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***Abstract* – Our aim is to develop a smart neural interface with transparent electrodes to allow for electrical monitoring of the site of interest during optogenetic stimulation of the spinal cord. In this work, we present the microfabrication process for the wafer-level development of such a compact, active, transparent and flexible implant. The transparent, passive array of electrodes and tracks have been developed using graphene, on top of which chips have been bonded using flip-chip bonding techniques. To provide high flexibility, soft encapsulation, using polydimethylsiloxane (PDMS) has been used. Preliminary measurements after the bonding process have shown resistance values in the range of kΩ for the combined tracks and ball-bonds.**

***Keywords* – Neural interface, optogenetic stimulation, active implant, graphene, PDMS.**

I. Introduction

Epidural spinal cord stimulation (ESCS) has been proven to promote locomotion recovery in patients affected by spinal cord injuries (SCIs) [1]. However, optimization of the specifications for such therapies is still under research and identifying the mechanism of action could greatly benefit from parallel monitoring of the response of the biological tissue during stimulation. Usually, in ESCS, energy is injected into the tissue in the form of electrical pulses, leading to activation. Alternatively, energy in the form of light can also be used to activate the tissue, in a more spatially specific manner, using optogenetics.

Already available electrode arrays for ESCS feature opaque electrodes which limit electrical monitoring of the tissue response during optogenetic excitation [2]. Therefore there is the need to develop optically transparent, conductive, and flexible electrodes. This will allow for capturing the electrical activity of the neurons below the activation site at the time of stimulation. One potential material for these electrodes is graphene, as the material is optically transparent, bendable, potentially biocompatible, and has excellent electrical properties [3, 4]. Graphene microelectrode arrays have been previously reported as passive implants with Au tracks to interface the electrodes with the outside active system [3]. However, for a smart implant, ultimately, active components, e.g. integrated circuits (ICs), have to be embedded with the electrodes to allow for signal acquisition, in-situ amplification and processing.

The aim of the current work is the development, by means of microfabrication, of a compact, flexible, graphene-based, active spinal cord monitoring implant for optogenetic stimulation.

II. Materials and Methods

A microfabrication process has been used to ensure repeatability and maintain the small size of the implant while achieving high resolution. Fig. 1 depicts the process steps of the current work.

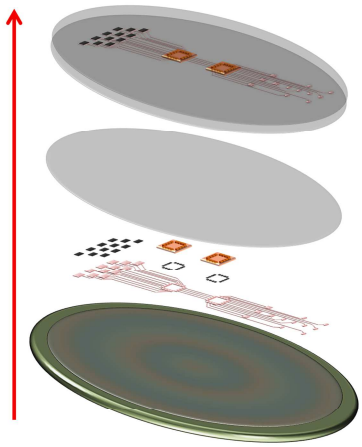

Fig. 1. Process steps of the proposed method. A molybdenum catalyst layer is used for graphene growth, on top of which chips are bonded and later, the complete structure is encapsulated in polymer.

Chemical vapor deposited (CVD) graphene tracks and electrodes have been microfabricated on a wafer level, using a pre-patterned 50 nm Mo layer as a catalyst, deposited on SiO_2 as described in detail in [5]. On top, 100 nm of Ti and 675 nm of Al have been deposited and patterned in order to create a bonding interface between graphene and the Au

stud bumps existent on the pads of the chips. Later, using a thermocompression flip-chip bonding technique, chips were bonded to the substrate. Next, 50 µm of Sylgard 184 polydimethylsiloxane (PDMS), 1:10 ratio, have been spin coated on top of the structure and cured at 90 °C for 1 hour. At this point, the complete structure had to be transferred or released from the original wafer in order to spin coat the final PDMS encapsulation layer. To do so, two approaches have been investigated, as illustrated in fig. 2.

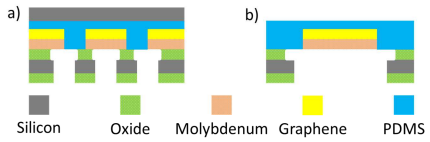

Fig. 2. Approaches used to transfer/ release the structure. In a), a wet transfer approach, using buffered hydrofluoric acid (BHF) 7:1 for oxide etching. In b), a "flex-to-rigid" (F2R) approach [6].

The approach in a) consists of the creation of through-silicon vias (TSV), before graphene growth, to increase the number of access points for the etchant, while the approach in b) consists of a deep reactive ion etching (DRIE) process for cm-size areas (that can later be coated with PDMS as final encapsulation), performed after having the complete structure on the wafer, an approach known as F2R [6].

III. Results and Discussion

First, Raman spectroscopy was performed to ensure the presence of graphene (fig. 3). The ratio I_{2D}/I_G suggests that a multilayer graphene has been grown on the substrate, while the ratio I_D/I_G estimates the amount of defects present in the graphene layer (the greater the ratio, the more defects can be found). The defects originate from the growth process [5].

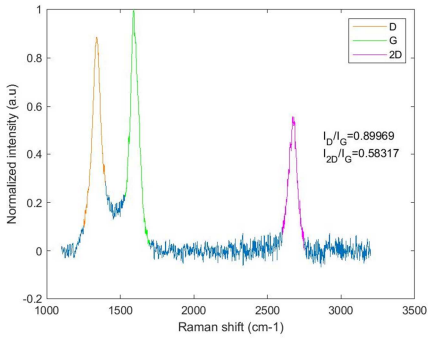

Fig. 3. Raman spectroscopy measurement, taken with a 633 nm laser. The 3 peaks indicate the presence of graphene.

A. "Wet" transfer of structure

BHF etching of the oxide layer from beneath the structure, has been tested. The expected etch rate was 150 nm/min and the total calculated etching time was 40 min. Yet, after 7 hours, no etching around the TSV has been observed. Possibly, DRIE of the TSV resulted in the deposition of a polymer layer which could not be removed by O_2-plasma treatment. To circumvent this, potassium hydroxide (KOH) etching was performed to widen the pathways for the BHF. However, after these long wet etching steps, the structures were highly damaged or even removed. Since later in the process, at this step, chips containing active components will also be present, wet transfer of the structure is to be avoided.

B. "Flex-to-rigid" approach

Fig. 4 shows the results obtained after the DRIE process, for silicon removal, in combination with wet etching of the remaining oxide layer. The complete area of the implant was successfully suspended and the membranes did not contain any significant wrinkles or damages.

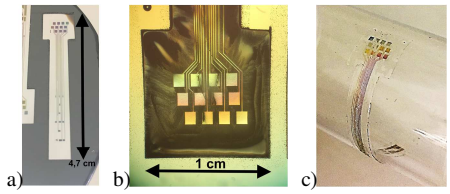

Fig. 4. Implant structure with Mo tracks after DRIE. In a), the complete suspended implant can be observed. In b), a detailed perspective of the PDMS membrane and tracks is presented. In c), the high flexibility of the structure is shown.

C. Flip-chip bonding

So far, no flip-chip bonding processes on graphene substrates have been reported in the literature and since our initial attempt showed that the adhesion between graphene and Au stud bumps is poor, the creation of a metal interface in between has been chosen as an alternative.

Fig. 5 a) depicts a visual representation of the structures before and after flip-chip bonding. In b), a computer tomography (CT) scan, after the bonding process is illustrated, while c) and d) show a preliminary 2-point measurement result. The resistance, ~7.6 kΩ, is the sum of the ball-bond resistance and 2 graphene tracks resistances (one graphene track indicated a resistance value of ~3.7 kΩ). Currently new devices are being fabricated which will allow the measurement of the flip-chip ball-bond only.

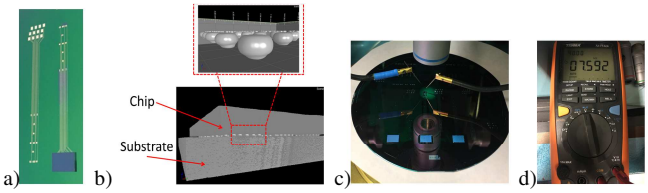

Fig. 5. Preliminary results after bonding. CT scans and 2-point measurements have been performed to ensure that the bonding process was successful.

IV. Conclusions

This work presents the process for developing flexible, active, graphene-based epidural spinal cord monitoring implant, by means of microfabrication only. It has been shown that F2R approach can be used to suspend large areas, thus avoiding as much as possible wet process steps that can damage the structures. Moreover, it has been demonstrated that flip-chip bonding of chips on a graphene substrate, using metal interfaces, is possible and initial measurements have shown that there is electrical conductivity after the bonding process. To the authors' best knowledge, this is the first reported graphene-based active implant.

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