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"Capacitive" Pulse Shapes for Platinum Cuff Electrodes

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Abstract— **Artificial electrical stimulation of the peripheral nervous system is an established therapy for several pathologies and motor impairments. Therapeutic outcomes can be improved with targeted patterns of neural activation, but the required signal amplitudes to achieve this response exceed the limits for safe stimulation. This can lead to electrode corrosion and tissue damage. In this paper, we present a novel approach to pulse shape design based on the properties of the electrode-electrolyte interface. We aim to improve electrode stability at higher voltages by exploiting the potential-independent mechanisms of charge injection. We identified signal parameters associated with capacitive current flow at the platinum interface and incorporated these features into the design of cathodal pulse shapes. A pulse shape comprising 4 high-frequency 'capacitive' harmonics demonstrated a 40-fold performance benefit compared to a conventional square pulse, but irreversible reactions could not be completely avoided during current flow. However, the enhanced electrode stability with the 'capacitive' pulse shapes suggests further optimization of pulse designs according to a surface 'stability function' might allow for safe stimulation with greater electrode voltages.**

I. INTRODUCTION

As the popularity and utility of implantable electrotherapies grows, so do the demands for smaller neural interfaces with improved portability and greater electrode densities for more targeted activation patterns. Research trends in waveform design have responded by focusing on maximizing the electrochemical limits for charge injection without structurally changing the stimulating surface. Efforts to achieve this aim with cuff electrodes include investigations with novel materials, improved material coatings and nonrectangular pulse shapes [1], [2], [3]. These approaches strive to maximize the output of a material for given area. The introduction of selective stimulation capacities into an implantable cuff platform poses a further challenge for the design of biocompatible electrical waveforms. The signal amplitudes required for blocking or selecting different electrophysiological responses exceed the charge limits for safe stimulation [4]. As a result, waveforms can either reduce the charge per phase [5], [6], [3] or manipulate the pulse shape to recruit processes that "protect" the electrode at higher charge densities [7].

This work aims to explore the electroprotective properties of pulse shapes suitable for nerve stimulation with cuff electrodes. This paper presents the derivation of pulse shapes based on the surface characterization of the platinum black electrodes. Signal parameters (e.g., amplitudes and frequencies) associated with reversible, capacitive conditions are identified and combined to construct novel pulse shapes. These pulses were configured with a charge density at the limit for safe stimulation and continuously applied to bipolar platinum cuffs to test the effect of pulse shape on surface stability. Electrical waveforms designed according to the intrinsic mechanisms of charge injection could offer improved electrode stability for therapeutic applications requiring higher signal amplitudes.

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II. MATERIALS AND METHODS

A. Electrodes

1) Thin-film cuff with platinum black: Symmetric tripolar cuff electrodes, with 2.5 mm inner diameter and 10 mm length, were used in this experiment. The cuffs were fabricated from two 5 μm layer of polyimide with a 300 nm layer of platinum (IMTEK, Germany). The contacts had platinum black deposited on their surface. The geometric area of each anode was 6.6 mm^2 and the total cathodal area was 1.8 mm^2. Thin-film polyimide cuffs offer a robust and stable interface for nerve stimulation experiments [8]. Surface characterization of these contacts was the starting point in our investigation, as these would be the appropriate for later investigations with tissue samples.

2) Platinum silicone cuffs: The destructive nature of corrosion testing required the fabrication of handmade cuff electrodes with similar dimensions as the thin-film cuffs. Bipolar cuff electrodes were created using platinum foil strips 1.0 x 6.5 mm, 0.025 mm thick (Goodfellow Cambridge Limited), insulated copper coil wire (0.25 mm in diameter) and silicone rubber compound (RS Components 692-542). Rings were formed from the platinum strips each attached to a wire with silver glue; the whole complex was then coated with silicone. The spacing between the platinum rings was 3 mm for the electrodes, each ring with an effective stimulating surface area of approximately 5 mm^2 ($\pm$ 0.25 mm^2).

B. Electrochemical Impedance Spectroscopy

Multi-amplitude impedance spectroscopy of the platinum black electrodes described their surface response under a variety of conditions. Cuffs were connected in the two-terminal, or psuedo-bipolar, connectivity and the frequency was swept over a range of 20 Hz - 2 MHz at unbiased potential with Agilent E4980A LCR Meter in a bicarbonate-buffered solution (mM): 112.0 NaCl, 2.0 KCl, 1.8 $CaCl_2 \cdot 2H_2O$, 2.4 $NaHCO_3$. Spectra were collected at AC amplitudes (V_{0-peak})

between 5 mV and 2 V. Additional impedance testing of the customized platinum cuffs assessed their performance quality under small-signal conditions (V_{0-peak} = 5 mV). These recordings were normalized to the small-signal platinum black measurements, as the harmonics for pulse shape construction were identified from the platinum black surface response. This metric provides a quantification of the differences between the platinum and platinum black responses. In particular, we looked for differences between the conditions required for the onset of reversible, capacitive processes.

C. Pulse Shapes

Signal conditions (i.e., amplitudes and frequencies) which were identified as recruiting reversible processes during the surface characterization were linearly combined for the construction of novel pulse shapes. A square pulse served as the control signal (1).

$$I_{stim}(t) = \begin{cases} 1 & \text{if } t \leq 0.5 \text{ ms} \\ 0 & \text{if } t > 0.5 \text{ ms} \end{cases} \tag{1}$$

Test pulse shapes were built as a weighted linear combination of sine waves using one harmonic (2), two harmonics (3) and four harmonics (4). These frequency bases were selected from the range associated with reversible, capacitive processes identified during surface characterization (approx. 1 - 100 kHz). The current-mode amplitude of these pulses was scaled such that the charge per phase corresponded to a charge density of 50 μC/cm^2 (geometric area) [9] for a duration of 0.5 ms. A longer duration than in [9] was selected for its potential to evoke a nerve blocking response.

$$V_{elec}(t) = sin(10\text{k} \cdot t \cdot 2\pi) \tag{2}$$

$$V_{elec}(t) = \left| \sum_{n=1}^{2} c_n \cdot sin(f_n \cdot t \cdot 2\pi) \right| \tag{3}$$

$$c_n = [1, 0.1]$$

$$f_n = [10, 100] \text{ (kHz)}$$

$$V_{elec}(t) = 1 - \left| \sum_{n=1}^{4} c_n \cdot sin(f_n \cdot t \cdot 2\pi) \right| \tag{4}$$

$$c_n = [0.5, 1, 1, 1]$$

$$f_n = [10, 11, 20, 50] \text{ (kHz)}$$

D. Equivalent Impedance Model

The pulses containing capacitive harmonics were converted from voltage-mode to current-mode for use with a programmable source. Their V-to-I conversion employed a Simulink® model of the interfacial electrode impedance. The equivalent electrical model was fitted to the small-signal surface impedance of the platinum black cuff connected in a psuedo-bipolar configuration. Its component values were computed with an unweighted least-squares regression in Matlab®.

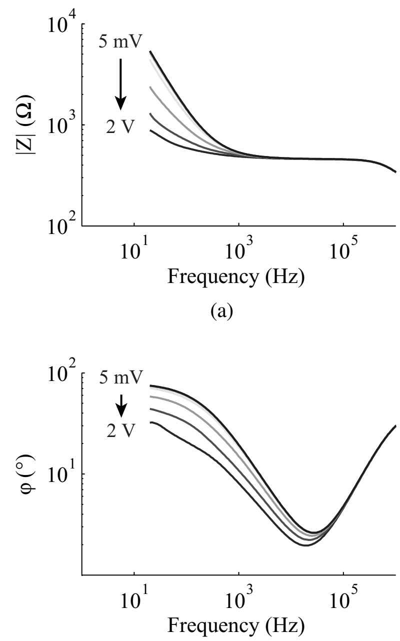

Fig. 1. Surface response of platinum black in a two-electrode configuration: (a) Impedance magnitude, and (b) phase for variable AC amplitudes (5 mV to 2 V).

E. Corrosion Test

The corrosion test involved the periodic monitoring of the electrode impedance following continuous stimulation. An ASCII file of each stimulus pulse was uploaded to custom control software and sent to an AC programmable current generator with a floating triax output (Keithley® 6221). Monophasic pulses were delivered at 1 kHz for 1-2 hours with a one minute pause for surface assessment at the specified intervals. The compliance voltage was adjusted over time to accommodate the changes in the electrode impedance. Bubbles lodged within the cuffs were released to ensure continual current flow throughout the testing period. The small-signal impedance spectrum was collected at intervals following periods of continuous stimulation using the apparatus and settings listed in Subsection II-B. These spectra were normalized to the initial conditions to describe its surface changes due to the stimulation protocol.

III. RESULTS

A. Surface Characterization

Multi-amplitude impedance spectroscopy of the platinum black cuffs elucidated the surface processes using a range of frequencies and AC perturbations. The impedance magnitude, $|Z|$, at the lower frequencies (20 Hz to 1 kHz) steadily declined with increasing frequency; higher potentials also had lower $|Z|$ values over this range. The phase (ϕ) decreased with increasing frequency between 20 Hz and 50 kHz, with higher amplitudes exhibiting lower phase values within this range. The potential-dependencies of both $|Z|$ and ϕ indicated the recruitment of nonlinear charge-transfer reactions. Higher frequencies (>5 kHz and >80 kHz

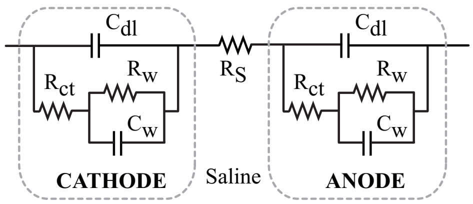

Fig. 2. Equivalent impedance model for V-I conversion of pulses.

TABLE I

FITTED VALUES FOR THE ELECTRODE-ELECTROLYTE INTERFACE MODEL TO THE SMALL-SIGNAL PLATINUM BLACK IMPEDANCE.

	R_s (Ω)	C_{dl} (nF)	R_{ct} (Ω)	C_W (μF)	R_W (kΩ)
Cathode	190.8	1.0 k	135.3	3.98	0.27
Anode		0.51	245.1	2.7	18.99

within the $|Z|$ and ϕ plots, respectively) showed potential-independent impedance behavior owing to the kinetic limits of the charge-transfer reactions. Within this parameter range, current flowed through the double-layer capacitance. Current injection via the surface capacitance allows for greater signal voltages without recruiting the chemical mechanisms; in effect, this circumvents the processes which can lead to corrosive changes in the electrode material.

B. Equivalent Impedance Model

Conversion from voltage-mode to current-mode required use of an equivalent electrical model of the interfacial impedance. The equivalent circuit model was based on Kovacs' description of the surface and arranged for a bipolar configuration [10]. Each electrode surface reflected four surface features: the double layer capacitance, the charge transfer resistance, the Warburg diffusion and the bulk resistance of the solution. This bipolar model with the fitted component values in Table I matched the experimental impedance under small-signal conditions for the platinum black (R^2 = 0.999).

C. Corrosion Test

1) Initial conditions: Performance validation of the custom platinum test electrodes demonstrated both their functionality and their differences relative to the platinum black material. The lower real surface area of the platinum contacts relative to the platinum black material per unit of geometric area suggested that the platinum contact would have lower relative impedance because of the decreased active area. This was confirmed for conditions of capacitive current flow (500 Hz to 2 MHz), where the interfacial reaction resistance was constant, in the case of the Test Cuffs 1, 2 and 3 (Figures 3(b), 3(e) and 3(h)). For frequencies <500 Hz, these cuffs exhibited variable impedance relative to the platinum black values because of the presence of chemical reactions. Cuff 4 demonstrated greater impedance magnitudes over the entire frequency spectrum (Figure 3(k)). All Test Cuffs displayed approximately constant relative impedance for frequencies

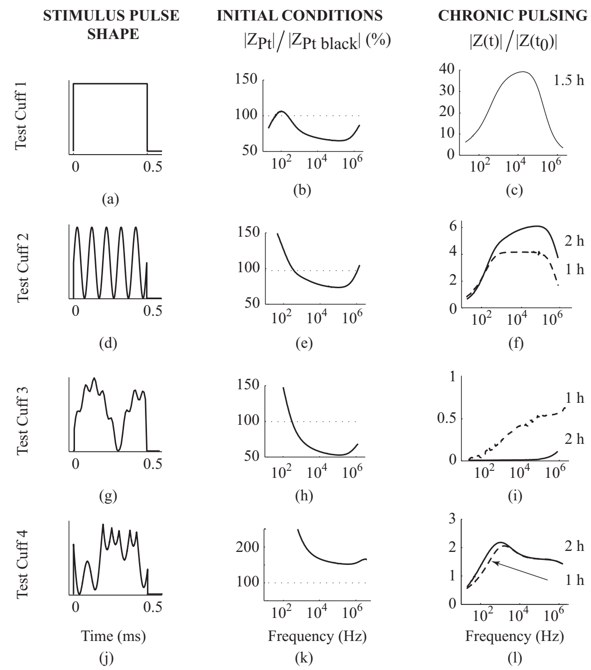

Fig. 3. Impedance changes during chronic pulsing with 'capacitive' pulse shapes. Initial conditions of the handmade platinum test electrodes ($|Z_{Pt}|$) are shown relative to the small-signal behaviour of a robust thin-film platinum black cuff ($|Z_{Pt\,black}|$), which is appropriate for later nerve experiments. Impedance changes following chronic pulsing with the custom platinum cuffs ($|Z(t)|$) are presented relative to their initial impedance small-signal behavior ($|Z(t_0)|$).

>10 kHz. This feature is consistent with the conditions identified for the potential-independent capacitive flow during the surface characterization of platinum black.

2) Chronic pulsing: Waveform comparison following 1.5-2 hours of pulsing revealed that the pulse shape containing four 'capacitive harmonics' had the lowest increase in impedance relative to its initial state (Figure 3(l)). Its relative impedance increased slightly between 100 Hz and 1 kHz with increasing stimulation time. This suggested a decrease in the interfacial capacitance, as current flow within this frequency band was primarily through faradaic mechanisms. The relative impedance between 10 kHz and 2 MHz remained steady, which indicates that the chemical mechanisms at the interface remained intact, though slightly greater than its initial state, following continuous pulsing. The normalized impedance for Cuff 3, which delivered pulse containing two 'capacitive harmonics', exhibited an impedance less than its initial conditions (Figure 3(i)). Two hours of pulsing displayed an impedance spectrum of nearly zero; this indicated a possible short between the contacts. The waveform pulse consisting of a 10 kHz sine wave, and delivered by Cuff 2, offered the second best performance in relative impedance changes (Figure 3(f)). However, the plateaus in the data between 50-300 kHz at a 4-fold increase after 1 hour and 2-200 kHz at a 6-fold increase after 2 hours of stimulation suggested changes to the faradaic processes. The capacitive properties of the surface between 20 Hz and 1 kHz remained stable during the stimulation protocol.

The square pulse delivered by Cuff 1 caused the greatest damage to the electrode surface (Figure 3(c)). Its impedance following 1.5 hours of stimulation showed an increase by a factor of 6.1 at 20 Hz to a maximum factor of 39 at 16 kHz before declining to a factor of 3.4 at 2 MHz.

IV. DISCUSSION

This paper examined whether enhanced electrode stability was possible with pulse shapes designed to minimize the recruitment of irreversible faradaic processes. We report that a pulse comprising 4 ‘capacitive’ harmonics offered the least damage to the stimulating surface. Continuous pulsing with a square pulse caused the most electrode deterioration. The electrode response to a pulse with an intermediate frequency composition (i.e., 1 ‘capacitive’ harmonic) was within the abovementioned extremes. The link between capacitive components and electrode performance was not fully elucidated due to a mechanical failure within an additional stimulating interface.

The motivation underlying the construction of the ‘capacitive’ pulse shapes was such that use of high frequency components would decrease the amount of time available for the progression of reaction kinetics. However, in perturbing the surface towards its previously published stability limit (e.g., 50 $\mu C/cm^2$, geometric area), the amplitude of the current-mode waveform was on the order of mA: 6 mA for the square pulse and 10-12 mA for the ‘capacitive’ pulses. The resulting electrode voltage had well exceeded the potential limits of hydrolysis. The presence of rapid bubbling during stimulus application demonstrated that irreversible reactions were being driven as a consequence of the surface potential. Thus, high-frequency content of the pulse shapes did not drive the potential low enough for the electrode to perceive each high-frequency period as an individual events, analogous to the sine waves during impedance spectroscopy. Rather, the higher amplitude to achieve the greater charge density effectively reduced the higher frequency content to a ripple-effect atop a transient dc bias during pulse application. This situation negated our attempt to exploit the kinetic-controlled surface behaviour at higher frequency because electrode voltage still remained at a level long enough for irreversible reactions to proceed.

The high-frequency rippling of the ‘capacitive’ pulse shapes did produce less damage to the electrode compared to a conventional square waveform. Further refinements to pulse shape design would involve amplifying the AC content relative to net DC effects within the duration of a cathodal pulse. The trade-off between these contributions within a cathodal pulse relate to a balance between injecting enough charge for nerve activation, which manifest as net DC effects, and injecting the charge in a manner which utilizes nonfaradaic mechanisms, which in this case, involves higher frequency harmonics. Optimization of pulse shapes according to a sort of ‘stability function’ derived from surface properties of the metal would allow for greater electrode voltages without structural damage and greatly aid advances towards implantable interfaces with selective stimulation capabilities.

V. CONCLUSIONS

Surface characterization of platinum black cuff electrodes revealed a frequency band of potential-independent impedance. We derived alternative cathodal pulse shapes based on the signal features associated with this surface behavior. These waveforms strived to minimize the potential-dependent chemical reactions and thus, to reduce electrode corrosion. A pulse shape comprising 4 high-frequency ‘capacitive’ harmonics demonstrated a 40-fold performance benefit compared to a conventional square pulse. Faradaic reactions were not completely eliminated from the charge injection mechanisms. However, the enhanced electrode stability with the ‘capacitive’ pulse shapes suggests further optimization of pulse designs according to a surface ‘stability function’ might allow for safe stimulation with greater electrode voltages.

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REFERENCES

[1] S. Cogan, “Neural stimulation and recording electrodes,” *Annu. Rev. Biomed. Eng.*, vol. 10, pp. 275–309, 2008.

[2] M. Sahin and Y. Tie, “Non-rectangular waveforms for neural stimulation with practical electrodes,” *J. Neural Eng.*, vol. 4, pp. 227–233, 2007.

[3] A. Wongsarnpigoon, J. Woock, and W. Grill, “Efficiency analysis of waveform shape for electrical excitation of nerve fibers,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 18, no. 3, pp. 319–328, 2010.

[4] A. Vuckovic, N. Rijkhoff, and J. Struijk, “Different pulse shapes to obtain small fiber selective activation by anodal blocking-a simulation study,” *Biomedical Engineering, IEEE Transactions on*, vol. 51, no. 5, pp. 698–706, 2004.

[5] A. Vučković and N. Rijkhoff, “Different pulse shapes for selective large fibre block in sacral nerve roots using a technique of anodal block: an experimental study,” *Medical and Biological Engineering and Computing*, vol. 42, no. 6, pp. 817–824, 2004.

[6] N. Accornero, G. Bini, G. Lenzi, and M. Manfredi, “Selective Activation of peripheral nerve fibre groups of different diameter by triangular shaped stimulus pulses.” *The Journal of Physiology*, vol. 273, no. 3, p. 539, 1977.

[7] A. M. Dymond, “Characteristics of the metal-tissue interface of stimulation electrodes,” *IEEE Trans. Biomed. Eng.*, vol. 23, no. 4, pp. 274–280, 1976.

[8] T. Stieglitz, H. Beutel, M. Schuettler, and J. Meyer, “Micromachined, polyimide-based devices for flexible neural interfaces,” *Biomedical Microdevices*, vol. 2, no. 4, pp. 283–294, 2000.

[9] T. Rose and L. Robblee, “Electrical stimulation with Pt electrodes. VIII. Electrochemically safe charge injection limits with 0.2 ms pulses,” *IEEE Trans. Biomed. Eng.*, vol. 37, no. 11, pp. 1118–1120, 1990.

[10] G. T. A. Kovacs, “Introduction to the theory, design, and modeling of thin-film microelectrodes for neural interfaces,” in *Enabling Technologies for Cultured Neural Networks*, D. A. Stenger and T. McKenna, Eds. Academic Press, 1994, pp. 121–65.