Impedance as guidance for electrode placement in intraoperative monitoring of nerve fibers

L Díaz Rodríguez¹, M Varga², J K Wolter², U Pliquett¹

¹ Institute for Bioprocessing and Analytical Measurement Techniques, Heilbad Heiligenstadt, Germany
² Department of Electrical Engineering and Information Technology, Technical University Dresden, Germany

E-mail: laura.diaz@iba-heiligenstadt.de

Abstract. Electrodes for intraoperative monitoring should be reliable and should not disturb the surgeon. Since any wiring could complicate the handling of other medical instruments, new developments of autonomic, wireless electrodes are preferred. This new generation of electrodes will not only stimulate the nerve, but will monitor the action potentials as well. A failure to read nerve potentials may be indicative not just of damaged nerves, but may also result from bad electrode-nerve contact. To resolve this, we have developed electrodes which are equipped with impedance measurement features, facilitating simple connectivity checks. Due to energy constraints, the system works in the time domain using a rectangular voltage wave excitation. The voltage at the electrode is sampled and transmitted via RF link to the host computer. The frequency range covered by the excitation and sampling is between 100 Hz and 50 kHz, which is sufficient not only for detecting contact failure, but also for detecting thick layers of connective tissue between the electrode and the nerve fibre. In either instance, the surgeon can be warned of a bad electrode placement.

1. Introduction

Intraoperative monitoring of nervous tissue or neuromonitoring is a well-established technique used in many neurosurgical interventions which might compromise the proper function of neural structures. Usually, the electrophysiological responses of nerves, namely action potentials, are aroused by a stimulation current, which is directly applied onto the surface of a nerve. Subsequently, the electrical response of the nervous tissue is recorded and interpreted. This kind of examination often ensures integrity of neural tissue, hence reducing the risk of post-operative deficits.

Neuromonitoring has also been used as a help to localize special structures in the brain, as guidance for electrode placement in Deep Brain Stimulation and for making lesions in specific structures in the treatment of movement disorders and pain therapy [1]. In this regard, neuromonitoring has become an important tool, which assists the neurosurgeon by detecting and alerting from probable nerve damage, thus providing the possibility of performing the necessary corrective measures immediately.

In order to ensure the success of neuromonitoring, the contact between nerve and electrode should also be guaranteed. The loss of action potentials or the inability to stimulate a nerve might not always be caused by nerve injury but also can be caused by a bad electrode-nerve-contact or by the misplacement of the electrode. Both failures can be discarded by using impedance spectroscopy as an additional security measure.
In general, different kinds of tissues possess different electrical properties. The differences are determined by the composition of the tissue, i.e. water content, proteins or lipid content. Specifically blood and brain are good conductors while skin, fat and bones are relatively poor conductors [2]. Although nerve tissue is a good conductor, the isolating property of the myelin sheaths increases the total impedance of myelinated nerves [3]. Keeping in mind these particularities and using impedance spectroscopy different kind of tissues might be distinguished from each other.

In this work we present a novel wireless organic electrode equipped with an impedance measurement system that might be useful for further assistance of electrophysiologists and neurosurgeons in the task of placing or implanting electrodes.

2. Materials and Methods

2.1. Carbon Electrodes
The electrodes employed in our measurements were fabricated and provided by the Institute for electronic packaging at the Technical University of Dresden. One of the pursued goals was to develop non-metal electrodes, which would improve biocompatibility and would enhance electrode-nerve contact. The first electrode-prototype was fabricated with a conductive carbon paste (SD 2843 HAL, black, matt ordered from Lackwerke Peters GmbH + Co KG), which was screen printed on a paper substrate [4]. The total electrode resistance was required to be as low as possible to ensure charge transfer to the nervous tissue and therefore to ensure successful stimulation. Using the conductive carbon paste we measured an electrode resistance between $360 \Omega - 450 \Omega$ before and after gas sterilization [4].

2.2. Wireless transmission of power and data
Power and data were delivered using an inductive link between coupled coils driven at 13.56 MHz. The design and optimization of the high-efficiency power amplifier and the power link are described in previous work of the authors [5], where energy transmission is achieved at a distance between coils of 15 cm. For this application, the relative large distance of transmission was required to interfere as less as possible with the maneuver-space of the surgeon during the intervention.

Data was transmitted to the electrode using the amplitude shift keying (ASK) modulation technique. Subsequently the received data was decoded in commands to setup the output signal. Once the excitation signal is enabled, the voltage drop over the tissue is sampled and digitalized to be sent back to the host computer, where further calculations can be realized.

2.3. Time domain impedance measurement
Since the designed electrodes are meant to be used for stimulation of nerve tissue, most of the energy received is going to be required for it. Due to this energy constraint the time domain approach was adopted as the lowest energy consumption alternative. Nevertheless, time domain impedance measurement offers the advantage of a fast measurement making use of a single square signal as excitation wave which is a broad-bandwidth signal. Taking advantage of this fact and using the fast Fourier transformation (FFT) it is possible to obtain amplitude and phase information over several decades of frequency in a short time.

2.3.1. Impedance measurement circuit. A two electrode configuration was used in this work. The square wave excitation signal is generated by internal circuitry and is then applied to the tissue by means of a differential output amplifier as shown in Figure 1 (b). The simplicity of the impedance measurement circuit was a deciding factor for future work in the integration into an ASIC.
In the circuit shown in Figure 1(b) the tissue impedance $Z$ is connected in series with the known impedance $Z_0$, which mimics the plain electrode and is composed by resistors and capacitors. Using Ohm’s law it can be shown that the voltage $V_x$ equals:

$$V_x = V_{out} \frac{Z_0 - Z}{Z_0 + Z} \quad (1)$$

From Equation 1 it can be observed that, when $Z = Z_0$, voltage $V_x = 0$, when $Z \rightarrow \infty$, $V_x = -V_{out}$. In the same way, when $Z_0 \rightarrow \infty$, $V_x = V_{out}$. Rearranging Equation 1 we can calculate the tissue impedance $Z$:

$$Z = Z_0 \frac{V_{out} - V_x}{V_{out} + V_x} \quad (2)$$

2.4. Material under test: Nervous and connective tissue
To validate the impedance measurement system, fresh nervous tissue was obtained from the spinal cord of a swine. The external membranes surrounding the spinal cord were peeled off to expose the white matter. Afterwards, the spinal cord was cut in several small pieces and these were conserved between measurements in Ringer’s solution. White matter is rich in myelinated nerve fibers; therefore it is a suitable tissue for the ex-vivo experiments. The connective tissue was obtained from the hypodermis of the swine. The hypodermis is rich in fibroblasts and elastin, feature that mimics the connective tissue around the nerves.

3. Ex-vivo experiments
In order to ensure tissue vitality during the whole experiment, the impedance measurement was realized within 2 hours after excision of spinal cord and hypodermis. The carbon electrodes were coated with a thin sheath of silicon, keeping a non-isolated area for measurement of 2 x 2 mm² (Figure 2(a)). The purpose of the silicon isolation is to imitate implantation conditions, in which the electrode surface is only in contact with nerves of around 1-3 mm total diameter. Figure 2(b) shows a segment of spinal cord on top of the carbon electrode. This arrangement ensures that only a small part of the white matter containing the myelinated nerve fibers gets in contact with the electrode. Since the contact between electrode and tissue is built only by smoothly positioning the electrode on top or below the nerve, neither pressure nor fixation was applied to the spinal cord or to the hypodermis.

**Figure 2** (a) - Carbon electrode coated with silicon, (b) - Swine spinal cord on top of the same silicon-coated carbon electrode.
4. Results and Discussion

All the measurements were done with a 1 kHz square wave. Figure 3 shows the results obtained from eight different measurements made with nerve and connective tissue. These results show the preliminary characterization of the carbon electrode. From Figure 3 (a), it can be observed that the distinction of both tissues is achieved at rather higher frequencies than expected. It can be seen that the clearest difference between tissues impedances is found in the range between 50 kHz and 1 MHz and the error bars show that different tissue can be clearly distinguished.

Figure 3 (b) shows the impedance phase of the measurements. In comparison with the standard error obtained from the impedance absolute value, the phase standard error between measurements is relatively high. An explanation of this phenomenon can be seen in Figure 3(c). Nerve tissue and connective tissue conserve a dispersion trajectory, but the capacitive contribution to the impedance is shifted in frequency. Consequently, the evaluation of multiple independent parameters yields a higher significance of the tissue discrimination.

In conclusion, although more experiments are needed, the developed impedance measurement system is a potential tool, which can be used for monitoring the electrode-nerve contact in neuromonitoring techniques.

![Figure 3](image-url)

**Figure 3** (a) - Impedance measurement magnitude from eight different measurements; (b) - Impedance phase of connective and nerve tissue; (c) - Nyquist plot of the frequency response of both tissues (only four measurements are shown).

References

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