Identification of a First Order Model of Implanted Electrodes on the First SUAW Patient

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Abstract

Impedance evaluation of implanted electrodes is necessary to ensure that the implant can deliver the specified current, and to survey any failure due to the electrodes. To perform the identification in the implanted hardware, a simple, but sufficiently descriptive model, must be used because of the poor excitation input signal and in order to avoid complex identification algorithms. Data recorded on the first implanted SUAW patient give information on the real impedance of both neural and epimysial electrodes and an opportunity to validate simple first order models.

1 Introduction

Measuring the in vivo impedance of implanted electrodes is of prime importance, mainly for three reasons. First, we want to look at the electrical behaviour evolution versus time and stimulation excitation. Secondly, such data in humans give important indications for the next implant generation design. Thirdly, this is an easy way to get information about the integrity of the electrode. This is a key point for implanted devices when dysfunctions occur. It's crucial to characterise as precisely as possible the origin of the problem. The repair surgery can be more accurately planned with a less invasive intervention. The impedance evaluation can be done by measuring both voltage and intensity - even if a current source is used to ensure that the delivered current is within specifications -, and then perform off line computations to fit the data with the model. Fitting algorithms - like Least Mean Square search - work well if there are few parameters in the model and a good excitation signal. For the moment implanted devices deliver quasi square current impulses so that the excitation is poor in a mathematical sense. Complex models need complex input waveforms, stochastic ones preferred. This paper discusses the validity of the model chosen and presents the values of the parameters found for each electrode.

2 Materials and Methods

2.1 Materials

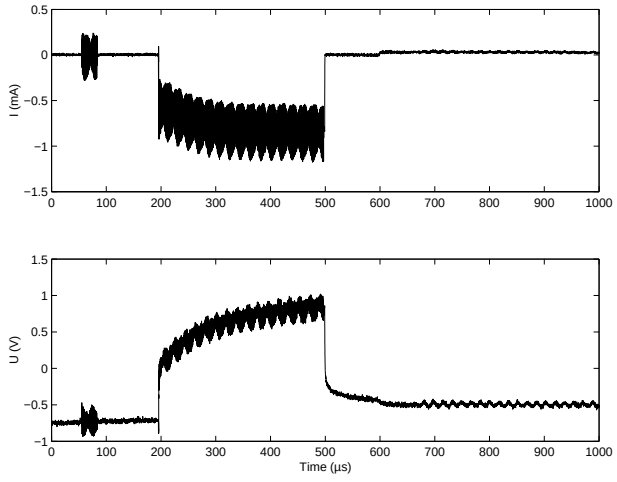

Figure 1: Example of raw data from a neural electrode : right quadriceps

During a surgery performed on the first patient in december 2001, the opportunity was given to do impedance measurements. He was first operated on september 1999 [1] with 4 neural electrodes (Atrotech),

and 8 epimysial electrodes + 2 reference electrodes (IBMT [3]). The measurement was done with a Tektronix TDS3012B oscilloscope with a shunt resistor placed on the anode of 1k for the neural I measurements, and 0.1k for the epimysial ones. Each set of sampled curves is 10000 points long, 9 bits resolution with a 100ns sampling period. The input stimulation signal was delivered by a SUAW implant [2] with the following characteritics : fixed frequency of 25Hz, and fixed pulsewidth of 300μs. Figure 1 shows raw data taken from one neural electrode with the RF signal superimposed. This phenomenon is amplified by the electronic context (long sterile wires between the patient and the oscilloscope). As regards intensity levels, they were chosen to provide a full contraction of muscles. The curves U and I were then stored for off line processing. This protocol was defined in order to avoid a long measurement procedure during the surgery.

2.2 Data Processing

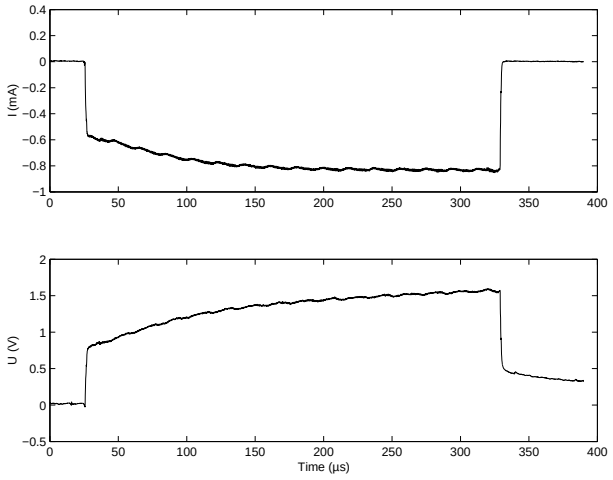

Figure 2: Extracted, normalised and filtered data

Before using the data for the identification of the model, we process them through 3 different blocks. The first performs a time calibration of the whole signal to get only the stimulation pulse with part of the zero current phase (phase during which no current is delivered just before the recovery phase). It is simply based on a voltage threshold discrimination on the U curve (less noisy) : 0.5V for the neural and 2V for the epimysial ones. The second block aims at RF and other fixed frequency signals rejection. Due to aliasing, the frequency produced by the implant transmission does not appear at the right place in the FFT transform, but it is still well identifiable. Two bands were to be eliminated by a comb filter tuned with a zero mean FFT transform. The last block achieved a zero phase second order Butterworth filtering to eliminate residual noise. It was tuned so that its cutoff frequency was far above the estimated cutoff frequency of the whole system i.e. roughly 700kHz.

2.3 Modelisation

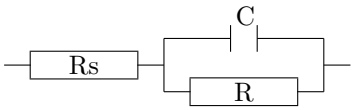

Figure 3: First order model

As we discuss in the introduction, we want to use a simple equivalent circuit to model the general electrical behaviour of the electrodes without detail due to the poor excitation input signal delivered by the implant. A classical Rs serial resistor, and R C parallel cell model was used because the 3 parameters are easy to interpret and to identify. The equivalent impedance can be writen :

$$Ze(p) = \frac{Rs + R + RsRCp}{1 + RCp} \tag{1}$$

Furthermore, if the input is a perfect rectangular current i0, the voltage response is of the form :

$$u(t) = i0[Rs + R(1 - \exp\frac{-t}{RC})] \tag{2}$$

With t=0s, u(t)=Rs*i(0) gives an easy way to identify Rs. It's the same for R if one considers that at the end of the pulse we get roughly (Rs+R)*i(300μs), given that 3*RC is lower than 300μs (error less than 5%). Nevertheless, a correction can be done on the value of R after the evaluation of the time constant. For the computation of C, a method based on the starting slopes of U and I can be used. From equation 1 we can compute the time constant RC and then C, extracted from the differential equation at t=0s :

$$C = \frac{u(0) - (R + Rs)i(0)}{R * [Rs * di/dt(0) - du/dt(0)]} \tag{3}$$

With this method Rs and the time constant RC can be evaluated using two times equations 2 and 3. A Least Mean Square algorithm was also used to fit the actual curve with the model so that both methods could be compared. All the algorithms were implemented with Matlab software.

3 Results and Discussion

The previously described data processing was applied to the 12 implanted electrodes. The residual error between models and actual data were not significant (figure 4 shows an error of less than 50mV compared to 1.5V of the U curve and mainly due to the residual noise) so that it's worth plotting both curves. A first order model cannot distinguish be-

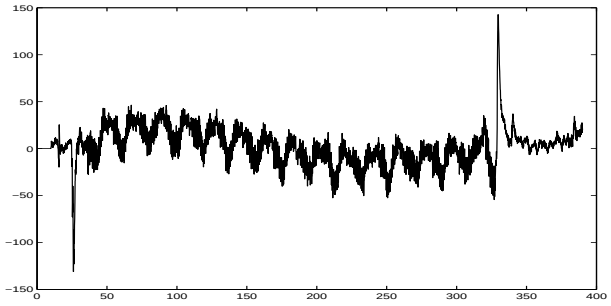

Figure 4: Error between recorded data and the modelised ones

tween the electrode model and the tissue model, but with a simple rectangular input signal, the results obtained on a second order model were not consistent. LMS algorithm gives more precise values reported in tables 1 and 2. Nethertheless, the method based on direct computation of Rs, R, and C gives closed results. The deviation is less than 5% on the resistors and the time constant value, inducing a deviation of less than 10% on the value of C. After two years of

Electrode	Rs Ω	R Ω	C nF
L. SPE	1210	552	259
L. Quad.	1110	608	228
R. SPE	1220	542	207
R. Quad.	1350	657	238

Table 1: Model parameters for neural electrodes

use, the impedances remain homogenous within electrode types. Thus, the differences among stimulation thresholds, and stimulation efficiencies cannot be explained by differences of the electrical behaviour. These results confirm the importance of the search of the motor point for the epimysial electrodes and the placement of the neural electrodes to obtain a good response regardless of the electrical behaviour. The values are all within the SUAW implant specifications that were based mainly on literature and thus confirm the general results obtained by other teams. As

Electrode	Rs Ω	R Ω	C nF
L. Ischio	491	108	121
L. Illiac	512	122	80
L. Glu. Max	538	122	76
L. Glu. Med	596	131	112
R. Ischio	488	108	113
R. Illiac	522	98	97
R. Glu. Max	554	117	103
R. Glu. Med	528	117	73

Table 2: Model parameters for epimysial electrodes

regards the comparison of both methods of identification, the differences remain in a sufficient interval (less than 10% deviation) allowing to use both methods for failure diagnosis. The advantage of the second method based on a direct computation of parameters, is that it can be easily implemented on the hardware of an implant for in line diagnosis avoiding complex algorithm implementation and large amounts of data transfer outside the body. The continuing work will focus on the implementation of impedance evaluation on hardware using the second method for future use on implanted stimulator devices.

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