Sensitivity and Selectivity of Intraneural Stimulation Using a Silicon Electrode Array

Wim L. C. Rutten, *Member, IEEE*, Harmen J. van Wier, and Johan H. M. Put

Abstract—**Artificial electrical stimulation of peripheral nerves needs the development of multielectrode devices which stimulate individual fibers or small groups in a selective and sensitive way. To this end, a multielectrode array in silicon technology has been developed, as well as experimental paradigms and model calculations for sensitivity and selectivity measures. The array consists of twelve platinum electrode sites (10 × 50 μm at 50 μm interdistance) on a 45 μm thick tip-shaped silicon substrate and a Si_3N_4 insulating glass cover layer. The tip is inserted in the peroneal nerve of the rat during acute experiments to stimulate α motor fibers of the extensor digitorum longus muscle. Sensitivity calculations and experiments show a cubic dependence of the number of stimulated motor units on current amplitude of the stimulatory pulse (recruitment curves), starting at single motor level.**

Selectivity was tested by a method based on the refractory properties of neurons. At the lowest stimulus levels (for one motor unit) selectivity is maximal when two electrodes are separated by 200–250 μm, which was estimated also on theoretical grounds.

The study provides clues for future designs of two- and three-dimensional devices.

Introduction

NEUROMUSCULAR control by artificial electrical stimulation of peripheral nerves for rehabilitation purposes is a long-standing subject of experimental and model study. The experimental studies comprise animal research into such aspects as design of implants, physiological demands for implants, bioacceptance of devices, electrochemical performance characteristics of electrodes, and the efficiency of the stimulus paradigm [1]–[5]. Models calculate and evaluate potential fields around electrodes, excitatory mechanisms, and choice of electrode sites [6]–[8], [15].

With regard to the choice of site and the paradigm of stimulation two approaches can be discerned. One approach has a premise in chosing extraneural electrodes in order to avoid possible intraneural damage [5], [9]–[11]. As the fibers are relatively far from the electrode sites, special paradigms are designed in order to reach sufficient selectivity. The other approach uses intraneural electrodes [12] thereby increasing selectivity of stimulation. Special care has now to be taken to minimize neural damage.

This report is a study on the sensitivity and selectivity of intraneural stimulation. The ideal neural stimulatory information transducer should be able to activate individual or small groups of neural fibers within a fascicular bundle, for example in a motor nerve. This would be of great importance for fine-tuned neural control of paralyzed muscles. Also, it would prevent muscular fatigue by alternate recruitment of groups of motor units.

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The authors are with the Department of Biomedical Engineering, Faculty of Electrical Engineering, Twente University, 7500 A. E. Enschede, The Netherlands.

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Basically, the activation process of a myelinated fibers starts by imposing a sufficiently large voltage gradient over the nodes in a fiber in the axial direction [7]. As the neural tissue is mechanically very flexible and is electrically an inhomogeneous, anisotropic, and capacitive volume conductor with variable internode distances, it is hardly possible to achieve stable, selective stimulation of fibres from outside the nerve trunk. Therefore, a local intraneural approach with sufficient redundancy of the number of electrodes is to be preferred. In a local approach using microelectrodes it is sufficient in principle to apply current pulses to one node in order to reach the stimulatory threshold of the so called activating functions [8].

Several ways may be chosen to realize intraneural excitation of individual fibers. Two of these use microtechnology fabrication techniques of silicon devices to meet the miniaturization demands which the individual nerve fiber approach imposes and to allow future incorporation of integrated electronics for multiplexing the large number of electrodes and still employing the minimum amount of leads to the inplanted device.

In one method the two ends of fibers in a deliberately sectioned bundle are allowed to regrow through metallized holes in a silicon base plate which is positioned perpendicular to the fiber direction [2].

In the second approach presented here, a multielectrode array in silicon technology is being inserted into a fascicle. This report focuses on the sensitivity and selectivity of the array, inserted in a motor nerve of a rat during acute experiments. In first instance, the array is one dimensional with 12 electrode sites. On basis of the performance of this array, two-and three-dimensional devices can be designed optimally.

Another important aspect is the bioacceptance of such a device. Promising preliminary test results have been reported elsewhere [13].

Materials and Methods

The Intraneural Stimulation Device

Fig. 1(a) shows a schematic drawing of the tip of the silicon device, the twelve platinum stimulation sites upon it (10 × 50 μm), and the positioning of the device in a fascicle. The positioning is such that the array direction is perpendicular to the longitudinal fiber direction, with the plane of the stimulation sites parallel to this direction. Not shown in the Si_3N_4 insulation cover layer over the tip (except over the platinum surfaces). The tip length is 840 μm, its thickness is 45 μm. The length dimension has been chosen on basis of the fascicle diameter of about 0.5 to 1.0 mm (in man as well as in rat). The interelectrode distance is 50 μm. How this measure follows from the fiber diameter of about 10 μm [14] and the number of about 350 efferent motoric α fibers in a total population of 1000–2000 fibers in an average fascicle of the peroneus communis nerve of the

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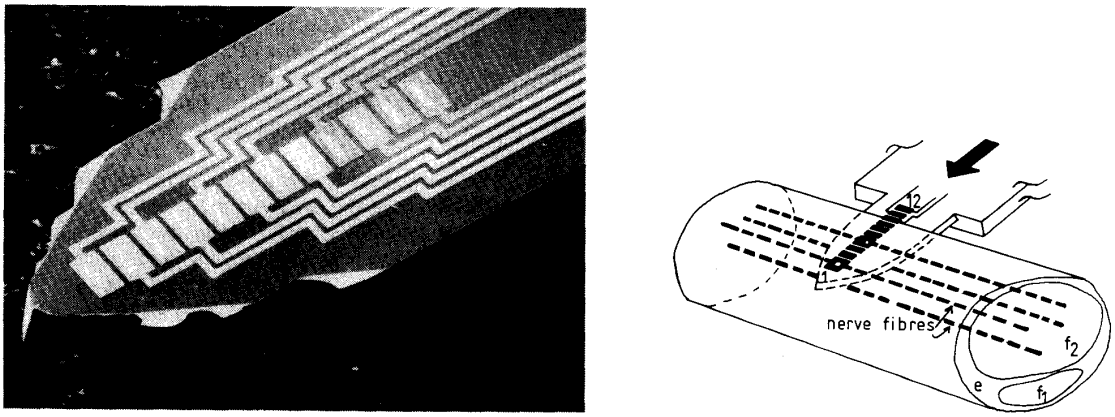
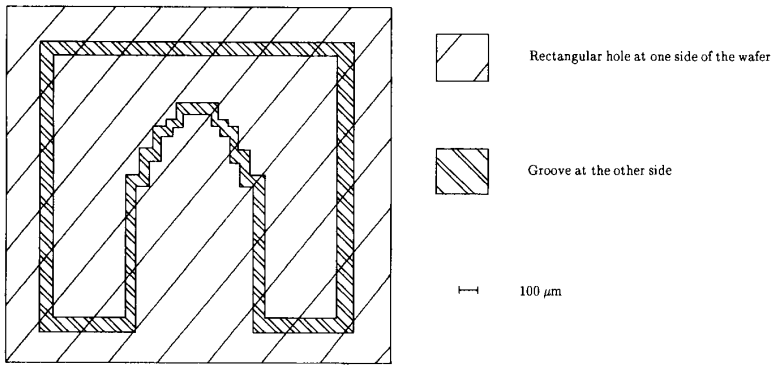


(a)



(b)

Fig. 1. (a) Scanning electromicrograph (left) of the silicon stimulation device and the way its tip is inserted into fascicle f_2 (right). The array direction is perpendicular to the longitudinal fiber direction while the plane of the tip is parallel to this axis. The array has 12 Platinum electrodes sites (10 × 50 μm) at interdistances of 50 μm. The silicon substrate of the tip is 45 μm thick and 840 μm long. Platinum layers of 0.5 μm are deposited, sandwiched between titanium adhesion layers of 20 nm. On top a 0.3 μm Si_3N_4 layer is deposited upon an intermediate 0.2 μm SiO_2 layer. (b) The two masking patterns used for etching the shape of the tip. The etching is stopped when the inner most hatched area in the crossprofile "falls out."

rat [16] will be explained further. Usually this nerve has only one fascicle.

Platinum is deposited on the oxidized silicon substrate [(100) wafers, 280 μm thick] using an intermediate titanium layer. On top of the platinum layer an insulating Si_3N_4 layer is deposited (by low pressure chemical vapour deposition), also with an adhesion layer of titanium. Si_3N_4 layers have excellent insulating properties [3]. In between these deposition steps the platinum electrode leads, electrode surfaces, and bonding pads (not shown in Fig. 1) were formed by photolithographic or plasma etching. Finally, the shape and thickness of the tip was obtained by double-sided anisotropic etching in KOH [17], [4], using the two masking patterns of Fig. 1(b). It was etched as sharp as possible, but rugged enough to pierce the epi- and perineurium.

Experimental Procedure

The tip was inserted (see Fig. 1) in the exposed peroneal nerve of the anesthetized rats during acute experiments in order to stimulate the extensor digitorum longum muscle (EDL) of the right hindleg isometrically. In four cases the insertion had to be aided by a small longitudinal preincision of the epineurium before piercing. After insertion the nerve was covered with parafin oil. Muscle and nerve temperature were kept at 38°C. Stable physiological conditions of the nerve could be maintained over periods of 4–6 h.

Rectangular cathodic current pulses were used as stimuli at a monopolar electrode (the common reference electrode was positioned in muscle tissue of the right hindleg) to elicit single twitch contractions of the EDL muscle under isometric conditions. Pulse patterns (see next section) had a repetition interval of 1127 ms. Each force response of the EDL muscle was the averaged result of 16–64 twitches. Each electrode had its own stimulatory channel. Pulse amplitude could be varied in steps of 0.1 μA, pulse width in steps of 1 μs.

Distribution of Nodes of Ranvier in α Motor Fibers of the Peroneal Nerve

1) Homogeneous Distributions of α Nodes: Swett *et al.* [16] report that the peroneal nerve in the rat contains 632 ± 27 motor fibers. No distinction can be made between α and γ motor fibers, but on basis of estimates from other authors it is assumed that about 55% of this number, i.e., 350 fibers, are α motor

fibers. This figure is of importance for the calculation of the density of α motor nodes of Ranvier. With a diameter of 0.5 mm for the peroneal fascicle, an internodal distance of 1 mm and 350 α motor fibers in the fascicle, the average nodal density N [18]. So, only 40 out of the aforementioned 350 α-motor geneous distribution of nodes (regular matrix) this density implies that a node "sees" an average territory of $1/N = 0.6 \times 10^{-12}$ m^3. An electrode with a stimulatory field of 0.6×10^{-12} m^3 would have a probability of one to stimulate an α motor node. This volume will be called the unitary stimulus volume. The value of 0.6×10^{-12} m^3 corresponds to a half spherical volume with a radius of 64 μm (because of the planar insulating substrate of the tip, the electrode "sees" a half spherical field for nearby distances). A point source electrode which is separated from its neighbor on the silicon array by a distance of $2 \times 64 = 128$ μm and which excites just one node will just not interfere with the unitary stimulus volume of the neighbor which excites just one other node with the same current.

The peroneal nerve in the rat innervates four muscles, the EDL muscle, the tibial muscle, the peroneal muscle, and the hallucis muscle. The average number of motor units in the EDL muscle is estimated at 40, with an average twitch force of 0.01 N [18]. So, only 40 out of the aforementioned 350 α-motor nodes belong to the EDL muscle. This leads to an EDL-α-nodal density N_e of 2×10^{11} nodes/m^3, a half sphere radius of 134 μm and a separation demand of 268 μm.

In case of full spherical stimulation the radius is 106 μm and the separation demand 212 μm. The actual electrode separation in the array of 12 was chosen 50 μm. Then, for example, by stimulation at electrodes 1 and 6 or 2 and 7 etc. a separation of 250 μm is achieved.

2) Poisson Distribution of α Nodes: The assumption of a regular matrix spacing of α motor nodes is true for the internode distance in the longitudinal direction in the same fiber (1 mm for fibers with a diameter of 10 μm) but not for the internode distance between different fibers. Node positions are not deterministic. Yet it is important for the design and use of multielectrode systems to know the probability of finding $0, 1, 2 \cdots$ α nodes (for the control of the EDL muscle) in a given stimulatory volume around an electrode site [24].

One may consider the probability of finding $N_n = 0, 1, 2$ nodes in a volume vol around a point electrode to be determined by a Poisson distribution function.

$$P(N_n = 0, 1, 2, 3, \cdots) = \frac{e^{-m} m^{N_n}}{N_n!}$$

in which $m = \text{vol} \times N_e \cdot N_e$ is the fixed average density of EDL α motor nodes (see section 1) above).

The probability to find no nodes in vol increases to one with vol decreasing to zero. $P(N_n = 0) = 0.368$ for $m = 1$, i.e., for vol $= 1/N_e$. The probability to find one node is maximal for $m = 1$, i.e., $P(N_n = 1) = 0.368$ for vol $= 1/N_e$. The probability to find no or one node equals the sum $P(N_n = 0) + P(N_n = 1) = 0.74$. An increase of vol lessens $P(N_n = 0)$ but also lessens $P(N_n = 1)$, at the expense of an increase of the total probability to find two or more nodes.

The analysis in this paragraph leads to the same choice for vol as the "average territory in the preceding paragraph. Choosing vol $= 1/N_e$ leads to the maximal probability of finding one node. The corresponding "unitary stimulus volume" is a half sphere with a radius of 134 μm, implying a separation demand between electrodes of 268 μm.

3) Anisotropy of Conductivity: The analyses above assume an isotropic conductivity of the neural tissue. It is worthwhile to consider the effect of the anisotropic conductivities σ_{er} and σ_{ez}.

The extracellular conductivities in the rat peroneal nerve are not known from the literature. In the cat spinal cord the ratio σ_{ez}/σ_{er} is about 9 [21]. This conductive anisotropy would lead to ellipsoidal stimulation areas instead of spherical with a ratio of the ellipse axes of 3. One may incorporate this anisotropy ratio in the probabilities P_r and P_z by a scaling operation. The effective ratio P_r/P_z then scales to 0.46 instead of 4. As this ratio results from estimated conductivity data (in another mammalian species) we will round this value to one, i.e., we assume that the average node probability is isotropic.

Local Stimulatory Field Analysis of a Single Electrode in the Array

One now many simplify an electrical field analysis of stimulatory field strength to a purely local approach because the "proximity" measure of 134 μm is smaller than the estimated internode distance of 1 mm. Local means that in the so called activating function f [8], [7]

$$f = V_{e,n-1} - 2V_{e,n} + V_{e,n+1}$$

one can put the two extracellular potentials at nodes $n - 1$ and $n + 1$ equal to zero (see the Appendix).

In a homogeneous half spherical, isotropic medium one has for the potential at distance R

$$V_{e,n} = \rho_e I/2\pi R$$

in which ρ_e is the resistivity of the medium and I is the stimulus current. Solving the RC-membrane network equation of McNeal [7]

$$dV_n/dt = 1/C_m \{G_a(V_{n-1} - 2V_n + V_{n+1} + f) - G_m V_n\}$$

for $V_{n-1} = V_{n+1} = 0$ and with the initial condition $V_n = 0$ at $t = 0$ yields

$$V_n(t) = \frac{-2G_a}{2G_a + G_m}(1 - e^{-T/\tau}) \frac{\rho_e I}{2\pi R} \qquad \tau = \frac{C_m}{2G_a + G_m}$$

in which C_m is the membrane capacitance, G_a is the axial internal conductance, G_m is the membrane conductance, and T is the pulse width. $V_n(t)$ is the membrane potential minus its rest value.

With the parameter values taken from McNeal [7] (see further) one has

$$V_n(t) = -0.38(1 - e^{-T/\tau}) I/R$$

(τ must be experimentally verified determined, McNeal [7] takes $\tau = 12.5$ μs.)

Excitation occurs when $V_n(t)$ reaches the threshold voltage V_d. Defining R_0 as the distance from the electrode to the node beyond which the excitation limit ($V_d = 30$ mV) is just not reached and chosing a pulsewidth $T = \infty$, one derives

$$R_0 = 2KI_0$$

with

$$K = \frac{-2G_a\rho_e}{(2G_a + G_m)\, 4\pi V_d} = 6.4 \text{ m/A} \qquad (1)$$

(K must be experimentally verified.)

I_0 is the rheobase at the distance R_0. The calculations use literature values [7] for ρ_e (3 Ωm), axoplasmatic resistivity ρ_i (1.1 Ωm), membrane capacitance/unit area c_m (0.02 F/m^2), membrane conductance/unit area g_m (300 Ω^{-1} m^{-2}), axial internal conductance G_a (1.43 × 10^{-9} Ω^{-1}), a nodal gap width l of 2.5 μm and an axon diameter of 10 μm. The value for K is in accordance with calculations of thresholds versus distance by Rattay [8].

The half spherical stimulation volume with radius R_0 contains $2\pi N_e R_0^3/3$ α-nodes. This number of nodes excites the same amount of motor units, yielding a total twitch force amplitude of $F_t 2\pi N_e R_0^3/3$. F_t is the average twitch force amplitude of a motor unit. With (1) and $I/I_0 = (1 - e^{-T/\tau})^{-1}$ this leads to a cubic force versus current relationship

$$F = 7.35 F_t N_e K^3 I^3$$

$$\text{for } T = 100\ \mu\text{s and } \tau = 70\ \mu\text{s}. \qquad (2)$$

A Selectivity Test for Intraneural Stimulation

Selectivity is maximal when each molopolar electrode excites one specific α-fiber. In practice, with increasing current the stimulus field from one electrode will expand and start to overlap with those of neighboring electrodes. A measure for selectivity, which is valid within the refractory period, is

$$S_{i,j} = \frac{F_{i+j} - F_j}{F_i} \quad \text{or} \quad S_{j,i} = \frac{F_{i+j} - F_i}{F_j}$$

in which F_i (or F_j) is the force due to electrode i (or j) separately and F_{i+j} is the force due to stimulation at both monopolar electrodes. In the latter case S will have a value between zero and one, provided the stimulus timing fullfills two requirements. First, the pulse at electrode j must have no temporal overlap with that at electrode i to prevent summation of the stimulus fields at nodes that are normally subthreshold. The pulse at electrode j must be separated in time long enough from the pulse at electrode i to let the membranes be fully discharged, i.e., beyond the so called "*RC*-overlap time interval." Secondly, the two pulses must arrive both within the absolute refractory period of 1 ms in order to not activate again nodes in a possible spatial overlap region which have already fired. The optimal separation can be derived from experiment by variation of the time separation between the two pulses.

If one stimulates again at j later than 1 ms after i nodes in an overlapping space are no longer refractory, so they will be active again. The muscle will now integrate mechanically the two twitches (staircase phenomenon).

Results

Sensitivity

Experiments were performed in six rats. A representative selection of five force versus current curves in four animals (seven data series) is given in Fig. 2. Maximum force of an EDL muscle is about 0.7 N. Fig. 2 shows saturation to this value or onset to saturation in six of the seven series. At the start stimulation begins at the lowest attainable force level, varing between 0.002 and 0.015 N. These levels are probably single motor unit levels [18]. Recruitment curves are sampled series, no attempt was made to find all possible (discrete) force levels. The lowest force levels correspond to widely differing currents (Fig. 2). For comparison the slope of the theoretically derived F versus I^3 relation (2) is drawn also in the figure. The three leftmost curves (five series) approximately obey this cubic relationship, while the two right-most curves are steeper in the force range below about 0.05 N.

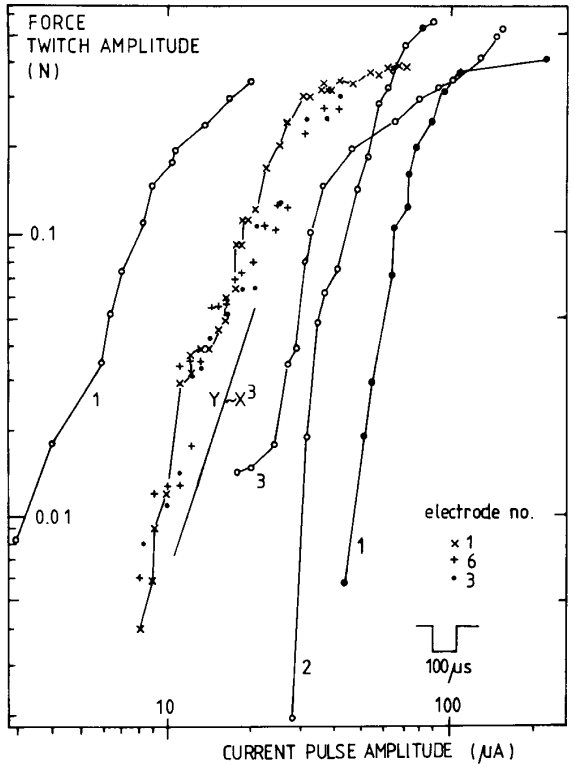


Fig. 2. Isometric twitch force amplitude versus amplitude of a current stimulus pulse. Recruitment curves are for monopolar cathodic stimulation with rectangular current pulses of 100 μs. Numbers indicate the electrode position in the array. Recordings were made in four animals. The two right most curves are from the same animal. The second curve from the left is drawn through the × symbols. Two other series in this animal (• and + symbols) are not interconnected for clarity. For comparison, a straight line has been drawn according to the relation $y \sim x^3$.

Other experiments (not shown) in which current was fixed and pulse width T varied, yielded that $\tau = 70$ μs is a realistic experimental value. The application of (2) to the most sensitive curve in Fig. 2 with $T = 100$ μs, $N_e = 2 \times 10^{11}$ m^{-3} and $F_t = 0.01$ N gives an estimate for the K factor: $K = 10$ m/A. This value is in the order of the estimated value of 6.4 m/A.

Selectivity

Fig. 3 shows the force as a result of stimulation with fixed currents at two electrodes of the array, numbers 1 and 4 (interdistance 150 μm) as a function of pulse separation T. For $T = 0$ the combined force is 0.13 N which is far more than just the summation of two times 0.02 N, but still below a factor $2^3 \times 0.02$ N. This indicates that the current fields of these two electrodes almost completely overlap. This is corroborated by the decrease of the force to the single electrode level at $T =$

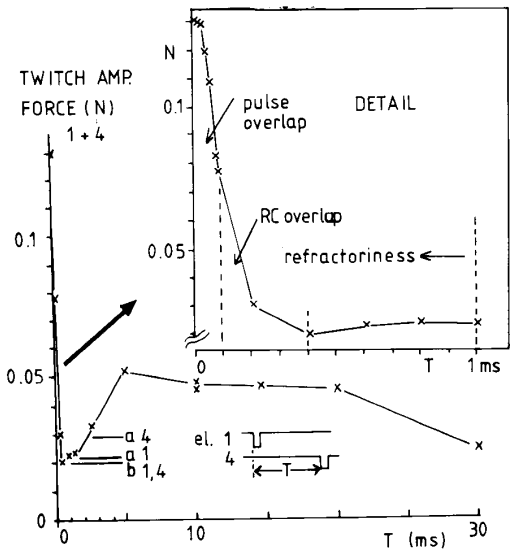


Fig. 3. Isometric twitch force amplitude versus the time interval T' between two cathodic pulses at electrodes 1 and 4. Force levels for stimulation at only one electrode (1 or 4) are also drawn, before ($b1$, 4) as well as after ($a1$, $a4$) the experiment. In the inset "Detail" the first millisecond is expanded; it shows three regions: the pulse overlap region below 100 μs, the "*RC* overlap" region between 100 and 400 μs, the refractory region below 1 ms. The decline down to the single electrode-force level at T = 400 μs indicates complete overlap of stimulus fields. The subsequent stable period up to about 1 ms is due to refractoriness, the increase after about 1 ms is due to mechanical integration of twitches, up to $T \approx 5$ ms (tetanus phenomenon). Desintegration starts at about 20 ms.

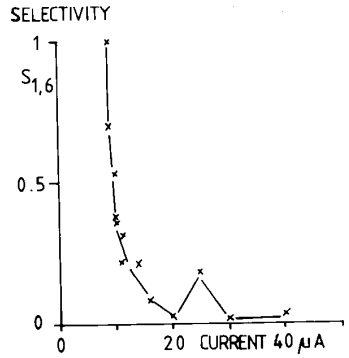


Fig. 4. Selectivity $S_{1,6} = (F_{1+6} - F_6)/F_1$ versus current pulse amplitude. Pulsewidth is 100 μs, pulse separation T = 400 μs. F_{1+6} means the force response to stimulation at electrode 1, followed after 400 μs by stimulation at electrode 6. The steep decline from S = 1 indicates that the stimulus fields of electrode 1 and 6 just do not overlap at a current of 10 μA, but start to overlap as soon as the current increases.

400 μs. Beyond about 1 ms the stimulated nodes are no longer refractory, so mechanical integration of twitches (staircase phenomenon) occurs up to the maximal possible tetanic value of 0.05 N (apparently the force levels drift upward during the experiment, see Fig. 3). Beyond a separation of 20 ms the integration is less effective as this time interval is that of the duration to the maximum of a twitch (so now the overlapping twitches start to desintegrate).

The inset right above in Fig. 3 shows that it is not sufficient to let the two pulses just not overlap. Beyond 100 μs and below 400 μs the membrane apparently still contains enough charge to enable currents to raise the membrane potential above excitation threshold. This leads us to a choice of 400 μs between two pulses. Fig. 3 is representative for the selectivity we obtained, also at low stimulus levels up to electrode separations of 200 μm.

Fig. 4 shows a selectivity curve of $S_{1,6}$ versus current for stimulation at wider separated electrodes, i.e., electrodes 1 and 6 which are 250 μm apart. The results in this figure indicate that at a electrode separation of 250 μm we were able to reach the maximal selectivity of 1, at minimal possible stimulus level.

Below this separation maximal selectivity was always below one. So it can be concluded that electrodes on the linear array must be separated by 200–250 μm for maximal selectivity. This compares favorably well to the calculated 268 μm (half spherical stimulus field) for EDL nodes.

Discussion

Absolute sensitivity of the intraneural array is satisfactory in so far that single motor units can be stimulated, as can be concluded from the lowest attainable force levels in the EDL muscle (Fig. 2): from $2 \times 10^{-3} N$ to $8 \times 10^{-3} N$ [18]. The current needed for the lowest level in each curve may however vary considerably, by a factor of about 14. A possible explanation is that in the most sensitive case (left-most curve in Fig. 2) the nearest node was right beside the electrode, while in least sensitive case (right-most curve) this distance was much larger. In that case the current ratio can be at most 134/3 = 45 (the calculated maximal radial distance to a node, 134 μm, divided by the short possible distance, i.e., the thickness of a myelin sheath, 3 μm). The observed ratio of 14 falls within the possible range. However, the observed curves not only differ in their lower parts, they also do not converge to one curve at the higher force levels. Instead, they are more or less parallel to each other at the higher force levels on this double logarithmic plot. This is probably caused by current shunting effects due to excess fluid leaks around the array during insertion, resulting in a fractional current loss. These effects can be partly overcome by embedding the nerve in paraffin or vaseline after the insertion and by starting the experiments only after a stabilization time of about an hour. Tissue reactions, the way of insertion, and the conservation of the nerve in these acute experiments, however, still disturb the ideal situation. Future experiments with chronically inplanted arrays will we hope avoid a number of these disturbing factors.

Another explanation for varying sensitivity might be the degradation of the Si_3N_4 layer. Although these layers are very dense and stable [3], this problem was avoided by taking a new array for each animal. Also, the layer was inspected visually under a highly manifying light microscope. No change before and after the experiment was observed.

The analysis in the methods section showed that statistically an electrode needs a half-spherical stimulus field radius of 134 μm to stimulate one α motor node. Twice this value must be interpreted as the minimal separation between two electrodes in order to avoid overlap of stimulus fields. The experiments showed that this value of 268 μm is corroborated by a measured separation of 250 μm. The discrepancy is surprisingly small and should be judged in view of the many parameters involved and the limited number of values in the literature.

In the statistical approach the average nodal density is an important parameter. It is derived from literature values of neural counts in spinal segments after retrograde labelling by horse

radish peroxidase (HRP) of nerve bundles. We have used the counts by Swett *et al.* [16], as in the study special care was taken to label specific bundles without leakage of HRP to other bundles. The authors claim that their precautions lead to the very accurate count of 632 ± 27 motor fibers in the peroneal nerve. The standard deviation is very small compared to other studies [19]. The recruitment data (Fig. 2) allow the experimental determination of the K factor: $K = 10$ m/A. A discrepancy is observed with the expected value of 6.4 m/A. As the factor $G_a/G_a + G_m \sim 1$, K is largely determined by the resistivity of the medium ρ_e [see (1)]. Literature values for the bundle resistivity vary between 100 and 8.2 Ωm (for toad [20] and cat [21], respectively) for the radial component of ρ_e and is about 1 for the longitudinal component [20], [21]. We have used the Frankenhauser standard value, $\rho_e = 3$ Ωm as an isotropic value, thereby neglecting anisotropy. Also, paraffin may have partially substituted the nerve fluid. Both neglections lead to an underestimate of ρ_e and thus of K. As parafin has a high ρ_e value (~ 25 Ωm) this may increase the effective K factor. As stated earlier, the uncertain effects of especially fluid/paraffin upon ρ_e may also explain the widely different shunting effects, seen in the parallel shift of recruitment curves in different insertions (note that the three symbols x, $+$, and $\cdot$ in the second leftmost curve in Fig. 2 belong to the same insertion).

It is to be noted that the selectivity measure $S_{i,j}$ is independent of the conduction parameters as long as these are equal at electrodes i and j. This implies that the derived density N_c is a rather robust parameter. As a consequence, only the K parameter remains as an adjustable parameter.

The activation functions favor excitation of fibers with longer internode distance which in general corresponds to thicker fibers. In a local approach it is not necessary to consider the difference in sensitivity of fibers with different diameter as the internode distance is assumed to be large enough in comparison to the distance from electrode to nodes. We have estimated an internode distance of 1 mm (corresponding to a fiber diameter of about 10 μm, which is the average mammalian motor fiber diameter [14]).

The local approach will be justified better at low stimulus levels than at full recruitment. Indeed, it is observed that a high levels the forces no longer satisfy the cubic dependence on current. Of course, the discrete, deterministic property of node positions is most overwhelming in the immediate surrounding of an electrode site. In principle, any statistical analysis fails in this region. Only the exact knowledge of electrode and the first available node positions would permit exact calculations. The premise of a homogeneous distribution of nodes in the local area around an electrode seems however reasonable, as the cubic dependence of force on current is observed. However, the density of EDL α-nodes is not high enough to give enough selectivity between two neighbour (at 50 μm distance) electrodes, just as was already numerically expected.

The linear array was inserted perpendicularly, i.e., the insertion track was perpendicular to the longitudinal fiber directions. Even with full employment of the minimal possible electrode separation of 50 μm one easily calculated that it will not be possible to reach selectively all 350 α nods using a linear or even a two dimensional planar device (oriented with its plane along the nerve cross section). One has to construct a real three dimensional type of device, employing also the longitudinal fiber direction. Supplementary, one should explore multipolar excitation techniques instead of monopolar, in order to compress stimulating fields towards more elongated ellipsoidal shapes.

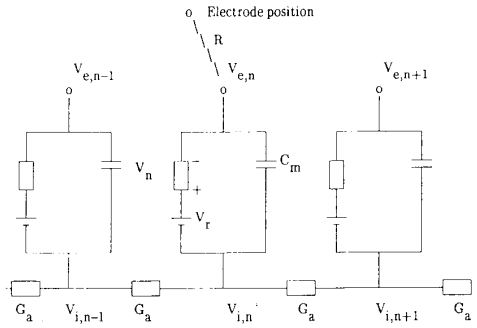


Fig. 5. Electrical network representation of a nerve fiber below threshold [7].

Apart from these geometrical considerations, the present array and stimualtion techniques should also be modified to suit better the demands for chronic application. First, the platinum electrode surfaces must be enlarged to lower the charge density of the pulses to the gassing limit of about 400 $\mu C/cm^2$ [22]. This can be achieved by diminishing the insulating Si_3N_4 fringe cover over the edges of the platinum electrode rectangles (see Fig. 1(a) left: the bare platinum appears paperwhite, the Si_3N_4 coverage appears gray, best seen at the four right most electrode pads). An increase of the area from the present 10 × 50 to 35 × 75 μ is possible, thereby lowering the charge density from 2000 to 380 $\mu C/cm^2$ (pulse amplitude 10 μA and pulsewidth 100 μs). Secondly, the use of other materials, such as activated irridium instead of platinum would bring down this value even further [23] and make safe operation possible for 100 μA/100 μs pulses. Thirdly, for chronic application it would be better to use actively charge balanced biphasic or passively charge balanced unidirectional pulses [22].

Appendix
Electrical Network Respresentation of a Nerve Fiber Below Threshold [7]

$V_{i,n}$	Potential inside nerve fiber at node n
$V_{e,n}$	Potential outside nerve fiber at node n
$V_{i,n} - V_{e,n}$	Membrane potential at node n
V_r	Resting potential (−70 mV)
V_n	$V_n = V_{i,n} - V_{e,n} + V_r$
R	Distance between electrode and node
C_m	Nodal capacitance
G_a	Axial internal conductance
G_m	Nodal membrane conductance.

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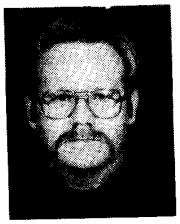
Wim L. C. Rutten (M'86) was born in The Hague, The Netherlands, in 1950. He received the M.Sc. degree in experimental physics in 1974 from the University of Leiden, Leiden, The Netherlands, and the Ph.D. degree in physics in 1979.

In 1971 he joined the Solid-State Magnetism Research Group at the Kamerlingh Onnes Laboratory of the University of Leiden. Since 1979 he has been engaged mainly in audiology and the physcis of hearing at the ENT Department of the University Hospital in Leiden. Since 1985 he has been Senior Staff member of the Biomedical Engineering Division, Department of Electrical Engineering, University of Twente, Enschede, The Netherlands. His current research interests are selective neural electrical stimulation, volume conduction in muscle and nerve and motor control using neural networks.



Harmen J. van Wier received the M.Sc. degree in electrical engineering from the University of Twente, Enschede, The Netherlands, in 1989.

From 1975 to 1980 he was employed as a system engineer with the Philips Company, first on processor controlled telephony equipment, later in the medical field on equipment for Computer Aided Tomography. His research interests include digital image coding and the analysis of biomedical signals. He is presently employed as a teacher at the Technical College Hogeschool Gelderland, Arnhem, The Netherlands.

Johan H. M. Put was born in Haaksbergen, The Netherlands, in 1955. He was trained as a veterinary practitioner at the University of Utrecht, Utrecht, The Netherlands.

In 1984 he joined the Biomedical Engineering Division, Department of Electrical Engineering, University of Twente, Enschede, The Netherlands. His main tasks were the microsurgical and anaesthetic support in the experiments on animals. Since 1989 he has been a full time veterinary practitioner.