of the PDMS rough substrate (e). Fig. 6(f) shows the schematic
of the deflected membrane which is actuated by suction through a thin air channel at the corner of the patterned area, as shown in Fig. 6(g).

![Diagram of fabrication process](image)

**Fig. 7** Fabrication process for a membrane device.

Fig. 8 shows the PDMS pillars and the suspended membrane imaged by scanning electron microscopy (SEM) and optical microscopy. The membrane thickness is about 1.5µm. To our knowledge, it is nearly one order of magnitude thinner than those reported in literature [10].
DEVICE TESTING

In the experiment, suction was applied on the membrane through the air path next to the patterned area, as shown in Fig. 7(g). In our process, it is critical to fabricate a thin suspended membrane since it offers the minimum structure dimension variation after actuation. Thin membrane is also easier to deflect. Fig. 9 shows the membrane device...
under testing. A color difference can be clearly observed when the membrane is actuated (deflected) by an applied suction.

![Actuation of a membrane device.](image)

As a result of the membrane deflection, contact angle was changed as shown in Fig. 10. First, the contact angle on rough and flat PDMS surfaces without membrane was measured as reference values, which are 144.5° and 113.2° respectively. Then the contact angle on the membrane device was measured when it was actuated and released as shown in Fig. 10. The experimental results show good agreement with the theoretical predictions. It is also confirmed that the droplet can move across the boundary of the patterned area.

![Contact angle measurement when membrane is actuated and released.](image)

**Fig. 9** Actuation of a membrane device.

**Fig. 10** Contact angle measurement when membrane is actuated and released.
FUTURE WORK

Since our final device targets addressable on-chip micro-droplet manipulation, electrostatic actuation will be used to control the membrane deflection. The device schematic is shown in Fig. 11. A micro droplet is squeezed between a top glass coated with hydrophobic film and a bottom microfabricated rough substrate with suspended hydrophobic membrane on top. Electrodes are embedded on top of the membrane and at the depressions in the bottom substrate. When the membrane is deflected by applied potential, the left side of the droplet is in contact with roughened surface which is superhydrophobic and the right side is on the flat surface which is medium hydrophobic. The pressure gradient induced by surface tension difference can move the droplet to the right direction in this configuration.

![Fig. 11 Cross section of transport mechanism.](image)

Fig. 12 shows the major fabrication steps for the device described above.

![Fig. 12 Fabrication process for membrane actuation.](image)
SIMULATION

Pneumatic actuation case
The current pneumatic actuated chip does not meet our requirement in terms of film deflection magnitude. With finite element simulation running on ABAQUS, we justified that the current air pressure is enough to pull in the PDMS film and raised the question on the structure of the pneumatic system.

1) Modeling

Due to the symmetry of the film-pillar structure, we only model one quarter of a unit to save CPU time. As shown in Fig. 13, a = b = 25 µm and the thickness is 1 µm. Boundary 1 and 2 are fixed while 3 and 5 are not allowed to move along y direction and 4 and 6 are not allowed to move along x direction. The air pressure is applied only on one side of the film. The PDMS material is modeled as elastic material with $E = 0.75$ MPa and $\nu = 0.49$. The target z displacement of the central part of the film is set to be 25 µm.

2) Solutions

ABAQUS-S4R reduced 4 node shell element was used. The total number of elements is 281 which guaranteed the deformation can be captured without too large mesh distortions. Nonlinear solution was used due to the geometric nonlinearity expected under large deformation.

The pressure vs displacement curve is as shown in Fig. 14 and Fig. 15 is the deformation shape under the pressure of 500 psi. ANSYS has also been used to verify ABAQUS’s solution. Their results match very well.

From the result, we see the air pressure of the atmosphere is large enough to pull in the film. However in our experiment, we only observed very little amount of

Fig. 13 Dimensions of the PDSM Film.
displacement, which indicated that there are fundamental limitations of the pneumatically actuated device.

**Electrical actuation case**

Due to the complexity brought by the vacuum system used in pneumatic actuation, the electrical actuated device has been proposed in previous section. One of the key control parameter is the applied voltage since high voltage may affect the biological solution being analyzed. To determine the appropriate voltage range, electrostatic/structural coupled finite element simulation was performed.

Since ABAQUS is not able to solve electrostatic/structural coupled problems, we only use ANSYS for this case. The ANSYS sequentially coupled solver was adopted. Results from structural analysis become loads for the next electrostatic analysis step. Then the result of the electrostatic analysis will change some input to the structural analysis. The iteration continues till the equilibrium is achieved. Before each iteration step, element morphing would be performed to guarantee the element compatibility along the boundaries.

1) Modeling

The force generated by electrical field decreases dramatically when the gap between two conductors increases. So we change the dimensions in Fig. 13 for electrical actuated chip. In this simulation, \( a = b = 2 \, \mu\text{m} \), the thickness is \( 1 \, \mu\text{m} \). The gap between film and ground conductor is \( 3.3 \, \mu\text{m} \). Boundary conditions remain same.

2) Solutions

ANSYS Solid122 and Solid95 elements have been adopted to model the film in electrical and structural analysis, respectively. Shell elements are not available for multiphysics analysis. Triangular meshing was used for air component and brick meshing was used for the film. To ensure the compatibility along the boundaries, their mesh sizes were set to be close. The nonlinear geometric option was turned on due to the large deformation. Time step increment has been specified to \( 1/10 \) of the total step time. At the simulation, the film cannot be deformed to contact the ground conductor otherwise the convergence would not be achieved. We obtained the closest z-displacement of \( 3.24 \, \mu\text{m} \).
Fig. 16 shows the critical height of 3.3 µm can be obtained with the applied voltage greater than 115V. The pull-in effect can be observed from the curve, as the displacement goes up rapidly when the voltage surpasses 100V. Fig. 17 shows the deformed shape under voltage of 90V. The suggested control voltage range from this simulation is 120±5V.

![Fig. 16 Voltage vs z Displacement](image1)

![Fig. 17 Deformed Shape by ANSYS](image2)

For the Pneumatic case, our simulation results indicated that there are fundamental limitations of the device structure. The reason, we speculate, is that the membrane above the air path (Fig. 7(g)) collapses first once the suction is applied. This will block the suction and stop the further deflection of the membrane. New design of pneumatic actuation structure is needed to provide enough membrane deflection.

For electrostatic case, our simulation predicted the appropriate voltage range to deflect the film. The structure optimization can be performed in future with the help of FEM simulations. The contact pair of the lower surface of the film and the upper surface of the pillar can be added to predict more accurate results. The fillet radius would be determined by the art of fabrication process. However, larger fillet radius does provide less stress concentration and less convergence problem for FEM simulation.

**CONCLUSION**

A novel surface wettability switching mechanism for on-chip micro-droplet manipulation based on surface roughness modification was proposed. A membrane device based on soft lithography technique with PDMS was tested and it was found that the surface wettability can be dynamically switched from medium hydrophobic to superhydrophobic by deflecting the membrane with pneumatic means. Finite Element Analysis software package, ANSYS and ABACUS, were used to simulate the membrane deflection under air pressure and electric/stress coupled field respectively. The appropriate voltage for membrane pull-in was obtained.
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REFERENCES


BIOGRAPHY

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