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Polymeric Packaging for Fully Implantable Wireless Neural Microsensors

Juan Aceros *Member IEEE*, Ming Yin *Member IEEE*, David A. Borton *Student Member IEEE*, William R. Patterson *Member IEEE*, Christopher Bull, Arto V. Nurmikko, *Fellow, IEEE*

Abstract— **We present polymeric packaging methods used for subcutaneous, fully implantable, broadband, and wireless neurosensors. A new tool for accelerated testing and characterization of biocompatible polymeric packaging materials and processes is described along with specialized test units to simulate our fully implantable neurosensor components, materials and fabrication processes. A brief description of the implantable systems is presented along with their current encapsulation methods based on polydimethylsiloxane (PDMS). Results from *in-vivo* testing of multiple implanted neurosensors in swine and non-human primates are presented. Finally, a novel augmenting polymer thin film material to complement the currently employed PDMS is introduced. This thin layer coating material is based on the Plasma Enhanced Chemical Vapor Deposition (PECVD) process of Hexamethyldisiloxane (HMDSO) and Oxygen (O_2).**

I. Introduction

The potential for treating a broad range of neurological disorders has brought forward the rapid development of implantable neural-interface devices. Further miniaturization, advances in packaging schemes, and the development of short-range wireless broadcasting methods have unlocked new opportunities for fully implantable, chronic devices [1]. These opportunities range from diagnostic monitoring to active treatment of a disease. Regardless of the application, multi-channel broadband (0.1 Hz – 7.8 kHz) recordings of neural activity are always necessary to capture the dynamics of the rich cortical landscape.

The challenge of acquiring these signals has traditionally been met using recording devices with all necessary active electronics located external to the body of a subject. Such devices are capable of recording *in-vivo* neural signals from large groups of interacting populations of neurons with high spatial and temporal resolution, and have successfully been used in pioneering human clinical trials, enabling severely injured patients (e.g. spinal cord injury) to control a "neural cursor", a prosthetic arm, and a multi-jointed robotic arm [2].

The neural signals for such degree of neuroprosthetic control are acquired at the single-cell level using intracortical microelectrode arrays (MEAs), and are sent directly to a percutaneous interface where they are linked to an external unit. Since there are no active electronics in the passive implantable unit to process the signals, the systems are only operational when the subject is directly connected to the external unit. These types of percutaneous recording devices are bulky and potentially unsafe due to mechanical stress on cabling and infection. Furthermore, this tethered arrangement restricts the mobility and independence of the users. Hence, removing the percutaneous connector and replacing it with fully implantable neural sensors is one key next step for the advancement of neural interface technologies.

A major challenge to all researchers attempting to create compact, subcutaneous, active chronic implantable neural sensors is the electrical isolation, or packaging. It was learned early on from the development of the first commercial pacemakers, that reliable packaging of fully implantable biomedical devices is critical to their success. Current long-term stable and biocompatible polymeric insulation processes offer erratic reliability against the wet, ion-rich, and warm interior of the human body. Hence, most packaging schemes for chronic applications have been limited to solid titanium hermetic packages or equally rigid ceramics. However, this type of packaging has a limited future as next generation implantable brain-interface devices continue to increase in functionality, number of components and sensors, while requiring smaller footprints e.g. for placement in the subcutaneous spaces available in the head. For these chronic implantable devices, a new type of packaging scheme using thin flexible materials is necessary. However, developing such a chronic, impermeable, and biocompatible thin film protection layer e.g. against electric field induced ion transport is a tall order, especially since ionic transport mechanisms alone in polymeric insulators are not well known.

In this paper we present our current neurosensor packaging techniques based on epoxy and silicone. We also introduce a new quantitative characterization tool for accelerated testing of packaging materials, and a novel hybrid PDMS/SiO_2 thin film encapsulation material via a deposition process using a combination of Oxygen and Hexamethyldisiloxane (HMDSO) plasma [3]. We consider that this new process has the potential to significantly augment the encapsulation to overcome many of the aforementioned challenges for implantable devices.

This paper is organized as follows: in Section II, we briefly describe the microelectronic system details of our current

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J. Aceros, M. Yin, D. A. Borton, W. R. Patterson, C.W. Bull, and A. V. Numikko are with the School of Engineering, Brown University, Providence, RI 02912 (phone: 401-863-1408; email: Juan_Aceros@brown.edu.)

implantable wireless units. Section III focuses on the neurosensor fabrication of a 100-channel wireless broadband system, followed by a description of the encapsulation quantitative evaluation methods. Section IV presents our conclusions.

II. Brief Implantable System Description

This paper will focus on the use of packaging methods for neural interface devices currently fabricated in our laboratories. The topology for these devices has been described in detail elsewhere [4], and is based on a dual-panel design consisting of a cortical "front-end" sensor unit and an epicranial "back-end" telemetric unit. This Kapton-based flexible substrate topology has been successfully employed through several generations of multiple neurosensors in our laboratories, including the single and dual sensor "front-end" devices shown in Figure 1.

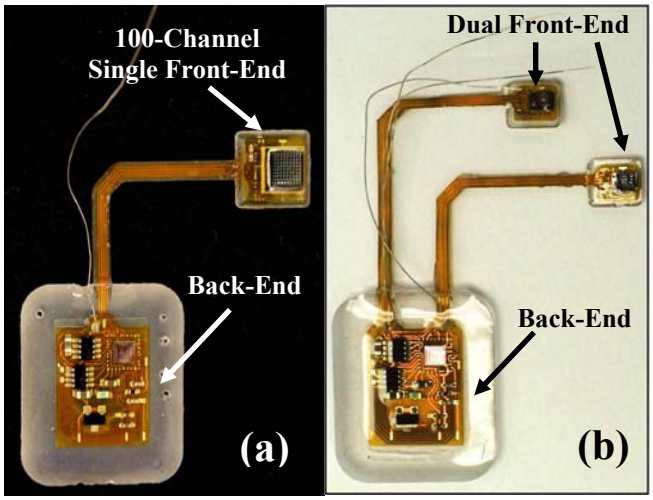

Figure 1. (a) 100-Channel Single Front-End Neurosensor, (b) 32-Channel Dual Front-End Neurosensor (access to two cortical sites).

A. Cortical Sensor: The "Front-End"

As previously mentioned, a crucial feature for the success of these technologies has been the detection of high quality neural signals with rich temporal and spatial information. These types of signals are sensed in our devices using microfabricated silicon MEAs (Blackrock Microsystems). These arrays are integrated with our custom designed Application Specific Integrated Circuit (ASIC) amplifier microchips to create the sensor unit "front-end". These high performance low-power and low-noise microchips require only five interfacing signals to the back-end (embedded within the common Kapton substrate): two power supply connections (*VDD* and *GND*), one sample clock, one synchronization and acknowledge signal, and one multiplexed analog output.

Each of the preamplifiers in the ASIC use a capacitive-feedback, folded cascode operational transconductance amplifier (OTA) configuration with a source follower output buffer. The details of the preamplifier circuit are discussed in [5]; it is important to underscore that our systems allow broadband data recording and telemetry on all 100 channels of low frequency field potentials (FPs) as well as single cell spiking activity.

B. Power and Telemetry: The Epicranial "Back-End"

The epicranial subunit consists of five major blocks: the LC tank for RF inductive power, a diode rectifier, two SAR-ADCs, a digital control ASIC, and a Vertical Cavity Surface Emitting Laser (VCSEL). The system is inductively powered at 13.56 MHz RF power carrier frequency. The digital control ASIC produces the clock and chip select signals for the ADCs from the power carrier, and generates the sampling clock for the preamplifier ASIC in the "front-end" sensor unit. The digitized neural data from the two ADCs are packaged and combined into a single serial bit stream to modulate a current source that drives the infrared laser (VCSEL). The VCSEL transmits the optical signal through the skin to an external infrared receiver, where the original neural data can be recovered [6].

III. Fabrication: Process Flow

A. Fabrication of a 100-Channel Small Footprint Implantable Neural Interface

The 10×10 (100 channel) intracortical microelectrode arrays are purchased from Blackrock Microsystems with electrode heights of 1.5 mm and Pt/Ir tip metallization. These arrays are flip-chip bonded to the previously described ASIC amplifier using our developed process described in [4]. The passive and active device components are populated on the "back-end" and "front-end" of the substrate using silver epoxy. The controller and amplifier ASICs are wirebonded to the substrate using an ultrasonic wedge bonder, and finally a 5mil thick Pt/Ir wire is attached to a pad on the "back-end" to serve as common voltage reference input to all the preamplifiers.

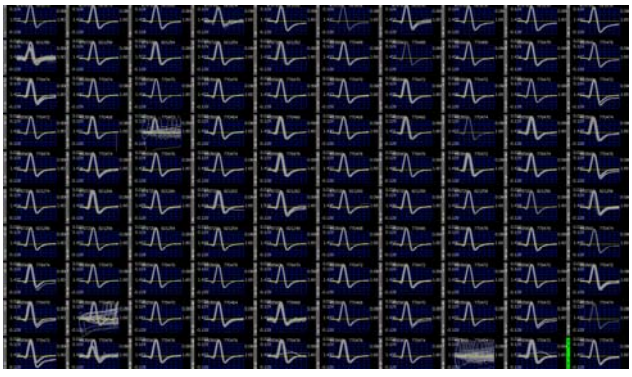

Figure 2. Artificial Spike Signals Sensed Using a 100-Channel Neurosensor.

The active neurosensors are bench top tested after fabrication and before their final encapsulation. The 100 channel system has a 45.6 dB gain, 1 Hz – 7.8 kHz bandwidth, 8.6 μV_{rms} rms input referred noise, and consumes only 52 µW of power. The power dissipation of each block is shown in Table I. Finally, a 100 Hz, 750 μV_{pp} artificial spike function (Figure 2) is used to test the device before silicone encapsulation.

TABLE I
POWER DISSIPATION DISTRIBUTION OF 100-CHANNEL MICROSENSOR

Blocks	Power Dissipation
Front-end Amp ASIC (100-ch)	6mW@3V
Digital control ASIC	10mW (9mW for the VCSEL driver) @3V
Two SAR-ADCs (LTC2366)	9.6mW@3V
Diode Rectifier	2mW
Secondary LC tank	3mW
Total	**30.6mW**

B. Electronic Packaging Methods

Packaging the system to ensure good isolation from the saline rich body environment (and vice versa) is the last step in completing the full fabrication of the device. For our systems, we developed a hybrid encapsulation method using a Hysol epoxy (M-31 CL) followed by complete system encapsulation with polydimethylsiloxane (PDMS, Nusil R-2188). The Hysol epoxy mechanically secures all components and wirebonds, while the PDMS provides a soft, biocompatible, and flexible protective barrier. Although we have successfully packaged devices for long term *in-vivo* applications, we emphasize that this process is highly sensitive to deposition conditions, adhesion, and surface treatment before application of the insulator, e.g. cleaning procedures. We have summarized in Table II the results from various device implantations with this encapsulation scheme into swine (Yorkshire; practical time limits due to animal growth apply) and non-human primates (rhesus macaques).

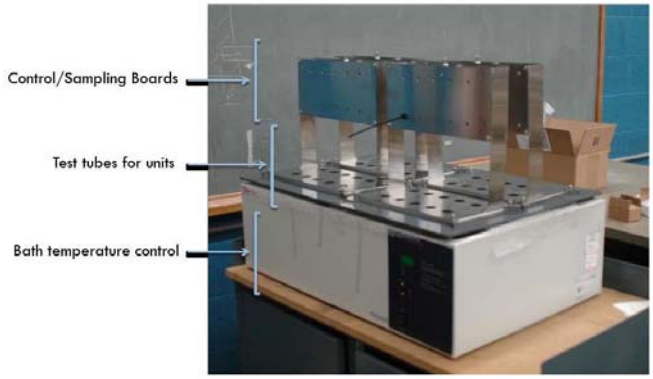

Figure 3. Test System for Accelerated Testing of Polymeric Materials.

We consider this initial encapsulation approach with PDMS a useful starting point with viable applications for short-term neural recording demonstration purposes (at least ~months). However, to create long-term, fully implantable devices more reliable packaging schemes and materials are necessary. To achieve significant statistical certainty on the quality of materials and packaging processes we have created a system capable of testing the encapsulation reliability of polymeric insulators. This system, shown in Figure 3, is capable of accelerated characterization of 100 test units simultaneously during periods of extended soaking in hot saline solution at variable temperatures.

The test units for this system are uniquely designed to simulate the topographical, thermal, mechanical and electrical stresses, as well as the fabrication procedures, materials and components that form the previously described fully implantable active neurosensors. With the test units we are capable of monitoring continuously the resistance between interdigitated conductors on the substrate surface as well as leakage current through the encapsulation material. A picture of these test units is shown in Figure 4.

TABLE II
SUMMARY OF POLYDIMETHYLSILOXANE INSULATED MICROSYSTEMS (EXTERNALLY POWERED MICROSYSTEM, EPM, AS DISCUSSED ABOVE).

Animal ID	Device Implanted	Days Implanted	# Units	Location	ID
Swine					
778	EPM16	N/A	N/A	L-PSC-UR	JU
779	EPM16	25	0	L-PSC-UR	P2
780	EPM16	72	6	L-PSC-UR	P3
787	EPM16	65	2	L-PSC-UR	P4
791	EPM16	36	0	L-PSC	SU
792	EPM100	44	N/A	L-PSC-UR	A1
793	EPM16	149	0	L-PSC-UR	P5
788	EPM16	27	2	L-PSC-UR	FA
Non-Human Primates					
99-4	EPM16	14	0	R-MI	PU
B002	EPM16 x2	508	7	R-MI, PMd	SC
RQ6065	EPM16 x2	226	2	L-MI, PMd	LU
Summary					
Total Device Implants (Pig)		**8**	***Total Days Implanted***		**>418**
Total Device Implants (NHP)		**5**			**>748**
Total Device Implants		**13**			**>1166**

** EPM16 refers to a 16-channel version of the microsystem while EPM100 refers to a 100-channel microsystem. L-PSC-UR = left primary somatosensory cortex, upper rostrum area, this was the area of implantation in swine, while MI = primary motor cortex: the location of implantation in non-human primates.

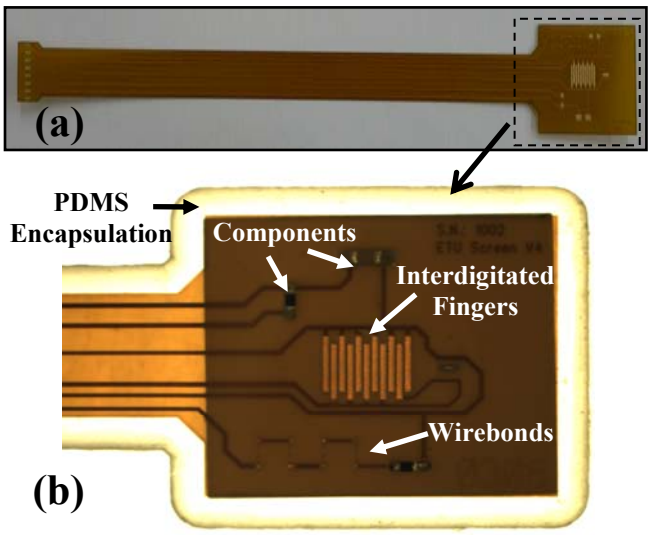

Figure 4. (a) Encapsulation Test Unit. (b) Details of Test Section.

These units are currently being employed to test multiple types of polymeric encapsulation materials, processes and configurations. It is our objective to build a trustworthy database of material properties to share with the scientific community. Work with PDMS (but also Polyimide, Parylene, LCP, and SU8 among others) is underway, as shown in Figure 5 for three of the many PDMS encapsulated test units (here showing development of ionic leakage over period of

months, and unacceptable cases). However, we are also interested in exploring other types of materials with augmenting potential for truly chronic implantable biomedical devices. Our approach is to include material strategies based on the combinations of organic and inorganic thin film multilayers. Here, we outline the adaptation to our implants of a recently developed packaging material based on a nanocomposite of PDMS and SiO_2 [7]. The thin conformal films are produced with a Plasma Enhanced Chemical Vapor Deposition (PECVD) system. This system deposits layers of material from a HMDSO and O_2 plasma mixture. The deposited materials have been reported to provide a reliable hermetic permeation barrier for highly water sensitive Organic Light Emitting Diodes (OLEDs) [7].

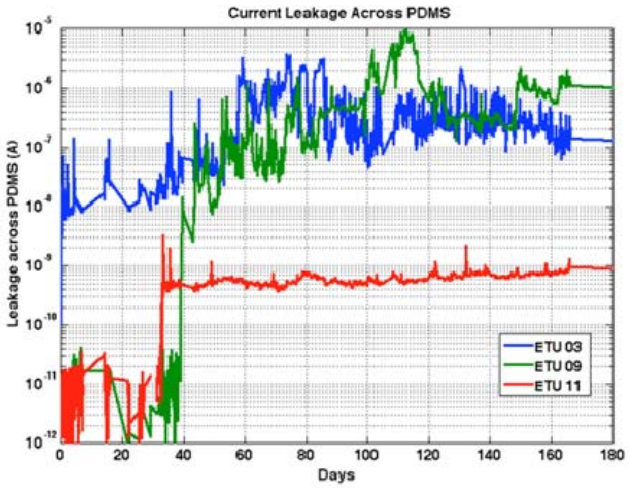

Figure 5. Leakage Current Characterization across the PDMS. Early Failure Appears for ETU 03 and at ~40 days for the others.

The films deposited by this process are a heterogeneous micro/nanoscale mixture composed of SiO_2 and silicone, where networked SiO_2 is viewed to play a role for successfully blocking diffusion paths within the silicone. It has been reported through infrared absorption techniques, wetting angle measurements, and indentation hardness experiments that these films are uniform on the macroscale. Hence, this supports the theory of a SiO_2 filled silicone hybrid film. In addition to the permeation barrier properties, the films are also highly flexible. They have been reported to remain intact with over 58,600 cycles of bending [3].

In our lab, we have successfully developed our own custom in-house PECVD system for device encapsulation. Preliminary infrared absorption photospectroscopy tests support the reported results of the hybrid SiO_2/silicon film. For example, monitoring the IR absorption by the O-Si-O stretch resonance with increase in O_2 flow with respect to HMDSO is one of the methods to define optimal process parameters as shown in Figure 6. Future work will investigate these thin (~100nm) films using AFM, hardness, and wet etching, as well as results from applying these coatings on implantable active neural sensors and test structures for accelerated lifetime verification and material analysis.

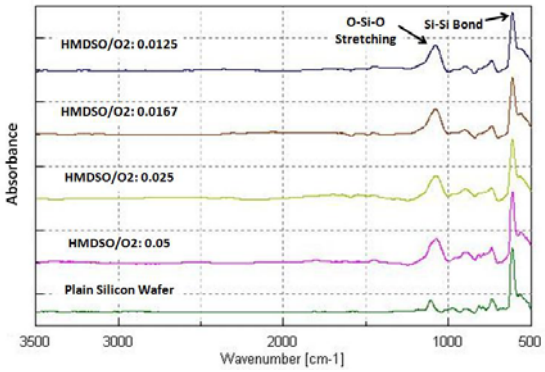

Figure 6. IR Absorbance for Various Flow Ratios (HMDSO = 1.0 sccm).

IV. Conclusion

We have presented an intracortical, wireless microsystem as an example of the packaging challenges for active devices. In addition, we presented current packaging techniques using PDMS and results from swine and monkey *in-vivo* system validation experiments. We described our custom instrumentation for accelerated characterization of polymeric materials and described the specialized test units currently employed for this task. Finally, we introduced a new packaging scheme for soft, flexible hermetic encapsulation based on PECVD deposition of organic and inorganic thin films. This new packaging technique is currently being transitioned for active circuitry insulation in our devices.

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