Simulation of Intrafascicular and Extraneural Nerve Stimulation

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Abstract—A model of nerve stimulation for control of muscle contraction and ensuing isometrical muscle force has been developed and implemented in a simulation algorithm. A description of nerve fiber excitation was obtained using probability distributions of a number of excitation parameters. The volume conduction model of the stimulated nerve incorporates both inhomogeneities and anisotropy within the nerve. The nerve geometry was taken cylindrically symmetric. The model of the nerve fiber excitation mechanism was based on McNeal, using the Frankenhaeuser-Huxley equations. Simulations showed that the diameter dependency of nerve fiber recruitment is influenced by the electrode geometry.

INTRODUCTION

ARTIFICIAL nerve stimulation is used in implantable systems for functional neuromuscular stimulation. These systems are designed to control the contraction of paralyzed skeletal muscles with the objective to obtain functional limb movements in stroke and spinal cord injured patients.

In the literature, models of nerve stimulation can be found [1]–[3]. They describe the excitation of a myelinated nerve fiber during nerve stimulation supposing all parameters are known. However, in practice, a number of them are unknown. In order to obtain a more accurate description of nerve fiber excitation during nerve stimulation, the probability distributions of these parameters can be incorporated in the model. Furthermore, the model should take into account the electrically inhomogeneous and anisotropic nature of peripheral nerves [4]–[8]. The nerve stimulation model described in this paper incorporates these aspects.

In functional neuromuscular stimulation, the relation between excitation of motoneurons and muscle force is of importance. A linear summation of force contributions of motor units to the isometric muscle twitch appeared to be a reasonable assumption [9]. Furthermore, it was found that motor unit force increases with increasing diameter of the innervating motoneuron [10].

THEORY

Volume Conduction Model of a Stimulated Nerve

A peripheral nerve consists of a number of nerve fiber bundles called fascicles. These are surrounded by con-

nective tissue, epi- and perineurium. Our volume conduction model of a stimulated nerve supposes a cylindrical symmetric nerve geometry with one fascicle (Fig. 1). The medium *i*, within the inner cylinder, represents the fascicle. Medium *o*, in between inner and outer cylinder, represents the connective tissue surrounding the fascicle. Medium *e*, external to the outer cylinder, represents the medium outside the nerve. The radius of the fascicle is r_i , while r_o is the radius of the nerve.

Media *o* and *e* are assumed to be homogeneous, with conductivities σ_o and σ_e , respectively. Medium *i* is supposed to be anisotropic with conductivity σ_{iz} in the *z* direction and σ_{ir} perpendicular to this direction [7], [8].

A cylindrical coordinate system z, r, θ is used. Two cylindrical symmetric stimulation configurations are considered: an intrafascicular point electrode in the origin $r = 0, z = 0$ [Fig. 2(a)] and an extraneural ring electrode $r = r_o, z = 0$ [Fig. 2(b)].

The electrical potential $V(z, r, \theta)$ fulfills the Laplace equation in cylindrical coordinates:

$$\frac{1}{r} \frac{\partial}{\partial r} \left(r \frac{\partial V}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 V}{\partial \theta^2} + \frac{\partial^2 V}{\partial z^2} = 0. \quad (1)$$

The solution of differential equation (1) in the case of a cylindrical symmetric geometry is given by [7], [8]

$$V(z, r) = \frac{1}{2\pi} \int_{-\infty}^{\infty} \left[g_0(k) \cdot I_0(|k| \cdot r) + h_0(k) \cdot K_0(|k| \cdot r) \right] e^{jkz} dk. \quad (2)$$

I_0 and K_0 are modified Bessel functions of the first and second kind, respectively, and of order 0.

Anisotropy within the fascicle can be included by the coordinate transformation [7], [8]

$$\begin{aligned} z_1 &= z \\ r_1 &= \sqrt{\left(\frac{\sigma_{iz}}{\sigma_{ir}} \right)} \cdot r \\ \theta_1 &= \theta. \end{aligned} \quad (3)$$

The functions $g_0(k)$ and $h_0(k)$ can be determined using the boundary conditions at the cylinder surfaces (continuity in the potential and normal current density), for $r = 0$ and $r \rightarrow \infty$, and at the current injection point rep-

Manuscript received April 27, 1987; revised August 10, 1987.
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IEEE Log Number 8718189.

Fig. 1. A two-cylinder model of a nerve. The inner cylinder represents the boundary of the fascicle *i*. In between the inner and outer cylinder, the connective tissue *o* surrounding the fascicle is located. Outside the outer cylinder, the extraneural medium *e* is situated.

Fig. 2. The electrode configurations considered. (a) An intrafascicular point electrode. (b) An extraneural ring electrode.

representing the electrode ($r = 0$ for the intrafascicular point electrode and $r = r_o$ for the extraneural ring electrode) [7], [8]. Thus, the boundary conditions can be written in a matrix form

$$\begin{bmatrix} 0 & A_{12}(k) & 0 & 0 & 0 & 0 \\ A_{21}(k) & A_{22}(k) & A_{23}(k) & A_{24}(k) & 0 & 0 \\ A_{31}(k) & A_{32}(k) & A_{33}(k) & A_{34}(k) & 0 & 0 \\ 0 & 0 & A_{43}(k) & A_{44}(k) & 0 & A_{46}(k) \\ 0 & 0 & 0 & 0 & 1 & 0 \\ 0 & 0 & A_{63}(k) & A_{64}(k) & 0 & A_{66}(k) \end{bmatrix}$$

$$\begin{bmatrix} g_{0i}(k) \\ h_{0i}(k) \\ g_{0o}(k) \\ h_{0o}(k) \\ g_{0e}(k) \\ h_{0e}(k) \end{bmatrix} = \begin{bmatrix} B_1(k) \\ 0 \\ 0 \\ 0 \\ 0 \\ B_6(k) \end{bmatrix} \quad (4)$$

The elements of matrix *A* are given by

$$A_{12}(k) = \sigma_{ir}$$

$$A_{21}(k) = I_0 \left(\sqrt{\left(\frac{\sigma_{iz}}{\sigma_{ir}} \right)} \cdot |k| \cdot r_i \right)$$

$$A_{22}(k) = K_0 \left(\sqrt{\left(\frac{\sigma_{iz}}{\sigma_{ir}} \right)} \cdot |k| \cdot r_i \right)$$

$$A_{23}(k) = -I_0(|k| \cdot r_i)$$

$$A_{24}(k) = -K_0(|k| \cdot r_i)$$

$$A_{31}(k) = \sigma_{ir} \cdot \sqrt{\left(\frac{\sigma_{iz}}{\sigma_{ir}} \right)} \cdot |k| \cdot I_1 \left(\sqrt{\left(\frac{\sigma_{iz}}{\sigma_{ir}} \right)} \cdot |k| \cdot r_i \right)$$

$$A_{32}(k) = -\sigma_{ir} \cdot \sqrt{\left(\frac{\sigma_{iz}}{\sigma_{ir}} \right)} \cdot |k| \cdot K_1 \left(\sqrt{\left(\frac{\sigma_{iz}}{\sigma_{ir}} \right)} \cdot |k| \cdot r_i \right)$$

$$A_{33}(k) = -\sigma_o \cdot |k| \cdot I_1(|k| \cdot r_i)$$

$$A_{34}(k) = \sigma_o \cdot |k| \cdot K_1(|k| \cdot r_i)$$

$$A_{43}(k) = I_0(|k| \cdot r_o)$$

$$A_{44}(k) = K_0(|k| \cdot r_o)$$

$$A_{46}(k) = -K_0(|k| \cdot r_o)$$

$$A_{63}(k) = \sigma_o \cdot |k| \cdot I_1(|k| \cdot r_o)$$

$$A_{64}(k) = -\sigma_o \cdot |k| \cdot K_1(|k| \cdot r_o)$$

$$A_{66}(k) = \sigma_e \cdot |k| \cdot K_1(|k| \cdot r_o). \quad (5)$$

For the intrafascicular point electrode, the elements of vector *B* are given by

$$B_1(k) = \frac{I}{2\pi} \quad \text{and} \quad B_6(k) = 0. \quad (6)$$

And for the extraneural ring electrode,

$$B_1(k) = 0 \quad \text{and} \quad B_6(k) = \frac{I}{2\pi \cdot r_o}, \quad (7)$$

I being the stimulation current. In the stimulation model, cathodic monophasic constant current pulses were simulated, having an amplitude *A* and a constant pulse width of 60 μ s:

$$I(t) = -A \quad 0 \leq t < 60 \mu s \\ = 0 \quad \text{otherwise.} \quad (8)$$

The functions $g_0(k)$ and $h_0(k)$ in media *i*, *o*, and *e* were calculated by inverting matrix (4). The electrical potential at any point was then calculated using (2).

Model of Nerve Fiber Excitation

A model for nerve fiber excitation was adopted from McNeal [1]. This author based his model on the work of Frankenhaeuser and Huxley [11] who described the electrical behavior of a node of Ranvier with a set of four nonlinear differential equations. McNeal represented a myelinated nerve fiber by an electrical network (Fig. 3). The nodes of Ranvier were modeled as lumped electrical components: membrane conductance G_m , nodal capacitance C_m , and the nerve fiber resting potential V_r . Inter-

Fig. 3. Network representation of a myelinated nerve fiber, as introduced by McNeal [1].

nally, the nodes were connected via a conductance G_a . These components depended on the external myelin diameter D , which McNeal assumed to be proportional to the internodal length L . The nerve fiber model consisted of 11 nodes of Ranvier, numbered -5 to 5 . Node 0 , in the current injection plane perpendicular on the nerve fibers, was described by the Frankenhaeuser-Huxley equations [11]. G_m of the other nodes was taken constant.

We used the model of McNeal in unchanged form and with the same parameter values with the following exceptions.

- Since we were interested in the effects of cylindrical inhomogeneities, we situated the nerve fibers in the inner cylinder of the volume conduction model as presented above and not in a homogeneous isotropic medium.
- Node 0 was not necessary in the current injection plane perpendicular to the nerve fibers. We defined node 0 as the node at the shortest distance L_0 from this plane (Fig. 4).
- Besides the membrane conductance of node 0 , nodes -1 and 1 were also described by the Frankenhaeuser-Huxley equations. The other eight nodes were described with constant G_m . Only nodes -1 , 0 , and 1 can have a membrane potential near threshold during threshold stimulation of a nerve fiber when L_0 varies between $-0.5L$ and $0.5L$. Consequently, only these nodes need to be modeled by the Frankenhaeuser-Huxley equations when determining excitation threshold of a nerve fiber [1].

Probability Distributions

In practice, when a nerve is stimulated, the exact position and characteristics of individual nerve fibers are unknown. In our model, we assumed nerve fiber excitation to depend on four parameters: the external myelin diameter D , the distance r to the axis of the fascicle cylinder, distance L_0 from node 0 to the current injection plane perpendicular on the fiber, and the stimulation pulse amplitude A . The internodal length L is assumed to be dependent on D . Because of cylinder symmetry, excitation probability does not depend on θ .

We define a discrete sample space Ω consisting of all the motoneurons in the stimulated nerve. On this sample space, we define four random variables: D , r , L_0 , and A . A is the excitation amplitude, defined as the stimulation pulse amplitude above which a nerve fiber is excited, and below which a nerve fiber is not excited. A is a function of D , r , and L_0 .

Fig. 4. L_0 is the distance from the current injection plane perpendicular to the nerve fibers to the node lying nearest to that plane (node 0).

The probability of nerve fiber excitation $P_{\text{exc}}(A)$ for a stimulation pulse amplitude A equals the probability distribution function $F_A(A)$. Using the probability density functions of D , r , and L_0 , $P_{\text{exc}}(A)$ can be expressed as follows:

$$\begin{aligned}
 P_{\text{exc}}(A) &= F_A(A) \\
 &= \int_0^\infty f_D(D) \cdot \int_0^{r_i} f_{r|D}(r|D) \\
 &\quad \cdot \int_{-L(D)/2}^{L(D)/2} f_{L_0|D,r}(L_0|D, r) \\
 &\quad \cdot F_{A|D,r,L_0}(A|D, r, L_0) \\
 &\quad \cdot dL_0 \cdot dr \cdot dD.
 \end{aligned} \tag{9}$$

$f_D(D)$ is the probability density function of D . It might be obtained from a statistical analysis of nerve fiber diameters in many nerves.

$f_{r|D}(r|D)$ is the probability density function of r for given nerve fiber diameter D . Assuming the geometrical distribution of motoneurons of a certain diameter over the fascicle to be uniform, we can write

$$\begin{aligned}
 f_{r|D}(r|D) &= f_r(r) = \frac{1}{\pi r_i^2} \cdot 2\pi r = \frac{2r}{r_i^2} \quad \text{for } r < r_i \\
 &= 0 \quad \text{for } r \geq r_i.
 \end{aligned} \tag{10}$$

$f_{L_0}(L_0)$ is the probability density function of L_0 . We assume a uniform distribution of L_0 between $-L(D)/2$ and $L(D)/2$, the internodal length L being a function of D :

$$\begin{aligned}
 f_{L_0|D,r}(L_0|D, r) &= f_{L_0|D}(L_0|D) \\
 &= \frac{1}{L(D)} \quad \text{for } \frac{-L(D)}{2} < L_0 < \frac{L(D)}{2} \\
 &= 0 \quad \text{otherwise.}
 \end{aligned} \tag{11}$$

$F_{A|D,r,L_0}(A|D, r, L_0)$ is the probability distribution function of A for given D , r , and L_0 . Because of the assumption that A only depends on D , r , and L_0 , this probability is 0 under a certain threshold $A_t(D, r, L_0)$ and 1 above this threshold.

When integrating over L_0 in (9), it must be noted that

nerve fibers of given D and r are excited for L_0 between $-L_{0r}(A, D, r)$ and $+L_{0r}(A, D, r)$ when stimulating with a pulse amplitude A . This is because node 0 is excited first for cathodical stimulation pulses [2]. $L_{0r}(A, D, r)$ can be determined using the nerve fiber excitation model and the volume conduction model described:

$$F_{A|D,r,L_0}(A|D, r, L_0) = 1, \quad \text{for } -L_{0r}(A, D, r) < L_0 < L_{0r}(A, D, r) \\ = 0 \quad \text{otherwise.} \quad (12)$$

A Model for the Amplitude of Isometric Twitch Force

We define another random variable T , representing the amplitude of the isometric twitch force of the muscle activated by the stimulated nerve. Using (9), the expectation of T for a stimulation pulse amplitude A can be expressed as

$$E[T](A) = \int_0^\infty M(D) \cdot f_D(D) \cdot \int_0^{r_i} f_{r|D}(r|D) \\ \cdot \int_{-L(D)/2}^{L(D)/2} f_{L_0|D,r}(L_0|D, r) \\ \cdot F_{A|D,r,L_0}(A|D, r, L_0) \\ \cdot dL_0 \cdot dr \cdot dD. \quad (13)$$

In (13), only the population of motoneurons within the nerve must be considered. A linear summation of the isometric twitch force of the motor units is assumed. The amplitude of the motor unit twitch force $M(D)$ increases with the diameter D of the motoneuron innervating the motor unit [10].

RESULTS

The nerve stimulation model described was simulated on an HP 9000/300 system programmed in Pascal.

Verification of the Model

The correctness of the computation algorithm of the potential field in the nerve was tested in the situation of all conductivities being the same. The result showed a good correspondence with the analytic solution for the potential field in a homogeneous medium for both electrode configurations considered. Furthermore, continuity in both the potential and the normal current density at the boundary surfaces were checked in a nonhomogeneous and anisotropic case.

The nerve fiber excitation model of McNeal [1] was tested as a special case of our model with a homogeneous isotropic medium. The results corresponded well with those found by McNeal.

Simulation Results

Most of the parameter values of the volume conduction model of the nerve were obtained from the literature. The conductivities $\sigma_{ir} = 0.08 (\Omega \cdot \text{m})^{-1}$ and $\sigma_{iz} = 0.5 (\Omega \cdot \text{m})^{-1}$ were obtained from Geddes and Baker [12].

The low extraneural conductivity $\sigma_e = 0.001 (\Omega \cdot \text{m})^{-1}$ models an almost electrical isolation of the nerve from the surrounding tissue, which corresponds to the use of an isolating cuff around the stimulated nerve. The conductivity of the connective tissue surrounding the nerve fascicle was not found in literature. We assumed $\sigma_o = 0.1 (\Omega \cdot \text{m})^{-1}$, somewhat higher than the conductivity of fat, which is about $0.04 (\Omega \cdot \text{m})^{-1}$, according to Geddes and Baker. The radius of the inner cylinder $r_i = 0.5 \text{ mm}$ is a realistic value in human nerves [13]. The thickness of the extrafascicular connective tissue sheath $r_o - r_i$ can vary much among species, but also in the same nerve within one individual [13]. In the simulations presented in this paper $r_o - r_i$ was taken to be 0.5 mm .

We simulated nerve stimulation and used as a criterion for nerve fiber excitation a membrane potential change of node 0 of more than 50 mV during the time period from the onset of stimulation until $200 \mu\text{s}$ after the stimulation pulse or a positive derivative of the membrane potential at the end of this period. The error in determining L_{0r} (12) due to falsely detecting an action potential in node 0 when there is none, or not detecting one when it is present is less than 1 percent of L .

We define the probability density of excitation as a function of r at a certain pulse amplitude A and for given D by

$$g_{A|D}(r) = f_{r|D}(r|D) \cdot \int_{-L(D)/2}^{L(D)/2} f_{L_0|D,r}(L_0|D, r) \\ \cdot F_{A|D,r,L_0}(A|D, r, L_0) \cdot dL_0. \quad (14)$$

Figs. 5 and 6 show $g_{A|D}(r)$ for the intrafascicular point electrode and the extraneural ring electrode at several values of A (Fig. 5) and D (Fig. 6). The diagonal line represents excitation of all nerve fibers and corresponds to the probability density function of r (10).

The excitation probability distribution $F_{A|D}(A|D)$ as a function of A for given D is presented by

$$F_{A|D}(A|D) = \int_0^{r_i} g_{A|D}(r) \cdot dr. \quad (14)$$

$F_{A|D}(A|D)$ is a recruitment curve for fibers with a diameter D . Fig. 7 shows this function for both electrode configurations.

DISCUSSION

The perturbation of the stimulating field in the presence of the myelinated nerve fibers was modeled by an anisotropic conductance inside the nerve fascicle. Like McNeal [1], we did not include the perturbation of the field by currents due to the generation of action potentials by the fibers. Although this seems a good approximation [7], [14], some influence during activity of many myelinated fibers, synchronized by stimulation, cannot be excluded. Clark and Plonsey [15] investigated this for nonmyelinated fibers.

$$\begin{bmatrix} g_{0i}(k) \end{bmatrix} \quad \begin{bmatrix} B_1(k) \end{bmatrix}$$

$$I(t) = -A \quad 0 \leq t < 50 \mu\text{s} \\ = 0 \quad \text{otherwise.} \quad (8)$$

(a)

(b)

Fig. 5. Excitation probability function $g_{A|D}(r)$ as a function of radius r at several stimulation pulse amplitudes A for a nerve fiber with a diameter of $10 \mu\text{m}$. (a) Intrafascicular point electrode: the stimulation pulse amplitude A increases from 8 to $62 \mu\text{A}$ in steps of $6 \mu\text{A}$. (b) Extraneural ring electrode: the stimulation pulse amplitude A increases from 43 to $63 \mu\text{A}$ in steps of $2 \mu\text{A}$.

We did not take into account the blocking phenomenon caused by hyperpolarization of nodes not lying closest to the electrode [16], [17]. The phenomenon could be incorporated in the model by simulating all nodes with the Frankenhaeuser-Huxley equations and taking propagation of the action potential as the criterion for nerve fiber recruitment. Experimental data suggest the blocking phenomenon to occur at stimulation pulse amplitudes of more than eight times threshold [16]. Considering the amplitude range in which $F_{A|D}(A|D)$ varied between 0 and 1 , no fibers were stimulated at more than eight times threshold for the extraneural electrode [Fig. 7(b)]. For the intrafascicular electrode, the phenomenon could cause an error in $F_{A|D}(A|D)$ of less than 3 percent in this range

(a)

(b)

Fig. 6. Excitation probability function $g_{A|D}(r)$ as a function of radius r at one stimulation pulse amplitude for four nerve fiber diameters. D : 5 , 10 , 15 , and $20 \mu\text{m}$. (a) Intrafascicular point electrode: the stimulation pulse amplitude is $20 \mu\text{A}$. (b) Extraneural ring electrode: the stimulation pulse amplitude is $29 \mu\text{A}$.

for $5 \mu\text{m}$ diameter fibers [Fig. 7(a)]. Above this range, a decreasing recruitment for increasing stimulation pulse amplitude could result.

The introduction of probability distributions made it possible to evaluate the excitation probability of myelinated nerve fibers of given diameter as a function of the position of the fibers in the nerve cross section and the stimulation pulse amplitude. Without the introduction of probability distributions, only thresholds for nerve fiber excitation could be determined for given parameters D , r , and L_0 [1]–[4], [6].

McNeal found a decreasing threshold of excitation for increasing nerve fiber diameter when the position of the nerve fibers was the same and when L_0 was 0 [1]. The model described in this paper gives more information on

Fig. 7. Excitation probability distribution function $F_{A|D}(A|D)$ as a function of stimulation pulse amplitude A for four nerve fiber diameters D : 5, 10, 15, and 20 μm . (a) Intrafascicular point electrode. (b) Extraneural ring electrode. The dashed vertical line marks the stimulation pulse amplitudes used in Fig. 6.

the diameter dependency of nerve fiber excitation. For every nerve fiber diameter, a stimulation pulse amplitude range was found in which recruitment varies between no recruitment and full recruitment (Fig. 7). The overlap of these recruitment ranges for different nerve fiber diameters appeared to depend on the electrode geometry (intrafascicular point electrode or extraneural ring electrode: Fig. 7). However, for both electrode configurations, the minimum stimulation pulse amplitude at which full recruitment occurred appeared to decrease for increasing nerve fiber diameter. Furthermore, the simulations showed differences in the range of radius r for which the excitation probability density function $g_{A|D}(r)$ is non-zero. This range increased for increasing diameter D (Fig. 6). However, when the electrode is close to part of the nerve fibers, as is the case for the intrafascicular point electrode, there is at low stimulation pulse amplitude a

certain range of radius r for which $g_{A|D}(r)$ is highest for thin nerve fibers. The reason is the decreasing internodal length for decreasing nerve fiber diameter which results in a decrease of the average value of L_0 . Because of high gradients in the potential field in the neighborhood of the electrode, the stimulation threshold is low for small L_0 and small distance between nerve fiber and electrode.

We presented preliminary experimental data for verification of the simulation results needed for further application of the model [18]. Experimental data on nerve fiber diameter distribution and the relation between nerve fiber diameter and motor unit force are needed, however, for evaluation of the relation between nerve stimulation and muscle force.

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