

Fabrication of platinum spherical electrodes in an intra-ocular prosthesis using high-energy electrical discharge

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Abstract

A 100-channel neurostimulation circuit comprising a complementary metal oxide semiconductor (CMOS), application specific integrated circuit (ASIC) and a 10×10 electrode array for interfacing with surviving neurons of the retina has been designed. We discuss the fabrication of the accompanying electrode array that is to be connected to the implant. Specifically, a technique is described that allows the consistent and accurate formation of platinum spheres.

The technique uses a controlled discharge across an ionised air gap. The magnitude of the discharge energy (controlled by pre-selecting different discharge capacitors), allows for the control of sphere diameter.

One hundred of the said spheres with attached $100 \mu\text{m}$ diameter platinum wires are then fabricated into an electrode array formed from a silicone base. We demonstrate that using our technique, it is possible to inexpensively fabricate uniform spheres for use in neuroprosthetic devices and that said spheres are suitable for both recording and stimulation of physiological functions.

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1. Introduction

The key to successful implementation of neuroprostheses lies with appropriate electrode–tissue interfacing. A number of novel and interesting techniques have been applied towards achieving this objective including pre-formed silicone elastomer and platinum electrodes for the cochlea that conform to the inner wall of the scala tympani [1], penetrating silicon electrodes for high density stimulation of the visual cortex [2], micro-machined sieve electrodes for interfacing with neurons in regenerated nerve fiber bundles [3,4], lithographically formed retinal interfaces [5], and TiNi shape memory fabrication techniques [6] for three-dimensional neural interfaces.

In addition to the ability to transfer charge across the electrode–tissue interface, a number of desirable or imperative features will contribute to the success or failure of a given electrode design. Of particular importance are the energy transfer limitations intrinsic to wireless communica-

tions for implantable electronics. This precludes the use of electrodes with particularly high electrical impedance, otherwise voltage compliance limitations reduce charge injection capabilities. Furthermore, despite the implementation of the most extraordinary fabrication techniques, the device will ultimately be judged on its ability to withstand the harsh and corrosive environment of the body and therefore must possess properties appropriate for such use.

Electrodes often are required to interface with three-dimensional structures such as the spherically shaped retina, spirally created cochlea, or cylindrically formed peripheral nerves. As such, despite the advantages of being lightweight, readily manufactured and apparent tolerance within the body, two-dimensional fabrication techniques such as lithographic machining may have limitations without supplementation by a third dimension as it is frequently the case that a conformal substrate is required. In the case of the electrode surface, limitations in material deposition, achievable thickness or plating techniques limit the versatility of the thin-film methods alone.

The present authors propose a hybrid approach wherein thin-film interconnects pass signals between remote stimulating electronics and the electrode–tissue interface where

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specifically formed electrodes of platinum or other suitable material, are placed within three-dimensional conformal substrates for intimate interfacing with neurons.

The World Health Organization (WHO) conservatively estimates that non-preventable blindness caused by the retinal dystrophy known as age related macular degeneration (AMD), affects more than 8 million people world-wide [7]. Retinitis pigmentosa (RP), most commonly inherited as an autosomal recessive disorder [8], affects a further 1.5 million people world-wide [9].

While these diseases often inflict devastating results upon those affected, an important attribute of both AMD and RP is their relative concentration on photoreceptors, leaving higher order neurons, notably the retinal ganglion cell layer and their axons that form the optic nerve, viable and capable of physiological excitation by way of electrical stimulation [10,11].

It has therefore been proposed, and indeed active pursuit has been initiated by several research teams [12] including the present authors [13], to develop a neuroprosthesis that interfaces with retinal neurons in an effort to restore useful vision to those blinded by the aforementioned disorders.

The present paper introduces and characterises a technique for fabricating spherical electrodes from bare wire comprised of electrode material. The approach has been taken as part of the development of a 100-channel complementary metal oxide semiconductor (CMOS), application specific integrated circuit (ASIC) based neurostimulation system that was designed, constructed and tested by the authors [13] and is presently undergoing animal studies.

The system comprises an externally worn transmitter and a body implantable stimulator. The purpose of the system is to communicate both data and power across tissue via radio frequency (RF) telemetry such that externally programmable, constant current, charge balanced, biphasic stimuli may be delivered to neural tissue at 100 unique sites.

The ASIC has been designed primarily as a treatment of degenerative disorders of the retina whereby the 100 channels are to be utilised in the delivery of a pattern of stimuli of varying intensity and or duty cycle to the neural tissue of the retina known to survive after the onset of disease [14]. However, it is conceivable that other fields of neurostimulation such as cochlear prosthetics and functional electronic stimulation may benefit from the employment of the system.

In the case of the retinal prosthesis, the purpose of the device is as a therapeutic treatment to those degenerative disorders as previously described. The device is to be implanted within the ocular anatomy [15] or in the retrobulbar space and interface with the surviving retinal neurons. Fig. 1 illustrates the proposed in situ placement of the prosthesis within the eye.

To address the difficult issues associated with the design of appropriate electrode–tissue interface technology for a visual prosthesis, we created an electrode array comprised of materials well known to be tolerated by the body and survive in this corrosive environment. The substrate for the array

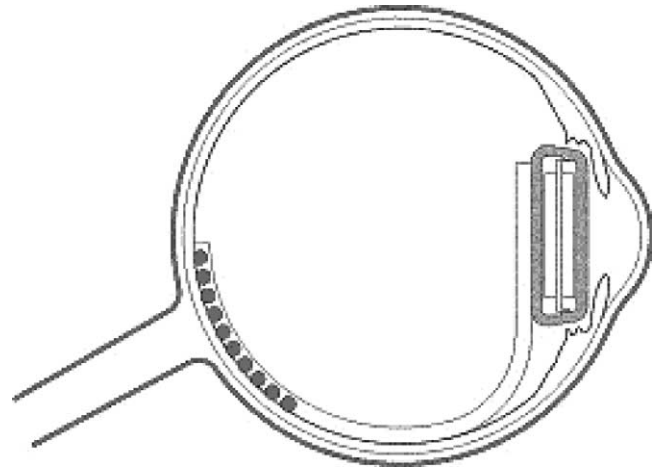


Fig. 1. In situ placement of implant and electrode array within the ocular anatomy. Proportions of this figure are in reference to the anatomy of the *O. aries* (sheep). The ASIC is fixed at the anterior of the eye with the electrode array closely approximating the retinal ganglion cell layer at the posterior of the globe.

is a silicone elastomer and the 100 electrodes are formed from 100 μm diameter platinum wires. While the silicone substrate shown herein is flat, moulding the elastomer into a wide-range of three-dimensional shapes is readily achieved.

Owing to charge-density considerations, it is not practical to directly connect the end of the platinum wire to the tissue. Furthermore, sharp or jagged edges may damage the fine retinal structures. In order to form an electrode, with an electrode–tissue interface of sufficiently low impedance ($<5\text{ k}\Omega$ at several kHz), typically an electrode of greater than 400 μm diameter is required. While this size is recognisably large relative to the neurons to be stimulated, there is mounting evidence that while physiological excitation of individual or small groups of retinal neurons is possible, psychophysical phosphene perception may only be possible with large electrodes. Greenberg has suggested that (at least for the electrode configuration used) the electrode separation distance (and hence the probable degree of discernment between sites) is likely to be not less than 0.25 mm, the distance at which a 3 dB rise in stimulation threshold is observed in the frog [16]. Grumet, however, has shown that (physiological) discernment between electrodes exists with an electrode pitch of 25 μm [17]. The differences in resolution between physiological stimulation and psychophysical perception appear to be in stark contrast to one another. Humayan and his co-workers reported that blind human patients could discern 435 μm electrode separation distances, a level of acuity corresponding to crude ambulatory vision [11]. A further study has indicated that patterned stimulation could be observed with electrode separations of similar magnitude [18].

Practical implementation of a retinal neurostimulator may indeed require that the electrodes are large in order that perception of distinct stimulation sites is achieved and charge injection limits are not exceeded. While this does constrain

the electrode quantities and thus the vision that can be achieved with such an approach, the observations of Rizzo and his colleagues offer compelling evidence that such large electrodes may facilitate useful perception of patterned vision by way of the centre-surround inhibition mechanisms that survive in the diseased retina [19].

In order to create electrodes with consistent charge injection characteristics it is important to fabricate the electrode tips into uniformly sized spheres. To our knowledge, there is no direct literature on methods for so doing.

In this paper we describe a technique for forming spheres at the tip of platinum wires in the context of using this technique in the design of an implantable intra-ocular neurostimulator. Further, we briefly demonstrate the application of the spheres in both stimulation and recording of physiological function within the vision system of the sheep.

2. Methods

The formation of the individual electrodes (platinum illustrated herein) that comprise the electrode array were fabricated using high-energy electrical discharge such that the shape and size of each was predictable and uniform.

Electrodes were formed using the adjustable energy discharge circuit as shown in Fig. 2. The circuit consists of two parts, a high-tension circuit that serves to ionise the air gap and a low-tension, high-energy circuit that discharges across the ionised air gap. The two points that form the air gap consist of a platinum wire from which the electrode is to be formed, and a tungsten carbide “return” electrode.

The high-energy circuit is comprised of a capacitor bank that allows for a selection of energy levels to be applied to the discharge thus enabling adjustment of the sphere size. The high-tension circuit is comprised of a high voltage transformer (10 kV) with the primary coil energised to 12 VDC by way of a switch. Upon release of the switch, the primary coil’s inductance generates a voltage to counteract the instantaneous change in current through it. As a result, the high-tension secondary coil discharges and ionises the air gap between the discharge point and ground. This ionisation temporarily completes the circuit to ground for the capacitor bank, thus causing the energy stored within the capacitor bank to discharge across the air gap.

With Pt wire connected to the discharge point, and an air gap established to ground, the energy discharge of the capacitors results in the liquefaction of the exposed tip of the Pt wire. Surface tension of the molten metal facilitates the formation of a spherical shape upon solidification. The size of the sphere was established by maintaining a uniform distance between discharge electrode and the platinum wire in addition to selecting the capacitance to be discharged upon ionisation of the air gap.

Different discharge energies in the range from 0.05 to 3.45 J were selected by switching in different capacitance values. For each discharge energy, five separate spheres were created and their diameters measured using a Vernier caliper (accurate to $\pm 6.4 \mu\text{m}$ ($0.25 \mu\text{in.}$)). Results are expressed as mean ± 1 standard deviation.

To demonstrate the electrodes in a physiological application, trans-retinal electrical stimulation of the vision system of the *Ovis aries* species (common name: sheep) is applied

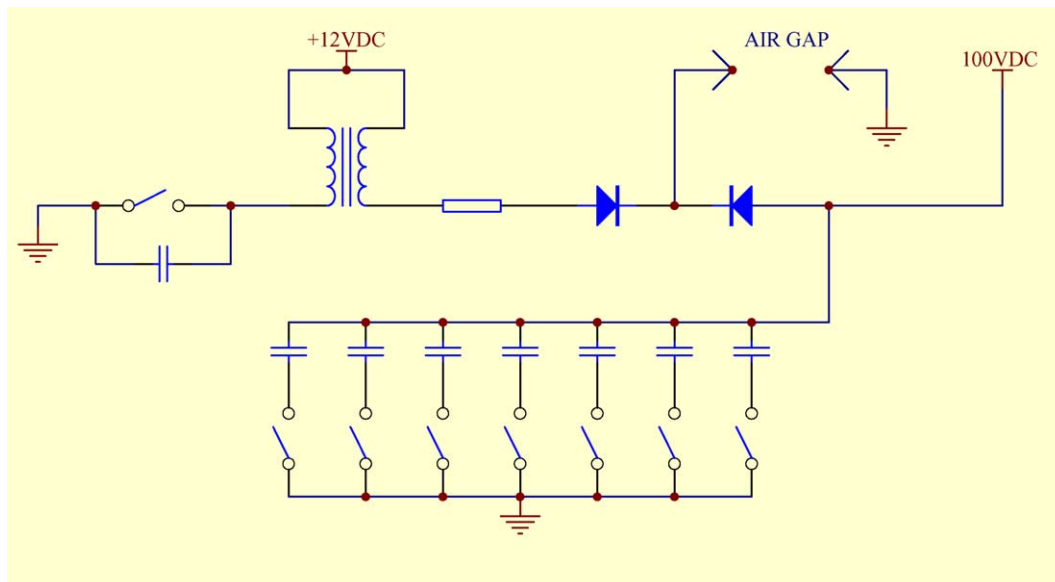


Fig. 2. Schematic diagram of a high-energy discharge circuit used to ionise an air gap separating the end of a $100 \mu\text{m}$ Pt wire from a tungsten carbide electrode. Upon ionisation, a pre-selected capacitance was discharged across the ionised gap, melting and forming the sphere on the end of the Pt wire. Note that the circuit used in obtaining the results presented herein contains potentially dangerous levels of voltage and energy that are capable of eventuating serious injury or death. Extreme caution when using this circuit is imperative for the safety of the user.

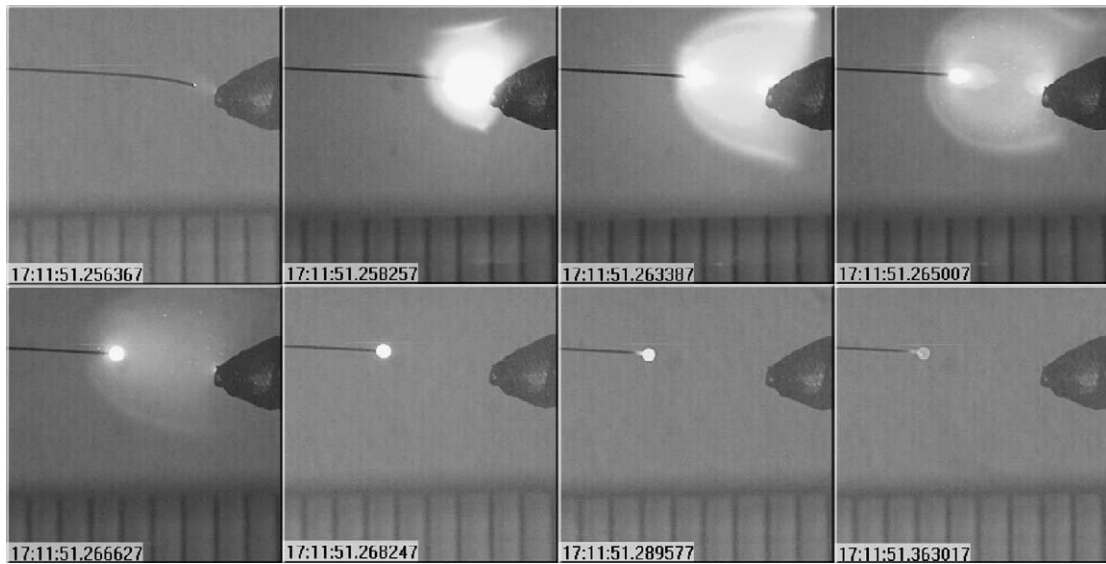


Fig. 3. Time lapse photographs of $450\ \mu\text{m}$ Pt sphere formation. Graticules are 1 mm apart and the time frame is given at the bottom left of each frame in units of seconds.

using the spherical electrodes, and the cortical response acquired using recording electrodes fabricated using the same procedure. The experimental protocol is described in detail within the literature [20]. Briefly, the anaesthetised animal is fitted with two recording electrodes, $450\ \mu\text{m}$ diameter Pt spheres, placed sub-durally upon the visual cortex with a grounding electrode (bone screw) located at the lambda formed by the intersection of the lambdoid and sagittal sutures. The positive recording electrode was placed 20 mm nasal and 10 mm lateral to the lambda. The negative electrode was placed 18 mm dorsal and 10 mm lateral to the lambda. Stimulus was derived from the ASIC described by the authors previously [7] and delivered to a $450\ \mu\text{m}$ spherical electrode placed in the retrobulbar space with reference to an identical electrode placed within the vitreous body. Identical stimuli (2.0 mA, 2.0 ms duration, balanced bipha-

sic) were delivered multiple times ($N = 200$) with the synchronised response ensemble averaged.

3. Results and discussion

Fig. 3 depicts the time lapse image series of the formation of the electrode spheres. The surface tension of the molten metal ensures a well-formed spherical shape. The surface is smooth and free of defects. It is envisaged that subsequent treatment may be appropriate so as to increase the surface roughness of these spheres thereby increasing their charge-carrying capacity.

Fig. 4 shows the characterisation of discharge energy versus sphere diameter for the following conditions: 100 V capacitor voltage; 1.25 mm air gap; $100\ \mu\text{m}$ Pt (99.99%) wire.

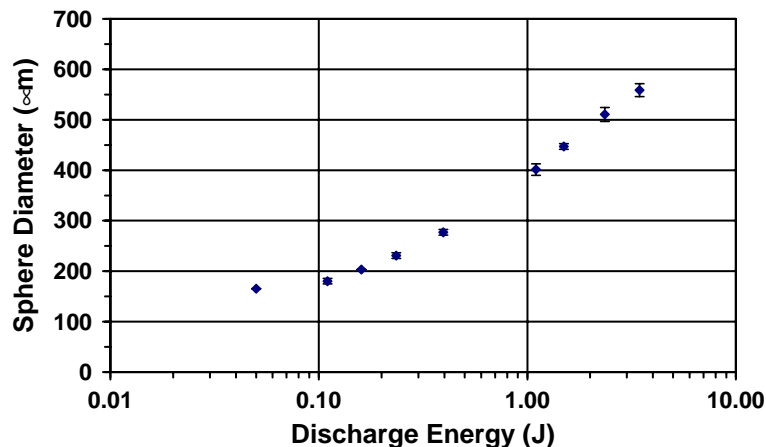


Fig. 4. Discharge energy versus resulting sphere diameter; 1.25 mm air gap, $100\ \mu\text{m}$ Pt (99.99% pure) wire diameter, 100 V capacitor voltage. $N = 5$ spheres per energy level.

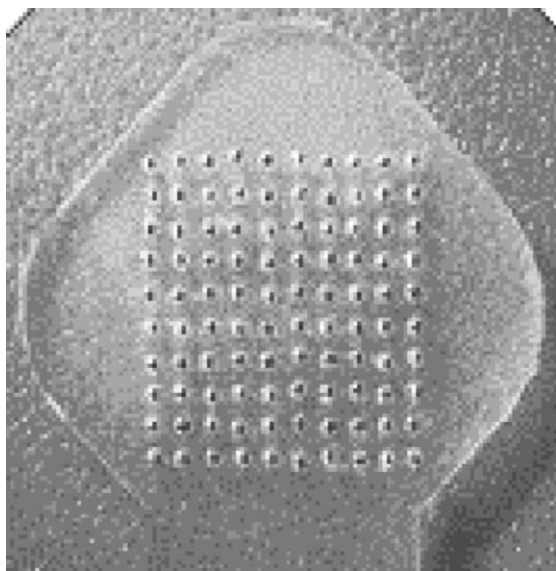


Fig. 5. Populated 10×10 electrode array for connection with the encapsulated ASIC. The platinum electrode spheres closely approximate the retinal ganglion cell layer. Platinum sphere diameters are $450 \mu\text{m}$.

Careful maintenance of the discharge parameters ensured an accurate and reproducible sphere diameter as is indicated by the error bars. Fig. 5 shows the populated electrode array containing 100 platinum spheres of $450 \mu\text{m}$ diameter within a cast polydimethylsiloxane (silicone elastomer) substrate. Fig. 6 shows the X-ray of the completed prosthesis with a reduced electrode count of four platinum spheres for use in animal testing.

The span of the air gap as well as the condition of the tungsten carbide return electrode are critical to the consistency

of the electrode sphere diameters. Variation in either, leads to diminished discharge of the high-energy capacitor(s) thus reducing the period during which the electrodes are formed in the Pt wire, resulting in a smaller and less predictable size. Consistency of the tungsten carbide return electrode requires monitoring of the tip, which erodes as a course of use, and the removal of soot resulting from impurities upon the Pt wire surface such as oil.

The discharge circuit itself must be fabricated from rugged and appropriate components. In particular, the diode that serves to prevent discharge of the high-tension circuit into the capacitors must be capable of passing substantial current over a short period of time while maintaining a zener voltage of several tens of kilovolts. The use of several diodes in series for this purpose is recommended. The diode that serves to isolate the 12 VDC supply from the 100 VDC supply may have a breakdown voltage substantially lower than that of its counterpart but as substantial voltage must pass forward through this diode, the use of several diodes in parallel is recommended.

The resistor shown in Fig. 2 serves to prolong the discharge of the high-tension circuit, thus allowing for a more thorough discharge of the high-energy capacitor bank. This resistor, and all switches within the circuit must have substantial capacity to withstand high voltages and high current impulses.

Fig. 7 illustrates the ability of the electrodes to evoke and record a response resulting from stimuli applied to the vision system of a high-order mammal. The figure shows an ensemble average of electrically induced cortical potentials from 200 stimulus events. The high frequency spike at $t = 0$ is the stimulus artefact. This is followed by the cortical response, most likely from a substantial proportion of the

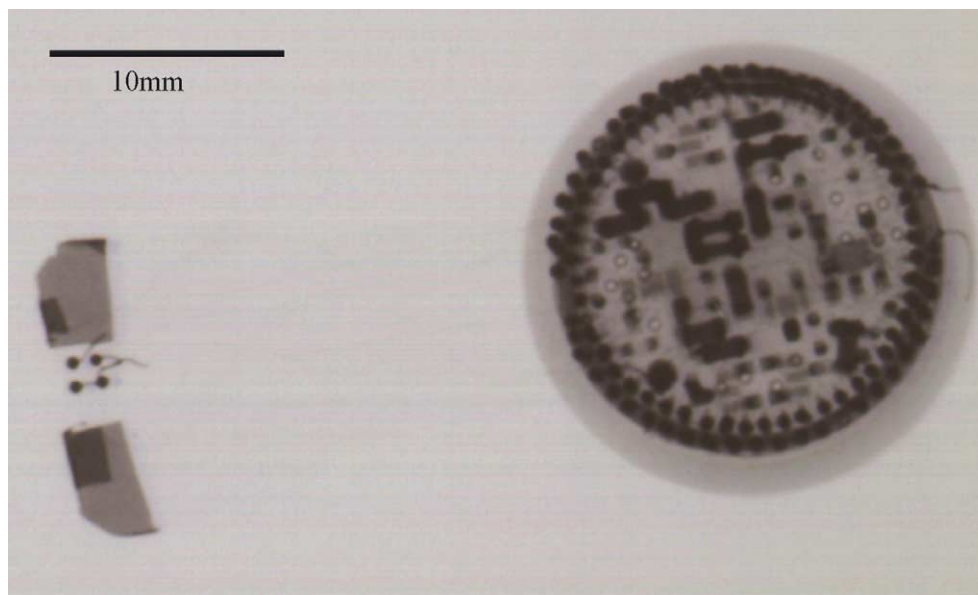


Fig. 6. X-ray image of functional implant shown after explanation from an animal. At left of image, four, $450 \mu\text{m}$ diameter spherical electrodes are shown adjacent to two plate electrodes comprised of Pt foil. The $25 \mu\text{m}$ wires connecting the electrodes to the stimulator body (right of image) are too fine to be visible in this image.

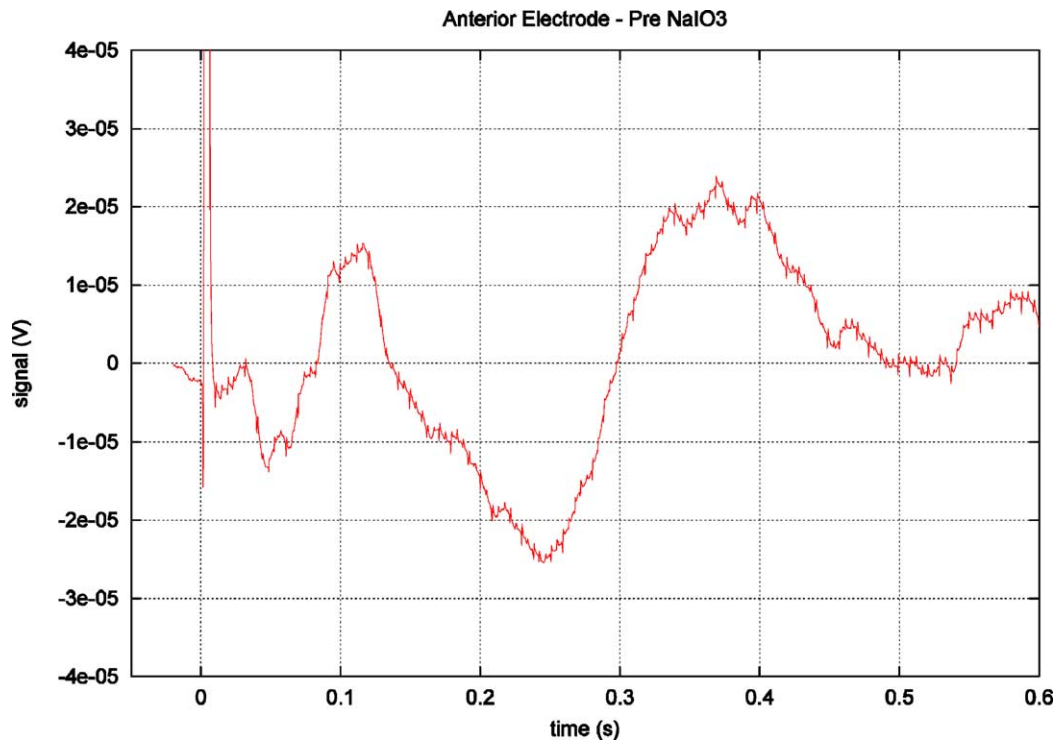


Fig. 7. Electrically induced cortical response to 2.0 mA stimuli, 2.0 ms per phase (balanced biphasic pulse). Ensemble averaged from 200 stimulation events with synchronised recording from the visual cortex of the sheep.

retinal neurons owing to the similarity of this response to light-evoked responses.

4. Conclusions

The fabrication of uniform and consistently sized platinum spheres can be readily achieved using the technique described. The electrodes are suitable for use in physiological applications involving both recording of biological signals and in the delivery of stimuli to excitable tissue.

In their application within a visual prosthesis, the topic of ongoing research by the authors, several of these electrodes will be formed in a patterned array and interfaced to lithographically fabricated thin-film interconnections leading to the stimulating electronics. Electroplating of platinum in thick (100 μm order) pads at the interface of each electrode and interconnect track will facilitate low impedance electro-mechanical bonding. Subsequent coating with a silicone or parylene insulating layer shall provide robust, independent electrical pathways between each stimulator channel and its associated electrode.

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Biographies

G.J. Suaning was born in San Francisco area of California. He received his BSc in 1986 and MSc in 1988. Both degrees are in Mechanical

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In 2003, Dr. Suaning completed his PhD from the Graduate School of Biomedical Engineering, University of New South Wales, Sydney, Australia. His field of study is in prosthesis design and the electrophysiology associated with retinal neuroprostheses. Dr. Suaning holds an appointment at the University of Newcastle, Australia in the Faculty of Engineering.

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Dr. Lovell's research work has covered areas of expertise ranging from cardiac modelling, home telecare technologies, biological signal processing and visual prosthesis design, having written over 100 refereed journals, conference proceedings and book chapters.

He is currently in the IEEE Engineering in Medicine and Biology Society (EMBS), Vice President for Member and Student Activities (2002–2003). Recently he was awarded the IEEE Millennium Medal for services to the EMBS in relation to its Web-based infrastructure. He is also the Conference Co-chair for the World Congress in Medical Physics and Biomedical Engineering to be held in Sydney in 2003.

C.Y. Kwok received the BSc, BE and PhD degrees in Electrical Engineering from UNSW, Sydney, Australia in 1971, 1973, and 1979, respectively.

In 1978 he joined the Engineering Staff of AWA Microelectronics. In 1980, he was a project scientist in Electrical Engineering at UNSW. From 1982 to 1988 he was a senior lecturer at the National University of Singapore. In 1988, he joined UNSW as a senior lecturer and he is currently an Associate Professor. His current research interests include semiconductor devices, thin insulating films, integrated circuit design and silicon micro-machining.