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Journal of Neuroscience Methods 150 (2006) 212–227

JOURNAL OF NEUROSCIENCE METHODS

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Matching geometry and stimulation parameters of electrodes for deep brain stimulation experiments—Numerical considerations

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Received 16 March 2005; received in revised form 17 June 2005; accepted 20 June 2005

Abstract

Deep brain stimulation, the electric stimulation of basal ganglia nuclei, is a treatment for movement disorders such as Parkinson's disease. The underlying mechanisms are studied in animals, e.g. rodents. Designs and materials of commercially available microelectrodes, as well as experimentally applied driving signals vary tremendously. We used finite integration modeling to compare the electric field and current density distributions induced by various electrodes. Current density or field strength "hot spots", which are located particularly at sites of high curvature and material interfaces coincided with corrosion and erosion at poles and insulation, respectively, as shown by scanning electron microscopy of stainless steel electrodes. Cell constants, i.e. geometry factors relating the electrode impedance to the specific medium conductivity, were calculated to determine the electrode voltage for a given stimulation current. Nevertheless, for electrodes of the same cell constant but of different geometry, current and field distributions may be very dissimilar. We found geometry-dependent limiting values of the stimulation current, above which electric tissue damage may occur. These values limit the reach of the stimulation signal for a given electrode geometry. Also, electrode geometries determine the shape of the stimulated tissue volume. This study provides tools for choosing the most appropriate geometry for targeting different-sized brain areas.

Keywords: Microelectrodes; High-frequency stimulation; Neuromodulation; Rat; Finite integration technique; Field distribution; Electrode impedance

1. Introduction

Before deep brain stimulation (DBS) became a therapy in itself, electric stimulation of basal ganglia had been used to guide neurosurgeons to the precise position for a surgical lesion, the ultimate therapy of late-stage Parkinson's disease (PD). Nowadays, DBS is well established as a symptomatic treatment for PD and other movement disorders. The main advantage of DBS over surgical lesions are its ability to modulate stimulation parameters and its reversibility (Benabid et al., 1987).

Different hypotheses exist to explain the mechanism of DBS. Both neuronal excitation and inhibition are being discussed (Vitek, 2002; McIntyre et al., 2004). It is still unclear which parameter determines whether neurons are artificially stimulated; the induced transmembrane potential, which is proportional to the tissue field strength, or the activating function (Rattay et al., 2003). Roughly speaking, the activating function considers sites in the vicinity of neurons or axons to be especially vulnerable to artificial stimulation when they are sources or sinks of field, i.e. the sites of charge induction. In a homogeneous external field, such sites will be represented by axon endings or bends. In an inhomogeneous field it will be, e.g. the axonal spots exposed to the strongest field inhomogeneity (Rattay et al., 2003). Considering only the field inhomogeneity but not the tissue and cell structures for the activating function both dependencies will result in certain spatial distributions of the efficiency of neural excitation. For a review of the current knowledge on medical

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0165-0270/$ – see front matter © 2005 Elsevier B.V. All rights reserved.
doi:10.1016/j.jneumeth.2005.06.013

DBS, please see Horch and Dhillon (2002), Kuncel and Grill (2004).

Unlike other therapies, DBS was not preceded by extensive work in animal models. This might be a reason for a lack of standardization in stimulation protocols and equipment in animal models such as rodents. In addition, no data has so far been published that describes the commercially available electrodes for rodent models, their properties and the field they induce in the target tissue—the central nervous system (CNS). As electrodes of different geometries are available, it is difficult for researchers to choose an appropriate electrode without having detailed information, e.g. on the size of the maximal applicable current, the stimulated tissue region, etc.

While DBS has been continuously applied in patients without signs of tissue damage for years (Haberler et al., 2000; Henderson et al., 2002; Moss et al., 2004), tissue damage has been observed following long-term stimulation in animals (Darbaky et al., 2003). This might be the reason why physiological data based on animal models are usually restricted to short-term stimulation. However, a recent study showed that tissue damage is also relevant in short-term stimulation (Harnack et al., 2004). These findings in animals may have a number of reasons, such as the use of less inert electrode materials, e.g. non-noble metals, the smaller electrode size and blunt edges (higher curvatures) resulting in higher local current densities leading to more intense local electrode reactions. Electrode reactions produce potentially toxic products like denatured proteins, gas, dissolved metal ions and erosion-products of the insulating materials, etc. These reactions are accompanied by so-called "overpotentials" corresponding to the energy that is dissipated in the electrode reactions and thus lost for the electric stimulation of the tissue. We have discussed the consequences elsewhere (Gimsa et al., 2005). Moreover, tissue damage may also be directly induced by electric cell membrane poration at sites of high electric field strength over a very short time (Suzuki et al., 1998). Potential sites are field hot spots near the electrode surface. The field strength at these hot spots thus imposes a criterion for the maximum current or voltage that can be applied to an electrode of given geometry.

The classical way of describing the field distribution induced by electrodes of complex geometries is to directly measure potentials on enlarged electrode models suspended in a water-filled trough; see Gimsa et al. (1988) for example. Today, numerical techniques are state-of-the-art. They allow for a fast and very accurate solution of Maxwell's equations even for very complex geometries. Maxwell's equations describe the properties of electromagnetic fields by a system of coupled partial differential equations for the electric and magnetic field quantities. There are different approaches for the numerical solution of electromagnetic field problems, e.g. Boundary Element Methods, Finite Elements Methods (FEM) or Finite Difference Methods. Each of these methods has its pros and cons. The Finite Integration Technique (FIT) (Weiland, 1977; Schuhmann et al., 1996) is used throughout this manuscript. Like FEM, FIT is a volume-oriented method, i.e. the whole space considered in the computation is filled by small finite volumes of tetrahedrons or cuboids. A special feature of FIT is that it consistently transfers Maxwell's equations into linear operator equations on the grid. Here, consistency means that all vector-analytical and physical properties of the fields still hold on the grid (see Appendix A). Thus, energy conservation and most of the other properties are correctly reflected by the discrete solution. For details, see Appendix A.

For fully three-dimensional electrode models, FIT allows for describing the spatial potential, current density, field distributions, etc. Our calculations have been aimed at describing commercially available concentric microelectrodes of different geometries which are or can be used in animal DBS experiments, especially the stimulation in rats. Our intentions were:

1. to describe the inhomogeneous current density distributions at the electrode surfaces causing a variation in the intensity of the electrode reactions and to localize probable hot spots of metal corrosion and erosion of the insulating parts;
2. to calculate field hot spots as the potential sites of electrically induced tissue damage by membrane poration;
3. to calculate the maximally applicable current (or voltage) still avoiding membrane poration in dependence on the medium conductivity;
4. to calculate cell constants which are the geometry factors relating the impedance of the electrodes to the specific conductivity of the surrounding medium;
5. to describe the spatial distribution of the efficiency of neural excitation by the spatial distributions of the electric field strength and the activating function around the various electrodes.

This manuscript is a continuation of our study on microelectrodes for animal DBS experiments, demonstrating their electrochemical particularities (Gimsa et al., 2005). Our current analysis provides information beyond that disclosed by the electrode manufacturers. This information is indispensable for researchers who want to choose the optimal electrode geometry for specific CNS targets.

2. Materials and methods

2.1. Electrode properties

In the following, we consider concentric bipolar microelectrodes of different geometries as summarized in Table 1.

An ideal electrode behavior (no potential drop at the metal/medium interface by electrochemical electrode processes) and a constant potential at each site of a metallic surface were assumed for calculations.