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An all-diamond, hermetic electrical feedthrough array for a retinal prosthesis

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ABSTRACT

The interface between medical implants and the human nervous system is rapidly becoming more and more complex. This rise in complexity is driving the need for increasing numbers of densely packed electrical feedthrough to carry signals to and from implanted devices. This is particularly crucial in the field of neural prosthesis where high resolution stimulating or recording arrays near peripheral nerves or in the brain could dramatically improve the performance of these devices. Here we describe a flexible strategy for implementing high density, high count arrays of hermetic electrical feedthroughs by forming conducting nitrogen doped nanocrystalline diamond channels within an insulating polycrystalline diamond substrate. A unique feature of these arrays is that the feedthroughs can themselves be used as stimulating electrodes for neural tissue. Our particular application is such a feedthrough, designed as a component of a retinal implant to restore vision to the blind. The hermeticity of the feedthroughs means that the array can also form part of an implantable capsule which can interface directly with internal electronic chips. The hermeticity of the array is demonstrated by helium leak tests and electrical and electrochemical characterisation of the feedthroughs is described. The nitrogen doped nanocrystalline diamond forming the electrical feedthroughs is shown to be non-cyctotoxic. New fabrication strategies, such as the one described here, combined with the exceptional biostability of diamond can be exploited to generate a range of biomedical implants that last for the lifetime of the user without fear of degradation.

1. Introduction

Implantable electronic devices must necessarily adhere to a very strict set of standards before they can be approved for clinical use. One of these standards is the requirement that electronically active components such as microprocessors must be encapsulated in a hermetically encapsulated [1], to protect the body from the toxicity of conventional electronic components and as well as to protect the components from the harsh environment inside the body which leads to accelerated device failure. Encapsulation materials such as titanium or ceramics have a long history of success in devices such as pace-makers and cochlear implants [1]. Such materials are impermeable to water or gasses and are very well tolerated by the body. The critical complication in most implantable devices is the need to cross the wall of the encapsulation with electrically conducting wires, commonly called a feedthrough. Very often wires are required to supply power and data to components sealed with in the hermetic capsule. In some cases, wires may also be required to carry electrical impulses to or from neural populations targeted by the device. Feedthroughs are understandably one of the most common failure points of hermetic capsules and therefore must be a carefully considered element in the design of any implantable device.

The feedthrough array discussed in this paper is designed to perform both as an electronics encapsulation and as a stimulation interface to the retinal tissue for an epi-retinal vision prosthesis. Recently there have been significant advances in the field of bionic vision [2–6] including the world's first commercially available implant [7]. Retinal prostheses consist of an array of stimulating electrodes that are surgically implanted against the surface of the

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retina. They are typically positioned on or close to the macular region of the retina, the area responsible for high acuity vision [8]. The devices are designed to treat diseases such as retinitis pigmentosa where the light detection cells within the retina are damaged but most of the inner middle retinal neurons such as ganglion and bipolar cells survive. These surviving cells can be electrically stimulated with an array of electrodes resulting in recipients perceiving an array of spots (or phosphenes) which can be used to form a crude image. Current state of the art epi-retinal devices tend to have around 60 electrodes positioned close to the retina providing the user with low resolution vision. An obvious potential precursor to achieving a higher resolution outcome for the patient is to employ higher numbers of electrodes positioned closer together. High numbers of stimulating electrodes require a large numbers of feedthroughs from the control electronics with the result that the reliability of an individual feedthrough must be extremely good. Very often arrays of stimulating electrodes are connected to the electronics capsule by a flexible leadwire. When hundreds or thousands of electrodes are called for, individual wiring of each electrode and routing wires to remote capsule becomes untenable, in particular in the eye where physical space is limited.

Fig. 1 shows an illustration of the approach currently employed by Bionic Vision Australia (BVA) to solve both the feedthrough reliability problem and the issue of connecting control electronics to a high number of stimulating electrodes. The illustration shows an array of diamond feedthroughs and electrodes. The feedthrough array (grey) is constructed from two types of diamond; a polycrystalline, electrically insulating diamond substrate containing many electrically conducting nitrogen doped ultra nano-crystalline diamond (N-UNCD) feedthroughs. On the flip side of the array (shown in Fig. 2(c)) the feedthroughs terminate in isolated N-UNCD pads which are employed as the stimulating electrodes of retinal prosthesis. We have previously shown that N-UNCD has appropriate electrochemical characteristics to act as a neural stimulation material [9,10]. The fact that the substrate and feedthroughs are made from the same material, i.e. diamond, minimizes the possibility of feedthrough failure through materials mismatch, resulting in increased reliability. Importantly for our application the mechanical strength and low density of diamond means the capsule can be made thinner and lighter than it could be if made from a ceramic or a metal such as titanium. The approach also takes advantage of diamonds established biocompatibility [11–13] and superb biochemical stability [14] offering the prospect of a long lasting implant. Finally, the pitch and shape of the array shown in Fig. 1 has been designed to be directly flip chip bonded to a purpose built ASIC capable of delivering electrical stimulation through 256 independently controllable channels. Direct bonding of the stimulator to the array negates the need for a high count lead to the electrodes and thus the technology is easily scalable to higher numbers of electrodes. The metal tracks on the inner side diamond array (Fig. 1, right and upper edge) connect to a small number of supply power leads, forward data leads, counter electrode lead, backward data lead and power decoupling capacitors. Following is a description of the method employed to fabricate the diamond arrays and results of hermeticity, electrical testing and cytotoxicity testing.

2. Materials and methods

2.1. Feedthrough fabrication process

Thermal management grade polycrystalline diamond wafers (TM100 grade, 10 mm × 10 mm × 0.25 mm, element six Ltd) were used as the feedthrough substrates. The diamond wafers were supplied with one smooth face (<1 nm RMS roughness) and one very rough face. The rough face was polished to <10 nm RMS surface roughness using a resin bonded wheel on a Coborn PL3 rotary polisher (Fig. 2(a)). Feedthrough arrays were fabricated within the polished diamond wafers according to the schematic shown in Fig. 2. 150 μm deep 80 μm square pits were milled into the polished PCD substrate in 150 μm pitch square arrays (Fig. 2(b)) with an Oxford Lasers Alpha series laser cutter fitted with a Nd:YAG laser operating at 532 nm. In a second step, 9 × 10 μm diameter feedthrough holes were cut into the bottom of each pit penetrating through the remaining 100 μm of substrate diamond (Fig. 2(c)).

After laser milling, graphitic cutting residues were removed from the substrate by boiling under reflux in a mixture of 10 mL H_2SO_4 (conc) containing 1 g of $NaNO_3$ for 60 min followed by 30 min of 50 W 3:1 Ar:O_2 plasma (Diener FEMTO LF). N-UNCD was grown within the feedthrough holes and over the array face by microwave chemical vapour deposition (CVD) (iPlas, Cyrannus 1 Plasma source, Fig. 2 (d)) using a gas mixture of 20% N_2, 79% Ar and 1% CH_4. CVD parameters were 1250 W microwave power, 100 Torr chamber pressure and a stage temperature of 800 °C. Full details of N-UNCD synthesis have been previously described [9]. A 100 nm thick, gold layer was deposited onto the N-UNCD surface with a Thermionics VE-180 e-beam Coating System (Fig. 2(e)) to protect the electrodes during subsequent plasma cleaning. Individual feedthroughs were isolated from one another by laser milling through the gold and N-UNCD films and slightly into the PCD substrate (Fig. 2(h)) followed by 4–8 h of 50 W 3:1 Ar:O_2 plasma to remove the electrically conducting, graphitic, laser cutting residues from between N-UNCD electrodes. To make

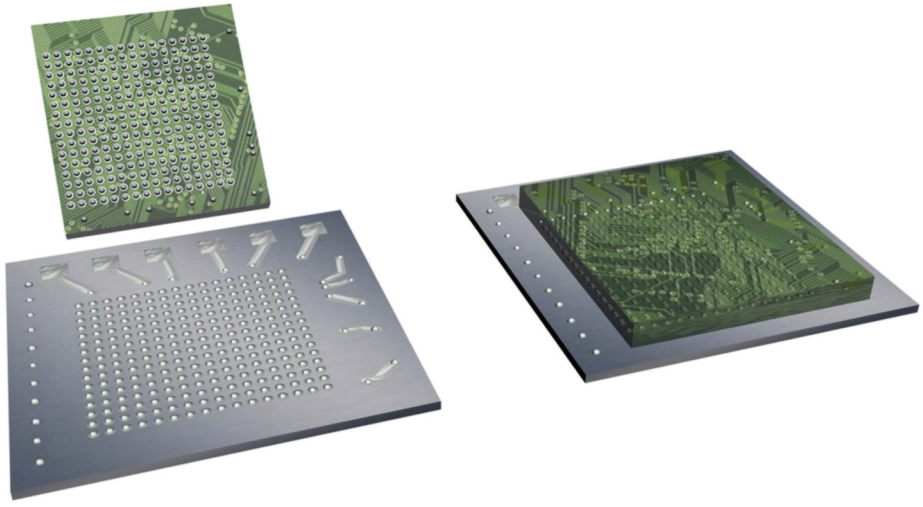

Fig. 1. Illustration of the Bionic Vision Australia diamond feedthrough and electrode array (grey) with 256 hermetic feedthroughs leading to N-UNCD stimulating electrodes. The array measures 4 × 4 mm square and 0.25 mm in thickness. The diamond feedthrough array and stimulator chip are shown separately on the left and again after flip chip bonding of the two components on the right.