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Optimising Electrode Surface Area to Minimize Power Consumption in a Cortical Penetrating Prosthesis*

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Abstract— **In the design of cortical stimulating prostheses for applications such as vision perception or motor control it is preferable to use wireless power and data transfer to maintain a biological barrier to infection. This puts a constraint on the amount of power that can be supplied to the implant, in turn limiting the power that can be used for stimulation. Design of electrodes for such prostheses have considered factors such as efficiency of stimulation and penetrating capacity; here we consider the design of electrodes from the power consumption perspective. We use the simple electrode geometry of a sphere to determine if the surface area of the electrode can be chosen in order to minimize the power consumed during stimulation. As is known to happen when an electrode is inserted to penetrate into brain tissue, we have assumed that, due to mechanical damage from electrode insertion and the response of brain tissue to the foreign body of an electrode, there will be a kill zone around the electrode where no viable neurons are present. Using realistic thicknesses for the kill zone from the published literature and our own unpublished work, we demonstrate that when the width of the kill zone is known, an electrode's surface area can be chosen so that the power consumed during stimulation is minimized.**

I. INTRODUCTION

Bionic vision has been the goal of many researchers for over 45 years including visual cortical prostheses [1] that have the promise of being useful in a broader range of blindness conditions than prostheses in any other part of the visual pathway. This has previously been successfully achieved for a visual cortex prosthesis, albeit in a very limited manner: in 1968 Brindley and Lewin implanted wireless electrodes on to the surface of the visual cortex of a 42-year-old blind woman [1]. Regardless of where it is implanted, in order to maintain a barrier to infection [2] and minimize the number of surgeries, wireless transmission of power and data to the prosthetic device is preferred.

For any implantable electronic device, wired or wireless, it is important that the power supply be adequate for the intended purpose [3]. One constraint of using a wireless device is that the electronics of the tiles must fit in an area less than the area of the electrodes, so that the tiles can be tessellated to provide several hundred stimulation points. This requires a very small receiving coil for power and data, which greatly restricts the amount of power that can be received. Monash Vision Group is currently developing a bionic vision system based on a number of implanted tiles, each with electrodes penetrating the visual cortex; thus we are keen to minimize the power consumption of the implanted electronics. This study applies a novel perspective on the reduction of power consumption by considering how the electrodes to be used can influence this factor.

To minimize the power consumption of the implantable device as a whole, both the power required to drive a current through the electrode and the power consumption of the electronic circuit driving the electrode must be considered. We will consider a perfect driver, using active devices to control the current that have a very low forward resistance when 'on', compared with the electrode's impedance (Fig. 1a) Regardless of the exact details of these devices, the maximum power required by the stimulating system is simply the maximum voltage drop across the terminals of the stimulating and return electrodes, V_{active}, multiplied by the maximum stimulating current, I_{elec}. That is, once a power supply voltage is chosen that is sufficient to drive the required current through the electrode, the power consumption is proportional to the electrode current. At first glance it may be assumed that in order to minimize power consumption, the current across the electrode should be minimized. However, the current which can be delivered by the electrode is restricted by the supply voltage, so that the latter becomes an important consideration.

In a previous study [4] using finite element analysis we found that the power required to drive the electrode depended on the electrode's surface area. Increasing the surface area of the electrode will greatly reduce its impedance; however, this is counteracted by an increase in the electrode current required to activate neurons [5, 6].

In this paper we attempt to quantify this phenomenon and calculate an optimal surface area for stimulating cortical electrodes. Importantly, we consider that neurons and axons close to the electrode may be damaged by electrode insertion, so that those that do survive are located some distance away from the electrode surface due to the formation of a scar tissue of neuroglial cells around the electrode [2, 7]. Using simple electrical engineering concepts, we predict that the thickness of the region of damaged axons and neurons directly determines the optimal size of the electrode in terms of power consumption using a very simple scaling factor.

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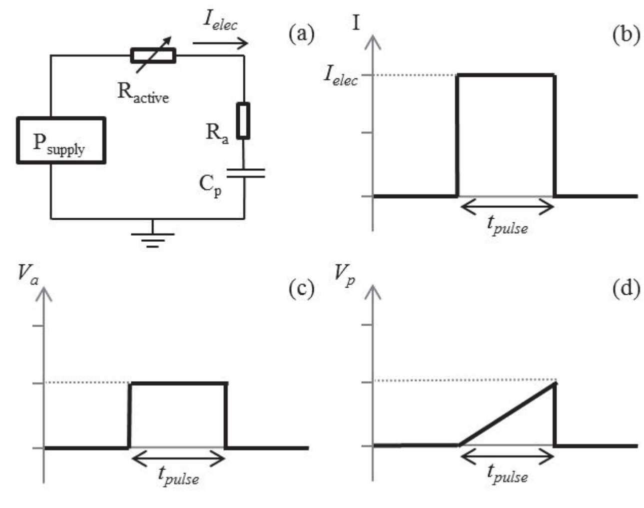

Figure 1. a) Equivalent circuit. The power supply voltage must be adequate to maintain the required current through the electrode, where the electrode is modelled by a capacitor, C_p, and the bulk of the brain by a resistance, R_a. b) Input current pulse where I_{elec} is the current required to activate neurons and t_{pulse} is the pulse length c) The minimum voltage required to maintain the current in (a) through R_a. (d) The minimum voltage required to maintain the current in (a) through C_p.

II. Methods And Results

A simple model of the electrode-electrolyte interface consists of an access resistance, R_a, due to the bulk resistivity of the brain, and a polarization impedance, Z_p, due to the creation of a leaky dielectric layer at the electrode surface. Z_p is usually modeled as a capacitor in parallel with a leakage resistor: here for simplicity we will consider Z_p as a capacitor only, C_p. Fig. 1a shows the simplified circuit which includes the drive circuit and the equivalent electrode circuit.

Fig. 1c shows the voltage across R_a required to maintain I_{elec} through R_a; note that this voltage required is independent of the pulse width, t_{pulse}. Fig 1d shows the voltage across C_p, required to maintain I_{elec} through C_p. Note that this voltage increases with t_{pulse}. A practical point is that to reduce the propensity for tissue damage and electrode corrosion it is important that voltage drop, V_p, across C_p is always within the water window, so that reduction and oxidation of water does not occur [8].

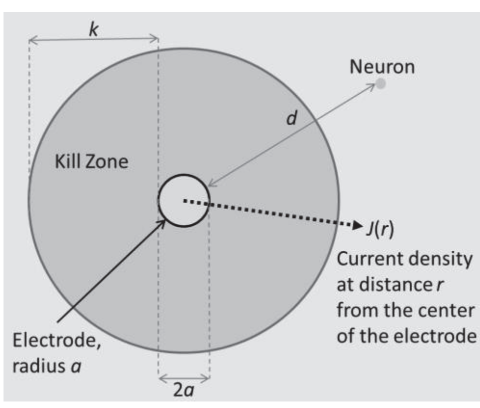

Figure 2. Definition of the problem, when an electrode is implanted into the body, the body mounts an inflammatory response which results in an increased distance between the electrode and viable neurons. The kill zone, k, is the region around the electrode where viable neurons are no longer present as they have died from mechanical trauma and cell death process triggered during implantation of the electrode into the cortical tissue. The surrounding cortical tissue is considered to be isotropic and homogeneous.

We will consider the simple geometry of a smooth spherical electrode (Fig. 2). This allows us to study the effect of surface area on current density distribution in the tissue and the impedance of the electrode using simple analytical approximations. The surrounding brain tissue is also assumed to be an isotropic and homogeneous conductive medium, as has been assumed in other studies [8-10]. The electrode itself is considered to be highly conductive.

Cell excitation occurs when there is enough charge on the membrane capacitance that the voltage drop across the membrane reaches a certain threshold value [11]. Here we will consider the current density at the surface of the cell membrane to be the trigger parameter. The current density, J_{elec}, passing through the surface of a spherical electrode, radius a, in a homogeneous medium due to I_{elec} is

$$J_{elec} = \frac{I_{elec}}{4\pi a^2}. \tag{1}$$

The current density travelling outwards from the electrode within the homogeneous medium at a distance d from the surface of the electrode is

$$J(d) = \frac{I_{elec}}{4\pi (d+a)^2}. \tag{2}$$

Implantation of an electrode will result in the death of cells within some distance from the electrode due to mechanical trauma to these cells, and from the cortical inflammatory response against the foreign material which results in the death of damaged cells [12]. This "foreign body response" also results in the formation of a scar of glial cells around the electrode, which will further distance viable cells from the electrode surface [2, 7]. This will result in an effective kill zone of thickness, k, where the density of viable neurons is greatly reduced around the electrode surface. Thus we are most interested in the current density at distance k from the electrode surface. Equation 2 can be rewritten in order to calculate I_{elec} required to provide a threshold current density, J_{th}, at distance k from the electrode surface

$$I_{elec}(k) = J_{th} 4\pi (a+k)^2. \tag{3}$$

Where, J_{th} is the current density at the cell surface required to evoke an action potential. It is obvious from Equation 3 that in order to minimize I_{elec}, a, should also be minimized. However, the power required for stimulation is also dependent upon the voltage required to ensure that the current is maintained through the electrode, which in part depends upon the electrode impedance.

Under normal operating conditions both polarization impedance and access resistance will be present. Thus the instantaneous power required to maintain the electrode current at the desired value is

$$P_e = P_{acc} + P_{pol}. \tag{4}$$

Where P_{acc} and P_{pol} are the powers required to maintain the desired current over R_a and C_p respectively. Let us first consider the instantaneous power consumption over R_a. For a spherical electrode in a homogeneous medium with bulk resistivity, ρ, R_a is given by

$$R_a = \frac{\rho}{4\pi a} \tag{5}$$

and the required P_{acc} is given by

$$P_{acc} = I_{elec}^2 R_a. \tag{6}$$

Substituting Equations 4 and 5 into 6 we have

$$P_{acc} = 4\pi J_t^2 \rho \left(\frac{(a+k)^4}{a} \right). \tag{7}$$

Now let us consider the instantaneous power consumption over C_p. The capacitance of a spherical electrode with an insulation layer far thinner than its diameter is given by

$$C_p = \frac{4\varepsilon_0 \varepsilon_r \pi a^2}{d_{debye}}. \tag{8}$$

The voltage across this capacitance at time, t, is given by,

$$V_p(t) = \frac{I_{elec} t}{C_p} = \frac{I_{elec} t d_{ebye}}{4\varepsilon_0 \varepsilon_r \pi a^2}. \tag{9}$$

The power required from the power supply to continue charging the capacitance at time, t, is given by

$$P_{pol}(t) = I_{elec} V_p(t). \tag{10}$$

The maximum instantaneous power occurs when t is equal to t_{pulse} shown in Fig. 1b. From here onwards we will consider minimizing power at $t = t_{pulse}$, because the electrode drive system has to be designed for this situation. This then sets the voltage required at the power supply for the system. Substituting Equations 4 and 9 into equation 10 leads to

$$P_{pol} = \left(J_{th} 4\pi (a+k)^2 \right)^2 \times \left(\frac{t_{pulse} d_{debye}}{4\varepsilon_0 \varepsilon_r \pi a^2} \right). \tag{11}$$

We can combine the results for P_{pol} and P_{acc} into Equation 4 and, the resultant equation can be differentiated with respect to electrode radius to find a minimum where

$$\frac{dP_e}{da} = \frac{d(P_{acc} + P_{pol})}{da} = 0. \tag{12}$$

Solving for when this equation equals zero yields three solutions for the optimal electrode radius,

$$a_{opt} = -k, \\ \pm \frac{\sqrt{4d_{debye}^2 t_{pulse}{}^2 + 20 d_{debye} \varepsilon_0 \varepsilon_r k \rho t_{pulse} + \varepsilon_0{}^2 \varepsilon_r{}^2 k^2 \rho^2} - 2 d_{debye} t + \varepsilon_0 \varepsilon_r k \rho}{6 \varepsilon_0 \varepsilon_r \rho}. \tag{13}$$

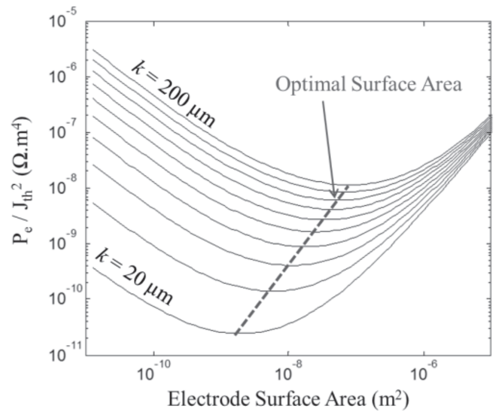

Figure 3. Scaled power required to maintain the threshold current density at the neuron's surface versus the electrode's surface area for increasing kill zones thicknesses k = 20, 40, 60, 80, 100, 120, 140, 160, 180, 200 µm.

Only the positive solution is physical and a minimum. This solution can be used to find the optimum electrode surface area, A_{opt}, where

$$A_{opt} = 4\pi a_{opt}^2 \tag{14}$$

$$A_{opt} = \\ 4\pi \left(\frac{\sqrt{\left(4d_{debye}^2 t_{pulse}{}^2 + 20 d_{debye} \varepsilon_0 \varepsilon_r k \rho t_{pulse} + \varepsilon_0{}^2 \varepsilon_r{}^2 k^2 \rho^2\right)} - 2 d_{debye} t + \varepsilon_0 \varepsilon_r k \rho}{6 \varepsilon_0 \varepsilon_r \rho} \right)^2. \tag{15}$$

Fig. 3 shows the scaled instantaneous power when t_{pulse} = 100 µs, against electrode surface area for different kill zone thicknesses, when the values from Table 1 are substituted into Equation 16. It can be seen that the optimal electrode surface area increases as the kill zone increases.

III. Discussion

A. Choosing Electrode Surface Area

We are currently developing a visual prosthesis with electrodes that penetrate into the visual cortex. In order to maintain a physiological barrier to infection, wireless transmission of power and data is to be used, where the size of the transmission coil is limited by the size of the tile. This greatly restricts the power which can be transferred across the coil. An electrode must therefore be chosen to minimize the power supply to drive all of the electronics.

Studies have shown that increasing electrode surface area minimizes impedance [13], and increases the current required to activate neurons [5, 6]. To the best of our knowledge, this is the first paper which quantifies these two findings to find

TABLE I. Definition of Constants

Constant	***Value***	***Definition***
ρ	10 Ω.m	Brain resistivity
ε_0	8.85 × 10^{-12}	Permittivity of vacuum
ε_r	81	Relative permittivity of Debye layer
d_{debye}	1 nm	Width of Debye layer

an optimum surface area where the power required from the supply to drive the electrode during stimulation, IV, is a minimum. We have studied this issue in the context of a smooth spherical electrode. We recognize that this electrode geometry is not practical for an implantable stimulator, however it provides a good basis for modeling and the results can be transferred to more complex electrode geometries using numerical methods, such as finite element analysis.

We have developed an equation which can be used to calculate the optimal electrode surface area to minimize the device's power consumption. For example, typical values of t_{pulse} lie in the range of 100 to 500 μs and from our data [14] the width of k is approximately 150 μm. If we take t_{pulse} = 100 μs and k = 150 μm and substitute these values into Equation 27 we find an a_{opt} of about 63 μm, corresponding to an A_{opt} of about 49 700 μm^2. This surface area is significantly larger than most current electrode designs which have surface areas less than 10 000 μm^2 [8]. Thus, having a larger electrode surface area than what is currently used for penetrating neural prosthesis may improve power efficiency; however this might be at the cost of specificity. In terms of a visual prosthesis, this may greatly reduce the resolution of the device and a smaller electrode may still be preferred.

Here we have examined how to minimize power consumption in the context of a cortical (visual) prosthesis where the target tissue will be separated from the electrode surface due to the death of neurons surrounding the electrode upon insertion, and by the later formation of a glial scar. However, this procedure could also be applied to any electrode where the target tissue is located a distance away from the electrode surface, for example if surface electrodes were used to target the visual cortex, the target tissue, cortical layer 4cb [15], would be located approximately 1.5 mm from where the electrodes would be positioned.

B. Kill Zone

The results of this study emphasize the importance of continuing research into reducing the kill zone around the electrode. While altering the electrode's surface area can reduce power consumption, reducing the electrode-to-neuron distance would dramatically improve power consumption of the device, regardless of electrode geometry. This is shown in Fig. 3. As biocompatibility of electrodes is improved, and kill zones reduced, the optimal electrode size to minimize power consumption will become smaller and smaller. Ultimately, the constraints in obtaining optimal electrode size may lie in manufacturing limitations as it may be too small to manufacture, and/or may become deformed during insertion.

C. Region of Tissue Activated

Reducing the surface area of the electrode will also reduce the column of tissue that is activated at threshold. This will have the benefit that the electrode will become more specific; however, it will also reduce the chance that enough neurons will be located in this region of tissue in order for stimulation to cause phosphene perception. Here we only considered the current density at one point a distance away from the electrode surface. It might be more accurate to consider a column of tissue to be activated. This has been examined in the context of deep brain stimulation by Butson and McIntyre [16] using finite element analysis. This would certainly be the case if a large number of neurons are needed to work together for phosphene perception.

D. Limitations

In this study a smooth spherical electrode was used to determine the optimal electrode surface area. While the current distributed at a reasonable distance from an electrode is substantially independent of the shape of the electrode [13] the spherical model is unable to predict the current density distribution of other geometries in the volume close to the electrode surface. The polarization impedance is also extremely susceptible to specific electrode geometry and is greatly influenced by current density on the electrode's surface [17]. An electrode with a rough surface might have a larger capacitance, so that the voltage across the capacitance would decrease. We have also approximated brain tissue as being isotropic and homogeneous; this has been shown to greatly influence predictions of neural activation.

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