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Design Of Recording and Stimulation Neural Electrodes in The Brain

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Abstract:

Emerging prostheses and treatments for spinal cord injury, stroke, sensory deficits, and neurological disorders are based on electrical stimulation of nerve tissue and recording of neural electrical activity. Understanding the electrochemical mechanisms underlying the behaviour of neural stimulation and recording electrodes is critical for the development of chronically implanted devices, especially those that use a large number of microelectrodes. Materials that support charge injection via capacitive and faradaic mechanisms are available for stimulation. These include titanium nitride, platinum, and iridium oxide, each with its own set of benefits and drawbacks. We describe and criticise the use of charge-balanced waveforms and maximum electrochemical potential excursions as criteria for reversible charge injection with these electrode materials. Techniques for characterising electrochemical properties relevant to stimulation and recording are described, along with examples of differences in electrode response in vitro and in vivo.

***Keywords* — Neural Prostheses, Charge-Injection, Microelectrodes, Electrochemistry, Electrode Characterization, Safe Stimulation.**

INTRODUCTION

Current and emerging neural prostheses and therapies based on nerve stimulation and recording use electrodes that are permanently connected to the central and peripheral nervous systems. Upper and lower limb prostheses for spinal cord injury and stroke; bladder prostheses; cochlear and brain-stem auditory prostheses; retinal and cortical visual prostheses; cortical recording for cognitive control of assistive devices; vagus nerve stimulation for epilepsy and depression; and deep brain stimulation (DBS) for essential tremor, Parkinson's disease, epilepsy, dystonia, and depression are just some of the applications. All require low impedance electrodes for recording or safe reversible charge injection electrodes for stimulation.

ELECTRICAL REQUIREMENTS:

STIMULATION ELECTRODES

Electrical stimulation causes a functional response by depolarizing excitable cell membranes. The flow of ionic current between two or more electrodes, at least one of which is close to the target tissue, causes depolarization. Electrical stimulation is typically delivered as a series of biphasic current pulses in most neural applications. Figure 1 depicts typical pulse waveforms with pulse parameters. [1]

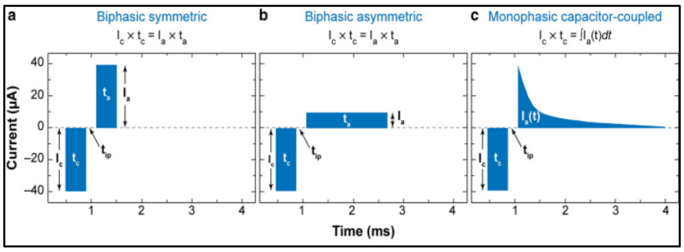

Figure 1: Charge-balanced current waveforms commonly used in neural stimulation. The parameters vary greatly depending on the application and electrode size. Ic (cathodic current), 1 μA−10 mA; Ia (anodic current), 1 μA−10 mA; tc (cathodic half-phase period), 50 μs – 4 ms; tip (interphase dwell), 0 - 1 ms; and ta (anodic half-

phase period), 50 µs - 10 ms are typical waveform parameters.

Each pulse has cathodal and anodal phases, as well as current amplitudes and durations that result in a net charge of zero for the pulse (charge-balance). A cathodal current is decreasing at the stimulation electrode, with electron flow from the electrode to the tissue. The term anodal refers to an oxidising current with electron flow in the opposite direction. For a cathodal constant-current pulse of magnitude ic and pulse width tc, the charge delivered is simply the time integral of the current, which is simply ictc. A capacitor discharge circuit can sometimes achieve charge-balance with intramuscular electrodes and electrodes that interface with the peripheral nervous system, resulting in a monophasic, capacitor-coupled waveform, as shown in Figure 1. [1]

Neural Recording:

In a closed-loop neural system, neural recording is required to sample local field potentials in addition to the stimulator that triggers the AP (LFPs). If the stimulator is regarded as the executor, the neural recording component is the system's digital back end. LFPs, as opposed to stimulation signals, are electric potentials in the extracellular medium surrounding neurons that have a very small amplitude (µV) and a low frequency (1–200 Hz). Because of the nerve signal's microvolt level, direct digital quantization before amplifying is not reliable. [2]

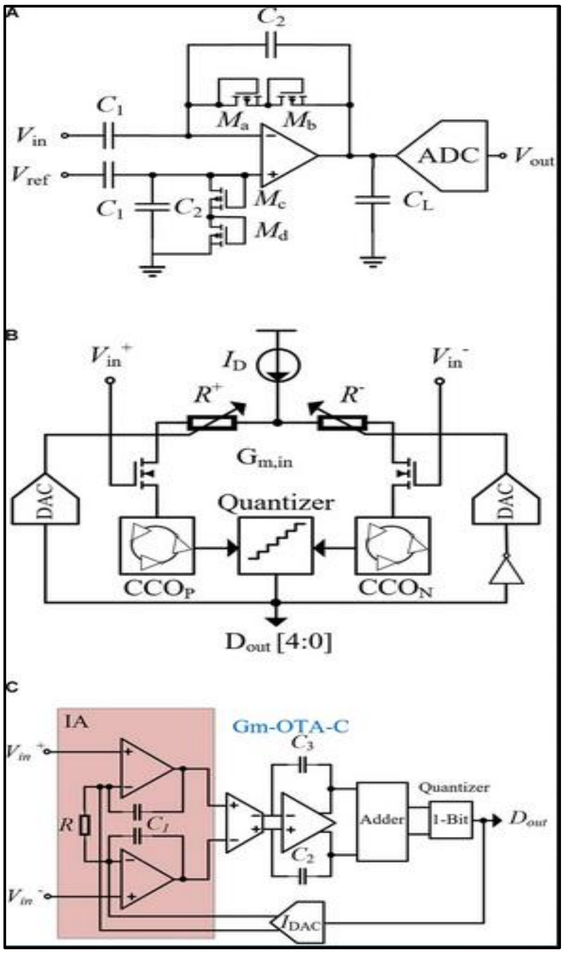

Figure 2: Artifact elimination circuit with blanking technique (A), active electrode discharge technique (B), iterative hardware loops (C), adaptive filtering technique in post-processing (D), chopper technique (E), track-and-zoom (TAZ) neural ADC, localised stimulation technique (G), dual electrode in-phase stimulation (H), and RTPPS technique (I).

OBJECTIVES:

1. To study of Design of recording and stimulation neural electrodes in the brain.
2. To study of properties of electrodes.
3. Implementation of neural recording technique.

REVIEW OF LITERATURE:

The present article reviews the electrochemical properties required for electrodes used in stimulation and recording of neural tissue with an emphasis on microelectrodes and their electrochemical characterization. Differences in the in vitro and in vivo properties of electrodes are discussed, and possible methods for the in vivo evaluation of electrodes are described. The review emphasizes those electrode materials that are presently employed in stimulation and recording , and identifies differences in these electrode materials based on their intrinsic electrochemical properties.

The neural prosthesis chip for biomedical use includes the neural/muscular stimulators and neural recording circuits. In these circuits, the stimulator has been widely used in biomedical applications for decades, such as cardiac pacemaking, cochlear/retinal prosthesis, and cell activation (Chen et al., 2010; Sooksood et al., 2011; Noorsal et al., 2012) [3-5]

The neural recording circuit is also involved in these applications to sense the neural signal or assess stimulation efficacy and the tissue status to enable closed-loop control in simultaneous neural recording and stimulation (Yoshida and Horch, 1996; Blum et al., 2007; Rolston et al., 2009, 2010; Venkatraman et al., 2009; Xu et al., 2012) [6-9]

Considering the safety of neural stimulation, the designed stimulator requires minimum residual charge in a single cycle, and the accumulated charge after multiple cycles also needs to be removed in time. The real-time monitoring of VE is necessary to eliminate the residual charge in time when the voltage does not return to the reference voltage at the end of the stimulation cycle (Ortmanns, 2007). A variety of the accumulated charge balance methods are introduced, such as the electrode short-circuit technology (Rothermel et al., 2009) [10-11]

RESEARCH METHODOLOGY:

Recent advancements in nerve prosthesis chip research necessitate high-quality data transmission from multiple neural electrodes. Wireless transmission is required when the data throughput in multiple-channel neural recording is high because it reduces the number of connected wires and simplifies the interface. The high transmission rate ensures stable multi-electrode recording, and the low output power has better anti-interference ability. A neural recording module and a wireless data transmission module are also implanted in the brain and abdomen, respectively. The modules are linked by a flexible coaxial subcutaneous cable capable of transmitting high data rate signals.

Books, educational and development journals, government papers, and print and online reference resources were just a few of the secondary sources we used to learn about the composition, use, and impacts of Design of neural electrodes for recording and stimulation in the brain.

RESULT AND DISCUSSION:

The important parameters in the design of neural stimulators are safety and efficiency. The injection and recovery of electric charge is the essence of stimulation. Excessive electric charge injection will cause irreversible damage to fragile nerves. As a result, we must limit the stimulation current and lessen the influence of residual charge in the tissue.

The CV response of electrodes in vitro and in vivo is also instructive. Figure 3 compares the CVs of an AIROF electrode in PBS, model-ISF, and subretinally in rabbit. The shift in the peak potentials of the Ir3+/Ir4+ redox wave (indicated by the arrows) from that obtained in highly buffered PBS is less pronounced in vivo than in the model-ISF. This implies that the in vivo buffering capacity is somewhere in the middle between model-ISF and PBS, and that the Ir3+/Ir4+ redox reaction is easier in vivo than in model-ISF.[12]

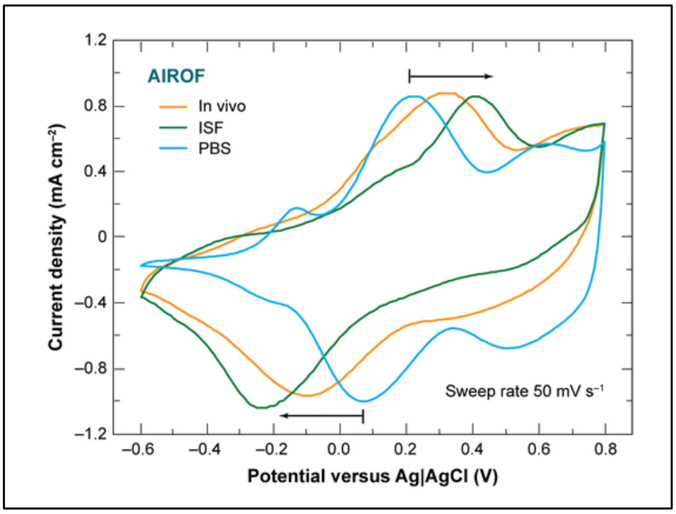

Figure 3: Comparison of the CV response of an AIROF electrode in PBS, model-ISF, and subretinally in rabbit.

Because this article introduces several advances in neural recording and stimulation integrated circuits, it is worthwhile to present a discussion about key design indicators that will assist the circuit designer in improving chip performance.

Accuracy is a critical parameter in neural recording design. The difficulty of sampling is greatly increased due to the small amplitude and low frequency characteristics of neural signals. Furthermore, the impact of stimulation artefacts, neural signal attenuation, and electrical signal crosstalk must be investigated further.

Table 1. Comparison of the parameters of wireless power supplies.

Power frequency (Hz)	**1.85M**	**13.56M**	**346.6M**	**13.56 M**	**1.5M**	**13.56M**	**2M**
Distance (cm)	0.5	20	18	18	15	18	-
Coupling	Ultrasound	4 coils	3/4 coils	4 coils	2 coils	3/4 coils	2 coils
Recording	N/A	N/A	EEG	Spike	EEG	Spike	EMG
Stimulation	CCS	CCS	N/A	CCS	CCS	N/A	CCS
Uplink data	LSK	BLE	2.4 GHz RF	OOK/ BLE	UWB/ FSK	FSK	LSK/ WiFi
Downlink data	OOK	BLE	N/A	BLE	ASK	N/A	DPSK/ WiFi
Area (mm^2)	3.1x 1.9 x 0.89	20 x 22 x 11	14 x 25 x 14	19 x 19 x 30	20 x 20 x 6.9	25 x 35 x 8	4.4 x 5.7
Power (mW)	0.15	43	6.4 – 13	35	6.9	51.4	-
Experiment area (cm^3)	N/A	20 x 46 x 20	61 x 61 x 30	24 x 46 x 20	26 x 45	30 x 28 x 18	0.5 x 0.5 x 0.5

A wireless power supply and communication system are required for the implanted neural chip. The parameters of the circuits with wireless power supplies are compared in Table 1. The power frequency is the frequency of the alternating current (AC) of the induction link, and external power supply is typically realised by using inductive coils. The transmission distance is proportional to the size of the inductive coil, and prior works have achieved transmission distances ranging from 15 to 20 cm. According to the comparison in Table 1, only a few designs incorporate both neural recording and stimulation functions in a single implanted neural chip. The data transmission mode between internal and external is related to the uplink and downlink data. The data transmission rate must be consistently improved as the number of channels increases. [13]

CONCLUSION:

The circuit structures and latest technologies of neural recording and stimulation circuits are summarised in this article. The key design directions of a closed-loop neural prosthesis chip are introduced, as well as advances in neural recording and stimulation integrated circuits. We discuss the critical parameters in the design process due to the differences between neural recording and neural stimulation. The various latest technologies mentioned in this article, as well as an analysis of the future trend, could assist designers in meeting their performance requirements in future biomedical device development.

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