A Model for Human Skin Impedance During Surface Functional Neuromuscular Stimulation

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Abstract—A new mathematical model for the bulk electrical impedance of human skin is presented. In particular this model describes the impedance of skin during surface functional neuromuscular stimulation (FNS) with square stimulation pulses. Experimental data are presented that illustrate the nonlinear dynamic properties of human skin during current and voltage controlled stimulation. Model predictions are compared to experimental data, measured under both constant voltage and constant current transcutaneous stimulation. It is found that this model captures a variety of nonlinear time-varying effects observed in the skin impedance when stimulating with either protocol. This model may be used as part of large neuromusculoskeletal models or in the more accurate modeling of transcutaneous FNS, which is currently the most common clinical implementation of FNS.

Index Terms—Impedance, modeling, neuromuscular stimulation, nonlinear systems, skin.

I. INTRODUCTION

FUNCTIONAL neuromuscular stimulation (FNS) has been employed widely to artificially activate skeletal muscles and so partially return motor function to the spinal cord injured (SCI) [1], [11], [28]. Prosthetic and orthotic devices using FNS have been developed to restore a SCI individual’s ability to stand, walk and grasp [1], [11]–[13], [28]. A variety of activation methodologies may be applied in situ to initiate neural action potentials and so provide the required force generation capability for these devices [14]. In a rehabilitation setting, rectangular constant current or voltage electrical pulses are typically applied transcutaneously to activate motor neurones and so initiate muscular contractions [1]. It has recently been shown that square subthreshold depolarising prepulses may be useful in closing the slow h gates in nerves to elevate the activation threshold of nerve fibers [7]. This may provide an effective strategy within FNS to disable superficial pain fibers [16]. A series capacitance is found to be negligible and the resistance \( R_s \) primarily attempts to model the resistance of deep tissues, while the parallel combination of capacitor \( C_p \) and linear resistor \( R_p \) is intended to model the skin impedance [2], [16], [31]. A series capacitance \( C_{pol} \) and resistance \( R_{pol} \) modeling the electrode-skin interface may also be included. However, for square dc stimulation pulses the capacitance \( C_{pol} \) is found to be negligible and the resistance \( R_{pol} \) is included in \( R_s \) [2]. The structure of Fig. 1(a) therefore reduces to the electrical equivalent circuit of Fig. 1(b). A number of simple methodologies for identifying the values of the elements in Fig. 1(b) have been proposed and validated [16]. As a result the qualitative properties of each component in this linear system model have previously been identified [2], [16], [31]. However, it has long been known that the impedance of skin in response to dc current pulses displays a variety of complex nonlinear behaviors which this linear model is unable to replicate [2], [16].
This paper aims to present a new model for human skin that captures the system dynamics observed experimentally under both constant current and constant voltage stimulation with typical FNS pulses.

II. METHODS
All experiments were performed using a custom designed computer controlled electrical stimulation unit and a high-speed PC based data acquisition system [20]. The stimulator used was capable of providing voltage controlled pulses in the range from 0 to 170 V and current controlled pulses in the range from 0 to 70 mA, with the ability to vary the amplitude, pulse width (PW) and interpulse interval (IPI) of each pulse as desired. This stimulator was controlled by custom software via a high speed real-time interface card. For the experiments presented here the stimulator was made monophasic by removing the charge balancing network. The voltage applied across the stimulation electrodes was captured using a high-speed digital storage oscilloscope connected to the PC, while the current flowing was similarly measured using a 1-Ω precision resistor in series with the stimulator leads. These signals were periodically downloaded to the computer for off-line analysis. All results presented here were obtained with self-adhesive surface gel electrodes (Stimtrode Model ST75D) of dimensions 4.5 cm × 2 cm as it’s a typical electrode size for transcutaneous FNS.

The experiments presented for intact human skin were performed over the quadriceps muscle group of four able-bodied volunteers over the course of 30–40 experimental trials. In order to guarantee uniform electrode-skin contact the site under the electrode was shaved. No other chemical or cosmetic preparation of the electrode site was performed. To ensure that the nonlinear behaviors observed were truly independent of the current direction experiments were repeated with current flowing in the opposite direction. Experiments were also performed using two other types of electrodes from other manufacturers. In both cases no significant differences were found in the dynamics observed. In the experiments performed to investigate the effect of removing the upper layer of dead skin, known as the stratum corneum, the sites under the electrodes were lightly abraded with emery paper.

III. MODEL DEVELOPMENT
A. Properties of Skin
While inadequate, the simple electrical circuit in Fig. 1(b) proves a useful starting point for the analysis of the skin’s response to an electrical stimulus. Fig. 2 shows the current flowing through the complete system when an 80-V voltage controlled stimulation pulse of PW 6.8 ms is applied. The resistance $R_s$ in Fig. 1(b) can be identified during constant voltage stimulation by dividing the applied voltage by the peak current flowing, as the capacitor is momentarily shorting the resistor $R_p$. Similarly, at the end of a stimulation pulse of sufficient width the same calculation yields a value for $R_s + R_p$, when the capacitor is fully charged. This in turn permits the estimation of $R_p$. Previous experimental evidence has shown that for untreated skin the value of $R_s$, which models resistance of the body core (and to a lesser extent elements of the stratum corneum and lower epidermis) is independent of the stimulation voltage, while there is an inverse relation between $R_p$ and the stimulation voltage [2], [21], [22]. As may be seen in Fig. 2 there is a renewed increase in the current flowing after the initial drop. Such a late increase is in agreement with the phenomena observed in previous qualitative studies [10], [22]. Van Boxtel [2] also found a similar depression in the current through the skin after the initial capacitive spike before it returned to the steady
state value. At the end of the voltage pulse there is a quasi-
steady-state relationship between the voltage applied and the
current flowing as shown in Fig. 2. The term “quasi” is used
as the current is still increasing at a very low rate at the end
of the pulse. It is also worth noting that the pulse widths
considered here are considerably larger than those typically
employed during FNS. Therefore a model capable of capturing
the dynamics over this longer time scale will certainly capture
the dynamics during the shorter more typical FNS stimulation
pulses.

Removal of the stratum corneum is shown to reduce the
value of $R_s$, while almost eliminating $R_p$ and the system’s
nonlinearities completely [2]. This demonstrates that in intact
skin the voltage or current dependent impedance of intact
skin modeled by $R_s$ resides almost exclusively in the stratum
corneum. In particular it has been shown that ionic conduction
within the stratum corneum is the main cause of the skin’s
nonlinear impedance properties [31], [32], and this premise
forms the basis for the model proposed here.

B. Model Proposed

Fig. 3 shows the skin’s response to a burst of six shorter
pulses, each of PW 1ms. These experimental results would
appear to indicate the presence of some sort of memory
element within the impedance considered. While the response
to a single voltage controlled pulse, as in Fig. 2, may be
modeled by a simple resistor-capacitor network, such as that
in Fig. 1(b), the response presented here to a train of six pulses
cannot. Such a memory effect could be incorporated, in part,
into the model in Fig. 1(b) by introducing an inductive ele-
ment. Yet, while resistive and capacitive sources abound in the
system under consideration there is no obvious physiological
source for an inductive component in the body. This is an
important consideration if the model is to reflect the physio-
logical reality and be of use (with larger neuromusculoskeletal
models) in the design of optimal FNS stimulation strategies or
pulses for disabling superficial pain fibers. However, models
of ionic conduction through biological membranes, such as
the well-known Hodgkin–Huxley (H–H) nerve model [8], [9],
[30], which contain only resistive and capacitive elements do
exhibit history dependent (i.e., memory) effects.

The system proposed here to model the skin bulk dc
impedance is shown in Fig. 4. It consists of the parallel ar-
rangement of a constant linear capacitor, $C_p$, with a nonlinear
time-dependent resistor $R_{lp}$, which is in series with a constant
linear resistor $R_s$. Modeling the memory component of the
impedance, demonstrated in Fig. 3, is achieved by employing

While such a model is similar to that used to capture the
behavior of a single ionic conduction channel [8] its basic
structure bears a similarity to the model proposed previously
(Fig. 1) [2], [16]. However, unlike the previous models devel-
oped, all the nonlinear characteristics of the impedance within
this model lie in $R_p$. In this way experimental findings that
the system nonlinearities cannot be attributed to $C_p$ or $R_s$ [31]
are incorporated into the proposed model for $R_p$. The precise
equations used to describe $R_p$ are

$$ R_p = \frac{1}{g_p} $$

$$ g_p = \frac{1}{\tau_g} [g_{p\infty} - g_p] + g_x $$

$$ \frac{1}{\tau_g} = \beta g + \alpha (c_1 - \beta g) $$

$$ g_{p\infty} = \frac{g_{p\infty}(\varphi) \cdot g_y}{g_y - g_{p\infty}(\varphi)} $$

$$ g_{p\infty}(\varphi) = \lambda_1 + \lambda_2 \cdot (1 - e^{\lambda_3 \varphi}) $$

$$ g_x = \frac{1}{\tau_x} [g_x(0) + g_x(t) - q] $$

where $g_p$ and $g_s$ are the conductances of $R_p$ and $R_s$, respec-
tively. $g_{p\infty}$ is the quasi-steady-state value of $g_p$ at the end of
the stimulation pulse and $g_{p\infty}(\varphi)$ is the function modeling the
variation of the quasi-steady-state conductance as a function of
the injected current, $i$. The injected charge is $q$, while $g_x$ mod-
els the time and charge dependent variation of the conductance
$g_p$ from the constant value of $g_{p\infty}$. The variation of $g_{p\infty}$ with the
current injected found in experiment is shown in Fig. 5, as is
the function $g_{p\infty}(\varphi)$ identified using the Levenberg–Marquardt
nonlinear least squares method [17]. The quantities $\alpha$ and $\beta$
in (3) determine the nonlinear current dependent dynamics of the conductance $g_p$, and are

$$\alpha_g = c_2 \cdot \left(1 - e^{-v/(v+c_v)}\right)$$

$$\beta_g = \frac{c_4}{e^{c_0(v+c_v+c_0)} + 1}$$

while $v$ is the voltage across the electrodes and $v_p$ models the half cell (or contact) potentials that occur at surface interfaces, $v_0 \in \mathbb{R}$ and $c_i \in \mathbb{R} \forall i$. For constant current stimulation the system model is precisely the same, the only difference being that the current $i$ forms the model input and the voltage $v$ then becomes the quantity calculated.

A mathematical model for the skin impedance during FNS has been presented in (1)–(8). The model structure is a significantly reduced form of the H–H equations, that possesses the desired property of history dependence, which is maintained by the presence of the first-order lag, without the complex structure.

IV. RESULTS

The model equations described in Section III have been implemented using a commercial simulation package and are readily simulated using a third-order Runge-Kutta integration method, which utilizes a second-order method for step size control. The sole inputs to the model system are perfectly rectangular voltage or current controlled stimulation pulses of the same pulse widths and amplitudes used in experiment. The parameters presented in the Appendix were used for all the simulations shown. These intact skin parameters found in Table I were identified using data other than that presented here (but which was recorded during the same experiment) in order that the system be validated against an ‘unseen’ data set. The determination of parameters to describe abraded skin is discussed later.

A. Response to a Constant Voltage Sequence

Fig. 6 illustrates a simulation of the model system carried out to ascertain whether the impedance structure proposed in Section III and illustrated in Fig. 4 is a viable model. The experiments presented in Figs. 2 and 3 were repeated in simulation. Such a test is significant for the model as its primary feature: the time dependent nature of the $R_p$ conductance, is being presented as an alternative approach to describing the skin impedance presented to the stimulator. In comparing the simulation results in Fig. 6 with the experimental results in Figs. 2 and 3 it may be seen that the model appears to capture the dynamics observed experimentally, both qualitatively and quantitatively. These results indicate that the model realizes successfully the ‘memory’ effect evident in the experimental data shown in Fig. 3.

As suggested by previous experimental evidence [2], [16], [31], the initial current spike in the model is a capacitive current as a result of the fast rising edge of the voltage pulse. The subsequent increase in the current after this capacitive spike occurs due to a reduction over time in the value of the parallel resistor $R_p$.

B. Intact Skin

Fig. 7 shows the measured current flowing through the system in response to voltage controlled stimulation pulses applied in 10 V intervals in the range from 10 to 100 V. As can be seen there is a nonlinear increase in both the maximum current amplitude and the dynamics of the system. Such a response is similar to those observed by others experimentally [2], [22]. Shown in Fig. 8 are the responses predicted by the model when using the parameters presented in Table I and voltage stimulation pulses of the same amplitudes and PW as in experiment. In comparing the simulation results of Fig. 8 with the experimental results in Fig. 7 it may be seen that the model successfully predicts both the quantitative and qualitative behavior of the system at different amplitude levels under constant voltage stimulation.
Fig. 7. Experimentally measured current waveforms in response to constant voltage stimulation. The stimulation voltage pulses ranged from 10 to 100 V in 10-V steps. The responses under 50 V may just be seen above the x-axis.

Fig. 8. Simulated current waveforms in response to constant voltage stimulation pulses ranging in amplitude from 10 to 100 V in 10-V steps when using the model proposed with the parameters in Table I.

An important test for this model, which was developed and identified for constant voltage stimulation, is how it performs when its predictions are compared to the system’s responses under constant current stimulation. Fig. 9 shows the system’s voltage responses measured in experiment to current controlled stimulation inputs of 0.14, 0.818, 2.3, 8.09, 14.36, and 18.54 mA. These stimulation current values are shown to illustrate the diverse range of responses observed with constant current stimulation at various current levels. They also serve to illustrate the difference between the system dynamics observed under constant current stimulation and those observed under constant voltage stimulation. Notice in particular the apparent decay in the system voltage at higher currents. This decay has been observed by others [25] and has been identified as a dynamic property of the system. Constant current stimulation pulses were used as inputs to the model, again using the parameters presented in Table I of the Appendix. Fig. 10 shows the responses predicted by the model for this activation.

From Fig. 10, it may be seen that the model appears to capture both the variation of the voltage amplitude with stimulation level and the decay in voltage with time. However, the initial voltage “spike” appears to be larger in simulation than it is in experiment. This may be explained by the fact that perfectly square stimulation pulses, which are not achievable in reality, were used in simulation. The constant current stimulation pulses applied in experiment were slightly rounded, similar to the voltage pulse illustrated in Fig. 2. In an effort to confirm that this is actually the case, similarly slightly rounded current pulses were used as inputs to the model in simulation and the voltage evoked was calculated. The results of these simulations are shown in Fig. 11, from which it may be seen that the model does indeed also capture the variety of effects observed during constant current activation of the system.

C. Abraded Skin

In order to test the physiological predictive nature of the model proposed, the system being considered was physically
altered. The parameters in the model changed to capture these new system dynamics were then investigated. If the model is a reasonable representation of the physiological system one would expect the resulting parameter changes to illustrate in some way the physical changes that occurred in the system.

Similar sites with identical electrode spacing, type, and size as in Section IV-A and IV-B were used. However, in this instance, the skin was also abraded with fine grade emery paper. This was performed to remove the upper layer of skin, the *stratum corneum*. The stimulation voltage levels required to obtain approximately the same current flow at the end of a stimulation pulse through the system as in Fig. 7 were identified and are shown in Fig. 12. Also shown is the current-voltage relationship for intact skin identified previously in the experiment. As can be seen both the impedance and the degree of nonlinearity in the system has been significantly reduced by removing the stratum corneum, as was expected from previous studies [2], [16], [22]. These voltage levels were then used as inputs to the abraded system. The set of current waveforms through the abraded skin in response to these lower stimulation voltages is presented in Fig. 13. From these results it may be seen that by removing the stratum corneum many of the nonlinear dynamics observed in nonabraded skin and shown in Fig. 7 are significantly reduced. In Fig. 14 the results of simulations carried out with the same model structure as previously are presented. The parameter values required to model the system with now abraded skin are presented in Table II of the Appendix. From these parameters it may be seen that reductions in the values of parameters $R_p$, $R_s$, and $C_p$ are required to model the now altered system. In particular, it is noticed that the values of $C_p$ and $R_s$ are reduced to 50 and 25%, respectively, of the values for modeling intact skin. Similarly, $v_r$ is reduced from 42 to 10 V and $R_p$ is effectively reduced to zero. These changed parameters reflect accurately both the changes the system has undergone and are of the same order reported in previous qualitative studies [2], [16], [32].
The control of FNS musculoskeletal systems is of particular interest to those using FNS in a rehabilitation setting. A central element in the design of control strategies is the possession of a good mathematical model for the system to be controlled. A mathematical model for human skin impedance has been developed and presented. This model while conceptually simple is nonlinear and time-varying, as is the real physiological system. The nonlinearities within this system are modeled as primarily occurring due to ionic conduction within the stratum corneum, as has been shown to be the case in experiment [31]. The system equations are compact, robust, numerically stable and easily implemented within most simulation packages. This model may be combined with others [3], [4], [5], [26] to form a more complete model of the neuromusculoskeletal system when activated under transcutaneous FNS. It may also be of use in the design of optimal FNS simulation strategies for the reduction of pain [7], [25], or “intelligent” controllers.

The results presented in Figs. 8–14 demonstrate the ability of this model to capture both qualitatively and quantitatively the variety of nonlinear time-varying effects characteristic of human skin’s electrical impedance. This has been demonstrated under both constant voltage and constant current stimulation conditions so illustrating the ability of this model developed for constant voltage stimulation to also “predict” the effects observed during constant current stimulation.

Both the experimental and simulation results presented also suggest that removal of the stratum corneum greatly reduces the influence of the parallel \( C_p \), \( R_p \) network shown in Fig. 4 on the injected current during voltage stimulation. This is in agreement with the experimental phenomena previously documented by others [2], [16]. It would appear, from the changes in the model parameters required to capture the observed effects, that this is due to a reduction in the value of \( R_p \) which effectively shorts the smaller skin capacitance \( C_p \). Nevertheless, this model, as with any other, is limited by the mechanisms built into it. While this model captures a number of behaviors it may be expanded or extended further. This may be achieved, at the expense of increased complexity, by adding a clearly identifiable electrode model, or by making this essentially unidirectional model bidirectional.

V. DISCUSSION

It is also worth noting that from the experimental results presented that it appears considerable savings could be made both on the technology used to achieve the reduced voltages required and on the energy consumption of a stimulator if the system properties could be changed by a less aggressive and less irritating method other than removal of the stratum corneum. Perhaps this will become possible with a new generation of conductive gels.

APPENDIX I

Presented in Table I are all the parameter values necessary for the simulation of the skin impedance model for intact human skin presented in the text. These parameter values have been determined from experiments over the quadriceps muscle group of able bodied volunteers.

Presented in Table II are the parameter changes necessary for simulation of the skin impedance model for abraded human skin. All other parameters are as in Table I.

REFERENCES


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