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Evaluation of High-Perimeter Electrode Designs for Deep Brain Stimulation

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Abstract

Deep brain stimulation (DBS) is an effective treatment for movement disorders and a promising therapy for treating epilepsy and psychiatric disorders. Despite its clinical success, complications including infections and mis-programming following surgical replacement of the battery-powered implantable pulse generator impact the safety profile of this therapy. We sought to decrease power consumption and extend battery life by modifying the electrode geometry to increase stimulation efficiency. The specific goal of this study was to determine whether perimeter or area had a greater effect on increasing stimulation efficiency. Finite-element method (FEM) models of eight prototype electrode designs were used to calculate the electrode access resistance, and the FEM models were coupled with cable models of passing axons to quantify stimulation efficiency. We also measured *in vitro* the electrical properties of the prototype electrode designs and measured *in vivo* the stimulation efficiency following acute implantation in anesthetized cats. The results indicated that: (1) area had a greater effect than perimeter on altering the electrode access resistance; (2) electrode (access or dynamic) resistance alone did not predict stimulation efficiency, because efficiency was dependent on the shape of the potential distribution in the tissue; and (3) quantitative assessment of stimulation efficiency required consideration of the effects of the electrode-tissue interface impedance. These results advance understanding of the features of electrode geometry that are important for designing the next generation of efficient DBS electrodes.

Keywords

electrode design; neural stimulation; deep brain stimulation; power efficiency; resistance

1. Introduction

Electrical stimulation of the nervous system is an effective therapeutic modality for treating neurological disorders. One of the most successful examples is deep brain stimulation

(DBS), where a surgically implanted metal electrode array is used to deliver 100–200 Hz electrical pulses within specific regions of the brain. DBS is effective in treating movement disorders, including Parkinson's disease (PD) and essential tremor (Limousin et al., 1999, Montgomery, 1999), and recent clinical studies have shown DBS to be a promising therapy for treating epilepsy and psychiatric disorders (Kuhn et al., 2010, Mayberg et al., 2005). Yet, despite the success of DBS, there are several areas where the therapy can be improved. The objective of the present study was to determine the effect of electrode geometry on the efficiency of stimulation, as this impacts the lifetime of the implanted pulse generators (IPG) used to deliver DBS.

The IPGs used in DBS have an average battery life of 2–5 years (Ondo et al., 2007, Halpern et al., 2011). Since the battery life of the IPG is less than the treatment duration, patients using DBS require repeated battery replacement surgeries, which are costly and obligate the patient to incur repeatedly the risks associated with surgery, including infection (Boviatsis et al., 2010, Bronstein et al., 2011) and misprogramming (Okun et al., 2005). Rechargeable IPGs (RPGs) are an alternative to primary cells: they have longer predicted battery lives but are more costly, need to be recharged on a regular basis, and are not well accepted (Lam and Rosenow, 2010). For example, in a cohort of 35 patients implanted with an RPG for spinal cord stimulation, 19 patients stated that they would have tolerated a 25 % larger RPG to relieve them of the burden of recharging, while 5 patients stated they would have tolerated a 50 % larger RPG (Lam and Rosenow, 2010). Therefore, RPGs are not a panacea for limited device lifetimes, and stimulation efficiency needs to be improved.

The source driving polarization of the neural membrane is proportional to the second order (centered) difference of the extracellular potentials ($\delta^2\Phi$) for myelinated axons and the second order spatial derivative of the extracellular potentials (Φ'') for unmyelinated axons and dendrites (Rattay, 1986). Because the magnitude of the extracellular potentials (and thereby $\delta^2\Phi/\Phi''$) is proportional to the applied current, and because the lumped access/series resistance of the electrode (R_a) is inversely proportional to the area of the electrode, stimulation efficiency can be increased by increasing electrode area (Butson et al., 2006), which reduces R_a .

R_a can also be decreased by increasing the perimeter of the electrode. An electrode inlaid in an insulating substrate generates a current density on its surface that is minimal at the electrode center and maximal at the edges (or perimeter) where the conductor and insulator meet. This phenomenon, referred to as “field fringing” or “edge effects”, occurs because charges on the edges, those which are not completely surrounded by neighboring charges, have the greatest access to the tissue. Therefore, in addition to an increased electrode area, an increased electrode perimeter may also increase stimulation efficiency (Wei and Grill, 2005, Wei and Grill, 2009a, Pendekanti and Henriquez, 1996).

The goal of this study was to determine which geometrical parameter, perimeter or area, more effectively increased stimulation efficiency. We modeled and tested eight prototype electrode designs with different areas and perimeters. First, we developed finite element method (FEM) models of the electrode designs to quantify the corresponding potential distribution, current density, and R_a of each design. Next, the FEM models were coupled

with cable models of passing axons to quantify stimulation efficiency. Finally, we quantified *in vitro* the electrical properties of our electrode designs and measured *in vivo* the stimulation efficiency following acute implantation in anesthetized cats. The outcome of this effort is a fundamental understanding of the features of electrode geometry that are the most important for designing the next generation of DBS electrodes.

2. Methods

2.1 Fabrication of electrode designs

Eight prototype electrode designs were fabricated by sputter depositing a chrome adhesion layer followed by a gold metal layer onto a planar polyimide substrate (Figure 1a). Planar substrates were wrapped around and bound to a steel cylindrical shaft to form cylindrical electrodes for stimulation *in vivo*.

Two different high-perimeter electrode designs were used to test the hypothesis that increased electrode perimeter from segmentation and/or increased edge curvature could decrease R_a . On one array, we tested a set of serpentine-edge or Serp (S) designs, consisting of two Serp electrodes, S1 and S2, and two banded electrodes, S3 and S4, which served as respective controls. The effect of perimeter was assessed by comparing S2 vs. S4 and S1 vs. S3; and we assessed whether the effect of area or perimeter was greater by comparing S1 vs. S2, S1 vs. S4, S2 vs. S3, and S3 vs. S4. Area was only reduced by factor of 1/3 to maintain the same edge-to-edge effective length between designs with different areas and the same type of edge (flat or serpentine), and increases in perimeter were capped at 300 % for a given area because increases greater than this could potentially generate charge densities per phase that are damaging to the electrode and neural tissue (Wei and Grill, 2009a).

On another array, we tested a set square-edge or Rook (R) designs, consisting of three Rook designs (R1, R2, and R3) and a banded electrode (R4) that served as a control. We were interested in rook electrodes because of their sharp edges, or areas of curvature approaching infinity, that could potentially enhance the “edge effect” while keeping the average charge density per phase within a range that has been deemed potentially non-damaging (Shannon, 1992). The effect of perimeter was assessed by comparing R1 vs. R4, and we assessed whether the effect of area or perimeter was greater by comparing R1 vs. R2, R1 vs. R3, R2 vs. R4, and R3 vs. R4. R2 and R3, which had an equal area and perimeter, were compared to assess the importance of the edge-to-edge electrode length.

2.2 Computational model of deep brain stimulation

We implemented a computational model of deep brain stimulation (DBS) consisting of a three dimensional FEM model of each prototype electrode within a volume conductor representing the tissue medium and cable-based models of passing myelinated axons (McIntyre and Grill, 2002). The FEM model, implemented in COMSOL Mutiphysics v3.4, included a single DBS electrode contact on an insulating shaft surrounded by brain tissue (Howell et al., 2014). Each electrode had a radius of 0.635 mm and longitudinal dimensions shown in Figure 1a. The volume conductor was large enough (60 mm × 60 mm × 60 mm cube) that doubling its volume resulted in a less than 5 % change in both the calculated potentials (Φ) and $\delta^2\Phi$. The electrical conductivity of the volume (0.2 S/m, (Gabriel et al.,

2009) was homogeneous, isotropic and purely resistive (Bossetti et al., 2008). The volume was discretized using ~ 0.5 million 3rd order Lagrangian elements, because interpolation of $\delta^2\Phi$ without discontinuities requires third order or greater elements (Howell et al., 2014). The mesh density was fine enough such that additional mesh refinement (i.e., subdivision of the elements) altered Φ and $\delta^2\Phi$ by less than 1 % and 2 %, respectively.

A normal current density of 0 A/mm² (i.e., a Neumann boundary condition) was imposed on the electrode shaft to model a perfect insulator, and fixed potentials of 1 V and 0 V (i.e., Dirichlet boundary conditions) were imposed on the electrode surface and outer boundary, respectively, to model monopolar stimulation. (Note, we do not distinguish between current-regulated stimulation and voltage regulated stimulation in the static case, when no interface is present, because the applied voltage and applied current are constant.) The electric potentials were calculated by solving Laplace's equation (Equation 1), and the applied current was calculated by integrating the current density (Equation 2) over the surface of the electrode. Changes in Φ and $\delta^2\Phi$ in the FEM model from mesh refinement and volume doubling resulted in < 1 % change in the calculated stimulation thresholds.

$$\nabla \cdot (\sigma \nabla \phi) = -\nabla \cdot \vec{J} = 0 \quad (1)$$

$$\vec{J} = \sigma \vec{E} \quad (2)$$

R_a was calculated by dividing the applied voltage by the resulting current. In an infinite medium, R_a can be broken into two parts: the distributed resistance of the modeled domain and the lumped resistance between an equipotential surface and infinity (Behrend et al., 2008). All electrodes produced equipotential surfaces that were approximately spherical well within the bounds of the modeled domain, so a point-source electrode was used to approximate the lumped resistance of the tissue medium. The distributed resistance of all electrode designs was > 80 times the lumped resistance between the outermost boundary and infinity (an error of < 2 %), so the distributed resistance calculated with the FEM model served as a suitable approximation for the true R_a .

Cable models of myelinated axons were implemented in the NEURON v7.1 simulation environment (Carnevale and Hines, 1997). Each axon was 2.5 μ m in diameter, 15 mm in length, and oriented either parallel or perpendicular to the electrode axis. The neural membrane at each node of Ranvier contained the parallel combination of a nonlinear sodium conductance (1.445 S/cm²), a linear leakage conductance (0.128 S/cm²), and the membrane capacitance (2.5 μ F/cm²), as these conductances (Sweeney et al., 1987) are sufficient for predicting activation thresholds (Schwarz et al., 1995). The myelin between nodes of Ranvier was assumed to be perfectly insulating.

The FEM and cable models were coupled to analyze the power efficiency of each electrode design. Populations were represented by 100 model axons uniformly distributed in an annulus around the electrode. Based on predicted volumes of tissue activated with the Model 3387 for typical DBS parameters, neural elements are expected to lie between ~0.9 and 2.9 mm from the surface of the electrode (Butson et al., 2007). To span this range, population of

axons were uniformly distributed in an annulus around the electrode with an inner and outer radius of 1 and 4 mm, respectively. Polar coordinates (r , θ) were picked using a Latin hypercube sampling, and r and θ were mapped to x and y using a coordinate mapping that maintained a uniform distribution in Cartesian coordinates.

The extracellular potentials at the nodes of Ranvier of the myelinated axons were determined via cubic interpolation from the potentials at the FEM grid points, and the interpolated potentials were used in NEURON to stimulate the population of passing axons with a 500 μ s monophasic rectangular pulse. Because the conductivity was linear, the potentials at a specific stimulation voltage were calculated by multiplying the 1V solution by a scalar. The threshold stimulation voltage to activate each model axon was calculated using a bisection algorithm (relative error < 1 %), and electrical power was calculated as the product of the stimulation threshold voltage and the corresponding current. Input-output (activation) curves of power versus the fraction of axons activated were used to assess power efficiency.

2.3 In vitro measurements

Measurements of voltage transients during applied current pulses and electrochemical impedance spectroscopy were conducted *in vitro* using the three-electrode technique in a single-compartment electrochemical cell. We used a large-area (4 cm²) platinum wire mesh as the counter electrode (CE), a silver/silver chloride (Ag/AgCl) electrode (Model RE-5B, BAS Inc., West Lafayette, IN, USA) as the reference electrode (RE), and the electrode design being tested was the working electrode (WE). All measurements were made at 37°C using an electrolyte model of interstitial fluid (2 mM Na₂HPO₄·7H₂O, 0.5 mM NaH₂PO₄·H₂O, 28 mM NaHCO₃, 7.5 mM KHCO₃, 110 mM NaCl, 0.5 mM MgSO₄, 0.5 mM MgCl₂, 0.5 mM CaCl₂) that was bubbled with a mixture of gas (5% O₂, 6% CO₂, and 89% N₂) to maintain a pH of 7.4 (Cogan, 2006). Before each experiment, metal electrodes were cleaned in a 2 M solution of sulfuric acid (H₂SO₄) for 30 minutes, sonicated in 70% ethanol for 30 minutes, and rinsed in deionized water (Fischer et al., 2009, Hudak et al., 2010, Izumi et al., 1991).

Voltage transients were generated by delivering 1 ms symmetric biphasic rectangular current pulses between the WE and CE at three amplitudes: 0.1 mA, 0.5 mA, and 1 mA. Measurements were repeated three times at each current level. The voltage transients were measured between the WE and RE using an isolated differential amplifier (100 M Ω input impedance, Model SR560, Stanford Research Systems, Sunnyvale, CA, USA) and recorded in MATLAB using the Data Acquisition (DAQ) Toolbox at a sampling rate of 40 kHz.

Electrochemical impedance spectroscopy was conducted using an electrochemical interface and frequency response analyzer (ECI Model 1287 and FRA Model 1252, Solartron Analytical, Hampshire, UK). The FRA applied a (10 mV) sinusoidal voltage superposed with a 0.3V DC bias voltage, and the ECI measured the resultant current between the WE and RE. The frequency of the applied sinusoidal voltage was varied between 1 Hz and 10 kHz.

2.3.1 Calculating Randles circuit values from voltage transients—The voltage (V_{cr}) across a three-element Randles equivalent circuit in response to a current pulse is described by Equation 3: where, I_{cr} is the applied current, V_0 is the initial voltage at $t = 0$, R_f is the Faradaic (charge-transfer) resistance, C_{dl} is the double-layer capacitance, R_a is the access (or series) resistance of the electrolyte/tissue, PW is the pulse width of the current pulse, and τ_{cr} is the time constant (Equation 4). In general, R_f and C_{dl} vary nonlinearly with the electrode potential, and R_f is not symmetric about the open-circuit potential. Therefore, R_a , R_f , and C_{dl} were determined by using nonlinear regression to fit Equation 3 to each phase of the recorded response: the first phase, $0 \leq t < \text{PW}$; the second phase, $\text{PW} \leq t < 2\text{PW}$; and the terminal phase, $t \geq 2\text{PW}$.

$$V_{cr}(t) = e^{-t/\tau} (V_0 - I_{cr}(R_a + R_f)) + I_{cr}(R_a + R_f) \quad (3)$$

$$\tau_{cr} = R_f C_{dl} \quad (4)$$

2.4 In vivo power efficiency analysis

Acute *in vivo* experiments were conducted in anesthetized cats using a protocol approved by the Institutional Animal Care and Use Committee of Duke University. Anesthesia was induced with ketamine HCL (Ketaset 35 mg/kg IM, supplemented as required at 15 mg/kg during surgical preparation) and maintained with alpha-chloralose (65 mg/kg IV, supplemented at 15 mg/kg). A 1 cm diameter craniotomy was made, the dura was reflected, exposing the cortex, and electrodes were implanted unilaterally in a vertical orientation (z direction) using stereotactic neurosurgical technique. We targeted the posterior internal capsule, which contains corticobulbar axons tracts projecting from the face region of the motor cortex (Duerden et al., 2011), and is located 7 mm anterior, 10.5 mm lateral and 10 mm above the interaural line (Nicholson, 1965, Snider and Niemer, 1961).

An analog isolated stimulator (ASI Model 2220, A-M Systems, Carlsborg, WA) was used to deliver stimulation between each prototype electrode and a subcutaneous 23 gauge needle at 0.5 Hz across a range of stimulus amplitudes between 0 and 5 mA. Stimulus waveforms were current-regulated, symmetric pulses with a cathodic (500 μ s) primary phase followed by a symmetric anodic (500 μ s) phase. The pulses were controlled with a high-speed digital-to-analog converter using the MATLAB Data Acquisition toolbox (Mathworks).

We measured the *in vivo* stimulation efficiency of each design by activating the posterior IC and recording evoked electromyographic (EMG) responses in the contralateral jaw muscles. Low-noise voltage preamplifiers (Model SR560, Stanford Research Systems, Sunnyvale, CA, USA) were used to record the applied current by measuring the voltage across a 100 Ω series resistor and the applied voltage by measuring the voltage across the stimulating electrode and a Ag/AgCl reference electrode (Model RE-4, BAS Inc., West Lafayette, IN, USA). Power efficiency was quantified by fitting the Hill equation (Equation 5) to the integral of the rectified EMG response as a function of the peak power, where f is a fraction of the maximum response, P is the stimulation power, P_{50} is the power needed to evoke a

half maximal response, and n is the Hill-fitting coefficient. The fitted expression was used to calculate the P_{50} for each electrode.

$$f = P^n / (P_{50}^n + P^n) \quad (5)$$

2.5 Sensitivity Analyses

Evoking an EMG response in the contralateral jaw muscles required a current-regulated waveform with a 500 μ s cathodic phase followed by a 500 μ s anodic phase, so to be consistent, we used the same waveforms in our modeling and *in vitro* analysis. However, in DBS, typical pulse widths are between 60–210 μ s (Kuncel and Grill, 2004), and waveforms can be either voltage-regulated or current-regulated. To address the effects of a smaller PW and the choice of a voltage-regulated waveform on the model results, we modeled the filtering effects of the ETI and conducted subsequent simulations at PWs of 100 μ s and 500 μ s.

We approximated the filtering effects of the ETI using the static model (i.e., with no ETI) to solve for the potential distribution in space and the Randles circuit to solve for the decay of potentials over time (Howell et al., 2014). For each electrode, R_a was set to the static model value, and the C_{dl} and R_f were set to the average values measured *in vitro*. The effects of the ETI on efficiency were quantified by determining the relative change in power efficiencies, compared to the static case, across the six combinations of Rook designs and the six combinations of Serp designs.

2.6 Statistical Analysis of Data

Data are presented as the mean $\pm$ standard error across five experiments, and all statistical tests were performed at a 95 % confidence level (i.e., $\alpha < 0.05$). Circuit element values from the model fitting were first analyzed with a repeated measures analysis of variance (ANOVA) to test the (null) hypothesis that all values came from the same distribution. A repeated measures approach was necessary because multiple electrode designs were tested per experiment. If the null hypothesis was rejected, post-hoc comparisons were made using a protected Fisher's least significant difference test.

Equation 5 was fit to the experimental data used to construct input-output curves of power vs. the integral of the rectified EMG, and a repeated measures ANOVA was conducted on the P_{50} values to assess if the effect of electrode design was significant ($\alpha < 0.05$). If the effect of lead design was significant, *post hoc* t-tests were run to compare the relative power performance of the tested electrode designs versus the control.

3. Results

We coupled potentials calculated with FEM models of eight prototype electrode designs with varying areas and perimeters to cable models of passing axons to quantify the effect of electrode geometry on DBS stimulation efficiency. Next, we conducted *in vitro* pulse transient analysis and quantified the electrical properties of the prototype electrodes by fitting the measured response to a three-element Randles equivalent circuit. Finally, we

measured *in vivo* the stimulation efficiency of select designs following acute implantation in the posterior internal capsule of anesthetized cats.

3.1 Model-based analysis of high-perimeter electrode designs

3.1.1 Effects of electrode geometry on access resistance—The FEM models were used to calculate R_a of each electrode design (Table 1). Increasing the perimeter of an electrode with a fixed area decreased R_a , but altering the surface area of the electrode had a greater effect on R_a than altering its perimeter. For example, the perimeter of R1 was 400 % greater than the perimeter of R4, and as a result, the R_a of the former was 10 % less than the R_a of the latter (Table 1). Yet, all electrodes with smaller surface areas (R1, R4, S1, and S3) had larger R_a than electrodes with large areas (R2, R3, S2, and S4).

3.1.2 Effects of electrode geometry on the power efficiency of stimulation—In electrode pairs where the former electrode had an equal area and greater perimeter than the latter (e.g., S1 vs. S3, S2 vs. S4, and R1 vs. R4), less power was required to activate axons oriented perpendicular to the electrode axis and (approximately) the 50 furthest axons oriented parallel to the electrode axis, but more power was required to activate the 50 closest axons oriented parallel to the electrode axis. Similar results were obtained in electrode pairs where the former had a greater area and perimeter than the latter (e.g., S2 vs. S3 and R3 vs. R4). Further, in the electrode pair where the former and latter had an equal area and perimeter, but different maximum edge-to-edge length (i.e., R2 vs. R3); relative power efficiency also depended on the distance and orientation of the axons. These relationships are summarized in Figure 3.

3.1.3 Shape of potential distribution impacts stimulation efficiency—We examined $\delta^2\Phi$ to quantify the dependence of stimulation efficiency on the orientation of the axons. In electrode pairs with equal areas (e.g., R1 vs. R4, R3 vs. R2, S1 vs. S3, and S2 vs. S4), the maximum $\delta^2\Phi$ parallel to the electrode axis was smaller in the former electrodes, which had the greater maximum edge-to-edge length or effective height (h_{eff}) (Figure 4). In these same electrode pairs, the maximum $\delta^2\Phi$ perpendicular to the electrode axis was larger in the electrodes with the greater h_{eff} . The dependence of $\delta^2\Phi$ on h_{eff} , however, was only observed in approximately the 50 closest axons, which were ~ 2 mm from the electrode surface. At distances greater than 2 mm, the magnitude of $\delta^2\Phi$ was greatest for electrodes with the smallest R_a . Therefore, close to the electrode, electrode geometry had an orientation-dependent effect on the shape of $\delta^2\Phi$ that markedly influenced stimulation efficiency; and further away, efficiency was better predicted by R_a .

3.2 Electrical properties of high-perimeter electrode designs measured *in vitro*

3.2.1 Three-element Randles circuit is an appropriate model—Electrochemical impedance spectroscopy was used to characterize the electrical properties of prototype electrodes *in vitro*. The measured impedance consisted of an electrode-electrolyte impedance, including a Faradaic resistance in parallel with a double-layer capacitance, and an access (electrolyte) resistance (Figure 5a). The real and imaginary components of the total impedance formed a partial semicircle for frequencies between 100 Hz and 10 kHz (Figure 5b), which was well fit with a Randles circuit (Bard and Faulkner, 2000). Below 100

Hz, the impedance diverged from a semicircle (Figure 5b), indicating diffusion limited transport of electrochemical species (Bard and Faulkner, 2000). Since greater than 96% of the power in a standard DBS pulse (500 μ s or less) is contained in the frequency band above 100 Hz (Bossetti et al., 2008), the Randles circuit is an appropriate model.

3.2.2 Randles circuit values of electrode designs—The equivalent circuit parameters (R_a , R_f , C_{dl}) for each electrode design were determined by fitting Equation 3 to each phase of the voltage transients measured *in vitro* (Figure 5c), and a repeated-measures ANOVA ($\alpha < 0.05$) was performed on the circuit parameters of the first phase to assess the effect of electrode geometry on the circuit values (Figure 6). For both serpentine and rook electrodes, there was a significant effect of electrode geometry that depended on the current amplitude, and this effect was present for all circuit parameters. Post-hoc t tests were performed to compare the values of the circuit elements between the individual electrodes.

Across all current amplitudes, electrodes with larger surface areas (R2, R3, S2, and S4) had significantly smaller R_a and larger C_{dl} than electrodes with smaller surface areas (R1, R4, S1, and S3), except at 1 mA, where C_{dl} of R1 and R3, and S3 and S4 were not significantly different. For rook electrodes, R_f was insensitive to changes in electrode area over the range of values tested, but for serpentine electrode, R_f depended on the surface area of the electrode. Unlike area, the effect of perimeter was not as consistent. As predicted by the FEM models, in electrodes pairs with equal area and unequal perimeter (e.g., R1 vs. R4 and S2 vs. S4), the electrodes with a greater perimeter had a significantly smaller R_a . However, unlike the models' predictions, R2 and R3, which had the same area and perimeter, did not have significantly different R_a ; and between S1 and S3, S1, despite having a larger perimeter, had a significantly greater R_a . In most cases, R_f and C_{dl} of electrodes with equal area but different perimeter were not significantly different.

3.3 Power efficiency of electrode designs measured in vivo

Since altering electrode area has a greater effect than altering electrode perimeter, we only tested two sets of electrode pairs with different areas *in vivo*: R3 vs. R4 and S3 vs. S4. Input-output curves of power dissipated versus the integral of the rectified EMG were constructed for the four electrodes (Figure 7), and the efficiency of stimulation was quantified by calculating the peak power required to evoke a half-maximal response, P_{50} . R3 consumed significantly (8 ± 3 %, $p = 0.029$) less power than R4, and S3 consumed significantly (21 ± 5 %, $p = 0.009$) more power than S4.

3.4 Impact of electrode tissue interface on model predictions

For current-regulated waveforms with PWs of 100 μ s and 500 μ s, the relative efficiencies of the 12 electrode pairs differed by 0.0021–3.9 % and 0.013–4.8 %, respectively, compared to the static case. In 11 of the 12 of the electrode pairs, electrodes that were more (less) efficient in the static case remained more (less) efficient when the filtering effect of the ETI was added.

During voltage-regulated stimulation, the effects of the ETI were more pronounced. With PWs of 100 μ s and 500 μ s, the relative efficiencies of the 12 electrodes pairs in the ETI case

differed by 0.092–3.4 % and 0.029–25 %, respectively, compared to the static case. The large changes observed with a PW of 500 μ s had the effect of reversing the trends predicted by the static model in 7 of the 12 electrode pairs. Therefore, electrodes that were more (less) efficient in the static case were less (more) efficient in the ETI case.

4. Discussion

The goal of this study was to determine how changes in electrode area and perimeter affected the efficiency of stimulation. The results yielded three important conclusions for designing more power efficient stimulation electrodes. First, the electrode access resistance can be reduced by increasing electrode area and/or perimeter, but the effect of increasing perimeter is not as pronounced as increasing area. Second, efficiency is dependent on the distribution of potentials in the tissue, so electrode (access or dynamic) resistance alone cannot be used to predict accurately the stimulation efficiency. And third, because the electrode-tissue interface (ETI) impedance is of the same order of magnitude as the access resistance and dependent on electrode geometry, the ETI impedance has a marked effect on stimulation efficiency.

4.1 Decreasing the access resistance of an electrode

Our modeling results demonstrate that R_a can be decreased by increasing the electrode area and/or perimeter. Increasing area by 50 % decreased R_a by as much as 23 %, while increasing perimeter by 400 % decreased R_a by only ~10 %. Therefore, increasing area had a much more prominent effect in decreasing R_a than increasing perimeter. These predictions were confirmed *in vitro* (Figure 6), with the exception of one case, S1 versus S3. S1 had a perimeter three times that of S3, and the model predicted that R_a of S1 would be less than R_a of S3. However, the *in vitro* measurements showed the opposite result (Figure 6b). Inspection of S1 under a microscope revealed that the surface of S1 was corroded (oxidized) at the tips of the serpentine edges, where the current density is highest (Wei and Grill, 2009a). A reduction in the surface area of S1 not only explained the unexpected result, but it also showed that high-perimeter designs with high-curvature edges (e.g., S1 and S2) may be more prone to electrochemical reactions that damage the electrode surface.

4.2 The effects of altering electrode geometry on the efficiency of neural stimulation

Increasing the perimeter of a cylindrical electrode with a fixed area and radius requires increasing its maximum edge-to-edge length, or h_{eff} . A cylindrical electrode whose height (h) is greater than its diameter (d) generates a potential distribution that is more uniform parallel to its axis than perpendicular to its axis, which reduces $\delta^2\Phi$ parallel to the electrode axis, so a long electrode (h or $h_{eff} > d$), despite having a smaller R_a , is expected to be less efficient than a short electrode in stimulating neural elements oriented parallel to the electrode axis.

Increasing h also brings the electrode surface closer to neural elements that are displaced longitudinally along the insulating shaft. The stimulation threshold for activating a neural element is inversely related to its distance from the electrode, so longer electrodes, because of their increased proximity to the targeted perpendicular elements, are expected to be more

efficient than shorter electrodes in stimulation of perpendicular neural elements. This explains why the high-perimeter designs (e.g., R3 and S2), compared to banded designs with greater R_a (e.g., R4 and S3), were less efficient in activating the closest axons oriented parallel to the electrode (Figure 3a) but more efficient in activating the closest axons oriented perpendicular to the electrode (Figure 3b). The orientation-dependent effects of electrode geometry on $\delta^2\Phi$ diminished with distance, so the efficiency of stimulating the furthest axons was best explained by a reduction in R_a .

4.3 The impact of the electrode tissue interface on the efficiency of neural stimulation

The area and perimeter of the electrode affected not only R_a but also C_{dl} (Figure 6). Because C_{dl} affects the rate at which the ETI charges, and because the extent to which the ETI charges depends both on the choice of PW and whether the waveform is voltage-regulated or current-regulated, sensitivity analyses were conducted to assess the impact of the ETI on our results. The model predictions were still valid for current-regulated waveforms with PWs of 100 μ s and 500 μ s, as well as voltage-regulated waveforms with a PW of 100 μ s; but when using voltage-regulated waveforms with a PW of 500 μ s, the presence of the ETI altered the model predictions in 7 of the 12 electrode pairs.

As the ETI charges during current-regulated stimulation, the applied current is kept constant by increasing the applied voltage over time (Equation 3). Because the applied current is constant, the magnitude of the potentials in the tissue is also constant, preserving any effects of electrode geometry on $\delta^2\Phi$ (Figure 4). Thus, during current-regulated stimulation, differences in the relative power efficiency between two electrodes over time arise from differences between their applied voltages. Recall from Equations 3 and 4, the applied voltage increases at a rate that is inversely proportional to both C_{dl} and R_f . Indeed, altering the area and perimeter had a significant effect on altering C_{dl} (Figure 6), which led to differences in the applied voltage over time between electrodes with different C_{dl} (e.g., S1 vs. S2). However, for $t = 500 \mu$ s, the differences in the applied voltage over time were small enough that relative power efficiencies between electrode pairs with different C_{dl} changed by $< 5 \%$.

During voltage-regulated stimulation, the applied voltage is held constant, and as the ETI charges, the applied current decreases over time. Equations 6 and 7 describe the applied current (I_{vr}) in an equivalent Randles circuit in response to a constant applied voltage (V_o).

$$I_{vr} = \frac{V_o}{(R_a + R_f)} \left(\frac{R_f}{R_a} e^{-t/\tau_{vr}} + 1 \right) \quad (6)$$

$$\tau_{vr} = C_{dl} R_f R_a / (R_a + R_f). \quad (7)$$

Increasing the area and/or perimeter of the electrode increases C_{dl} and reduces R_a (Figure 6), which have competing effects in altering τ_{vr} . Because R_a and R_f are typically on the same order of magnitude in brain tissue (Wei and Grill, 2009b), τ_{vr} is more sensitive to changes in C_{dl} than R_a and R_f , and the overall effect of increasing area and/or perimeter is an increase in τ_{vr} . Therefore, between a pair of electrodes with different τ_{vr} , the power efficiency of the

electrode with the larger τ_{vr} will increase over time with respect to the power efficiency of the electrode with the smaller τ_{vr} because the magnitude of the potentials generated by the former decay less rapidly over time than the potentials generated by the latter.

The difference in the rate of decay of I_{vr} between electrodes with different C_{dl} explains why in seven electrode pairs (i.e., R3 vs. R1, R3 vs. R2, R3 vs. R4, S2 vs. S1, S1 vs. S3, S2 vs. S3, and S2 vs. S4), where the former has a larger C_{dl} and longer h_{eff} than the latter, the relative efficiency between the two was dependent on PW. Consider, for example, S2 and S4. S2 had a greater C_{dl} than S4, so as the ETI charged, I_{vr} of S2 declined more slowly over time than I_{vr} of S4. At $t = 100 \mu s$, the difference in I_{vr} between S2 and S4 was not large enough to impact markedly the effect of electrode geometry on $\delta^2\Phi$ (Figure 4), so like the static case, S2 was less efficient than S4 in activating axons parallel to the electrode. However, at $t = 500 \mu s$, I_{vr} of S4 had declined far enough below I_{vr} of S2 that unlike the static case, S2 was more efficient than S4 in activating parallel axons.

4.4 Limitations

The methodology in this study made use of a linear representation of the ETI to study the effects of electrode geometry on R_f and C_{dl} . Because R_f and C_{dl} vary nonlinearly with the applied potential, simplification of the ETI requires consideration. The nonlinear dependence of R_f and C_{dl} on the applied potentials is described by The Butler-Volmer and Gouy-Chapman-Stern models, respectively (Bard and Faulkner, 2000). For typical stimulation parameters used in DBS, R_f and C_{dl} vary slowly enough in time that they can be approximated as constant during each phase of the pulse (Howell et al., 2014, Wei and Grill, 2009b). In other words, one can approximate the nonlinear ETI using a piecewise linear representation, where R_f and C_{dl} during each phase are chosen by evaluating their respective nonlinear expressions at the beginning of each phase.

A piecewise linear model of an equivalent Randles circuit was indeed accurate in describing the filtering effects of the nonlinear ETI both *in vitro* (Figure 5c) and *in vivo* (data not shown). However, it is not always valid to use an equivalent circuit model to describe the effects of the ETI, as this assumes the current density (J) is uniform throughout the conducting volume. Not only is J highly non-uniform on the electrode surface, but the non-uniformity of J depends on electrode geometry (Wei and Grill, 2005). Further, heterogeneities in the tissue conductivity, such as cerebral-spinal fluid filled spaces (Astrom et al., 2006) and encapsulation tissue (Butson et al., 2006), can alter J in the bulk tissue. In cases where J is highly non-uniform, the charging of the ETI cannot be described with a single τ (Oldham, 2004). Thus, future studies looking at the impact of the ETI on the stimulation efficiency of electrodes should consider explicit representation of the ETI.

4.5 Designing the next generation of deep brain stimulation electrodes

The future objective is to design electrode geometries that are optimal for stimulating white or grey matter regions of the brain so that electrodes used in DBS are better suited to their intended anatomical target. For example, tracts of myelinated axons within white matter regions are the primary targets in treating psychiatric disorders, such as obsessive-compulsive disorder and depression (Gutman et al., 2009, Mayberg et al., 2005); and axon

afferents within select brain nuclei of grey matter regions are the primary targets for treating movement disorders, including essential tremor (Birdno and Grill, 2008) and Parkinson's disease (Gradinaru et al., 2009). Our results highlight three important considerations for designing future DBS electrodes.

1. Area has a greater effect than perimeter on altering the electrical properties of the electrode and thereby impacting stimulation efficiency. Previous studies have proposed increasing efficiency by using electrode arrays to increase perimeter (Pendekanti and Henriquez, 1996, Wei and Grill, 2005) or by changing the dimensions of the cylindrical electrode contact to increase area (Butson and McIntyre, 2006). Since our modeling and experimental results support the latter, we recommend increasing efficiency by altering the height and/or diameter of the cylindrical electrode.
2. Stimulation efficiency depends not only on the (access and dynamic) resistance of the electrode but also on the shape of the potential distribution. High-perimeter designs, which had a greater effective height than a corresponding banded electrode with equal area, were more efficient at activating neural elements perpendicular to the electrode but less efficient at activating elements parallel to the electrode, despite decreases in R_a . Therefore, in therapies like DBS for the treatment of obsessive-compulsive disorder, where the target thalamocortical fibers connecting the mediodorsal thalamus to the prefrontal cortex (Lujan et al., 2008) are perpendicular to the implanted electrode shaft, performance can be increased by increasing the height of the cylindrical electrodes.
3. Electrode geometry has a marked impact on the filtering effects of the ETI. Pulse widths commonly used in DBS are between 60 – 210 μ s (Kuncel and Grill, 2004). Within this time window, the filtering effects of the ETI have a marked impact on the applied waveform (Wei and Grill, 2009b), as well as the stimulation thresholds (Butson and McIntyre, 2005), so we recommend that model-based design of stimulation electrodes consider the filtering effects of the ETI.

5. Conclusion

We theoretically and experimentally assessed the power efficiency of eight prototype electrode designs with varying areas and perimeters. The results of this analysis yielded three important conclusions. First, the electrode access resistance can be reduced by increasing electrode area and/or perimeter, but the effect of increasing perimeter is not as pronounced as increasing area. Second, efficiency is dependent on the distribution of potentials in the tissue, so electrode (access or dynamic) resistance alone cannot be used to predict accurately the stimulation efficiency. And third, because the impedance of the ETI is of the same order of magnitude as the access resistance and dependent on electrode geometry, it has a marked effect on stimulation efficiency. These results advance understanding of the features of electrode geometry that are important for designing the next generation of efficient DBS electrodes.

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References

- ASTROM M, JOHANSSON JD, HARIZ MI, ERIKSSON O, WARDELL K. The effect of cystic cavities on deep brain stimulation in the basal ganglia: a simulation-based study. *J Neural Eng.* 2006; 3:132–8. [PubMed: 16705269]
- BARD, AJ.; FAULKNER, LR. *Electrochemical Methods - Fundamentals and Applications.* Wiley; 2000.
- BEHREND MR, AHUJA AK, WEILAND JD. Dynamic current density of the disk electrode double-layer. *IEEE Trans Biomed Eng.* 2008; 55:1056–62. [PubMed: 18334397]
- BIRDNO MJ, GRILL WM. Mechanisms of deep brain stimulation in movement disorders as revealed by changes in stimulus frequency. *Neurotherapeutics.* 2008; 5:14–25. [PubMed: 18164480]
- BOSSETTI CA, BIRDNO MJ, GRILL WM. Analysis of the quasi-static approximation for calculating potentials generated by neural stimulation. *J Neural Eng.* 2008; 5:44–53. [PubMed: 18310810]
- BOVIATISIS EJ, STAVRINOUC LC, THEMISTOCLEOUS M, KOUYIALIS AT, SAKAS DE. Surgical and hardware complications of deep brain stimulation. A seven-year experience and review of the literature. *Acta Neurochir (Wien).* 2010; 152:2053–62. [PubMed: 20658301]
- BRONSTEIN JM, TAGLIATI M, ALTERMAN RL, LOZANO AM, VOLKMANN J, STEFANI A, HORAK FB, OKUN MS, FOOTE KD, KRACK P, PAHWA R, HENDERSON JM, HARIZ MI, BAKAY RA, REZAI A, MARKS WJ JR, MORO E, VITEK JL, WEAVER FM, GROSS RE, DELONG MR. Deep Brain Stimulation for Parkinson Disease: An Expert Consensus and Review of Key Issues. *Arch Neurol.* 2011; 68:165. [PubMed: 20937936]
- BUTSON CR, COOPER SE, HENDERSON JM, MCINTYRE CC. Patient-specific analysis of the volume of tissue activated during deep brain stimulation. *Neuroimage.* 2007; 34:661–70. [PubMed: 17113789]
- BUTSON CR, MAKES CB, MCINTYRE CC. Sources and effects of electrode impedance during deep brain stimulation. *Clin Neurophysiol.* 2006; 117:447–54. [PubMed: 16376143]
- BUTSON CR, MCINTYRE CC. Tissue and electrode capacitance reduce neural activation volumes during deep brain stimulation. *Clin Neurophysiol.* 2005; 116:2490–500. [PubMed: 16125463]
- BUTSON CR, MCINTYRE CC. Role of electrode design on the volume of tissue activated during deep brain stimulation. *J Neural Eng.* 2006; 3:1–8. [PubMed: 16510937]
- CARNEVALE, NT.; HINES, ML. *The NEURON Book.* Cambridge: 1997.
- COGAN, SF. In vivo and In vitro Differences in the Charge-injection and Electrochemical Properties of Iridium Oxide Electrodes. *Engineering in Medicine and Biology Society, 2006. EMBS '06; 28th Annual International Conference of the IEEE; Aug. 30 2006–Sept. 3 2006.* 2006. p. 882–885. Year
- DUERDEN EG, FINNIS KW, PETERS TM, SADIKOT AF. Three-dimensional somatotopic organization and probabilistic mapping of motor responses from the human internal capsule. *J Neurosurg.* 2011; 114:1706–14. [PubMed: 21375376]
- FISCHER LM, TENJE M, HEISKANEN AR, MASUDA N, CASTILLO J, BENTJEN A, ÉMNEUS J, JAKOBSEN MH, BOISEN A. Gold cleaning methods for electrochemical detection applications. *Microelectronic Engineering.* 2009; 86:1282–1285.
- GABRIEL C, PEYMAN A, GRANT EH. Electrical conductivity of tissue at frequencies below 1 MHz. *Phys Med Biol.* 2009; 54:4863–78. [PubMed: 19636081]
- GRADINARU V, MOGRI M, THOMPSON KR, HENDERSON JM, DEISSEROTH K. Optical Deconstruction of Parkinsonian Neural Circuitry. *Science.* 2009; 324:354–359. [PubMed: 19299587]
- GUTMAN DA, HOLTZHEIMER PE, BEHRENS TE, JOHANSEN-BERG H, MAYBERG HS. A tractography analysis of two deep brain stimulation white matter targets for depression. *Biol Psychiatry.* 2009; 65:276–82. [PubMed: 19013554]

- HALPERN CH, MCGILL KR, BALTUCH GH, JAGGI JL. Longevity Analysis of Currently Available Deep Brain Stimulation Devices. *Stereotactic and Functional Neurosurgery*. 2011; 89
- HOWELL B, NAIK S, GRILL WM. Influences of Interpolation Error, Electrode Geometry, and the Electrode-Tissue Interface on Models of Electric Fields Produced by Deep Brain Stimulation. *Biomedical Engineering, IEEE Transactions on*. 2014; 61:297–307.
- HUDAK EM, MORTIMER JT, MARTIN HB. Platinum for neural stimulation: voltammetry considerations. *Journal of Neural Engineering*. 2010; 7:026005.
- IZUMI T, WATANABE I, YOKOYAMA Y. Activation of a gold electrode by electrochemical oxidation-reduction pretreatment in hydrochloric acid. *Journal of Electroanalytical Chemistry and Interfacial Electrochemistry*. 1991; 303:151–160.
- KUHN J, GRUNDLER TO, LENARTZ D, STURM V, KLOSTERKOTTER J, HUFF W. Deep brain stimulation for psychiatric disorders. *Dtsch Arztebl Int*. 2010; 107:105–13. [PubMed: 20221269]
- KUNCEL AM, GRILL WM. Selection of stimulus parameters for deep brain stimulation. *Clinical Neurophysiology*. 2004; 115:2431–2441. [PubMed: 15465430]
- LAM CK, ROSENOW JM. Patient Perspectives on the Efficacy and Ergonomics of Rechargeable Spinal Cord Stimulators. *Neuromodulation: Technology at the Neural Interface*. 2010; 13:218–223.
- LIMOUSIN P, SPEELMAN JD, GIELEN F, JANSSENS M. Multicentre European study of thalamic stimulation in parkinsonian and essential tremor. *J Neurol Neurosurg Psychiatry*. 1999; 66:289–96. [PubMed: 10084526]
- LUJAN JL, CHATURVEDI A, MCINTYRE CC. Tracking the mechanisms of deep brain stimulation for neuropsychiatric disorders. *Frontiers in bioscience: a journal and virtual library*. 2008; 13:5892. [PubMed: 18508630]
- MAYBERG HS, LOZANO AM, VOON V, MCNEELY HE, SEMINOWICZ D, HAMANI C, SCHWALB JM, KENNEDY SH. Deep Brain Stimulation for Treatment-Resistant Depression. *Neuron*. 2005; 45:651–660. [PubMed: 15748841]
- MCINTYRE CC, GRILL WM. Extracellular Stimulation of Central Neurons: Influence of Stimulus Waveform and Frequency on Neuronal Output. *Journal of Neurophysiology*. 2002; 88:1592–1604. [PubMed: 12364490]
- MONTGOMERY EB JR. Deep brain stimulation reduces symptoms of Parkinson disease. *Cleve Clin J Med*. 1999; 66:9–11. [PubMed: 9926625]
- NICHOLSON PW. Specific impedance of cerebral white matter. *Exp Neurol*. 1965; 13:386–401. [PubMed: 5847284]
- OKUN MS, TAGLIATI M, POURFAR M, FERNANDEZ HH, RODRIGUEZ RL, ALTERMAN RL, FOOTE KD. Management of referred deep brain stimulation failures: a retrospective analysis from 2 movement disorders centers. *Arch Neurol*. 2005; 62:1250–5. [PubMed: 15956104]
- OLDHAM KB. The RC time “constant” at a disk electrode. *Electrochemistry Communications*. 2004; 6:210–214.
- ONDO WG, MEILAK C, VUONG KD. Predictors of battery life for the Activa Soletta 7426 Neurostimulator. *Parkinsonism Relat Disord*. 2007; 13:240–2. [PubMed: 17379565]
- PENDEKANTI R, HENRIQUEZ CS. Spatial potential and current distributions along transvenous defibrillation electrodes: variation of electrode characteristics. *Ann Biomed Eng*. 1996; 24:156–67. [PubMed: 8669713]
- RATTAY F. Analysis of Models for External Stimulation of Axons. *Biomedical Engineering, IEEE Transactions on*. 1986; BME-33:974–977.
- SCHWARZ J, REID G, BOSTOCK H. Action potentials and membrane currents in the human node of Ranvier. *Pflügers Archiv*. 1995; 430:283–292. [PubMed: 7675638]
- SHANNON RV. A model of safe levels for electrical stimulation. *Biomedical Engineering, IEEE Transactions on*. 1992; 39:424–426.
- SNIDER, RS.; NIEMER, WT. A stereotaxic Atlas of the Cat Brain. University of Chicago Press; 1961.
- SWEENEY, J.; MORTIMER, J.; DURAND, D. Modeling of mammalian myelinated nerve for functional neuromuscular stimulation. *IEEE 9th Annual Conference of the Engineering in Medicine and Biology Society*; 1987; p. 1577-1578. Year

- WEI XF, GRILL WM. Current density distributions, field distributions and impedance analysis of segmented deep brain stimulation electrodes. *J Neural Eng.* 2005; 2:139–47. [PubMed: 16317238]
- WEI XF, GRILL WM. Analysis of high-perimeter planar electrodes for efficient neural stimulation. *Front Neuroeng.* 2009a; 2:15. [PubMed: 19936312]
- WEI XF, GRILL WM. Impedance characteristics of deep brain stimulation electrodes in vitro and in vivo. *J Neural Eng.* 2009b; 6:1741–2560.

Figure 1. High-perimeter electrode designs for increasing the power efficiency of electrical stimulation. (a) Photomicrograph of the prototype electrode designs. (b) The area and perimeter of each electrode design as a fraction of the electrode dimensions of Serp 4, the lead presently used in DBS for Parkinson’s disease. The area (A) and perimeter (P) of Serp 4 are in units of π .

Figure 2.

The FEM electrode model and the surrounding neuronal elements. (a) The electrode and shaft were positioned such that the center of the electrode was at the origin of a conducting cube ($60 \text{ mm} \times 60 \text{ mm} \times 60 \text{ mm}$), and the axis of the shaft was oriented in the z direction. (b) Examples of randomly positioned passing axons distributed either perpendicular or parallel to the electrode surface.

Figure 3.

The relative power efficiency between six pairs of Rook and Serp designs. (a) The relative percent power consumed when stimulating parallel model axons versus the fraction of the population activated. Positive values indicate more power consumption and less efficiency, while negative values indicate the opposite. (b) The same format as in (a), except for perpendicular model axons.

Figure 4.

An example of the effect of electrode geometry on the shape of the potentials and their centered second difference ($\delta^2\Phi$) at an applied stimulus voltage of -1 V. *Top figures*, the potentials (left) and their respective $\delta^2\Phi$ (right) generated by S2 (open circles) and S4 (solid dots) at the nodes of a parallel axon 1.6 mm away from the electrode. *Bottom figures*, the potentials (left) and their respective $\delta^2\Phi$ (right) generated by S2 (open circles) and S4 (solid dots) at the nodes of a perpendicular axon 1.6 mm away from the electrode.

Figure 5.

In vitro measurements characterizing the electrical properties of the prototype electrodes. (a) The Randles circuit included the parallel combination of a Faradaic resistance and double-layer capacitance in series with an access (electrolyte or tissue) resistance. (b) *In vitro* impedance spectrums of serpentine electrode 4 (S4) calculated for frequencies between 1 kHz and 10 kHz (solid line + open circles), and 1 Hz and 100 kHz (dashed line). (c) Example of a voltage transient (bottom) in response to a current regulated biphasic square pulse (top). The analytical expression for a Randles circuit voltage transient was fit to each of the three phases of the voltage transient.

Figure 6.

Randles circuit element values (R_a , R_f and C_{dl}) calculated from model fitting to voltage transients recorded *in vitro* for (a) Rook and (b) Serpentine electrode designs. B, C, and D denote a statistical difference ($\alpha < 0.05$) with electrode numbers 2, 3, and 4, respectively. No asterisk denotes a p value of 0.01 $p < 0.05$; one asterisk denotes $p < 0.01$; and two asterisks denote $p < 0.001$.

Figure 7.

Input-output curves of peak power to stimulate the posterior internal capsule vs. the integral of the rectified EMG (IRE) of the contralateral jaw muscle. Raw data and fitted sigmoid functions from 5 separate experiments comparing the efficiency of (a) S3 to S4 and (b) R3 to R4. S3 consumed more power than S4 in eliciting a half-maximal response, while R3 consumed less power than R4.

Table 1

The theoretical access resistance (R_a) of the electrode designs

Electrode	Area (mm ²) $A = 1.905\pi$	Perimeter (mm) $P = 2.540\pi$	Access Resistance (Ω)	Relative R_a^b (%)
Rook 1	2A/3	4P	672	-10
Rook 2	A	2P	598	-19
Rook 3	A	2P	573	-23
Rook 4^a	2A/3	P	742	0
Serp 1	2A/3	6P	637	5
Serp 2	A	3P	569	-6
Serp 3	2A/3	2P	686	13
Serp 4^a	A	P	608	0

^a Bolded are the respective controls for the rook and serpentine electrodes

^b Relative change in R_a versus respective control