UNIVERSITY OF CINCINNATI

Date: May 21, 2004

I, Chunyan Li,

hereby submit this work as part of the requirements for the degree of:

Master of Science

in:

Electrical and Computer Engineering and Computer Science

It is entitled:

Polymer Flip-chip Bonding of Pressure Sensors on Flexible Kapton Film for Neonatal Catheters

This work and its defense approved by:

Chair: Dr. Chong H. Ahn

Dr. Ian Papautsky

Dr. Joseph H. Nevin

Polymer Flip-chip Bonding of Pressure Sensors on Flexible Kapton Film for Neonatal Catheters

A thesis submitted to the

Division of Research and Advanced Studies of

the University of Cincinnati

In partial fulfillment of the requirements for the degree of

Master of Science

In the Department of Electrical and Computer Engineering and Computer Science of

the College of Engineering

2004

by

Chunyan Li

B.S., YanBian University of Science & Technology, 2001

Committee Chair : Dr. Chong H. Ahn

ABSTRACT

The object of this thesis is to develop a new packaging method to mount silicon micro pressure sensors into 1.67 mm diameter neonatal catheters to measure blood pressure. A new polymer flip-chip bonding method on a flexible Kapton film has been developed and applied to a dual lumen neonatal catheter integrated with silicon micro pressure sensors.

Flip chip bonding technique has inherent advantages of miniaturization, improved reliability and cost reduction. Conductive polymer flip-chip bonding technique using micromachined conductive polymer bumps showed a very low contact resistance, simple processing steps, a high bumping alignment resolution (< ±5 μm) and a lower bonding temperature (~ 170 ºC).

In this work, Kapton film (Polyimide) has been chosen as a substrate material due to its biocompatible, flexible, good moisture resistance and low cost characteristics. A new polymer flip-chip bonding technique has been developed and successfully characterized in terms of conductivity and bonding strength, achieving a low temperature process and improved tolerance of thermal stress, which are very desirable for mounting micro sensors or micro actuators on lower-cost flexible polymer substrates for medical applications. Finally, silicon micro pressure sensors have been fully mounted on the flexible Kapton film with metal lines for neonatal catheter and then successfully characterized in both gas and liquid environments, which shows the bonding and packaging techniques developed in this research are suitable for the neonatal catheters.

ACKNOWLEDGEMENT

Firstly, I would like to thank the Lord for the big grace that He has provided me and all the so good people that He has given in my life.

Above all, I would like to express my great appreciation to my advisor, Dr. Chong H. Ahn for his advice, encouragement, inspiration, guidance, prayers and support throughout my graduate study. I am always very grateful for the God to have known him and can study under his guidance in my life. He has been and would be my lifelong mentor and role model in research and my personal life.

I would like to express my deep appreciation to Dr. Ian Papautsky for his valuable suggestions and comments, guiding me throughout this project. Also, I would like to deeply thank to Dr. Richard Azizkhan from the Department of Pediatric Surgery at the Children's Hospital Medical Center in Cincinnati, who provided most information desired for neonatal catheters and financially supported this research work.

There are so many people who made this work possible. I would like to thank all the members of Microsystems and BioMEMS Lab for their tremendous support. Special thanks go to Frank. Sauser for testing bonded sensors, Nikhil and Dongsung Lee for bonding strength measurement. Jeff Simkins has made many equipments worked out, Xiaoshan and Robert Cole has helped me to learn and fix laser micromachine. Rhonda has provided me many materials I needed for my experiments. Many thanks go to Xiaoshan Zhu, Chuan Gao, Junhai Kai, Aniruddha, Red Hong, Jungyoup Han, Rong

Rong, Sehwan Lee, Jaephil Do, Soohyun Lee, Alok, Phalgun, Shilpa and Dongsung Lee for their various contributions in my work.

I deeply thank Dr. Duke H. Park, Dr. In Yong. Park and other professors in YUST for their guidance and strong prayers for me. I deeply thank all members at the Korean Power Mission Baptist Church and Cincinnati Chinese Church for their love and prayers.

Finally, I would like to express my deep thanks to my parents, dear sister and Hong's family for their support, love and prayers. They always stand with me.

List of Figures

CHAPTER 1. INTRODUCTION

MEMS (Micro-Electro-Mechanical Systems) technologies have been considered as the most influential enabling technologies for the integration of almost any phenomena – motion, light, sound, chemical detection, radio waves and computation, but on a single chip. MEMS combine and integrate complete systems, but one major cost driver in today's MEMS is its packaging, which has also been one of the most difficult challenges. The advanced MEMS devices are not only small and highly complex, but many must communicate with the outside world by methods other than electronic signals. Further, MEMS devices are so specific that each can represent a novel problem to solve. These factors imply that new ways of packaging have to be developed. In order to measure blood pressure using neonatal catheters, in this thesis, a new packaging method to mount silicon micro pressure sensors on a flexible Kapton film has been developed and characterized for a dual lumen neonatal catheter.

1-1. FLIP-CHIP TECHNOLOGIES IN MEMS PACKAGING

Flip-chip attachment is a well-known technology since the early 1960s, but now it is experiencing a huge development due to the commercial interest in high density packaging. Flip-chip microelectronic assembly is the direct electrical connection of face-down (flipped) electronic components onto substrates, circuit boards, or carriers, by means of conductive bumps on the chip bond pads. In contrast, wire bonding, the older technology which flip chip is replacing, uses face-up chips with a wire connection to each pad. Flip-chip technology has been of much attention in the microelectronics industry in recent years because they not only make products thinner, smaller and lighter but also offer good electrical performance, improved reliability, lower costs, and higher input/output (I/O) density due to short interconnection length [1], [2]. Flip-chip technology achieved a higher level of acceptance for many different applications. For the Flip-chip assembly, many different technologies have been developed. They may be categorized into three divisions; namely, solder bump based, stud bump based, and adhesive based.

Recently, electrically conductive polymers have received considerable interest as they offer numerous advantages over traditional solders in flip-chip packaging because there is a significant reduction in pitch, weight and volume, flexible, simple process, fluxless formulations and also has excellent environmental compatibility [3]. Polymer flip-chip also allows the use of lower-cost substrates that cannot tolerate the high processing temperatures required for solder reflow. Electrically conductive polymer provides both electrical and mechanical interconnects between chip and the supporting substrate as like solder alloys. There are isotropically conductive polymer bonding and anisotropically conductive polymer boding. The polymer flip-chip bonding technique utilizes silver-filled conductive epoxy to electrically connect the contacts of electronic components [4], [5], [6]. A key factor in the realization of direct sensing and low-cost flip-chip technology is the use of isotropic electrically conductive adhesives. Figure 1-1 illustrates flip-chip bonding system using an isotropically conductive polymer. In the system polymer matrix provides the mechanical joint and metallic pads and metallic particles provide the electrical connection for the flip-chip boding.

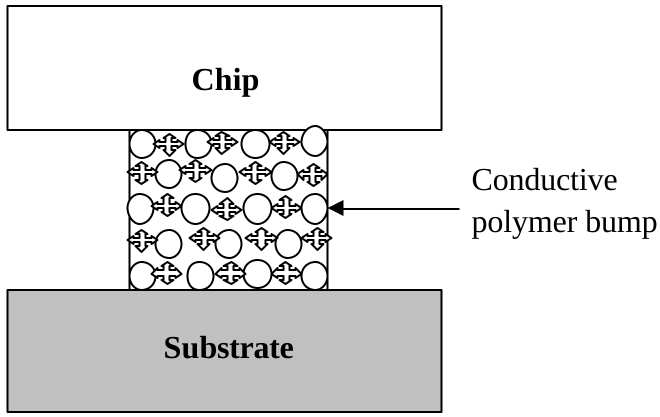


Figure 1-1. *Schematic diagram of flip-chip bonding system with isotropically conductive polymer.*

1-2. RESEARCH OBJECTIVE

The object of this thesis is to develop a new packaging method to mount silicon micro pressure sensors into 1.67 mm diameter (5 French) neonatal catheters to measure blood pressure. A new polymer flip-chip bonding method on a flexible Kapton film has been developed and applied to dual lumen neonatal catheters integrated with silicon micro pressure sensors.

Catheters have been used for long time in the medical field for treating dysfunction in the human circulatory system. In medicine, the blood vessels are routinely used as pathways for catheters to reach places within the body for both diagnostic monitoring and therapy. The spectrum of its use includes simple intravenous catheters that are used to deliver fluids and medications and arterial lines catheters that are used for blood sampling and continuous blood pressure monitoring. In the past decade new devices such as endovascular filters, stents and grafts have been developed and utilized for coronary artery disease, aortic aneurysms and atrial-septal defects. Trans-vascular approach permits ready access to the patient's internal anatomy and by avoiding an open surgical approach, damage to healthy tissue is minimized and discomfort is reduced.

In critically ill neonates, there are particular issues associated with vascular access and monitoring due to the extremely small size of the patients. Currently available monitoring capabilities for the neonates remain relatively crude and limited. Frequently, the clinician is only able to determine the physiological state of the infant with limited direct measurements and serial physical examinations. Specifically, drawing blood is the major cause for transfusion requirements in these infants. Many tests have to be either limited in frequency or cannot be performed at all because of the small patient size and limited blood volume. Furthermore, the ability to calculate cardiac outputs and measure pulmonary and systemic vascular resistance is almost non-existent and is usually only extrapolated based on overall clinical condition and course. The opportunity to successfully intervene may be lost until a significant deterioration is recognized.

To provide medical staffs with more timely diagnostic information, miniature sensors are needed to collect data for diagnostics and monitoring with enough sensitivity. These sensors are required to be small enough to be fitted in a catheter with an outer diameter of 1.67 mm (5 French).

1-3. PREVIOUS WORK REVIEW

Currently two kinds of packaging method, mounting sensors in both a guide wire and catheter are used in the previous work. One method is that printed circuit board (PCB) frame is used to do the wire bonding of every connecting wire, as a partial support for the sensor chip and as a connection to the rest of the instrument [7], requiring bigger packaging space. The other method is that using solder flip-chip bonding technique on the silicon carrier [8], requiring a high processing temperature.

1-4. NEW APPROACH

The benefits of flip-chip bonding technique compared to wire bonding techniques include miniaturization, improved reliability and cost reduction. Comparing polymer flip chip bonding technique with solder flip chip bonding technique, polymer flip-chip gives added benefits on environmental cleanliness, manufacturing efficiency, lower temperature process and due to compliant rather than rigid solder, the polymer bumps also have improved tolerance of thermal stress caused by mismatched coefficients of thermal expansion (CTE) between the chip, substrate, and passivation layer.

Conductive polymer flip-chip bonding techniques using micromachined conductive polymer bumps showed a very low contact resistance, simple processing steps, a high bumping alignment resolution ($< \pm 5$ μm), and a lower bonding temperature (~ 170 °C) [9], [10]. These characteristics allow polymer flip-chip bonding techniques compatible with the lower cost polymer substrates that cannot be processed at a high temperature needed for the solder flip-chip bonding.

This research presents a new approach to flip-chip bonding pressure sensors on a flexible Kapton film using micromachined conductive polymer bumps and passive alignment techniques with laser cut wall [11]. Some characteristics of the conductive polymer bumps between gold surface on the Kapton film and Aluminum surface on the Silicon have been characterized. These polymer bumps have a low contact resistance around 100 mohm, an average failure stress around 0.0044 N/mm^2. Piezoresistive pressure micro sensors have been flip-chip bonded for a neonatal catheter and put into the water at 27 °C and 36 °C for two weeks and there're slight contact resistance differences. And these packaged pressure sensors also have been fully characterized in both air and liquid environments, indicating that the packaging was satisfactory rugged for catheter use and will lead to "smart" catheters with additional sensors.

1-5. OUTLINE OF THESIS

The research objective of this thesis, some previous work and our new approach are described in Chapter 1. In Chapter 2, the design, fabrication method and bonding method are presented. In Chapter 3, experiment setup to measure contact resistance, humidity test for two weeks, bonding strength measurement and pressure sensor outputs in gas and liquid environments and the experiment results are presented. Chapter 4 consists of the conclusion and contribution of this research, and also suggestions for future development.

CHAPTER 2. DESIGN AND FABRICATION

The whole package is composed of a piezoresistive pressure sensor, ribbon cable, and catheter, as described in Figure 2-1. The ribbon cable includes contact metal pads to be flip-chip bonded with the pads of the pressure sensor, long electrical metal lines working as a cable, and large pads to bond wires outside the catheters.

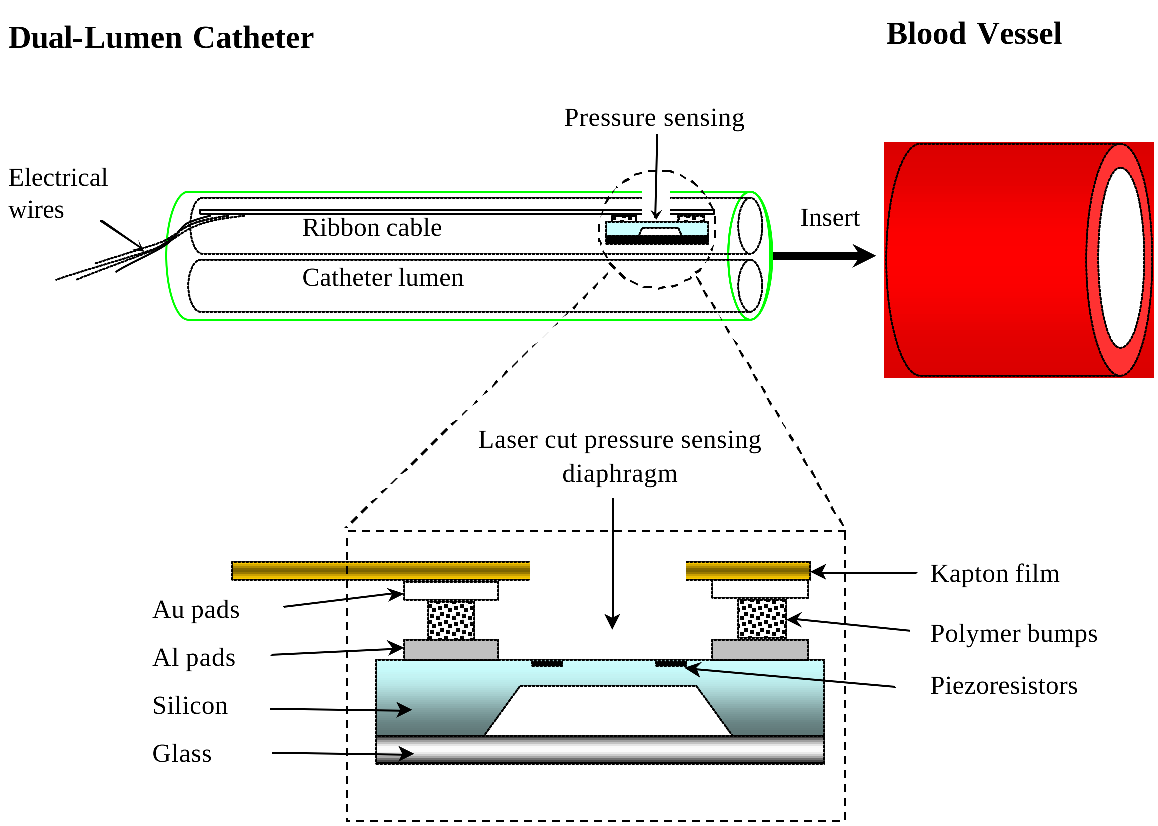


Figure 2-1. *Schematic diagram of a polymer flip-chip bonded pressure sensor on a flexible Kapton film for neonatal catheters.*

2-1. RIBBON CABLE

A flexible, biocompatible and long ribbon cable was designed as a connector between micro pressure sensors and outside wires and also a substrate to make conductive polymer bumps for flip-chip bonding. A ribbon cable could reduce the time and money required in interfacing the array by a large extent. A ribbon cable could essentially simplify the bonding process and could be manufactured using standard micormachining fabrication techniques. Kapton film, type HN (Dupont) with thickness of 125 µm has been chosen. First, Kapton film were bonded to the PDMS spin-coated silicon wafer on 100 °C hot plate to prevent any air bubbles inserted between PDMS surface and the Kapton film and easily peeling off after fabrication. After cleaning of the Kapton film, to prevent the thin and long metal line from easily peeled off, Kapton film were sent to the RIE machine and done surface modification at Oxygen Plasma, 150 V for 1 minute. Then 150-Å Cr seed layer and 2000-Å Au layer have been deposited in Temescal FC-1800 e-beam evaporator. Three different length of ribbon cables were developed in this work. The traces in these cables are 50 µm wide with 50 µm trace-to-trace spacing. The bonding pads on the flip-chip bonding side of the cable are 150 µm × 300 µm and 500 µm × 1000 µm on the outside wires connector side. Table 3-1 gives the total resistance data for three different lengths of the traces plus bonding pads two sides.

Table 3-1. Measured resistance of the ribbon cable metal traces.

Length of ribbon cable (cm)	1 cm	3 cm	5 cm
Measured average resistance (ohm)	19 ± 2	49 ± 4	70 ± 5

The bridge impedance of the pressure sensor is around 3 Kohm and it is clear from measurements that the resistances of the traces are far lower than the bridge impedance, so it would not contribute significantly to the impedance of the system.

2-2. PIEZORESISTIVE PRESSURE SENSORS

Pressure sensors represent fairly mature and commercially available components of MEMS technologies. Most commonly, microscale pressure sensors use either capacitive [12], [13] or piezoresistive measurement schemes [14], [15]. Piezoresistive detection technique is more favorable than capacitive detection for miniaturized sensors due to better scaling characteristics for our case. Another important advantage is that the amplifier circuitry can be placed far away from the sensors, e.g. outside the body.

Piezoresistive sensors rely on the piezoresistive effect in which sensor resistance changes in proportion to the changes of sensor dimensions, which might result from changes in pressure. A very small silicon micromachined absolute piezoresistive pressure sensor die (Model-SM5106) from Silicon Microstructures, Inc., is shown in Figure 2-2. The pressure sensor has four piezoresistors positioned over the four edges of a n-type silicon membrane. Since the edges are aligned to <110>, two sets of resistors are in the transversal mode and other two sets are in the longitudinal mode. So, the four resistors can be combined using the Wheaston's bridge concept, and the variation of pressure over the membrane can be easily detected using a parametric measurement. The output voltage due to the unbalance of four resistors in the Wheaston's bridge will be proportional to the pressure.

This pressure sensor die has a dimension of 1.56 mm × 1.56 mm × 0.9 mm and pad size of 130 μm × 300 μm. Thus, the sensor is small enough to fit inside the catheter of 1.67 mm in diameter.

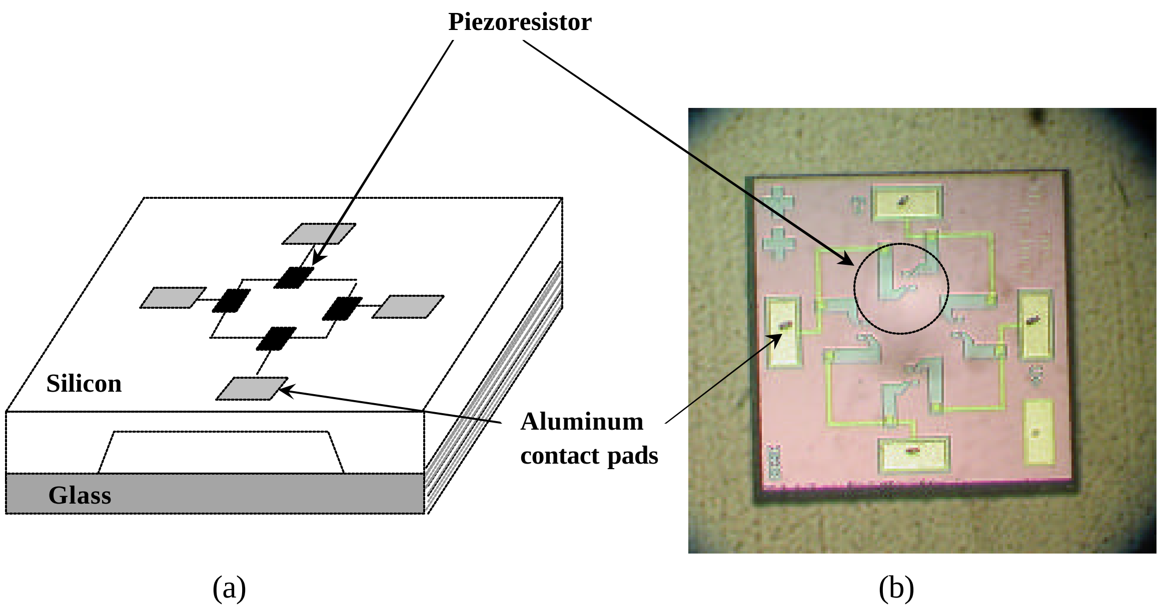
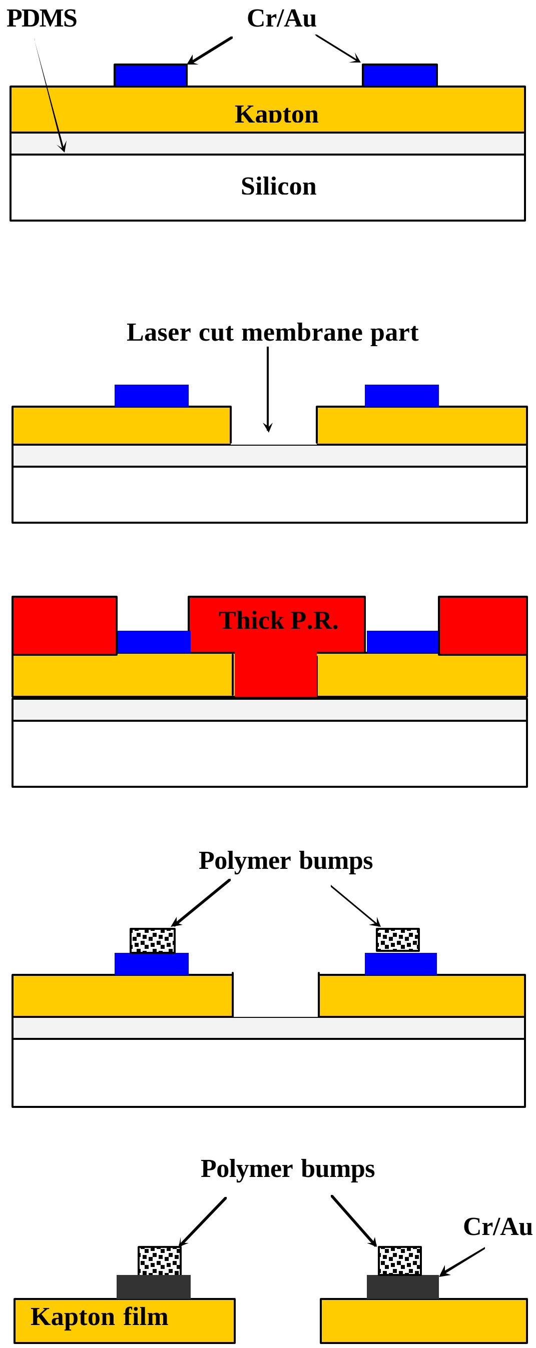


Figure 2-2. *Micro pressure sensor: (a) schematic view of pressure sensor and (b) photograph of SM5106 piezoresistive pressure sensor.*

2-3. MICROMACHINED CONDUCTIVE POLYMER BUMPS ON KAPTON FILM

The conductive polymer flip chip bonding technique utilized silver-filled, thermoplastic conductive epoxy to electrically connect the contacts of electric components. Polymer bumps are micomachined and cured. Figure 2-3 summarizes the fabrication steps of the micromachined bumps on a Kapton film.



(a) Contact metal layer deposition and lift-off

(b) Cut membrane part using laser micromachining

(c) Thick PR spin-coat and patterning for bump-holes

(d) Conductive polymer filling and photoresist curing

(e) Peel off from the PDMS and ready for bonding

Figure2- 3. *Summarized fabrication steps for the formation of conductive polymer bumps on a Kapton film.*

The Kapton film was attached to a silicon wafer using spin-coated PDMS (Polydimethylsiloxone) on the 100 °C hot plate to get a flat surface and was easily peeled off after fabrication. Several methods have been tried before to handle flexible Kapton film during fabrication, such as attach Kapton film on the silicon wafer using Shiply1818 photoresist, double sided tapes and sealing around the Kapton film using epoxy in Vacuum chamber, but have met several problems. Then, deposit 150-Å Cr seed layer and 2000-Å Au conducting layer on the Kapton film in Temescal FC-1800 e-beam evaporator. Wires and contact pads for the polymer cable were formed by patterning gold.

Section of the Kapton film covering the sensor membrane was cut out (relative voltage level 5, pulse repetition rate 100) using laser micromachining (model LMT – 4500, KrF excimer laser, 248 nm) to membrane part direct expose to the environment and get a better sensitivity.

Thick photoresist (AZ4620) was spin-coated for 5 seconds at 500 rpm and 20 seconds at 800 rpm and patterned for bump holes.

The conductive polymer (Epo-Tek-K/5022-115BE, Epoxy technology inc.) was pipetted into the bump-hole patterns, and overflowing conductive polymer was removed by a raser blade to get the flat and same height polymer bumps and selective cured in a convection oven at 100 °C for 15 min, then thick photoresist (AZ4620) molds were stripped away in the conventional photoresist stripper solution, leaving the polymer bumps on the contact metal pads.

The wafer was then cured in a convection oven at 150 ºC for 1 hour, to achieve a better conductivity for the conductive polymers and then peeled away from the silicon wafer carrier. Figure 2-4 shows micromachined thermoplastic conductive polymer bumps which have a flat surface morphology with a thickness of 25 µm and an area of 80 µm × 250 µm on gold contact pads and laser-cut sensing membrane area.

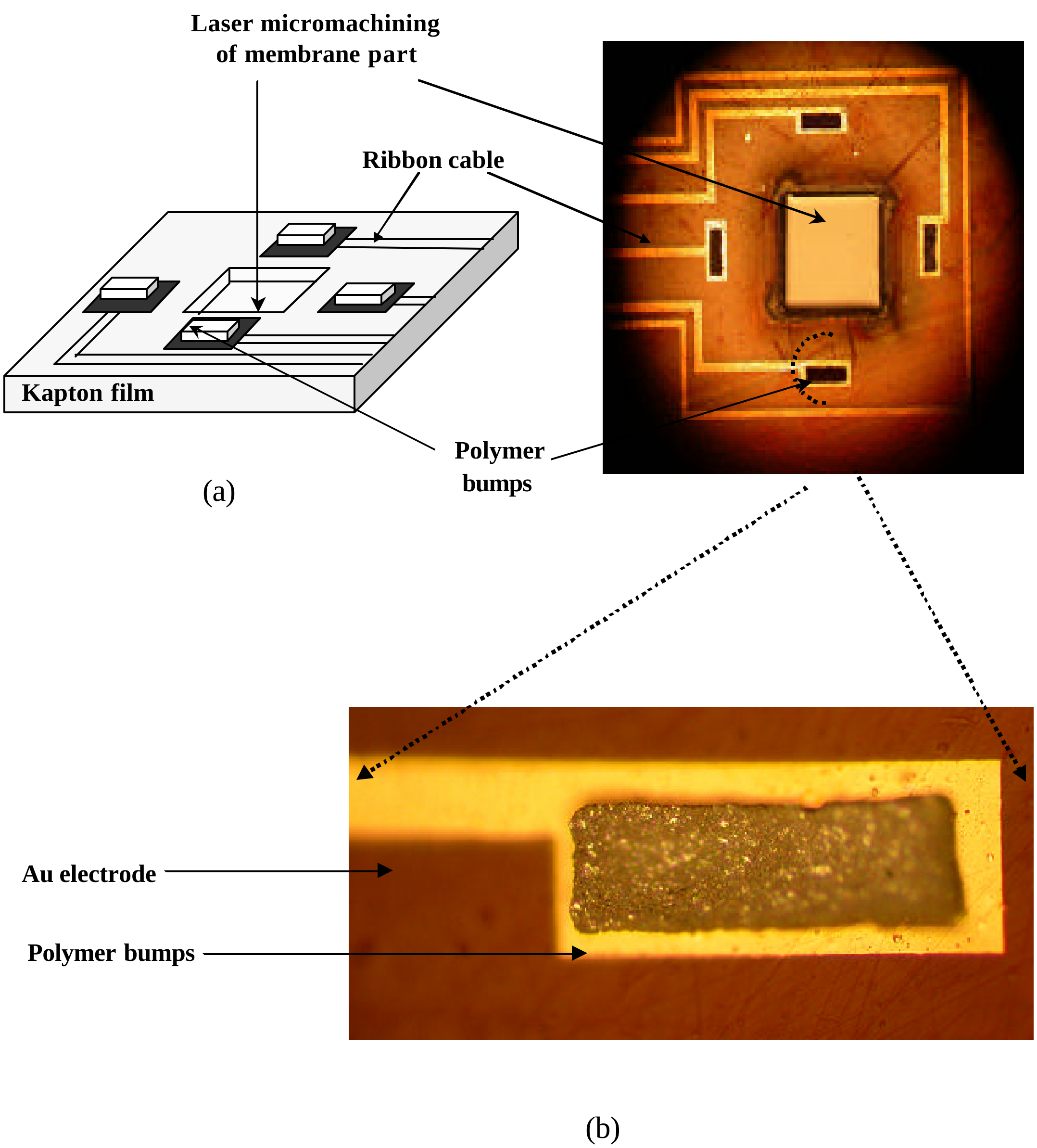


Figure 2-4. *Polymer bumps: (a) schematic view of polymer bumps and (b) photograph of micromachiend polymer bumps.*

2-4. ALIGNMENT WALL

The alignment of pressure sensor on the flip-chip bumps with acceptable alignment accuracy has been considered as one of the most difficult tasks in polymer flip-chip bonding. In order to overcome the difficulty, simplify the fabrication steps and also reduce the overall packaging size, we have fabricated alignment walls on the Kapton film using laser micromachining, which allows as easy and accurate alignment as shown in Figure 2-5.

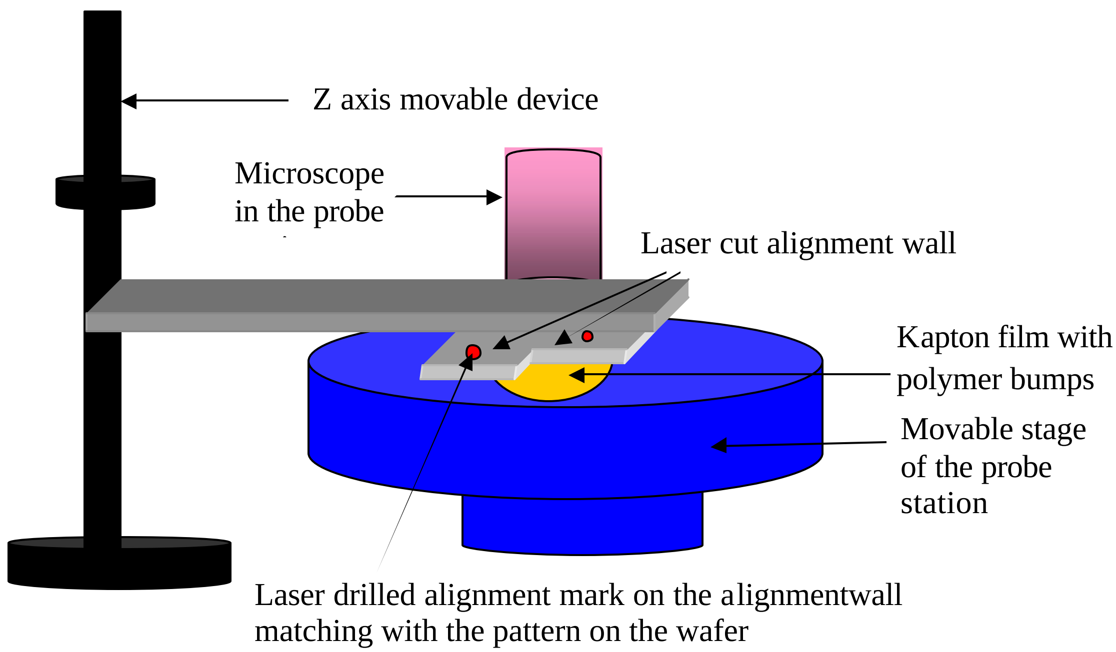


***Figure 2-5**. Schematic view of laser cut alignment wall for flip-chip bonding*

First, Kapton film with conductive polymer bumps was placed on the stage of the probe station and fixed using the Vacuum. Then Z axis movable device attaching laser cut alignment wall came close to the Kapton film and two inner sides of the alignment wall aligned with the pattern on the Kapton film. Two laser drilled alignment marks helped to make alignment accurate. After alignment, the alignment wall touched with the surface of the Kapton film. Pressure sensor was inserted into the wall and pressure added using needles in the probe station. After bonding, the alignment wall was slowly removed upside a bit to separate two touched surfaces, and then the stage of the probe station was moved from the opposite direction of the alignment wall.

Metal or SU-8 alignment pedestals [16], [17] were problematic when applied on the Kapton film since both pedestals are easily peeled off when the film flexed. Also due to small dimensions of the pattern and small distances between pedestals and pads, it is hard to get uniform polymer bumps because the pedestals will interrupt uniform spin coating of thick photoresist for bump hole patterns. Some errors may occur when dicing the pressure sensors. Such kind of error can be reduced by measuring the accurate pressure sensor dimensions first and then aligning and flip-chip bonding because laser cut alignment wall can be easily repositioned. In addition, the silicon and glass substrates, anodically bonded together to compose pressure sensor different dimensions following dicing. Our observations of the SM5106 sensors show that silicon substrates are smaller than glass substrates. The alignment wall was made from Kapton film with 125 µm thickness, which is much thinner than the height of the silicon substrate so it will not damage the sensor when repositioning and aligning.

2-5. FLIP-CHIP PACKAGING FOR MINIATURE PRESSURE SENSORS

The piezoresistive pressure sensors and ribbon cables were assembled using flip-chip bonding techniques with micromachined conductive polymer bumps and passive alignment techniques with laser micromachined alignment walls. Figure 2-6 illustrates the flip-chip bonding technique for the mounting of pressure senor on the ribbon cable. First, the laser micromachiend alignment barrier was aligned with the pattern already made on the ribbon cable. When the Kapton cable was pre-heated to approximately 20 °C above the thermoplastic conductive polymer melting temperature (~ 150 °C), the pressure sensor was flipped, positioned next to the alignment wall and contacted onto the contact pads of the ribbon cable under small pressure which using the needle of the probe station. The thermoplastic bumps then melted onto the contact pads of the substrate. The physical and electrical bonds were established as the Kapton film cooled below the melting temperature of the thermoplastic materials about two minutes later. Then, the alignment wall was slowly removed upside direction to separate alignment wall and Kapton film. In order to increase the bonding strength between the pressure sensor and the ribbon cable, a viscous 5000 cP UV-cured flexible adhesive (3321, Loctite Corp) was carefully coated around the pressure sensor. A commercially available UV-lamp (118V, 0.35 A, 60 Hz) preheated to 15 W cured the adhesive for 6 min. Finally 38 AWG wires were bonded to the gold pads using silver epoxy and covered with the same UV-cured adhesive as a protective layer for the ribbon cable and under-fill for the flip chip bonding to strengthen the bonding. After that, resistances between two electrodes were measured using probe station and recorded shown in Figure 2-7. The entire assembly was then cut to size using laser micromachining for insertion into the catheter for testing. As shown from Figure 2-8, the entire assembly was flexible enough to follow the twists and turns of the blood vessels.

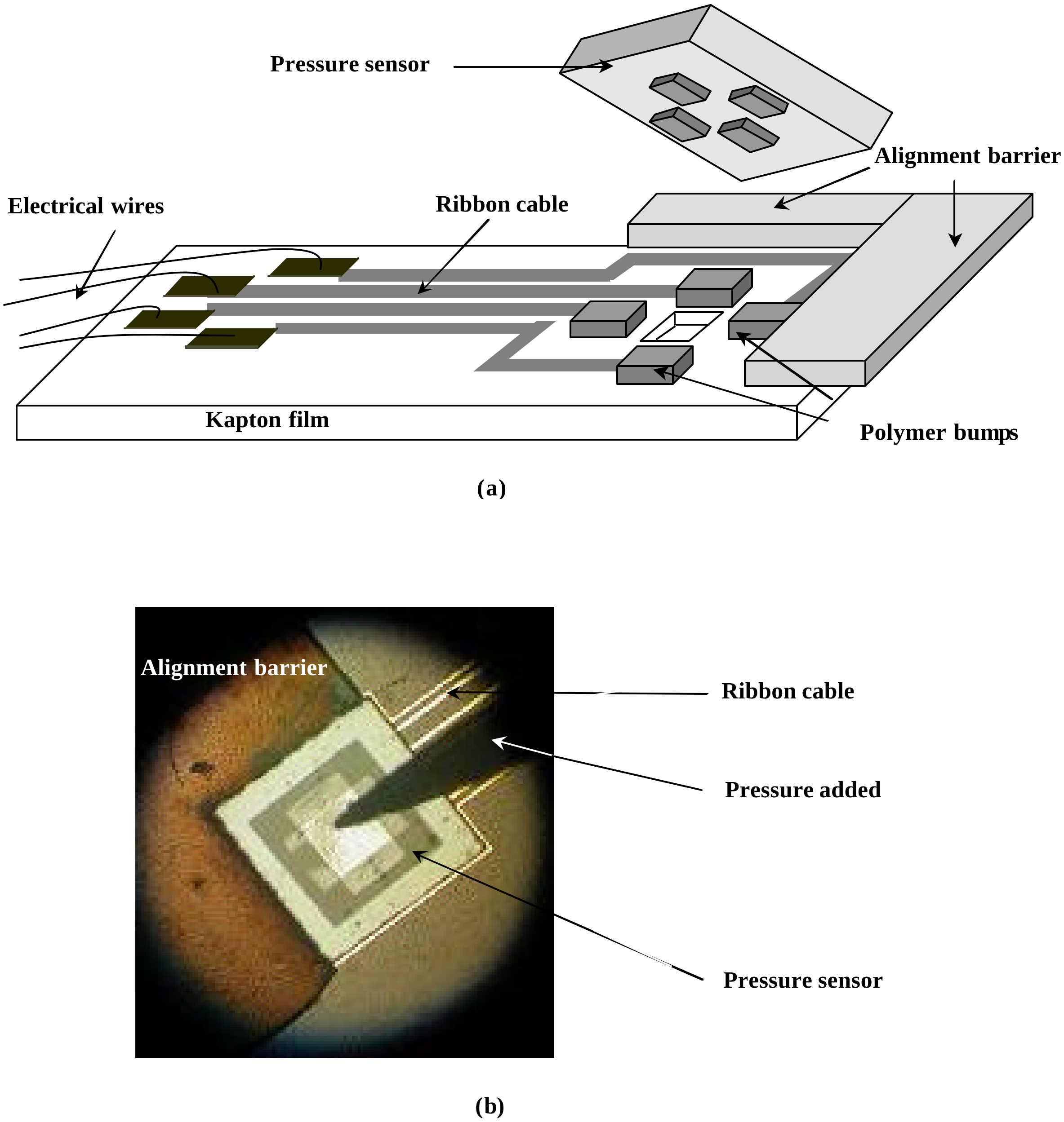


Figure 2-6. *Flip-chip bonding process: (a) schematic view of polymer flip chip bonding (b) photograph of polymer flip chip bonding*

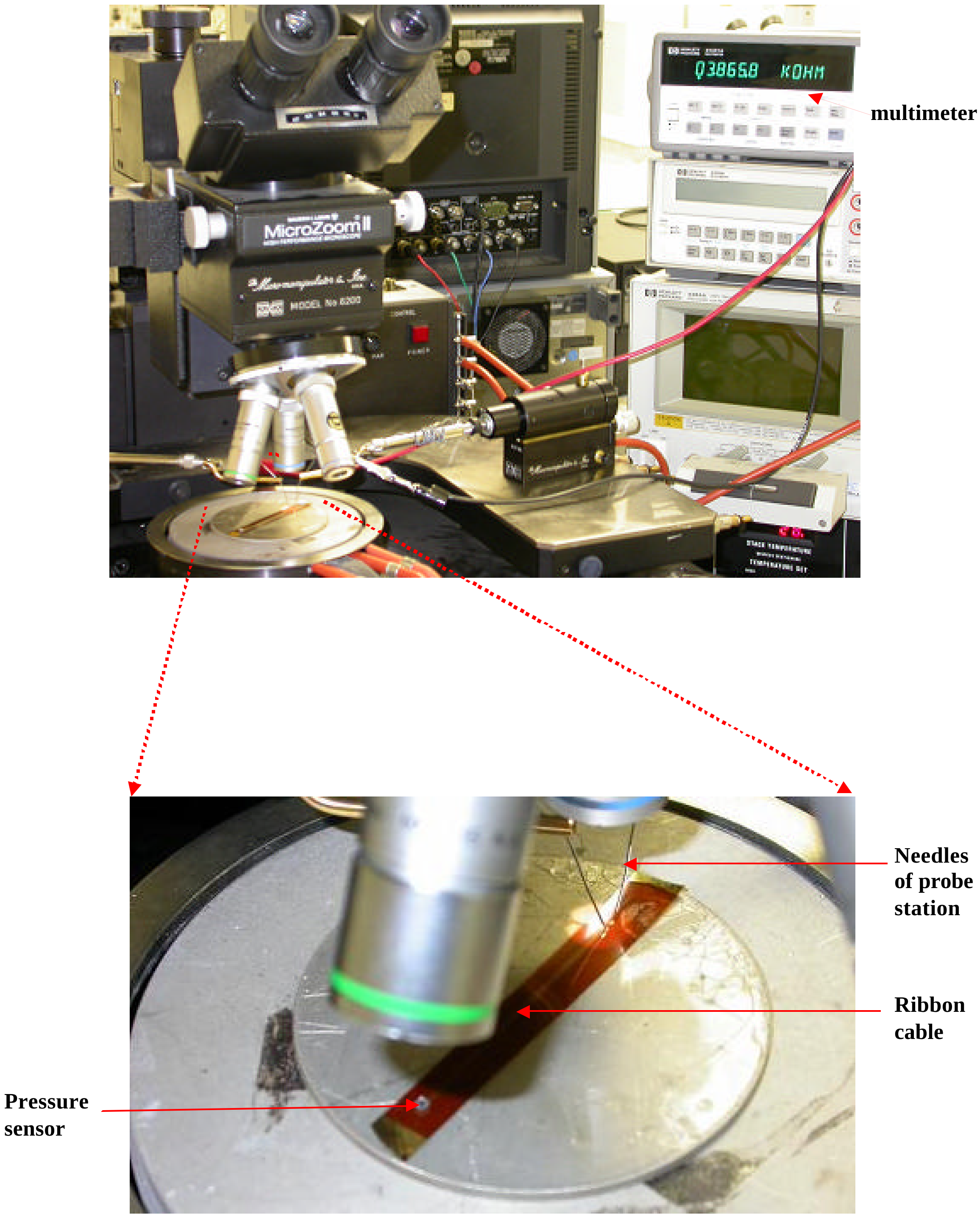
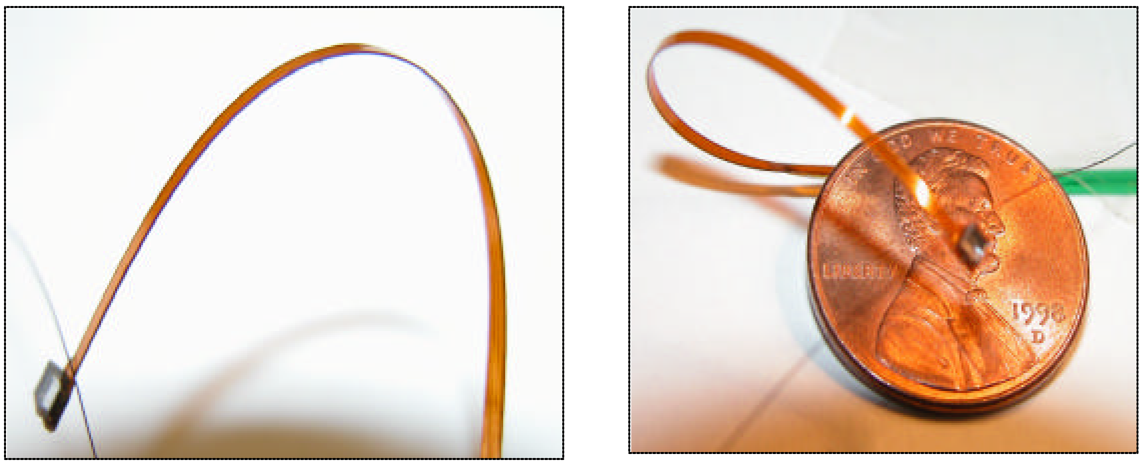
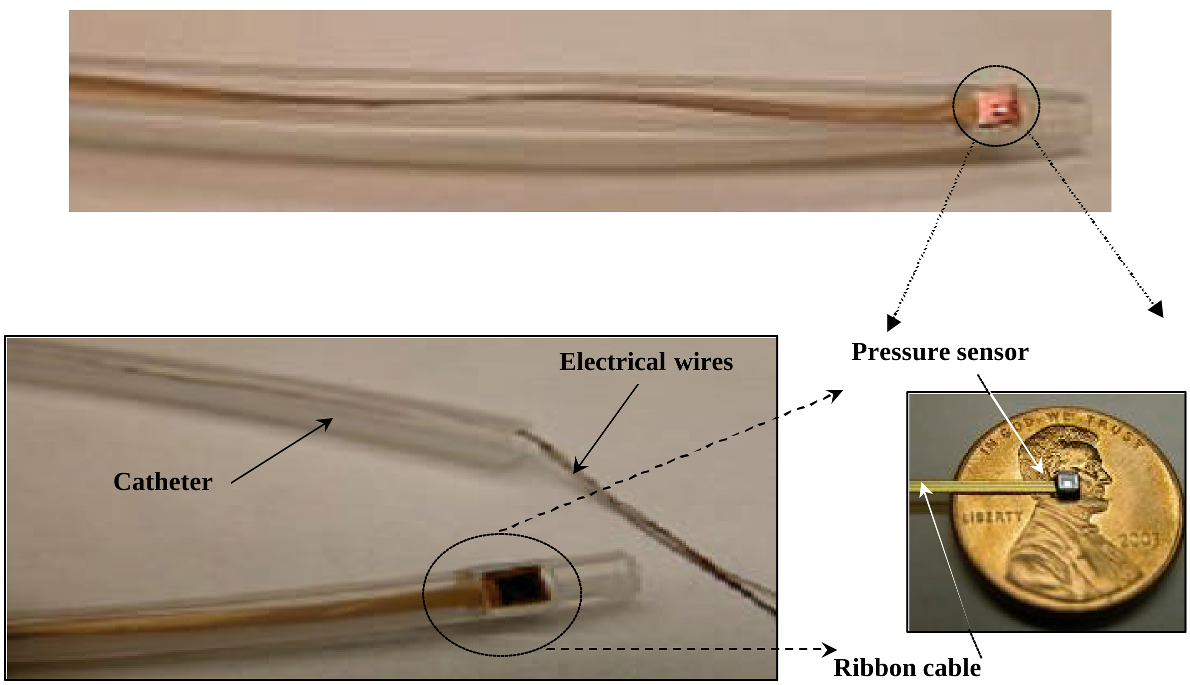


Figure 2-7. *Resistance measurement after flip chip bonding using probe station.*



(a)



(b)

Figure 2-8. *Flip-chip bonded sensors: (a) pressure sensor bonded on the Kapton flim with enough flexibility; (b) bonded sensor inside the catheter.*

CHAPTER 3. EXPERIMENTAL RESULTS AND DISCUSSION

The electrical properties of micromachined conductive polymer bumps have been characterized by using a four-terminal method. Humidity test has been done in the water environment both in 27 ºC and 36 ºC for two weeks. Bonding strength measurement has been done to characterize the bonding strength of the polymer bumps between gold surface and aluminum surface. In addition, the flip-chip bonded pressure sensors were also fully characterized in gas and water environments.

3-1. CONTACT RESISTANCE MEASUREMENTS

A conductive polymer bump between the flip-chip and substrate has a certain amount of contact resistance. Contact resistance is a function of the contact area between the conductive particles and the interconnection pads in accordance with Ohm's law. Good electrical contact is characterized by low contact resistance. The four-terminal is the typical normal measurement technique, which is always applied in measuring low-lever resistance and contact resistance to prevent the lead resistance from being added into the measurement [18], [19], [20], [21]. Using the four-terminal method, the contact resistance between aluminum pads on the silicon pressure sensor and gold pads on the Kapton film were measured. A schematic diagram of a flip-chip bonding for the contact resistance measurement using a four-terminal method is illustrated in Figure 3-1.

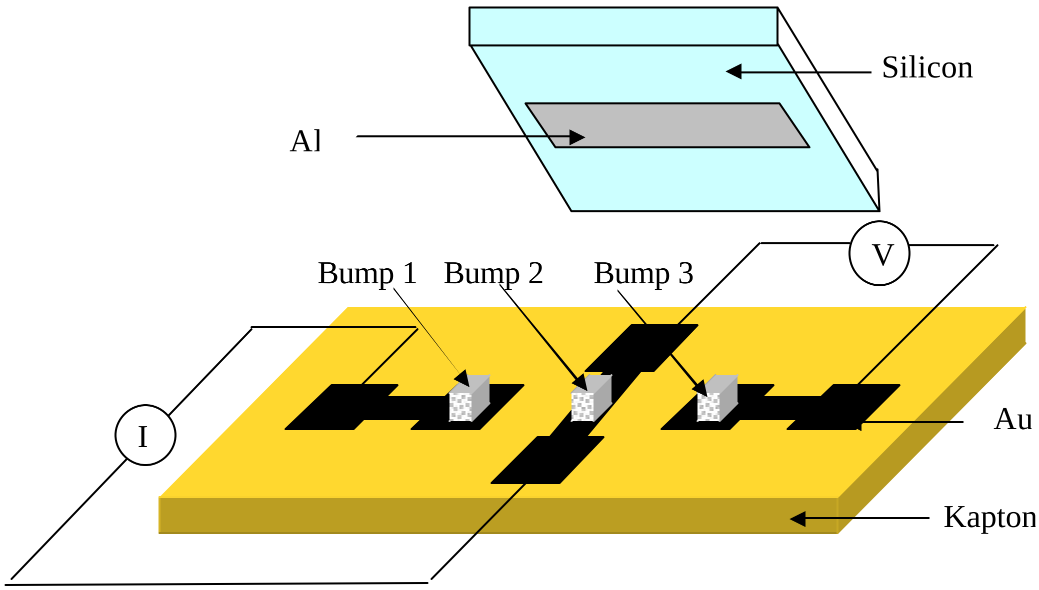


Figure 3-1. *Schematic view of contact resistance measurement using four-terminal method.*

A constant direct current is made to flow through two of the contacts from each end of the metal pad lines, and the voltage across the other two contacts is measured with a high-impedance-voltmeter. Using this method, the contact resistance of the middle bump 2 has been measured. The I-V characteristics measured for 25 μm high bumps of various sizes are plotted in Figure 3-2. The voltage drop was measured as the constant direct current was forced through the bumps up to 50.8 mA. The voltage drop obtained was linearly proportional to the current through the contact resistance. The slopes of the I-V graphs were linear with different driving currents, which confirm the achievement of ohmic contact through the polymer flip-chip boding. The contact resistances measured for 25 μm high bumps with 400 μm × 400 μm, 300 μm × 300 μm, 200 μm × 200 μm, and 100 μm × 100 μm areas were 17 m? , 39 m? , 77 m? and 149 m? respectively.

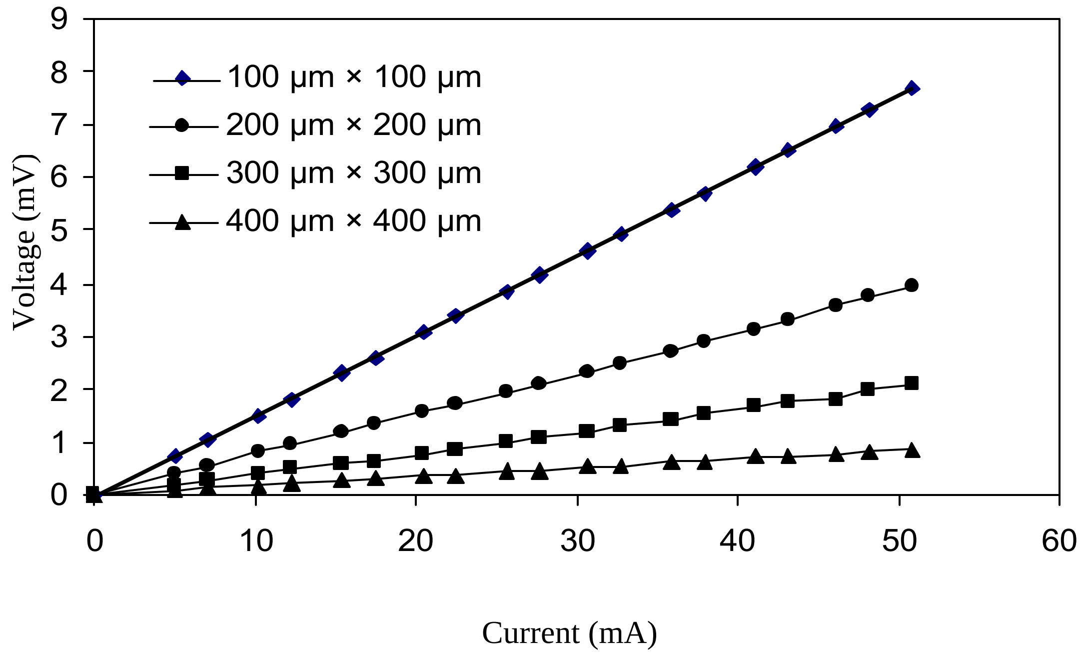


Figure 3-2. *I-V characteristics measured for the various dimension of bumps with a contact thickness of 25 μm.*

In Ohm's law, the contact resistance decreases as a function of contact area. Figure 3-3 shows the trend as a function of the nominal contact area. The specific contact resistance, defined as the contact resistance multiplied by contact area, for the micromachined conductive polymer bumps was on the order of ~ 10^{-5} ohm-cm^2 and the average value is about 2.71 × 10^{-5} ohm-cm^2 shown in Figure 3-4. In reference [22], metal-semiconductor ohmic contacts, typical ohmic contact with specific contact resistance of approximately 10^{-5} ohm-cm^2 has contact resistance of ~11 mohm for an area of 300 μm × 300 μm and is considered a good quality contact. The flat surface morphology of the micromachined polymer bumps suggests that modeling the area as a simple rectangle may not suffice. The actual effective areas between the bumps and metal pads may not be the same as the calculated square areas due to the large grain size of the polymers. In addition, there may be variation from bump to bump with respect to the fraction of polymer matrixes used as adhesive and silver flakes used as conductive filler. Nevertheless, the contact resistance of the micromachined conductive polymer bumps measured here is comparable to the values achieved by solder metal bumps and shows satisfactory values over a wide range of bump sizes and applied currents.

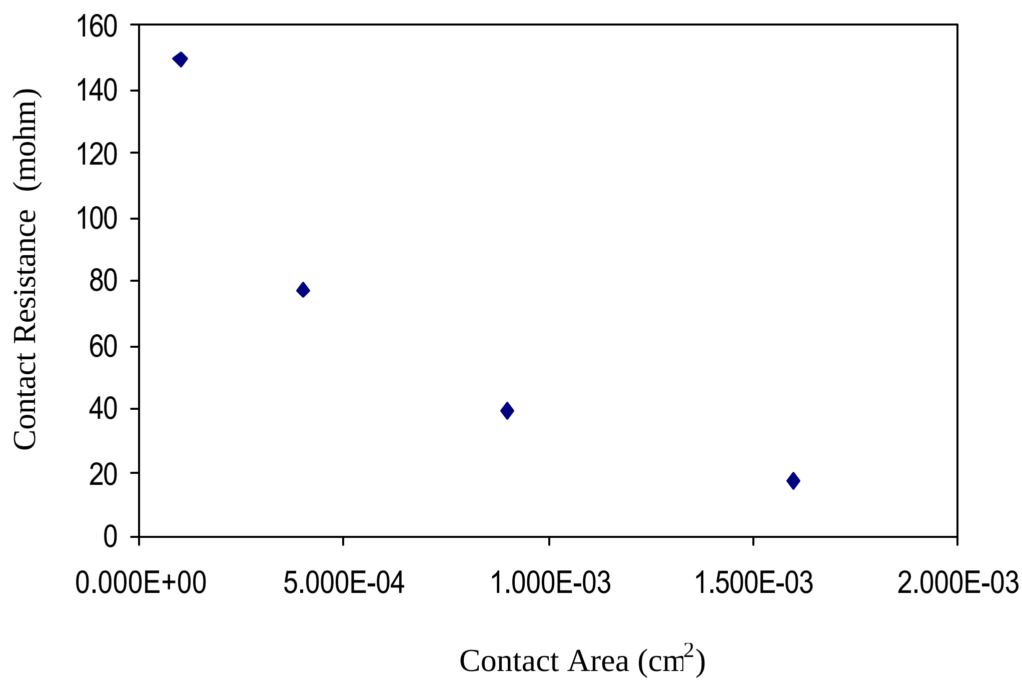


Figure 3-3. *Contact area dependence of contact resistance.*

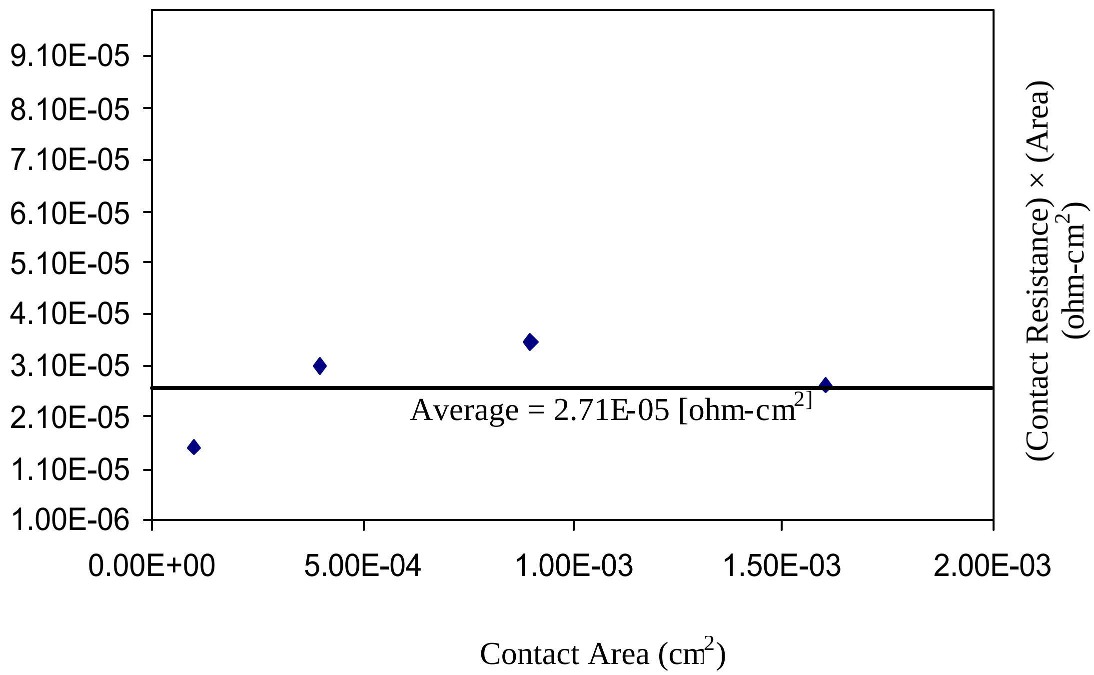


Figure 3-4. *Area times contact resistance dependence of contact resistance.*

Comparing the contact resistance (~ 100 mohm) with the bridge impedance of the pressure sensor (~ 3 Kohm), this contact resistance is far lower than the bridge impedance and it would not contribute significantly to the function of the pressure sensors.

3-2. Bonding Strength Measurement

The bonding strength test was carried out by using INSTRON 4465 Tester with a cross-head speed of 1 mm/min. The experimental setup is shown in Figure 3-5. The bonding strength was measured at breakage when chip is detached form the substrate. First we tried to measure the bonding strength of bump dimensions of 400 μm × 400 μm, 300 μm × 300 μm, 200 μm × 200 μm, and 100 μm × 100 μm , but the strength of these dimensions were too small to detect. So we made polymer bumps with bigger dimension of 10 mm × 10 mm, 9 mm × 9 mm, 8 mm × 8 mm and 6 mm × 6 mm. To do bonding strength measurement using INSTRON 4465 Tester, these samples should be mounted in the tester with proper sizes. First, to make aluminum surfaces, 5000-Å thickness aluminum layer was deposited on the Silicon wafer and was cut in 11 mm by 11 mm size. 10 mm × 10 mm, 9 mm × 9 mm, 8 mm × 8 mm and 6 mm × 6 mm size conductive polymer bumps were made on the Kapton film with the same method described in Chapter 2 and then cut in size 11 mm by 11mm. To hold these two substrates, thin glass covers were cut in 11 mm by 11 mm size and thick steels which conduct heat well were also cut in size 11 mm by 11mm. Then attach silicon pieces with steels and Kapton pieces with glass using double sided tapes. If bonding silicon substrates and Kapton films first, then bonding steels and glass on the up and down sides of the bonded samples, during bonding these two substrates always break the weak bonds of the polymer bumps. So first bonding these two holders first, then do polymer flip-chip bonding. After bonding, thin needles were attached under the steel using epoxy and to prevent breaking the weak bonds, light weight and flexible thread inserted into the tube were attached using epoxy. The tests were repeated five times for each condition using the same experimental setup and the average results have been calculated.

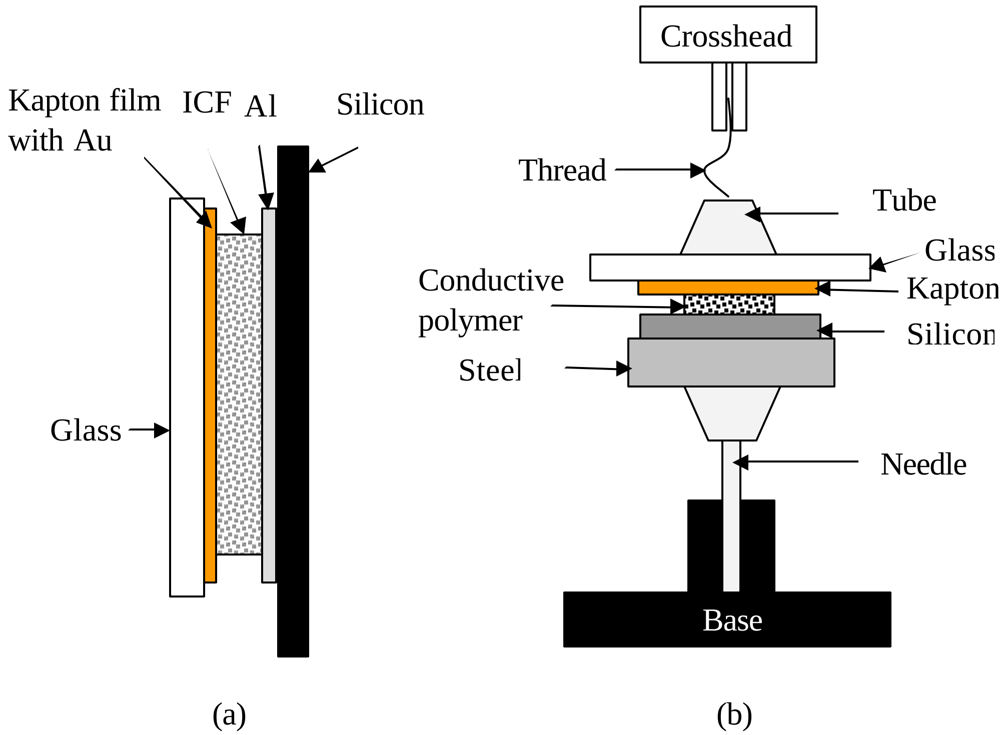


Figure 3-5. *Side view of a chip package prepared for bonding strength test and experimental setup.*

The test results of the bonding strength are shown in Figures 3-6, Figure 3-7 and Figure 3-8**.** Failure Stress for the various dimensions of bumps with a contact thickness of 25 µm is shown in Figure 3-6. Average failure stress for the various dimensions of bumps with a contact thickness of 25 µm is shown in Figure 3-7. The average failure stress for the conductive polymer bumps with a thickness of 25 µm is 0.004409 N/mm^2 and every measurement this bonding shows pretty stable bonding strength value. Figure 3-8 shows I-V characteristics measured for the various dimension of bumps with a contact thickness of 25 µm. From the figure we can see that this bonding shows pretty linear responsibility with bonding area vs. force.

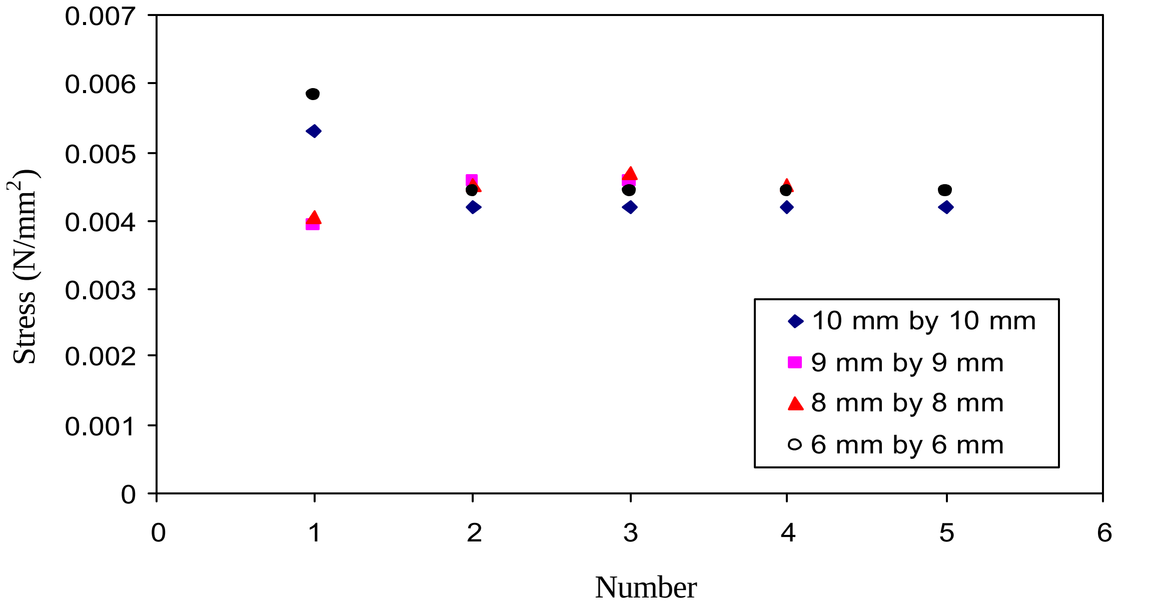


Figure 3-6. *Failure stress measured for the various dimension of bumps with a contact thickness of 25 µm.*

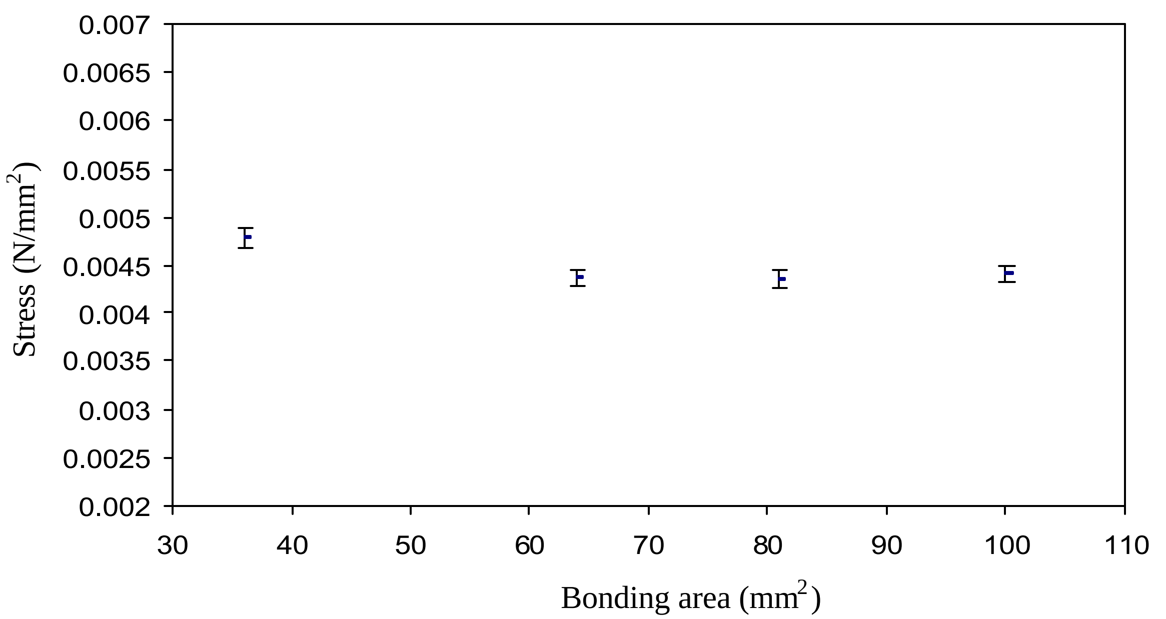


Figure 3-7. *Average failure stress measured for the various dimension of bumps with a contact thickness of 25 µm.*

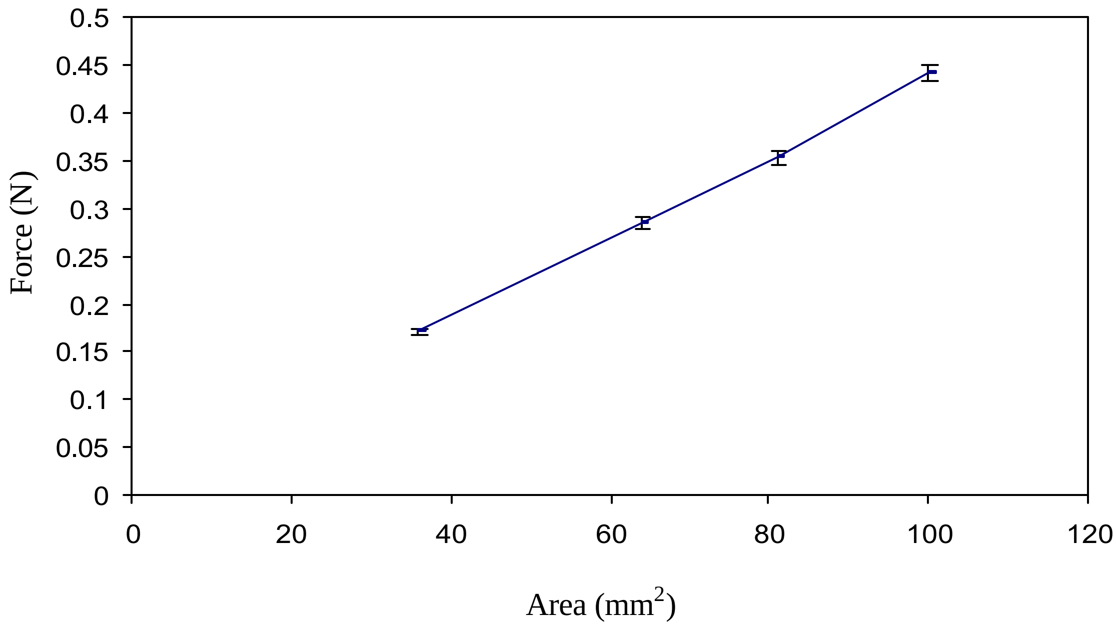


Figure 3-8. *Area vs. force characteristics measured for the various dimensions of 10 mm, 9 mm, 8 mm, 6 mm bumps with a contact thickness of 25 µm.*

Table 2 compares thermoset epoxies and thermoplastic polymers used in the polymer flip-chip bonding process. Both polymers consist of long polymer chains, with silver flake filler providing conductivity, and a rheology suitable for stencil printing of bumps as small as 70 µm in diameter.

Table 2. Data sheet information for the bonding strength of the adhesives

Adhesive	Polymer matrice	Shear strength (N/mm^2)	Elastic modulus (kN/mm^2)
Adhesive A	Epoxy	11.3-43	11
Adhesive B	Epoxy	8.3-22	1.9
Adhesive C	Polysulfone	>20.7	4.1
Adhesive D	Polyurethane	>12	>0.4
Adhesive E	Thermoplastic	>17	>2.4

Comparing with these adhesives for flip-chip bonding, the bonding strength of the polymer bumps made by lift-off method we used in this experiment, 0.004409 N/mm^2, is very small. Under-fill material with high bonding strength between polyimide surface and silicon surface is needed to get the stable and higher bonding strength.

3-3. HUMIDITY TEST

Eventually, to measure the blood pressure, packaged micro pressure sensors will be operated in the liquid environment. Humidity experiment should provide some information about the aging and corrosion properties of conductive polymer and under-fills of biocompatible UV-cured epoxy for the packaged micro pressure sensors for neonatal catheters. For that reason the bulk and contact resistances of the packaged pressure sensors were recorded for two weeks to access their behavior.

The sample which pressure sensor has been bonded on the Kapton film and all the exposing areas except the membrane part were all covered using UV-curable epoxy was placed in a cylinder of sufficient size filled with water controlled temperature of 27 °C and 36 °C. Necessary wires were attached using silver epoxy in advance.

There are four bonding pads connecting wires to the outside as shown in Figure 3-9. While placing in the water, the resistances between these 1-, 2-, 3- and 4-point pads are measured. The time in between the measurements is intervals of one day.

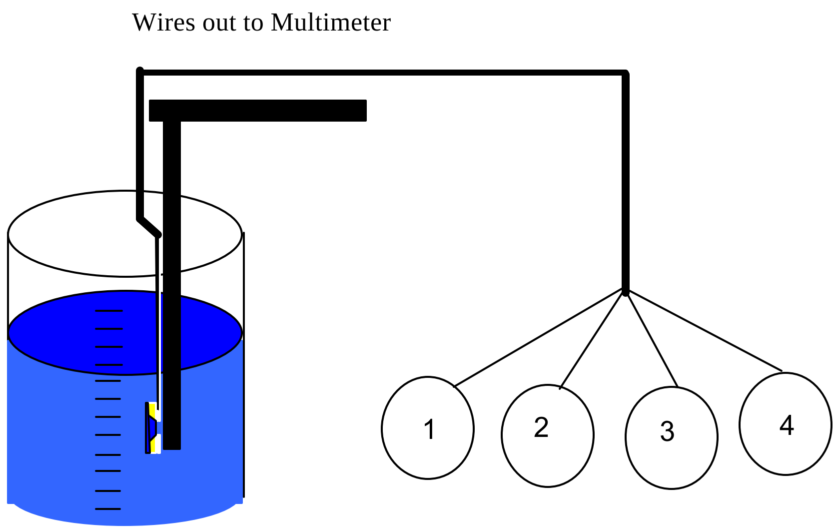
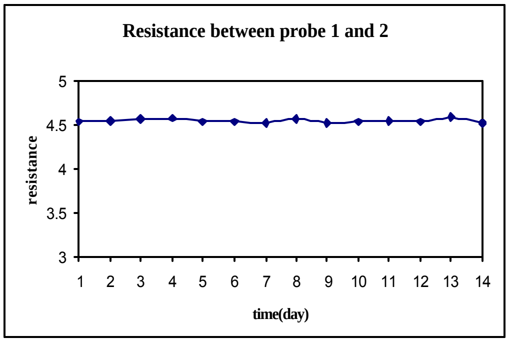


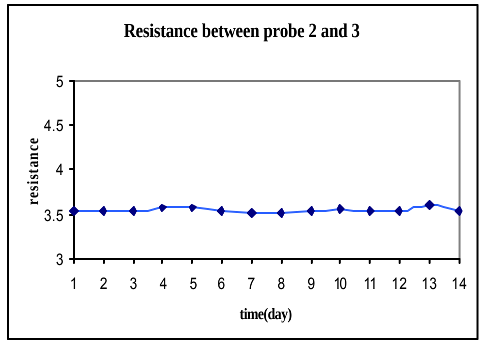
Figure 3-9. *Schematic view of experiment setup for the humidity test for two weeks.*

The measured resistance values are plotted over time (day) in Figure 3-10 and Figure 3-11. From the sample data it can be seen that after immersing into the liquid environment, there's a slight resistance change for two weeks, which indicates the bonding rugged enough for catheter use.

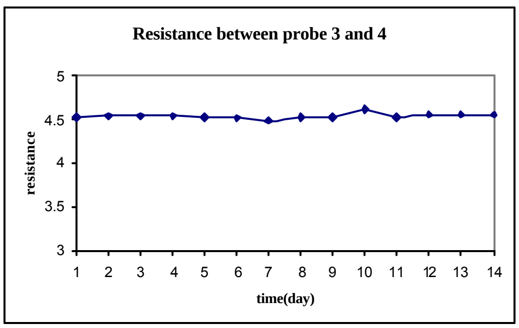
At 27°C



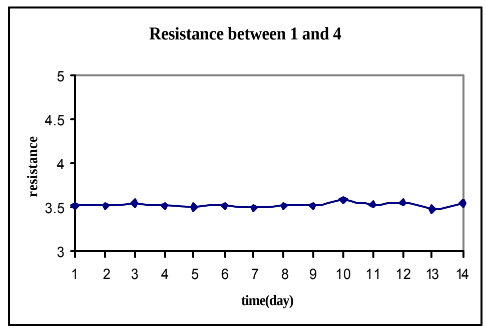
(a)



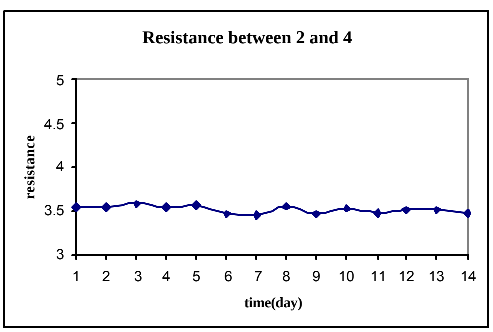
(b)



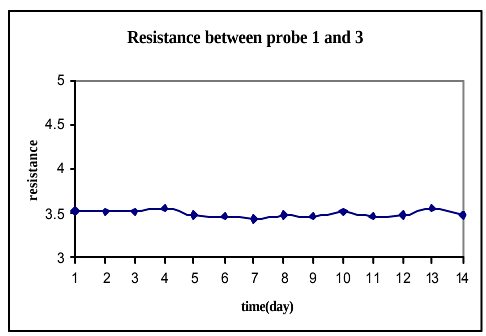
(c)



(d)



(e)



(f)

Figure 3-10. *Changes of resistances between four pads directly exposed to relative 100 % humidity at 27 ºC*

AT 36°C

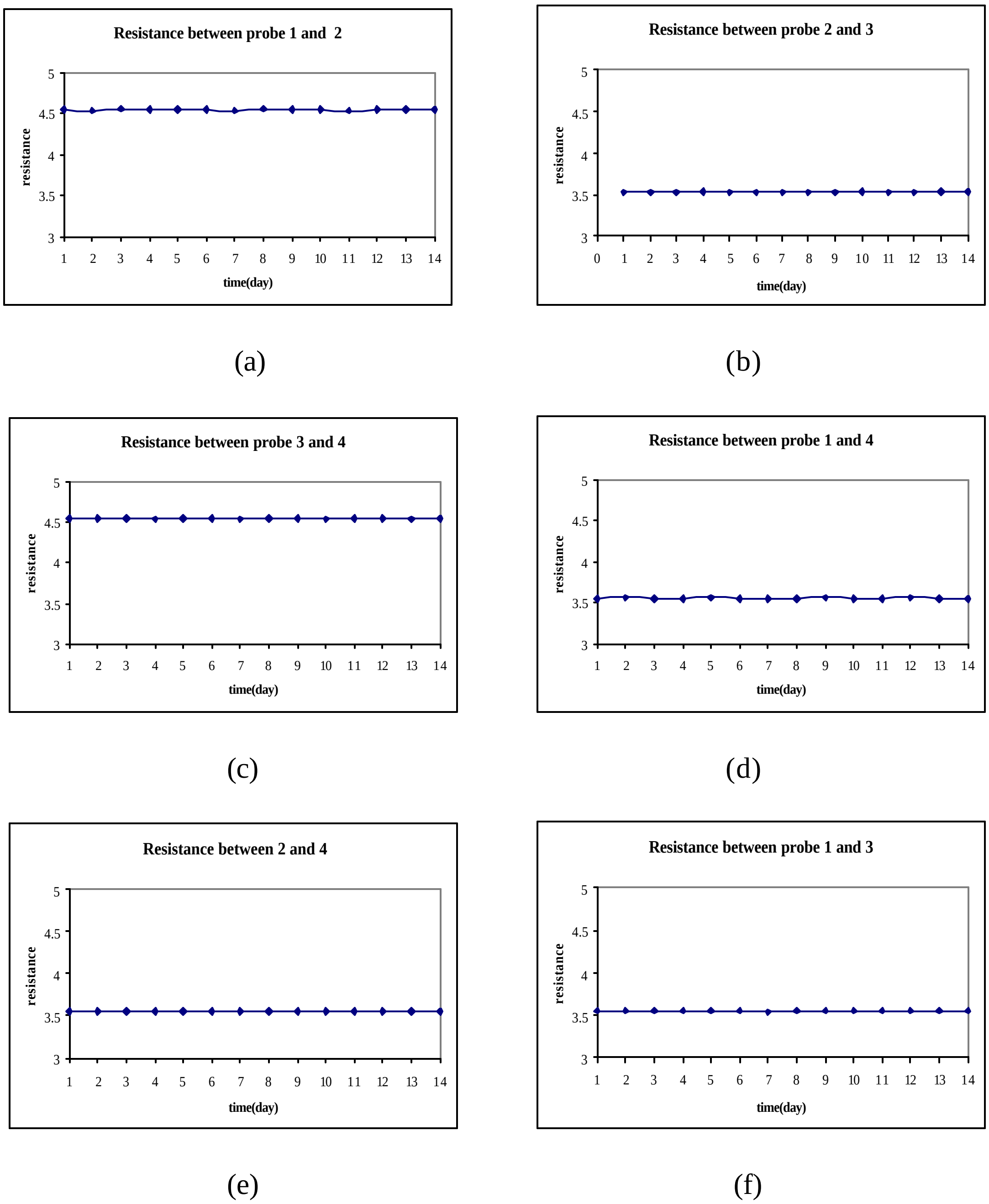


Figure 3-11. Changes of resistances between four pads directly exposed to relative 100 % humidity at 36 ºC

3-4. FLIP-CHIP BONDED PRESSURE SENSOR

The pressure micro sensors were successfully bonded to the polymer bumps and then characterized using the measurement setup as described in Figure 3-12.

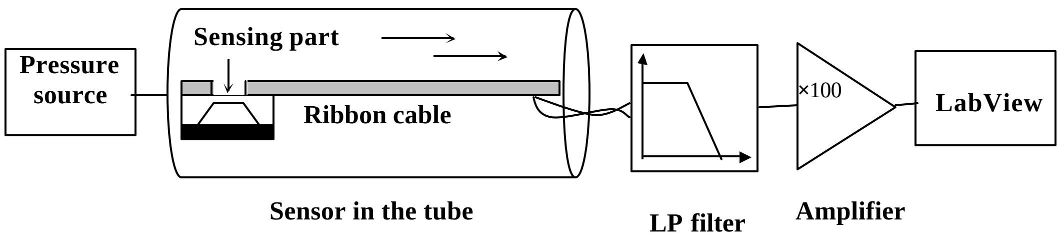
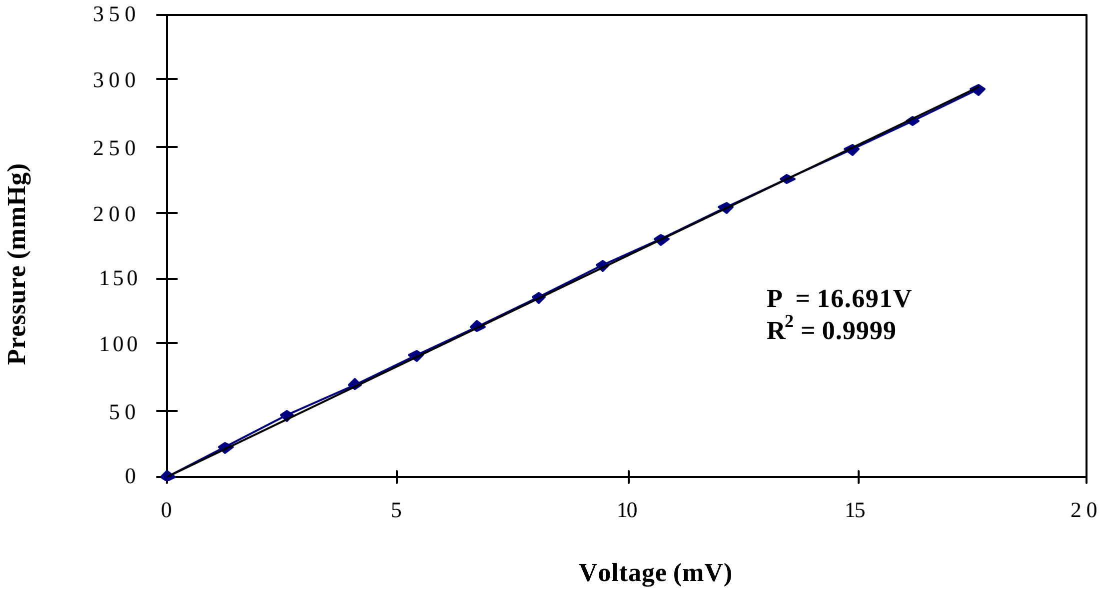


Figure 3-12. *Schematics of experimental setup for pressure measurement.*

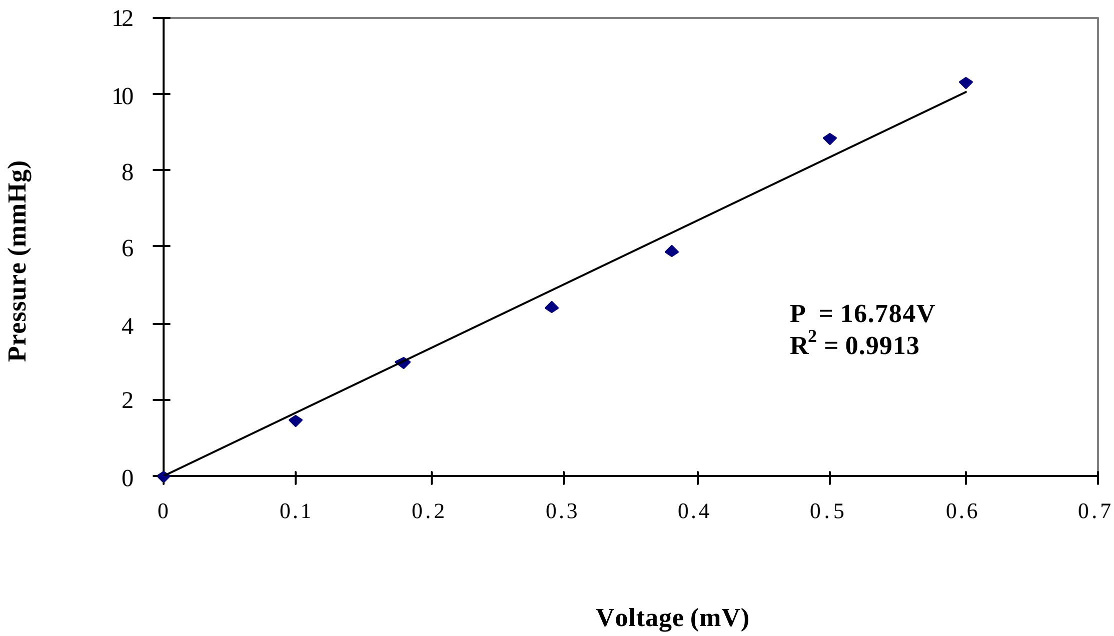
The experimental test setup was comprised of LabView 7 software with NI 6036E DAQ card. A 2345 signal conditioning platform was used to reduce noise and amplify the signal. The signal conditioning modules used to do this were LP01 (0-100 Hz) low pass filter and a SG04 strain gauge amplifier with a gain of 100. The excitation voltage used on the silicon pressure sensor was 2.5 V. The characterization of the sensor was performed in a Nitrogen chamber that has a commercial pressure sensor as a reference for pressure. The first step in the characterization of the sensor was to remove the offset voltage. Then slope of voltage vs. pressure (mmHg) vs voltage was found to give the conversion from mV to pressure for DAQ system. Figure 3-13 (a) shows the linear relationship between pressure and voltage for the packaged pressure sensor in nitrogen. The sensor was also tested in a cylinder of water shown in Figure 3-13 (b). The packaged sensor was linear in nitrogen and water with a slope of approximately 16.7 mmV/mHg for both environments.

The sensor was further tested in water filled compressible container to simulate blood pressure at room temperature. The container was compressed and released in the ranges of normal blood pressure (80 – 120 mmHg) shown in Figure 3-14. The characterized packaged pressure sensor operated within 0.4 mmHg error over the range of 0 - 400 mmHg.

These results show these packaged sensors have a good linearity and have enough sensitivity to measure the blood pressure even with added resistance of metal lines and contact resistance.



(a)



(b)

Figure 3-13. *Measured pressure vs. voltage: (a) nitrogen test (regulated nitrogen pressure to measure change in voltage from the sensor with 2.5V_{exe}) and (b) water test (varied depth of sensor to change pressure and measured change in voltage from the sensor).*

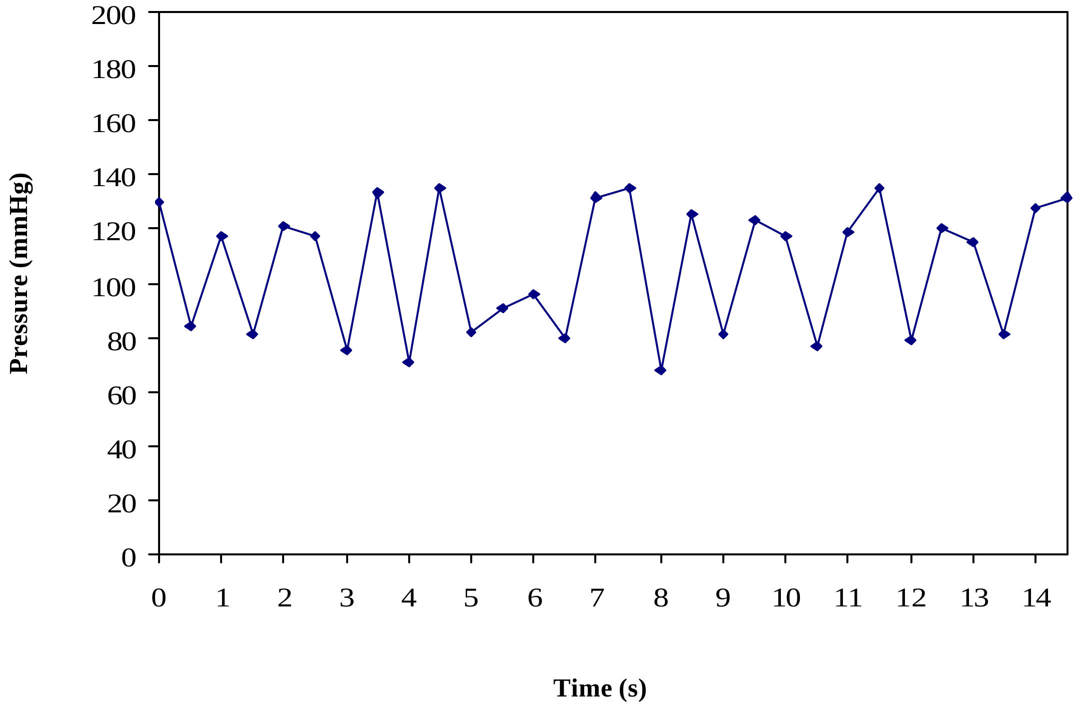


Figure 3-14. *Dynamic pressure measurement in water chamber, where the water chamber was compressed and released to simulate human blood pressure.*

CHAPTER 4. CONCLUSION

4-1. SUMMARY

In this work, piezoresistive pressure micro sensors were successfully packaged using flip-chip bonding on flexible Kapton films for dual lumen neonatal catheters. The whole packages were flexible and small enough to follow the twists and turns of the blood vessels. After flip-chip bonding with the conductive polymer bumps, the contact resistances measured by a four-terminal method for 25 µm high bumps with 400 µm × 400 µm, 300 µm × 300 µm, 200 µm × 200 µm, and 100 µm × 100 µm areas were 17 m?, 39 m?, 77 m? and 149 m? respectively. The humidity test for the packaged pressure sensor for the neonatal catheters shows pretty linear results for two weeks in 27 °C and 36 °C. In the bonding strength measurement, the average failure stress is around 0.0044 N/mm^2 and every bonding strength measurement shows pretty close value which shows pretty stable bonding techniques. The packaged pressure sensors were operated linearly with slope of 16.7 mmV/mmHg even with the added resistances of the gold wires to external control panels in both gas and liquid environment for over two weeks. The flip-chip bonding and packaging developed in this work ensured the durable operation in both gas and liquid environments, which indicate the developed technique is rugged enough for the neonatal catheter.

4-2. SUGGESTIONS FOR FUTURE WORK

Thinner thickness of flexible substrates can be used for more flexible application. In the humidity test, after two weeks, the under-fill of UV-Cured adhesive became white and expanded. Better moisture-resistance under-fills can be used for the longer implantation. For simple application, differential pressure flow sensors, which are bonding two pressure sensors on one Kapton cable can be made.

The polymer flip-chip bonding on the flexible substrates method developed in this work will lead to “smart” catheters integrated with a full array of metabolic and physiological sensors. In addition, the polymer flip-chip bonding on the flexible substrates technologies can have numerous applications for other BioMEMS applications.

REFERENCES

[1] L. Li, J. E. Morris, J. Liu, Z. Lai, L. Ljungkrona, and C. Li, "Reliability and failure mechanism of isotropically conductive adhesive joints," in *Proc. 45th Electron. Comp. Technol. Conf.*, pp. 114–120, 1997.

[2] S. Takeda, T. Masuko, Y. Miyadera, M. Yamazaki, and I. Maekawa, "A novel die bonding adhesive—Silver filled film," in *Proc. 47th Electron. Comp. Technol. Conf.*, pp. 518–524, 1997.

[3] Liu J, "Overview of conductive adhesive joining technology in electronic packaging applications," In: *Proce 3rd Adhesive Joining and Coating Technlogy in Eledtronics Manufacturing Conf,* Binghamton, NY, pp. 1-18, September 1998.

[4] F.W. Kulesza and R.H. Estes, *Adv. Packaging*, vol.6, pp. 26, 1997.

[5] U.S. Patent d5,074,947, December 1991.

[6] T. LeClair et al., *Proceedings of the International Symposium on Microelectronics (IMAPS)*, pp. 758, 1998.

[7] J.F.L. Goosen, "Design Considerations for Silicon Sensors for Use in Catheters and Guide Wires," *Smart Mater. Struct*, vol. 11, pp. 804-812, 2002.

[8] J. Sorrell and A. Bhattacharya, "Mounting of MEMS Sensors on Coronary Guide-Wires," University of Colorado at Boulder, 1998.

[9] K. W. Oh and C. H. Ahn, " Development of an Innovative Flip-Chip Bonding Technique Using Micromachined Conductive Polymer Bumps," *Proc. IEEE Solid-State Sensor and Actuator Workshop*, Hilton Head, SC, June 8-11, pp. 170-173, 1998.

[10] K. W. Oh, C. H. Ahn and K. B. Roenker, "Flip-Chip Packaging Using Micromachined Conductive Polymer Bumps and Alignment Pedestals for MOEMS," *J.Sel. Top. Quantum Electron.*, vol. 5, pp. 119-126, 1999.

[11] C. Y. Li, F. E. Sauser, R. Azizkhan, C. H. Ahn, I. Papautsky, "Polymer Flip-chip Bonding of Pressure Sensors on Flexible Kapton Film for Neonatal Catheters," *Proceedings of the 17th IEEE MEMS Workshop (MEMS '04)*, pp. 749-753, 2004.

[12] H. Chau and K. Wise, "An ultraminiature solid-state pressure sensor for a cardiovascular catheter," *IEEE Trans. Elec. Dev*, vol.35, pp.2355, 1988.

[13] B. Ziaie and K. Najafi, "An implantable Microsystem for Tonometric Blod Pressure Measurment," *Biomedical Microdevices*, vol. 3, pp. 285-292, 2001.

[14] L. Christel and K. Petersen , "A catheter pressure sensor with side vent using multiple silicon fusion bonding," *Proc. Int. Conf. Solid-State Sensors and Actuators (Trandsucers'93)*, pp. 620-623, 1993.

[15] E. Kalvesten, L. Smith, L. Tenerz, and G. Stemme, "The First Surface Micromachined Pressure Sensor for Cardiovascular Pressure Measurments, " *MEMS'98*, pp.574-579, 1998.

[16] B. Acklin, J. Bellermann, M. Schienle, L. Stoll, M. Honsberg, and G. Muller, "Self-aligned Packaging of an Optical Switch Array with Integrated Tapers," *IEEE Photon. Technol. Lett.*, vol. 7, pp. 406-408, 1995.

[17] W. Hunziker, W. Vogt, H. Melchior, P. Buchmann, and P. Vettiger, "Passive Self-aligned Low-cost Packaging of Semiconductor Laser Arrays on Si Motherboard," *IEEE Photon. Technol. Lett.*, vol. 7, pp. 1324-1326, 1995.

[18] STANDARD of Electric Power Industry of China DL/T596-1996, "The preventability test procedure for electric power facility".

[19] Mary Anne Tupta Hone, "your contact-resistance measurements," *Quality Magazine*, October 1995. (www.qualitymag.com).

[20] China National STANDARD GB/T15248-1994, "Measuring method for contact resistance of precious metal electrical contact materials".

[21] EIA STANDARD TP-06B MAY 2000, "Contact Resistance Test Procedure for Electrical Connector," EIA-364-06B

[22] T. C. Shen, G. B. Gao, and H. Morkoc, "Recent development in ohmic contacts for III-V compound semiconductors," *J. Vac. Sci. Technol. B*, vol. 10, no. 5, pp. 2113-212, 1992.