High Density Interconnects and Flexible Hybrid Assemblies for Active Biomedical Implants

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***Abstract*—Advanced microtechnologies offer new opportunities for the development of active implants that go beyond the design of pacemakers and cochlea implants. Examples of future implants include neural and muscular stimulators, implantable drug delivery systems, intracorporal monitoring devices and body fluid control systems. The active microimplants demand a high degree of device miniaturization without compromising on design flexibility and biocompatibility requirements.**

In need for integrating various microcomponents for a complex retina stimulator device, we have developed a novel technique for microassembly and high-density interconnects employing flexible, ultra-thin polymer based substrates. Pads for interconnections, conductive lines, and microelectrodes were embedded into the polyimide substrate as thin films. Photolithography and sputtering has been employed to pattern the microstructures. The novel "MicroFlex interconnection (MFI)" technology was developed to achieve chip size package (CSP) dimensions without the requirement of using bumped flip chips (FC). The MFI is based on a rivet like approach that yields an electric and mechanic contact between the pads on the flexible polyimide substrate and the bare chips or electronic components. Center to center bond pad distances smaller 100 μm were accomplished.

The ultra thin substrates and the MFI technology was proved to be biocompatible. Electrical and mechanical tests confirmed that interconnects and assembly of bare chips are reliable and durable. Based on our experience with the retina stimulator implant, we defined design rules regarding the flexible substrate, the bond pads, and the embedded conductive tracks. It is concluded that the MFI opens new venues for a novel generation of active implants with advanced sensing, actuation, and signal processing properties.

***Index Terms*—Biomedical microdevices, electromechanical contact, flexible substrates, high density interconnects, high density packaging, microassembly, microimplants, neuroprosthetics, polyimide, retina implant.**

I. Introduction

ACTIVE biomedical implants employ electronic and technical systems to restore physiological functions that were destroyed or impaired either by disease or accidents. Smallest footprint, high functionality, highest reliability and biocompatibility are one of the most important requirements for active medical implants. The smaller the device the less invasive is the procedure of implantation. On the other hand, devices need to be packaged to have a biocompatible tissue interface and to withstand biodegradation in the body. Packaging consumes most space in biomedical implants. Established biocompatible materials for housing electronics for example in pacemakers are titanium, glass and specific ceramics. The packaging is rigid and dimensions exceed sizes of several centimeters. Flexible structures like electrode wires and lead outs are insulated and covered with silicone rubber (PDMS), polyurethane and other polymer materials which have been extensively tested for their biocompatible material properties. The development of flexible polymer carriers for holding and interconnecting chips and miniaturized components offer the possibility to develop microelectronic and microoptical systems which are in direct contact with delicate soft tissues and biological structures. Instead of standard housed IC components bare or "naked" silicon chips and dice are used for hybrid integration to minimize component dimensions. The combination of mold array process (MAP) and ball grid arrays (BGA) yield package sizes from 1.6 to 1.3 mm [1]. Chip scale packages (CSP) combine flip-chip (FC) technology with surface mounting technology (SMT) and BGA [2]. Wafer level CSPs (CSP-WL) lead to smallest package footprints since no additional interposers, lead frames and polymer layers are needed [3]. The process involves a six step redistribution and bumping procedure to prepare the CSP chip for SMT. The CSP chips can be mounted on rigid FR4 boards [2] as well as on rigid-flex [3] or flexible substrates and carriers [4], [5].

This paper introduces a "MicroFlex Interconnection (MFI)" technology. The MFI technology employs photolithographic processes to prepare an ultra-thin (5 μm) film on which unprepared single silicon dice with conventional Al pads can be mounted. No additional solderable layers and bumps are needed to prepare the chip. The chip is mounted on the flexible substrate in a rivet-like approach employing gold ball studs with thermosonic bonding. The footprint of the interconnect is smallest, corresponding to the size of the chip. MFI allows high-density interconnects with center to center pad distances less than 100 μm. The applied materials and material compositions have been extensively tested for biocompatibility since the method was invented for biomedical applications. Passive light-weight polymer devices with integrated microelectrodes have been described previously for interfacing regenerating peripheral nerves [6]. The MFI technology is described in this paper. The MFI technology was invented for hybrid integration of a retinal implant. The retina implant has been developed for stimulating the ganglion cells of the retina. In the eye, space for implantation is very limited. The MFI technology has served as a high density interconnection and packaging technology to integrate microelectrode arrays and silicon chips for telemetry and nerve stimulation on a single, highly flexible substrate with smallest footprint. The layout of the MFI flexible substrates were designed in a way that surface mount devices (SMDs) and other microcomponents could be soldered

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TABLE I
PHYSICAL PROPERTIES OF POLYIMIDE PYRALIN PI 2611

	tensile strength	elongation	density at 25 °C	Young's modulus	glass transition	melting point	coefficient of thermal expansion (CTE)	dielectric constant @ 1 kHz, 50 % RH	dissipation factor @ 1 kHz, 50 % RH	dielectric strength	volume resistivity
units	kg/mm²	%	g/cm³	kg/mm²	°C	°C	(°C) x 10^{-6}	(ε_r)	(tanδ)	V/cm	Ω·cm
PYRALIN PI 2611	35	25	1.07	854	> 400	none	3	2.9	0.002	$2 \cdot 10^6$	> 10^{16}

directly onto the flexible carrier. In the following, we describe the requirements for implant biocompatibility in regard to materials and mechanical performance, the manufacturing of the ultra-thin flexible polyimide carriers with embedded metallization layers, and the MFI process including reliability testing. Design rules are presented which were derived from the requirements for active retina implants.

II. BIOCOMPATIBILITY OF MATERIALS

Biomedical implants have to fulfill different requirements for a long-term implantation. They have to be stable in the physiologic environment, i.e., no degeneration should occur. All parts of an implant have to be non toxic for the cells or have to be encapsulated with a non toxic material that serves as a diffusion barrier for these substances. The devices have to be designed with a "smooth" geometry to prevent induced trauma of tissue and nerves by sharp edges and corners. Additionally, the material itself should not be brittle or heavy weighted. The implant substrate should be flexible to conform to the natural soft tissue structure. A retinal implant is a good example. The shape of the implant structure has to adopt to the concave shape of the inner eye. Among other polymers, polyimide (PI) was identified as a potential candidate which serves as substrate and carrier for microelectrodes, silicon chips and embedded cable strands. Certain polyimides have proven to be biocompatible when interacting with blood [7]. Polyimide has also been used as an insulator on flexible 125 μm thin Kapton carriers with vacuum evaporated thin film multielectrode arrays [8]. The polyimide was patterned photolithographicly and etched with photoresist developer. The whole microelectrode structure had been applied to record cardiac signals inside the wall of the heart. Various polyimide types are available on the market. We performed cytotoxicity tests according to ISO 10 993 to confirm the biocompatible performance of the polyimides. Most of the examined polyimides turned out favorable to be used as biocompatible implant material. Inflammatory reactions of Polyimide in cochlea implants were modest, when present [9]. We have chosen the polyimide PYRALIN PI 2611(HD Microsystems). PI 2611 was particularly selected because of its relatively low water uptake and its thermal expansion coefficient near to the one of silicon nitride (Si_3N_4). The PI 2611 exhibits excellent insulation characteristics and dielectric strength at a low density as well as high material flexibility. A summary of the physical properties of PI 2611 is shown in Table I.

PI 2611 is processed with standard cleanroom equipment for microelectronics. It is patterned using reactive ion etching (RIE). The processing technology of PI 2611 is shown in the following section.

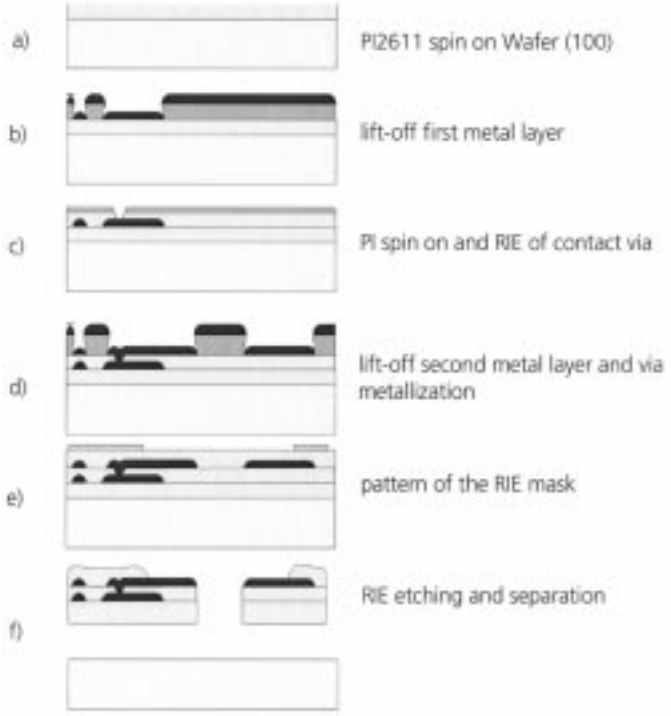


Fig. 1. Process for micromachining ultra-thin, flexible polyimide substrates with patterned microelectrodes, tracks, and interconnection pads.

III. FLEXIBLE MICROMACHINED SUBSTRATES AND EMBEDDED METAL STRUCTURES

We have developed a process technology for manufacturing ultra-thin highly flexible substrates on which metal microelectrodes, conducting tracks, and interconnection pads are deposited. The final PI design provides for interconnecting silicon chips and SMDs directly on the thin PI film when using the MFI technology. The thin PI ribbon films and the MFI are manufactured with standard thin film and micromachine technologies. To reduce interfaces and expensive packaging techniques the electrodes and substrates were designed using the same material in a single process. The multi-layer process for integrating metal tracks, micro vias and MicroFlex interconnections is illustrated in Fig. 1. At first, a layer of polyimide is spun onto a silicon wafer. The metallization (Ti/Au, Pt, Ir) for interconnection pads, connecting lines, and electrode sites was deposited in a subsequent step using thin film sputtering. Struc-

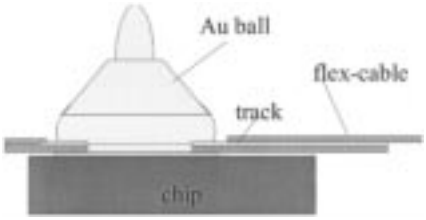


Fig. 2. Schematic illustration of the "rivet-like" electronic and mechanic interconnection between the thin (15 μm) polyimide flexible ribbon substrate and the chip underneath. The interconnection technique has been termed MicroFlex interconnection (MFI).

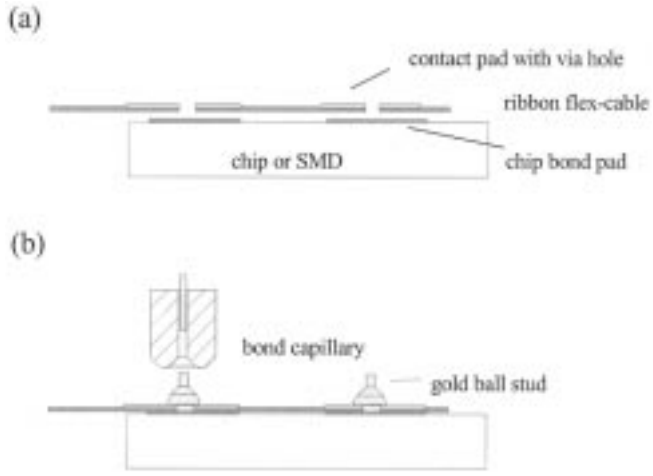


Fig. 3. Graphical illustration of the modified ball-wedge bonding process for MicroFlex interconnects. (a) Alignment of the PI ribbon substrate with the chip underneath. (b) Bonding and fixation of the chip to the PI substrate.

tures were patterned applying a lift-off technique [Fig. 1(b)]. After metallization, a second PI layer for insulation was spun onto the wafer [Fig. 1(c)]. RIE was used for patterning vias (10 μm) into the second PI layer [Fig. 1(c)]. The top and the bottom metal layer (Ti/Au, Pt, Ir) were interconnected through the vias by deposition of a second metallization layer [Fig. 1(d)]. In the final step, a polyimide layer of 5 μm was spun on for insulation. RIE was used to open the electrode sites and connection pads and to separate the devices by etching the outer shapes down to the support wafer [Fig. 1(f)]. Subsequently, the PI substrates with the embedded metallization layers were stripped from the wafer. The single devices were lifted manually from the wafer with tweezers. The PI ribbon substrates exhibited a total thickness not exceeding 15 μm. The substrates were highly flexible and bendable.

IV. MicroFlex Interconnection (MFI) Technology

When devising active neural implants it is mandatory to integrate passive and active electronic components close to the microelectrodes. We developed a new assembly method to interconnect ICs and surface mount devices (SMDs) with our highly integrated flexible ribbon substrates achieving smallest packaging dimensions. We termed this technique MicroFlex interconnection (MFI). One of the key characteristics of the MFI is the introduction of connection pads that have a central via. The via is used to place a stud on top of the metal pad, which connects through the via of the ribbon cable to the bottom chip. The

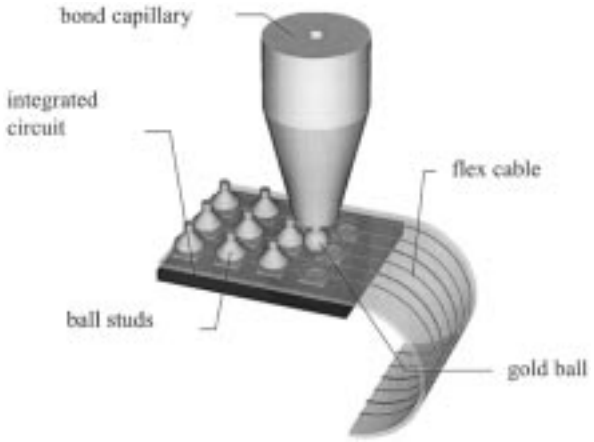


Fig. 4. Three-dimensional graphic presentation of the MFI process connecting patterned metal structures embedded in polyimide films with aligned chips at the bottom.

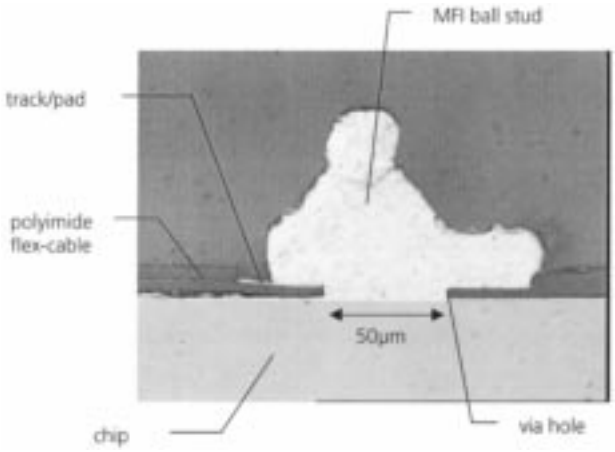


Fig. 5. Metallurgic micrograph showing a vertical cut through the interconnected microstructures.

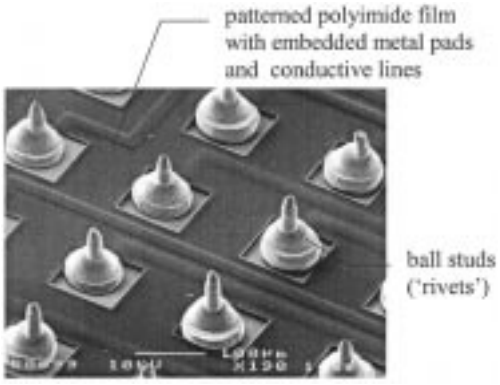


Fig. 6. SEM micrograph of the patterned polyimide/metal film with "rivet-like" ball studs placed in arrays.

"rivet-like" interconnection is performed utilizing a common thermosonic ball bumping process. A gold ball is bonded onto a metal pad by applying force, temperature, and ultrasonic energy (Fig. 2). The gold ball acts as a stud or metal "rivet" that electrically and mechanically interconnects the metallized PI carrier with the chip or with other microcomponents or substrates underneath. After aligning the PI ribbon pads and vias with

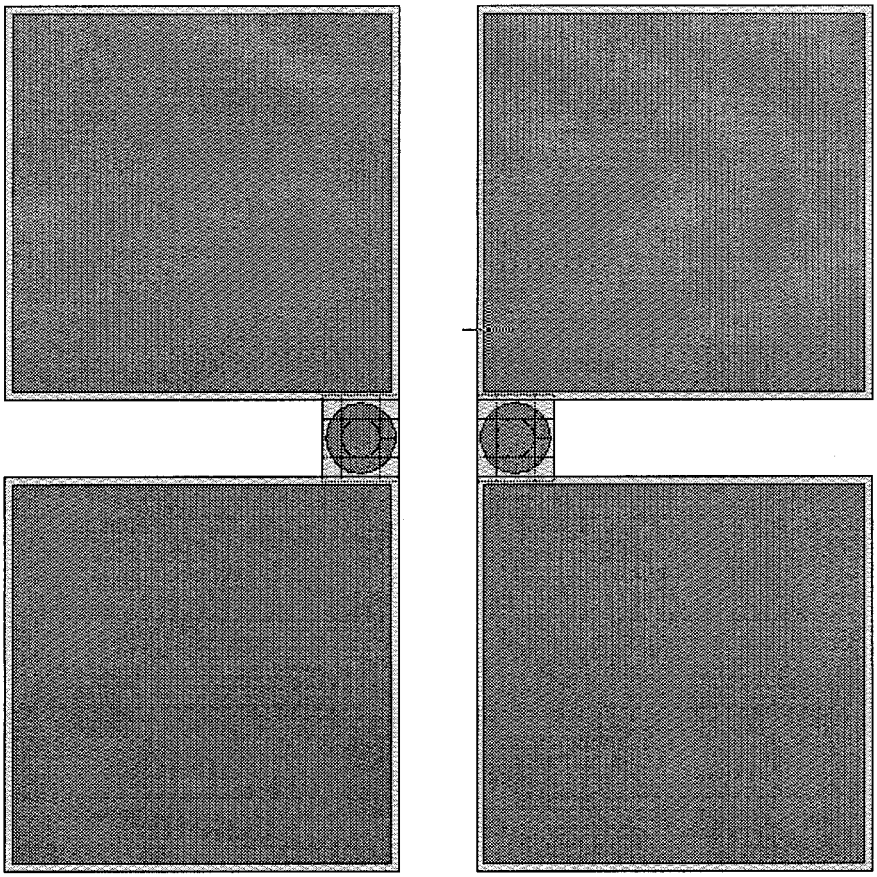


Fig. 7. Metal test structure embedded in polyimide to measure the resistance of the MicroFlex interconnects. Via holes have a diameter of 50 μm.

the metal pads of the bottom structure [Fig. 3(a)] a ball-wedge process is initiated. The capillary is lifted [Fig. 3(b)] and the wire is separated from the bond. Since the via holes of the PI structure are slightly smaller than the gold stud, but large enough to achieve a gold ball bond through the via hole onto the substrate pad, a robust mechanical and electrical interconnection is achieved. A graphical three-dimensional representation of the MFI process is given in Fig. 4. Fig. 5 shows a metallurgic photo through the ball stud which connects the metal pads of the polyimide flex-cable with the bottom chip. A tight connection of the ball stud with the polyimide foil and the bottom substrate is visible. In Fig. 6, an array of the rivet-like interconnects is shown.

V. Testing of Interconnection Properties

A test structure was designed for investigating possible resistance changes of ball studs under various conditions (Table II). Four large gold pads (3, 5 $\times$ 3, 5 mm^2, 300 nm layer) and two small gold pads with 50 μm vias were embedded in a 10 μm polyimide film. Each of the small pads was connected to two large pads Fig. 7). The pads were patterned to perform 4-wire resistance measurement. The rivet connected the polyimide structure through the holes in the small pads to the aluminum substrate below. Aluminum is a common material for contact pads on bare chips. A 1.5 mm thick aluminum substrate was used to keep the resistance of the substrate as low as possible for not interfering with the resistance measured for the bull studs. The resistance was measured over two contact bonds [Fig. 8(a)]. The

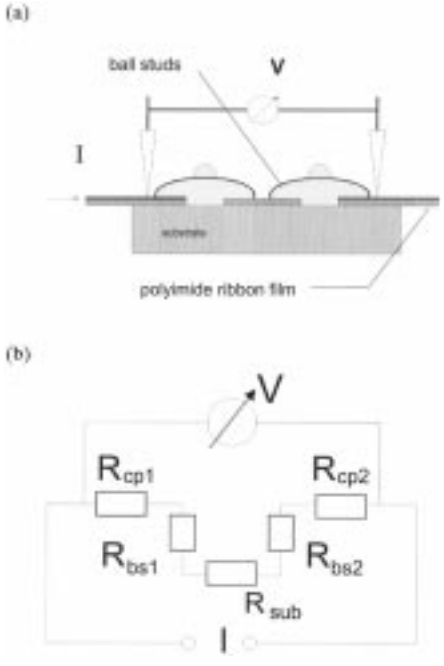


Fig. 8. Design of the (a) four-wire resistance measurement setup and the (b) corresponding analog model to measure the resistance of MicroFlex interconnects. R_{cp} = resistance contact pads, R_{bs} = resistance ball studs, R_{sub} = resistance of Al substrate.

corresponding model of the measurement is shown in Fig. 8(b). Various tests were performed according to MIL standard 883.

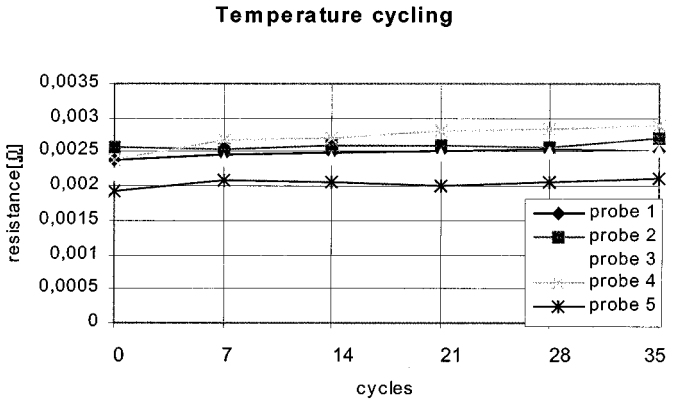


Fig. 9. Resistance measurements of MicroFlex interconnects when exposed to temperature cycles from −40 °C to 150 °C over a period of 111 h corresponding to 35 temperature cycles.

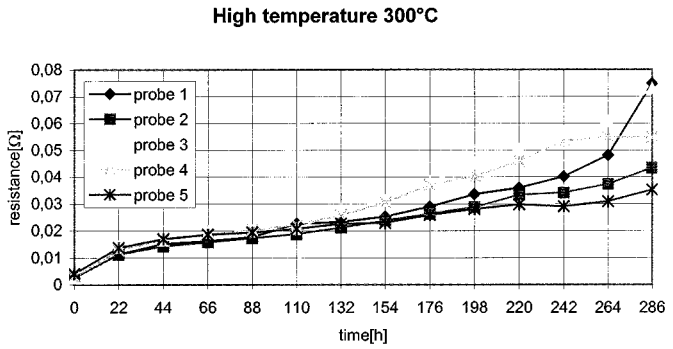


Fig. 10. Effect of high temperature (300 °C) during 12 days on the resistance of MicroFlex interconnects.

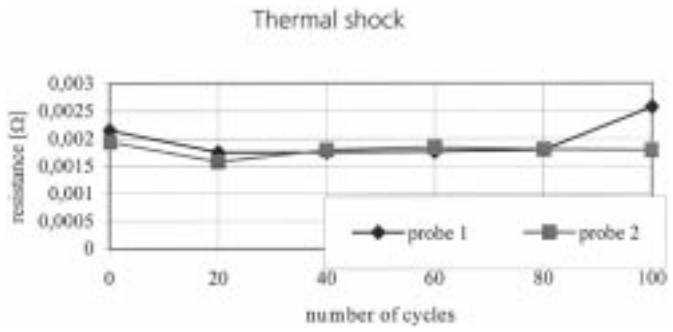


Fig. 11. Resistance measurements of MicroFlex interconnects as a function of thermal shock cycles.

The contact resistance was measured before and after exposure to the testing procedure. Measurements were taken on several samples and average values were calculated. The results on temperature cycling, high temperatures, temperature shock and humidity exposure are seen in Figs. 9–12, respectively. The results are summarized in Table III. No major change of resistance was observed during temperature cycling, temperature shock, and exposure to high humidity. A significant increase of resistance was observed when the testing structure was exposed to high temperatures of about 300 °C. The resistance changed from less than 5 mΩ to more than 70 mΩ.

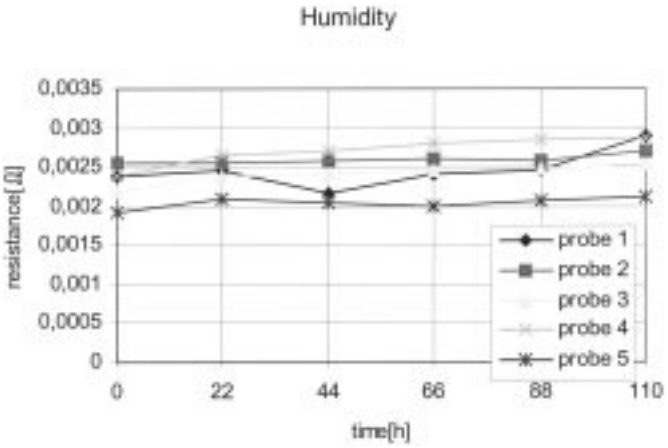


Fig. 12. Effect of 95% relative humidity on the resistance of MicroFlex interconnects during five days.

TABLE II
PROCEDURES FOR TESTING THE RELIABILITY OF THE MFI TECHNIQUES. THE PROCEDURES WERE ADOPTED FROM MIL STANDARD 883

Temperature cycle	-40°C to 150 °C in air, 5 min dwells for 110 h
Temperature shock	–180°C and 100°C liquid temperatures, 100 cycles
Biased humidity and temperature	70°C / 95% relative humidity for 110 h
High temperature operational life	300°C, 240 h
Vibration	50- 2000 Hz, g_{Peak}=20 m/s², 48 h

VI. MFI TECHNOLOGY APPLICATION IN IMPLANTABLE NEURAL STIMULATORS

The MFI technology was developed to design an implantable retina stimulator. The retina stimulator comprised a flexible substrate with integrated microelectrodes, two bare silicon chips, one diode, one capacitor, and a receiver coil. The application specific receiving chip decodes the inductively transmitted energy and data signals. The other silicon chip controls the addressing of one of the 24 microelectrodes of the stimulator array. All components were assembled on an individually designed polyimide substrate. The bare chips were mounted and interconnected on the flexible PI substrate using the MFI technology. The SMD capacitor and SMD diode were soldered onto 300 nm thick Au pads, which were sputtered on the flexible substrate. Solder paste (Sn62 Pb36Ag, GLT, Pforzheim Germany) was dispensed on the Au pads. After SMD components were positioned onto the paste.

The PI carrier including the SMD was placed on a hot plate and heated to 230 °C until the melted solder homogeneously formed around the SMD and on the Au pads. A sketch of the assembly is shown in Fig. 13. Fig. 14 shows the front end of the retina stimulator device with microelectrodes placed on a double ring polyimide structure and the stimulator chip mounted and interconnected using the MFI technology. In Fig. 15, the energy and signal receiving part of the retina stimulator is shown. All bare chips have been assembled and interconnected with the MFI technology. A complete version of an MFI assembled retina stimulator is seen in the photo of Fig. 16. The flexibility of the polyimide substrate with the hybrid assembled components is clearly visible. In a final step, the electronic components and the polyimide substrate are encapsulated with medical

TABLE III
RESULTS ON RESISTANCE CHANGES OF THE "MICROFLEXED" INTERCONNECTS BEFORE AND AFTER EXPOSURE TO TEST CONDITIONS

test conditions	resistance [mΩ] before	standard deviation [mΩ] before	resistance [mΩ] after	standard deviation [mΩ] after
Temperature cycle	2,29 (n=45)	0,22	2,54 (n=45)	0,26
Temperature shock	2,04 (n=18)	0,11	2,20 (n=18)	0,39
Biased humidity and temp.	2,23 (n=45)	0,19	2,08 (n=45)	0,29
High temp. operational life	2,88 (n=45)	0,58	56,15 (n=45)	15,74
Vibration	2,54 (n=18)	0,157	1,79 (n=18)	0,14

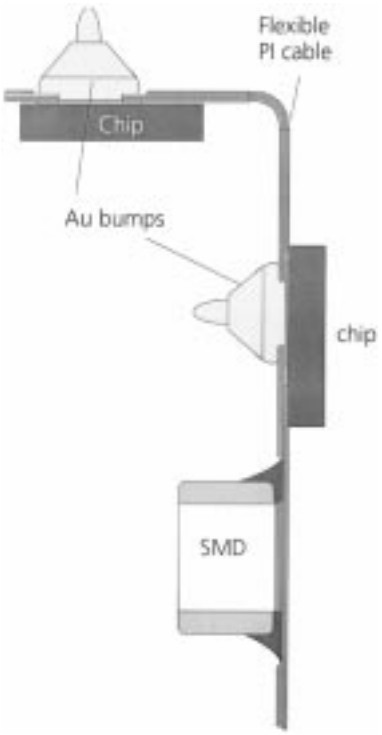


Fig. 13. Sketch for the MFI assembly of silicon chips and SMDs on either side of the flexible polyimide substrate.

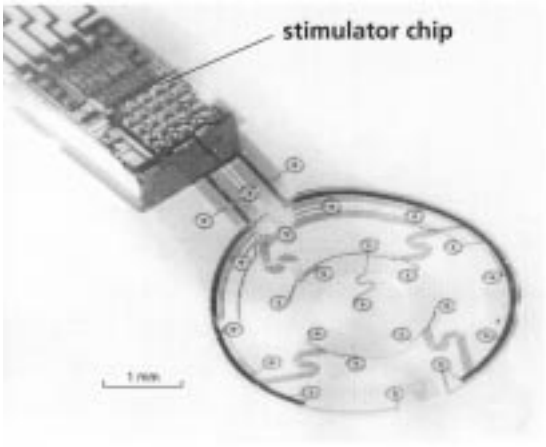


Fig. 14. Micrograph showing the front end of the retina stimulator device. The chip was assembled and interconnected using the MFI technology.

grade silicone (e.g., from NuSil Technology, Carpinteria, CA). Currently, we are working on advanced devices for implantable neural stimulators. Fig. 17 pictures an MFI assembled and interconnected generic multiplexer system. The multiplexer was

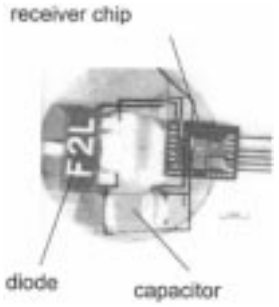


Fig. 15. Energy and signal receiving part of the retina stimulator containing MFI assembled microparts.

placed into a medical grade silicone tube (25 mm length, 3.5 mm inner diameter, 4 mm outer diameter), filled with two part medical grade silicone (NuSil Med-6015), and sealed with a silicone adhesive (NuSil Med-1000). A modified MFI technology was used to connect the stimulator site of the multiplexer with an active stimulator structure, e.g., a cuff electrode for peripheral nerve stimulation.

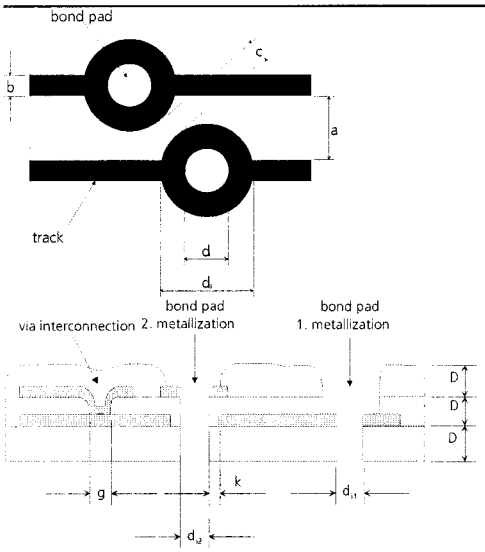
VII. DESIGN REQUIREMENTS FOR HIGH DENSITY FLEX SUBSTRATES AND INTERCONNECTS

From our experience with the retina stimulator we have derived design rules and interconnection density requirements. A summary of the requirements is given in Table IV.

VIII. DISCUSSION ON SUBSTRATES AND INTERCONNECTS FOR BIOMEDICAL IMPLANTS

In need for a new assembling and interconnection technology for a retina implant, we have developed the MicroFlex Interconnection (MFI) technology which enables us to assemble and interconnect bare dice and SMT components on ultra-thin, highly flexible polyimide substrates. Interconnection densities smaller than 100 μm have been achieved. The MFI technique requires minimal interconnection space, corresponding to small space requirements in flip chip technology and chip scale packaging. The MFI technology requires only one bonding process to mount and connect bare chips and other microcomponents to a flexible substrate. Chips and substrates do not need any bumping procedure before assembling and interconnecting. The MFI utilizes simple ball studs from wire bonding as rivets to interconnect the discrete components directly onto the thin polymer substrate. No fiber push connection bonders are needed that go through substrates [4] since our substrates are thinner than 20 μm.

TABLE IV
DESIGN RULES AND INTERCONNECTION DENSITY REQUIREMENTS FOR MICROIMPLANTS



Description		Value
substrate		
thickness	D	4..5µm
track		
thickness		≤0,6µm
width	b	≥10µm
gap	a	≥10µm
bond pad		
inner diameter	d_{i1} d_{i2}	single layer: 30µm ≤ d_{i1} ≤ 80µm multilayer: 50µm ≤ d_{i2} ≤ 80µm typ. d_i=60µm
outer diameter	d_a	round: ∅ d_a ≥ (d_i+30µm) square: □ d_a ≥ (d_i+30µm) x (d_i+30µm) typ. d_a= (d_i+50µm)
gap bond pad/bond pad	c	≥10µm
gap from tracks	k	≥10µm
via		
diameter (multilayer)	g	≥10µm
diameter (single layer)	h	≥20µm

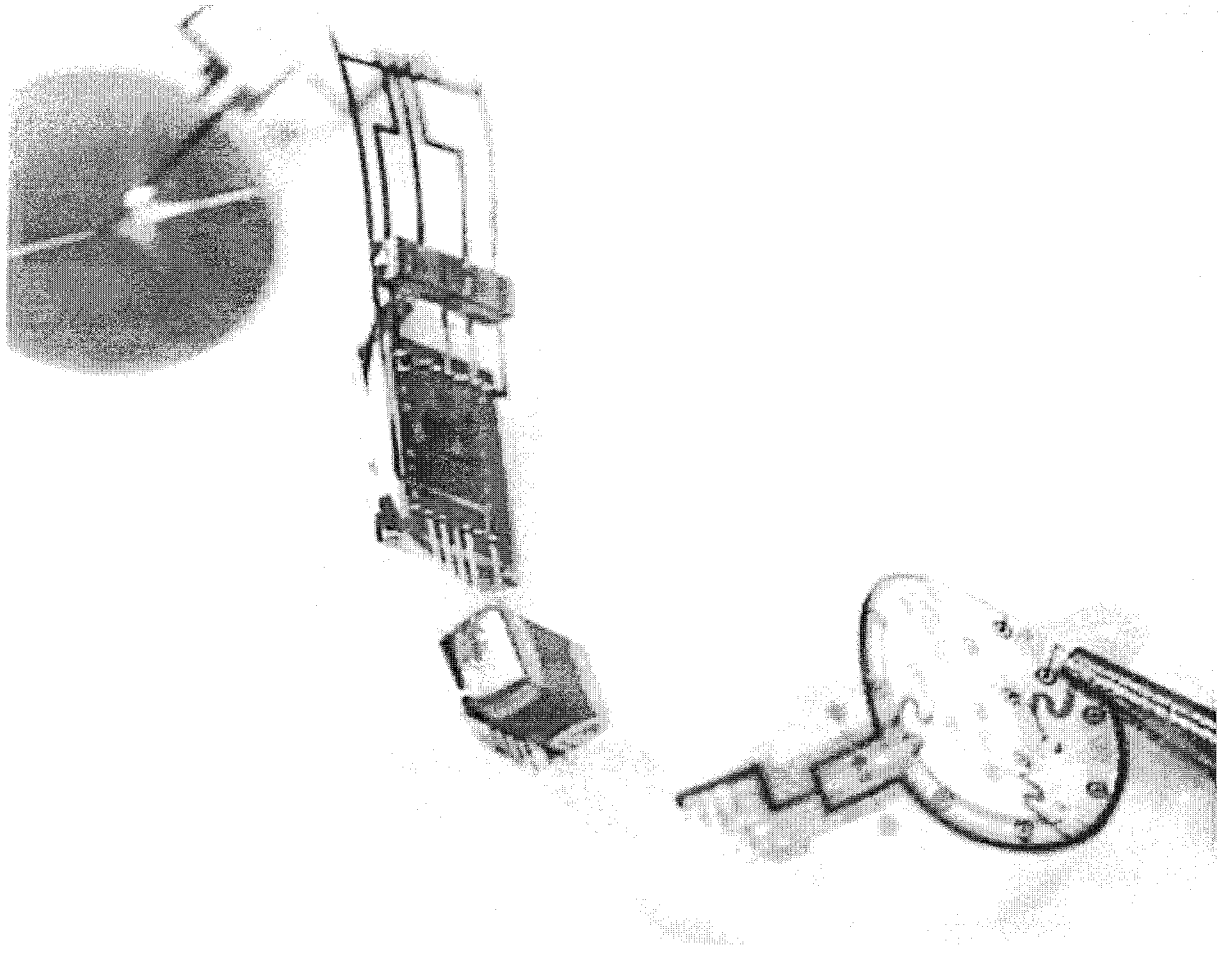


Fig. 16. Photo of the complete assembled retina simulator. The photo demonstrates the mechanical flexibility of the assembled device.

The MIL adopted tests of the MFI technology have demonstrated that MFI is a rather robust and reliable interconnection technology. The increase of resistance during 300 °C temperatures might be explained by the "Kirkendall" effect. The reduced conductivity is most likely due to the diffusion of gold into the aluminum. The Kirkendall effect can be neglected at tempera-

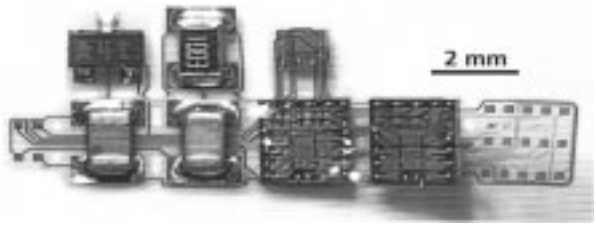


Fig. 17. Generic multiplexer on unfolded polyimide substrate for biomedical applications suited to interconnect to various types of microelectrodes at one end and to multistrand cables on the opposite site.

tures below 40 °C. Implants in contact with living tissue always operate at temperatures below 40 °C, therefore no adverse effects due to diffusion of this type are expected. For other applications, the diffusion of gold into the aluminum can possibly be avoided with diffusion barriers in the IC contact pads.

Polyimide has been used as a major material for the ultra-thin carrier films and insulation layers. Present trends in micropackaging indicate that copper/benzocyclobutene (Cu/BCB) is used for the low resistivity of Cu and for the well established process to pattern BCB [10]. Copper is a widely used conductor material, also in polyimide substrates [4], [5]. Copper is known to be toxic in contact with living tissue causing severe immunological reactions [11]. We have used platinum and gold as rather inert materials known for their biocompatibility [11]. It should be stated that when copper is appropriately isolated from body fluid, copper is used successfully as implant material, despite its toxic reactivity to body tissue. Donaldson *et al.* have achieved excellent protection of thick film microelectronic assemblies using one-part silicone adhesives as an encapsulant [12]. We have used nontoxic platinum as conductors since packing of flexible polyimide substrates with silicone rubber has not been tested extensively enough to be sure that body fluids get in conduct with the conductor. The disadvantage of platinum and gold is their higher resistivity in contrast to the one of copper. We are currently looking into electroplating procedures to increase the conductivity in our devices without the use of copper.

Our microelectronic implants on flexible polyimide substrates have been encapsulated in medical grade silicone rubber. Silicone has been reported as suitable encapsulant for implantable microelectronics including solder material [13], [14]. In order to provide further environmental and dielectric insulation, the use of additional Parylene C coatings will be investigated in future applications. Coatings with paryline have proven to be an excellent choice for brain probes or cochlea ear implants [15].

BCB may prove to be well suited as an alternative material to the polyimide PI 2611. Modified BCB is used as a curing material for certain biomedical implants. Preliminary cytotoxicity tests have shown that the material is nontoxic when used for curing synthetic polyolefins [16]. Extensive tests on the nontoxic effects of BCB are needed. In particular, the tests have to demonstrate nontoxic effects when the material is in contact with highly sensitive neural structures. Furthermore, BCB needs to be proven biocompatible after it has been processed. We learned from our experience with patterning the polyimide 2611, that alterations, for example in gas compositions for reactive ion etching, may lead to a toxic behavior of the patterned material.

Preliminary experiments have shown that polyimide sieve structures with microelectrodces are functional after six month of implantation in animals [17]. The MFI assembled and interconnected retina stimulator is currently tested in animal experiments under operating conditions. It is hoped that the favorable *in-vitro* and *ex-situ* results of the retina stimulator device are confirmed in the animal studies. We also apply the MFI technology successfully in other applications, e.g., for cuff-type neural stimulators, for interconnecting miniaturized ultrasound array sensors and for assembly and interconnection of pressure sensors and control chips in active catheters. It should be stated that the presented approaches for interconnections and assembly definitely do not suit for all active biomedical implant applications. The MFI technology is particularly suited for devices that need to be implanted into biological structures with complicated shapes and extreme confinements in available space. The eye is a good example for such stringent requirements.

The present MFI technology relies on a skilled operator to align the structures, and to place the bonds. In order to establish the MFI technology as a packaging and interconnection technology in the semiconductor industry collaborations are needed to automate the procedures and to develop cost-reduced procedures for high-volume production of MFI assembled devices.

IX. Conclusion

A new technique for interconnecting microdevices with flexible substrates was developed, termed MicroFlex interconnection (MFI) technology. The technique was applied to devise a retina implant microsystem as demonstrated in this paper. Other implantable devices are currently developed, among which are cuff-shaped stimulators with integrated multiplexer, novel miniaturized ultrasound arrays and controller chips for endoscopic applications, and pressure sensor chips and cables for application in catheters.

The MFI technology allows hybrid integration and assembling of bare silicon dice (ICs) and soldered SMD components at either side of the same flexible substrate. The dice are directly bonded onto the substrates with a “rivet” that interconnects through the via in the polyimide film onto the chip. The procedure of screen-printing bumps onto the chip or onto the substrate is avoided. Testing according to MIL standards has demonstrated that the MFI technology is robust and reliable. Future work is directed toward the automation of the MFI technology in order to establish a high-density interconnects and packing technology suited for high-volume applications.

We conclude that the MFI technology, comprising highly flexible substrates, high density interconnects, and simple assembly of bare chips and other microcomponents, opens new venues for a novel generation of active biomedical implants with advanced sensing, actuation, and signal processing properties.

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