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# Subfascicle Stimulation Selectivity with the Flat Interface Nerve Electrode

DANIEL K. LEVENTHAL and DOMINIQUE M. DURAND

Neural Engineering Center, Department of Biomedical Engineering, Case Western Reserve University, Cleveland, OH

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**Abstract—**The flat interface nerve electrode (FINE) is an alternative to cylindrical nerve cuffs for functional electrical stimulation (FES). By elongating the nerve in cross section, the FINE places more stimulating contacts around the nerve, and moves central axons closer to the electrode surface. Previous experiments have demonstrated that the FINE can activate selectively each fascicle in the cat sciatic nerve, and modeling studies have indicated that it should be possible to selectively activate groups of axons within individual fascicles. This hypothesis is tested using a combination of experimental and modeling techniques. Pairs of contacts stimulating the same fascicle were tested for subfascicle level selectivity, defined as the fraction of fibers activated by one contact but not by the other. It was possible to achieve greater than 90% selectivity with the FINE, but there was considerable variation in the results. The modeling studies showed that the selectivity achievable with a given contact pair depended strongly on the relative locations of the electrode and fascicle. Therefore, reshaping the cross section of a nerve can provide selectivity at the subfascicular level, but the electrode design must be optimized to improve selectivity across different nerve geometries. © *2003 Biomedical Engineering Society.*
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**Keywords—**FES, Functional, Electrical, Peripheral, Neural, Prosthesis.

## INTRODUCTION

Lost function may be restored to patients with central nervous system damage by electrically activating intact tissue distal to the neural lesion. This activation can be realized via electrodes placed on the body surface, in or on the muscles near the motor points, or directly on the motor nerves. Direct stimulation of nerve trunks offers many advantages over other methods, including the potential for controlling many muscles with a single implant, lower power requirements, and the ability to place the electrode far from contracting muscles.[8] To be effective in neuroprosthetic applications, peripheral nerve electrodes must be capable of targeting selected axons for activation without stimulating others. Furthermore, the electrode must be safe, and the stimulation characteristics stable over many years.

Several peripheral nerve electrode designs have been proposed. Intrafascicular electrodes place stimulating elements inside the fascicles, in close proximity to the axons.[2,4,13,14,17,21,23] These electrodes can provide selectivity, but it is unclear whether damaging the perineurium can cause long-term nerve injury.[2,12] On the other hand, extraneural electrodes place stimulating contacts outside the nerve, often within an insulating sheath.[7,15,20,21] These electrodes are less invasive to the nerve, but their stimulation selectivity suffers from the large amount of tissue interposed between the stimulating contacts and the target axons.

Recently, the flat interface nerve electrode (FINE) has been introduced in an attempt to improve the stimulation selectivity of extraneural electrodes.[20] All cuff electrodes impose their geometry on the nerve—the goal with the FINE is to create a geometry that optimizes stimulation selectivity. In contrast to cylindrical electrodes, the FINE either reshapes the nerve into, or maintains the nerve in, an ovoid geometry. Additional contacts can, therefore, be placed around the nerve, and central fibers are moved closer to the neural surface. The larger the circumference to cross-sectional area ratio, the greater the number of contacts, and the shorter the distance between stimulating contacts and target axons.

Chronic studies in rats have demonstrated that nerves and fascicles can be safely reshaped.[19] Also, acute experiments have demonstrated that it is possible to selectively activate individual fascicles in the cat sciatic nerve using this electrode.[20] Furthermore, finite element models have suggested that it is possible to selectively activate groups of fibers within individual fascicles using the FINE.[5] This could be important in both reducing fatigue and selectively activating individual muscles.[4,22] Subfascicle level selectivity, defined below as the fraction of fibers stimulated by one contact that are not activated by another, has not yet been demonstrated experimentally with the FINE.

Address correspondence to Dominique M. Durand, Neural Engineering Center, Department of Biomedical Engineering, Case Western Reserve University, Wickenden Blvd., Room 147, Cleveland, OH 44106. Electronic mail: dxd6@po.cwru.edu

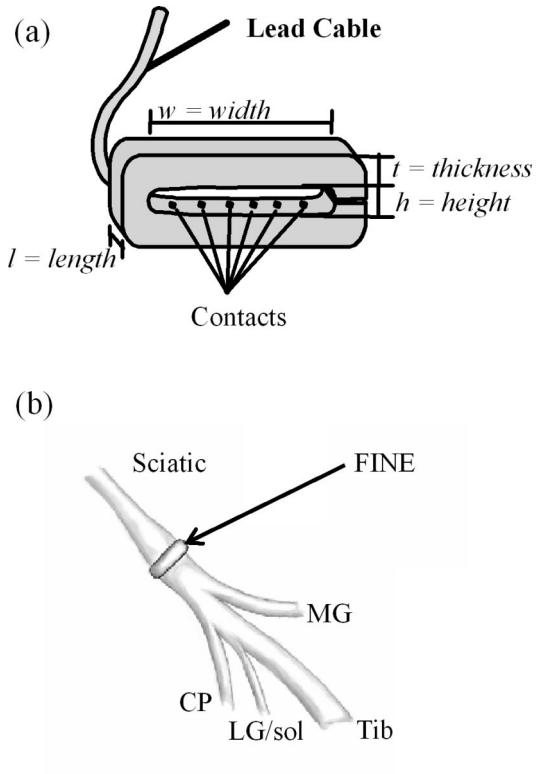


**FIGURE 1. (a) Design of the FINE electrode, and (b) site of implantation along the sciatic nerve.**

The hypothesis that subfascicle level selectivity is achievable was tested experimentally using joint torque measurements and numerically using the finite element method. The experiments provided a direct test of the hypothesis, while the model was used to (1) determine how closely selectivity calculations based on joint torque match selectivity calculations based on counting activated fibers, and (2) determine the effect of the location of the stimulating contacts with respect to the fascicles. These results provided design guidelines to improve the selectivity of the FINE.

## METHODS

### *Electrode Design*

A schematic diagram of the FINE is shown in Fig. 1(a). The electrode is made of molded silicone rubber (Dow Corning MED4-4210 elastomer), with platinum contacts spot welded to stainless steel wires embedded within. $h$, $w$, $l$, and $t$ were 1.3, 6.0, 6.4, and 1.0 mm, respectively. The contacts were approximately 0.5 mm wide and spaced approximately 0.5 mm apart. Five of the electrodes had 12 contacts, one had 11 contacts, and one had 13 contacts.

### *Experimental Procedure*

All animal procedures were approved by the Institutional Animal Care and Use Committee of Case Western Reserve University. Seven adult cats were anesthetized initially with ketamine hydrochloride ($30\ \mathrm{mg\,kg^{-1}}$, IM) and given atropine sulfate ($0.044\ \mathrm{mg\,kg^{-1}}$, IM) to reduce salivation. 5% glucose in normal saline was administered IV at a rate of $1\ \mathrm{cc\,min^{-1}}$. The animals were intubated and ventilated for the duration of the experiment. Sodium pentobarbital was administered in IV boluses as needed to maintain a surgical level of anesthesia, which was monitored by heart rate as well as paw-pinch and blink reflexes. A circulating water heating pad maintained body temperature.

In each cat, a FINE was implanted on the right sciatic nerve just proximal to its bifurcation [Fig. 1(b)]. In addition, single contact spiral cuff electrodes were placed on each of the four major branches of the sciatic nerve—the branch to the medial gastrocnemius (MG), the branch to the lateral gastrocnemius and soleus (LG/sol), the common peroneal nerve (CP), and the tibial nerve (Tib). A hypodermic needle in the back of the cat served as the distant anode.

The cat was placed in a stereotactic frame.[8] The knee joint was clamped with atraumatic, rounded cuffs, and the paw was secured to an aluminum shoe with a molded plastic tongue and plastic wire wraps. The shoe was connected to a three dimensional force and torque transducer (JR$^3$, Inc., Woodland, CA). The ankle, knee, and hip joint angles were held at 90°.

Current controlled monophasic pulses were delivered to the electrodes using a custom built stimulator controlled by a Dell Optiplex GX1p Pentium II 450 MHz computer. The same computer recorded the forces and moments at the transducer, and these measurements were converted to plantarflexion/dorsiflexion (P/D), medial rotation/lateral rotation (MR/LR), and eversion/inversion (E/I) torques about the center of rotation of the ankle joint.

### *Determination of Fascicle Activation*

Fascicle level selectivity was determined by comparing joint torques resulting from sciatic nerve stimulation with those obtained by activating the branches. Stimulation was carried out using 10 $\mu$s monophasic current controlled pulses at 25 Hz. Pulse amplitude (PA) was linearly increased during each trial from 0 to 2.5 mA, and at least 1 min was allowed between trials. The duration of each pulse train was 5 s. Three trials were averaged for each contact.

*Experimental Determination of Subfascicle Selectivity*

Torque addition experiments were performed using contacts stimulating the same branch to determine if they stimulated independent fiber populations. Contact pairs were stimulated both separately and simultaneously with short pulse trains. If the fiber populations activated by each contact are separate, the output from simultaneous stimulation is a linear summation of the torques generated by the contacts individually. Any deviation from this sum indicates that there is an overlap region where fibers are activated by both contacts.

For the linear addition of joint torques to hold, the stimulation thresholds for one contact must not change when the other is activated. This can occur if the contacts are pulsed too close together in time.[4,18,21,23] The stimuli were, therefore, interleaved during simultaneous stimulation. This may alter force generation by the overlap region, however, which experiences twice the stimulus frequency of other fibers. Because the relationship between tetanic tension and stimulus frequency is sigmoidal,[11] though, stimulating at a high enough frequency eliminates this problem.

To determine the stimulation frequency, 16 torque versus frequency curves were collected using branch electrodes. Only one curve was generated for each branch tested, and the activation levels at which the curves were collected ranged from 10% to full activation of the branch. All of these curves had achieved their plateau at frequencies below 75 Hz. This frequency was therefore selected for subfascicle selectivity measurements.

PA was adjusted so that the torque magnitudes generated by each contact were within 10% of each other. The duration of the pulse trains was 0.5 s, at least 1 min was allowed between stimulation runs, and the stimulus PW was 10 $\mu$s. Joint torque was averaged over the last 0.25 s of stimulation. Each measurement was repeated three times, and the results were averaged.

If there is no overlap between the stimulated fiber populations, $\boldsymbol{\tau}_s = \boldsymbol{\tau}_h + \boldsymbol{\tau}_l$, where $\boldsymbol{\tau}_s$ is the torque generated by simultaneous stimulation, $\boldsymbol{\tau}_h$ is the higher torque generated by single contact stimulation, and $\boldsymbol{\tau}_l$ is the lower of the two torques generated by single contact stimulation. The closer the vector $\boldsymbol{\tau}_s - \boldsymbol{\tau}_h$ is to $\boldsymbol{\tau}_l$, therefore, the less overlap there is in the stimulation regions. As a measure of proximity for these two vectors, $\boldsymbol{\tau}_s - \boldsymbol{\tau}_h$ was projected onto $\boldsymbol{\tau}_l$, and the magnitude of the resulting vector was divided by the magnitude of $\boldsymbol{\tau}_l$. The experimental selectivity, $S_{\text{exp}}$, was defined as

$$S_{\text{exp}} = \frac{(\boldsymbol{\tau}_s - \boldsymbol{\tau}_h) \cdot \boldsymbol{\tau}_l}{|\boldsymbol{\tau}_l|^2}. \quad (1)$$

$S_{\text{exp}}$ should vary as a function of the fraction of the fascicle activated. This was estimated by the activation level, defined as

$$AL_{\text{exp}} = \frac{|\boldsymbol{\tau}_l|}{|\boldsymbol{\tau}_{\max}|}, \quad (2)$$

where $\boldsymbol{\tau}_{\max}$ is the magnitude of the maximum torque obtained by stimulating the branch of interest.

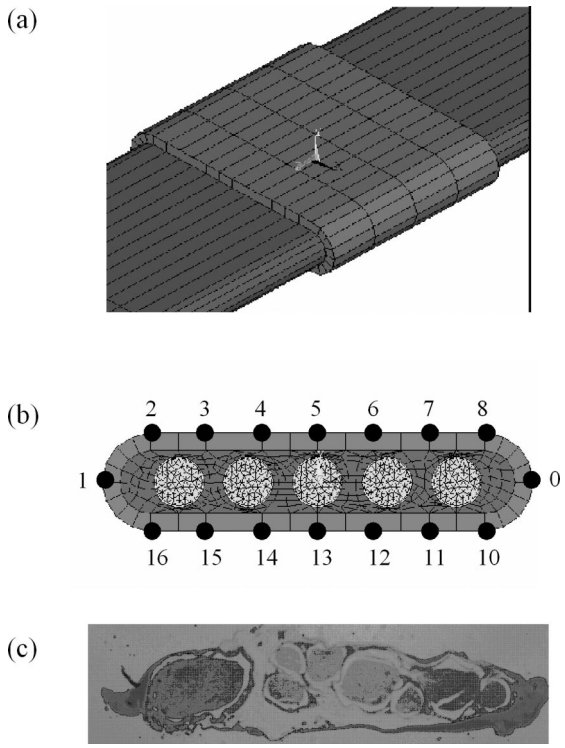


**FIGURE 2. Finite element model of the FINE and nerve, and a sample nerve cross section from the experiments. (a) Three-dimensional view, and (b) cross-sectional view through the center of the cuff showing the locations of the stimulating contacts. (c) Sample nerve cross section.**

*Finite Element Model*

A finite element model of a peripheral nerve and FINE was generated using the software package ANSYS (SAS IP, Inc., Houston, PA). Figure 2 shows three-dimensional and cross-sectional views of the model, as well as a sample cross section from an experimental nerve. The model is described in detail elsewhere.[5] Extracellular voltages were calculated within five identical, uniformly spaced fascicles for current pulses delivered at each of 16 contacts. Sixty axons with diameters chosen from a normal distribution were placed in each fascicle, with their nodes of Ranvier located randomly along the length of the fibers. The distribution mean was set at 15

$\mu$m with a standard deviation of 3 $\mu$m, approximating published alpha efferent fiber diameter distributions for cats.[3]

Force generation for each motor unit was estimated as an exponential function of fiber diameter:

$$F_i = \exp[k(CV_i) + b], \tag{3}$$

where $F_i$ is the force in grams generated by motor unit $i$, $CV_i$ is the conduction velocity of axon $i$ in $\mathrm{m\,s^{-1}}$, and $k$ and $b$ are empirical constants. Conduction velocity was obtained by multiplying the fiber diameter in microns by $6.0\,\mathrm{m\,s^{-1}}\,\mu\mathrm{m}^{-1}$.[9] $k$ and $b$ were estimated to be $-0.1\,\mathrm{s\,m^{-1}}$ and $-7.0$, respectively, from published data,[10] and yielded force values within the physiologic range.

Subfascicle level selectivity was quantified for a pair of contacts by the parameters

$$S_{\text{count}} = \frac{n_s - n_h}{n_l} \tag{4}$$

and

$$S_{\text{force}} = \frac{F_s - F_h}{F_l}. \tag{5}$$

The $n$'s represent numbers of fibers activated, and the $F$'s represent estimated force generation. The subscript $s$ stands for simultaneous stimulation of the two contacts, $h$ stands for the higher fiber count/force resulting from activation of one of the contacts, and $l$ stands for the lower fiber count/force resulting from activation of one of the contacts. $S_{\text{count}}$, therefore, represents the selectivity calculated based on counting the number of fibers activated, while $S_{\text{force}}$ represents the selectivity calculated based on force generation.

The stimulus amplitudes were set so that the same number of fibers were activated by each contact, and the selectivity was determined as a function of the fraction of the fascicle activated. The activation level was defined as

$$AL_{\text{count}} = \frac{n_l}{n_{\max}} \quad \text{or} \quad AL_{\text{force}} = \frac{F_l}{F_{\max}}, \tag{6}$$

where $n_{\max}$ is the total number of fibers in the fascicle (60 in this model), and $F_{\max}$ is the force generated by full activation of the fascicle.

Simulations for each contact pair were repeated 1000 times with fiber diameters randomly reassigned for each simulation.

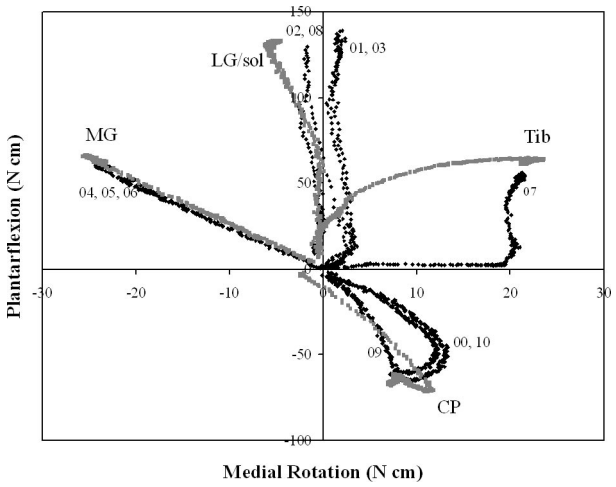


**FIGURE 3. Moment space plot of FINE and branch recruitments. The gray trajectories are from branch stimulation, and the dark trajectories are from different FINE contacts. In this example, contacts 1, 2, 3, and 8 are assigned to the LG/sol fascicle; 4, 5, and 6 are assigned to the MG fascicle; 0, 9, and 10 are assigned to the CP fascicle; and contact 7 is assigned to the Tib fascicle.**

## RESULTS

### *Experimental Selectivity*

*Fascicle Level Selectivity.* In order to test for subfascicular selectivity, at least two contacts must be able to activate a single fascicle. Therefore, fascicle level selectivity was determined first by comparing the torques resulting from FINE stimulation with those generated by stimulating the branches of the sciatic nerve. Figure 3 shows the results of FINE and branch stimulation from one experiment. Each contact is able to fully and independently activate a branch of the sciatic nerve, as indicated by trajectories in the plantarflexion/medial rotation moment space that arrive at the same point as full activation of a branch. The end points are important, as opposed to the entire trajectories, because some fascicles serve more than one muscle. If the order of activation for these muscles is different for the FINE and the branch electrode, the trajectories will diverge but ultimately arrive at the same point.

Seventy-four out of all 84 contacts in the seven cuffs fully activated a branch before stimulating another one. Furthermore, there was at least one contact that fully and independently activated every branch tested.

*Subfascicle Level Selectivity.* Eighty-one contact pairs stimulating the same branch were identified in 7 cats. Of these, only 18 pairs (at least one pair in each cat) were tested for subfascicle level selectivity due to time constraints. Selectivity was assessed by measuring the joint torque generated by activating two contacts concurrently.

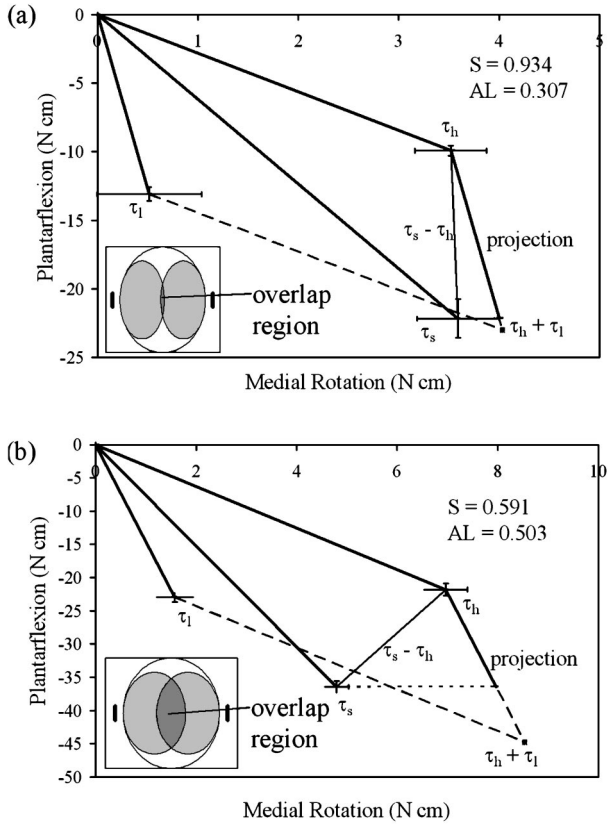


**FIGURE 4. Raw data showing sample calculations of $S_{exp}$ in the plantarflexion/dorsiflexion—medial rotation/lateral rotation moment space. The vector $\tau_s - \tau_h$ is projected onto $\tau_l$ (though they do not appear to be, the projections are orthogonal if the eversion/inversion moments are included) and the magnitude of the projection is divided by $|\tau_l|$. This provides a measure of how close $\tau_s$ is to the sum $\tau_h + \tau_l$. (a) High $S_{exp}$ at low $AL_{exp}$. (b) Lower $S_{exp}$ at higher $AL_{exp}$. The insets are schematic diagrams of the overlap region.**

The closer the results of dual contact stimulation were to a linear addition of the single contact results, the greater the selectivity. Figure 4 shows typical torque data for a single contact pair at two activation levels ($AL_{\text{exp}}$ =0.307 and $AL_{\text{exp}}$=0.503). At lower $AL_{\text{exp}}$, there is less overlap between the regions activated by each contact, $\tau_s$ approaches $\tau_h + \tau_l$, $\tau_s - \tau_h$ is close to $\tau_l$, and the selectivity is high. At higher $AL_{\text{exp}}$, the overlap region is larger, decreasing $S_{\text{exp}}$.

Figure 5 shows the selectivity measurements as a function of $AL_{\text{exp}}$ for three individual trials, as well as the results from all of the experiments. Figure 5(a) shows a typical $S_{\text{exp}}$ vs. $AL_{\text{exp}}$ curve with a large selectivity at low activation level, which decreases monotonically with increasing activation level. These results can be understood by looking at Figs. 4(a) and 4(b). As activation level increases, the overlap region grows, decreasing the selectivity.

Figure 5(b) shows an example where the selectivity is uniformly low. These contacts recruit the same fibers regardless of activation level. The most surprising result, however, is shown in Fig. 5(c), where the selectivity increases as a function of activation level. This implies that activation begins for both contacts with the same group of fibers, but as the stimulus amplitude is increased, activation spreads to different parts of the fascicle. Figure 5(d) shows all of the selectivity values as a function of activation level for all of the animals. Figure 5(d) shows that the selectivity varied widely across all of the animals. The remainder of this paper focuses on the results of modeling studies aimed at understanding the factors influencing the selectivity of the FINE.

### *Modeling Results*

The experimental results show that subfascicle level selectivity, estimated using torque measurements, is achievable with the FINE. Using the model, the relationship between force and fiber count based selectivity calculations is examined, as well as the effect of contact location on the selectivity of the FINE.

*Model Validation.* Five different contact pairs (contacts 5 and 13; contacts 4 and 14; contacts 2 and 16; contacts 2 and 3; and contacts 2 and 16—see Fig. 2) were tested for selectivity in the model, representing different orientations of the contacts with respect to the activated fascicle.

The model was validated by comparing individual modeling trials with individual experimental trials. Figure 6 shows three $S_{\text{force}}$ vs. $AL_{\text{force}}$ curves, each one generated by a different arrangement of fiber diameters within the target fascicle (see the Methods section). As expected, the case with high selectivity at low activation level [Fig. 5(a)] was reproduced by the model [Fig. 6(a)]. The case of uniformly low activation level [Fig. 5(b)] was also generated by the simulation [Fig. 6(b)]. The high selectivity at very low activation level is an artifact of the model, generated because the selectivity must be either 0 or 1 when only one fiber is activated by each contact. If the experimental measurements were sensitive enough to detect the activation of single fibers, Fig. 5(b) would likely appear nearly identical to Fig. 6(b).

Figures 5(c) and 6(c) are of particular interest because $S$ increases as a function of AL over part of the curve. These results were unexpected in the experiments, and the model not only reproduced this effect, but provided an explanation for it. A cluster of large fibers equidistant from two contacts may be activated before smaller fibers closer to the contacts. $S$ is small when only the large fibers are activated, and increases as smaller fibers closer to each of the contacts are recruited.

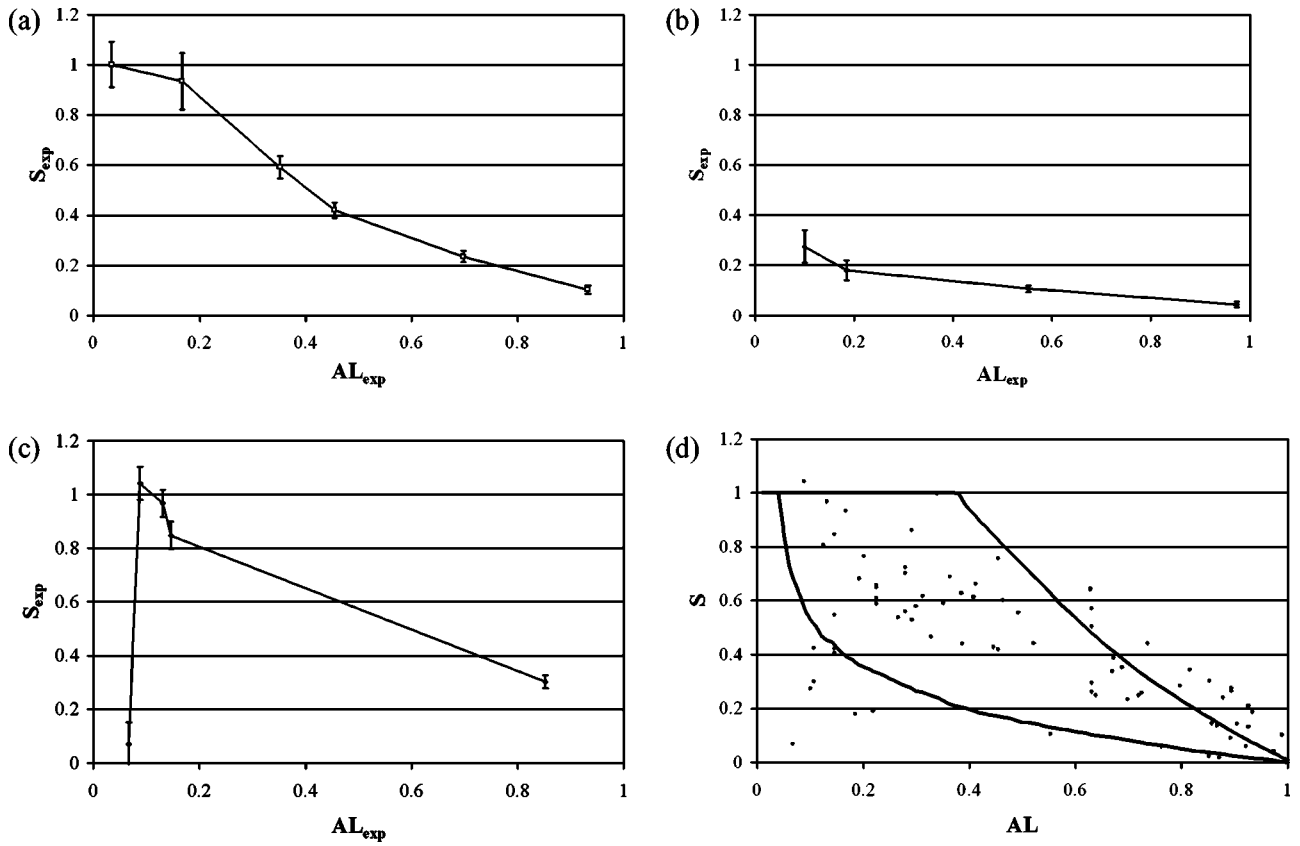


**FIGURE 5. (a)–(c) $S_{exp}$ vs. $AL_{exp}$ curves showing three representative curve shapes. (d) All of the experimental results superimposed on modeling results. The points are experimental measurements, the solid lines are the 5th and 95th percentiles of modeling results at each activation level.**

The model can be further validated qualitatively by comparing its output with experimental results. Figure 5(d) shows all of the experimentally obtained points, as well as the 5th and 95th percentiles for $S_{force}$. Of 77 experimental points, 46 (60%) fell within the 90% confidence interval for the model results [shown in Fig. 5(d)] and 60 (78%) were within the range covered by all model results. All of the experimental points outside the modeling range occurred at high $AL_{exp}$, indicating that there may be a systematic error in either the model or experimental methods in this region.

*Comparison of Force and Fiber Count Selectivity.* Selectivity has been defined as the fraction of fibers activated by one contact not activated by another. In the experiments, the number of fibers activated was estimated using joint torques. Because force generation is a function of fiber diameter, however, selectivity calculations based on these torques may not accurately reflect fiber activation. In the model, it was possible to directly compare selectivity determinations based on force generation and on fiber activation.

Figure 7(a) is a sample plot of both $S_{force}$ and $S_{count}$ as a function of the appropriate activation level, showing that the two values closely mirror each other. The relationship between these values is quantified in a histogram of the difference $S_{force} - S_{count}$ across all 5000 modeling trials [Fig. 7(b)]. There is a large peak at 0, with the distribution slightly skewed towards negative values. Therefore, $S_{force}$ tends to slightly underestimate $S_{count}$.

*Effect of Contact Location.* The effect of contact location with respect to the fascicles was analyzed in the model by comparing the selectivities generated by five different contact orientations: contacts 5 and 13; contacts 4 and 14; contacts 2 and 16; contacts 2 and 3; and contacts 2 and 16 [Fig. 2(b)].

Figure 8 shows histograms of $S_{force}$ at $AL_{force} = 0.25$ for different contact geometries. The model results show that selectivity depends strongly upon the arrangement of the contacts with respect to the fascicles. If two contacts are located directly over a fascicle [Fig. 8(a)] or diagonally across a fascicle [Fig. 8(b)], the selectivity is nearly always perfect. If these contacts are moved off center, the expected value of the selectivity decreases [Fig. 8(c)], and can become quite small if the contacts are located on the same side of the cuff [Fig. 8(d)] or moved away from the fascicle [Fig. 8(e)].

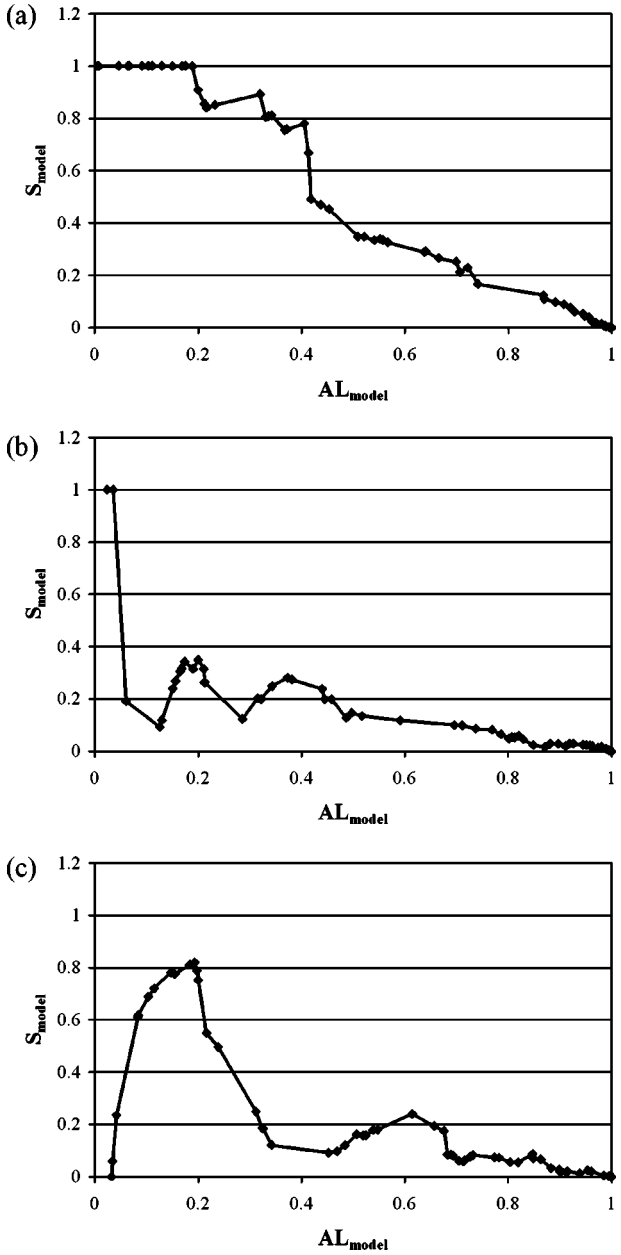


**FIGURE 6. (a)–(c) $S_{force}$ vs. $AL_{force}$ curves showing three representative curve shapes. These are individual trials representing one contact pair with one fiber diameter distribution.**

As the median selectivity decreases for a contact geometry, the variation in selectivity values increases. These results explain the wide variation in experimental selectivity values. Because the contacts were not aligned with the fascicles during implantation, it is likely that few, if any, of the contacts were located in optimum positions for selective stimulation.

## DISCUSSION

The primary goal of this study was to investigate the potential of the FINE to activate selectively groups of

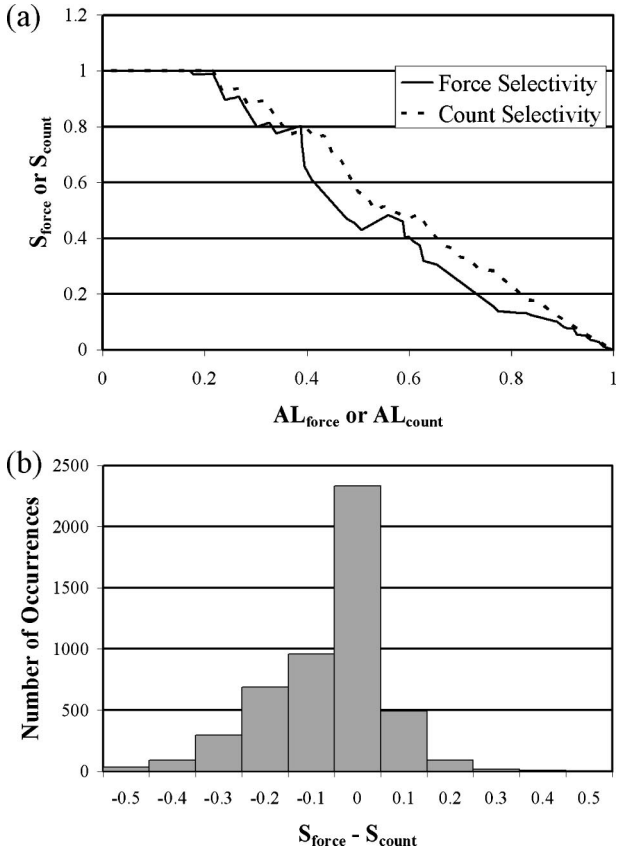


**FIGURE 7. (a) Sample plot of $S_{force}$ and $S_{count}$ against the appropriate activation level. These results were generated for contacts 4 and 14 in the model. (b) Histogram over all contact geometries of the difference $S_{force} - S_{count}$.**

fibers within individual fascicles. The strategy employed in the experiments was to identify contact pairs activating the same fascicle, then test these pairs for subfascicle level selectivity. It was found that such selectivity was achievable with the FINE in a small but substantial percentage of experiments.

Fascicle level selectivity was determined by comparing branch activation to FINE activation. This method assumes that a single fascicle entered each branch. Though definitive data do not exist to validate this assumption, it has been our experience that the CP, MG, and LG branches correspond to single fascicles at the level of implantation, and that the tibial component of the sciatic nerve may or may not be a single fascicle. This assumption should therefore be valid in the vast majority of cases, and does not affect our basic conclusions.

It was possible to selectively activate groups of fibers within individual fascicles, though the selectivity varied greatly between experiments. The model provides insight into the factors affecting the selectivity, as well as how to design future electrodes to insure that optimum selec-

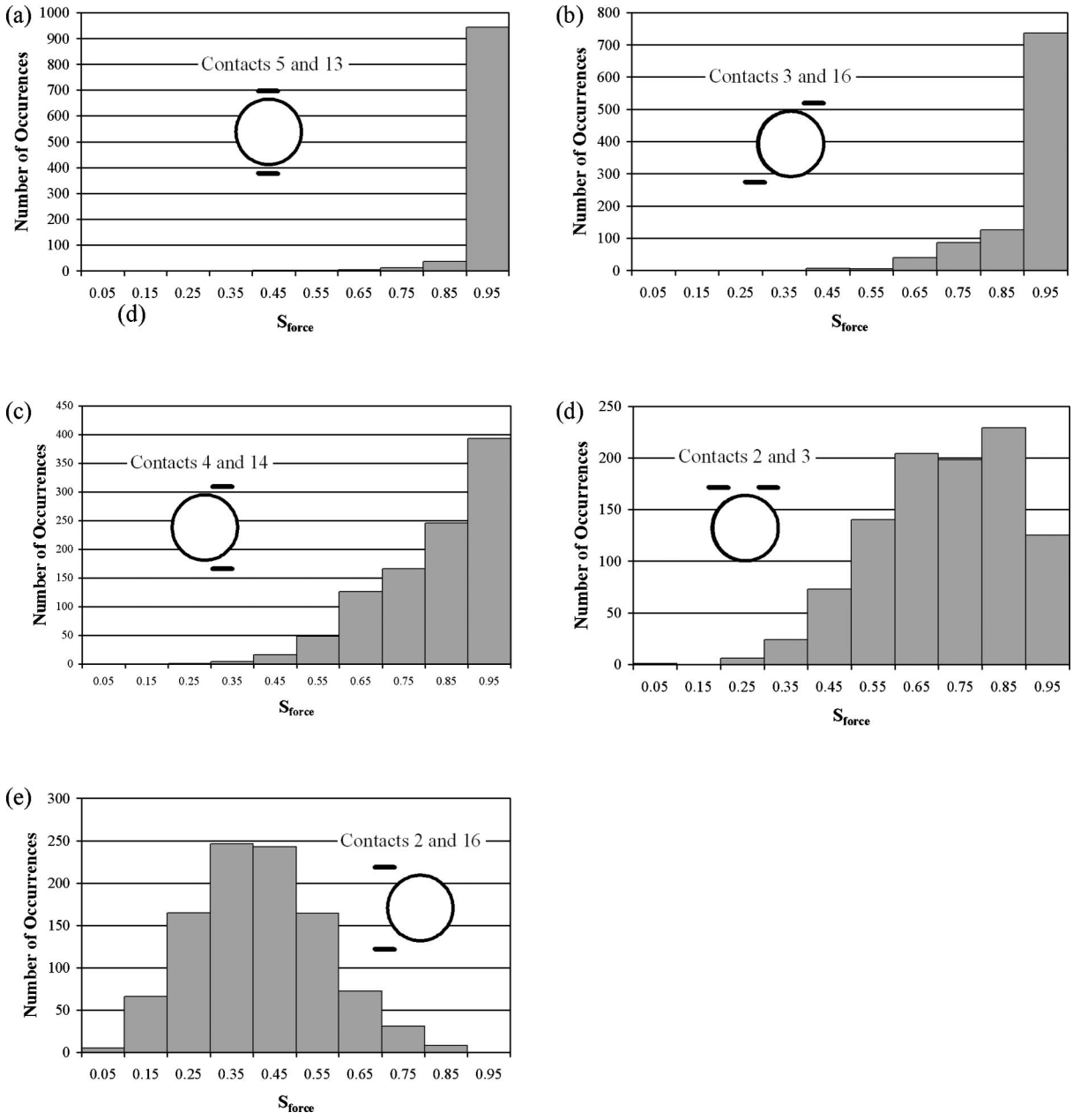


**FIGURE 8. Histograms of $S_{force}$ at $AL_{force}=0.25$ for different contact geometries. (a) Contacts 5 and 13 (transverse, centered), (b) contacts 3 and 16 (diagonal), (c) contacts 4 and 14 (transverse, off-center), (d) contacts 2 and 3 (adjacent), and (e) contacts 2 and 16 (transverse, off fascicle).**

tivity is obtained. The orientation of the contacts with respect to the fascicles not only affected the median selectivity, but also the variation in possible selectivities depending on the arrangement of fiber diameters within the fascicle. Contacts located across the nerve from each other, centered over a fascicle, provided the highest, most consistent selectivity in the model, with the variability increasing as the contacts were moved off center.

The exact locations of the contacts with respect to the fascicles were not known in the experiments. The likelihood of a pair of contacts being centered over a fascicle can be estimated, however. The electrodes were manufactured by hand with contacts spaced approximately 1 mm apart, which is approximately the diameter of a fascicle. The probability that a pair of contacts was located optimally in these experiments, therefore, was not very high. New manufacturing methods are required so that the contact density can be increased, guaranteeing that there will be at least one contact pair centered over each fascicle.

Also, it may be possible to reshape the nerve further than was done in this study. Here, the nerve was reshaped, but the geometry of the fascicles was not altered to prevent an acute rise in intrafascicular pressure. It is possible, however, to slowly reshape the fascicles,[19] which would allow more area for contact placement and reduce the contact to fiber distances.

Similar experiments have been performed using the force addition of two contacts as a measure of selectivity. Veltink *et al.*[21] compared the selectivity of intrafascicular and extraneural electrodes stimulating rat common peroneal nerve. In contrast to this study, they found no correlation between force generation and selectivity. This may be because they did not require that each contact generate the same force. The selectivity obtained by stimulating one contact strongly and another weakly could be different from the selectivity obtained by activating both contacts at a moderate level. They also found that the mean selectivity for the intraneural and extraneural electrodes was approximately the same, but that the intrafascicular electrodes generated a wider range of selectivities. Because data were not provided concerning the level of activation at which each of these measurements were made, however, it is difficult to compare their results directly with those presented here.

Rutten *et al.* used a one dimensional silicon intrafascicular electrode array to stimulate rat common peroneal nerve.[18] They found that it is possible to achieve perfect selectivity at only the lowest stimulation level, and only if the two stimulating contacts were separated by at least 250 $\mu$m. As the current was increased, the selectivity dropped steeply.

Recently, similar experiments were performed using the Utah Slanted Electrode Array (USEA), which consists of an array of needles on a silicon substrate that penetrate inside the fascicles.[4] This electrode was tested in the cat sciatic nerve. Using the USEA, selectivity greater than 0.8 (by the definition used in this paper) was achievable at up to 20% activation for individual muscle groups, with a sharp decline in selectivity at higher activation levels. Even at low activation level, though, selectivity varied greatly, as in the experiments presented here. One major difference between these electrodes is the very high contact density of the USEA, making it more likely that there will be a pair of contacts with high selectivity for any given fascicle.

There are also some fundamental differences between the selectivity calculations made in the above studies and those made for the FINE. By measuring selectivity in individual muscles, the issue of torque vectors that point in different directions was avoided. Those measurements should therefore be somewhat more accurate than those presented here. The torque addition method has the advantage that it is noninvasive, however, and can be used in chronic experiments.

A finite element model was used to address several issues concerning the experimental results. The first is the validity of using joint torques to estimate fiber activation. An exponential function was used to relate fiber diameter to muscle force. This relationship has been debated in the literature,[1,6,10,16] but provides a useful approximation to examine the qualitative relationship between the two methods of calculating selectivity. Determinations based on force closely approximated calculations based on fiber counts, but slightly underestimated selectivity. Large fibers have lower thresholds than small fibers, and are more likely to be activated by both contacts. Fibers in the overlap region therefore tend to generate larger forces than those that are only activated by one contact, reducing the selectivity value.

The model did not account for the directions of the torque vectors because of a lack of data describing the distribution of fibers serving different muscles. To determine the significance of this omission, Eq. (1) can be written in an alternative form:

$$S_{\text{exp}}=\frac{|\boldsymbol{\tau}_s|\cos(\theta_{sl})-|\boldsymbol{\tau}_h|\cos(\theta_{hl})}{|\boldsymbol{\tau}_l|}, \tag{7}$$

where $\theta_{sl}$ is the angle between the vector from simultaneous stimulation and the lower single contact vector, and $\theta_{hl}$ is the angle between the higher and lower vectors arising from single contact stimulation. As the cosines approach 1, this Eq. (7) becomes identical to Eq. (5). The median value for the cosines was 0.99, with a minimum of 0.91. Therefore, the orientation of the torque vectors affects the selectivity by at most approximately 10%.

Another deficiency of the model is its idealized geometry. Real nerves do not consist of identical cylindrical fascicles evenly spaced within a homogeneous medium. In more realistic geometries, fascicles may be ''stacked'' on top of each other [Fig. 2(c)] so that selective activation of each one is possible from only one side of the electrode. This could negatively affect selectivity, since the model indicated that optimum selectivity is achieved by contacts located across the nerve from each other. By further reshaping the nerve, however, it should be possible to place contacts on both sides of each fascicle.

The model predictions compared well with the experimental results, though at high $\text{AL}_{\text{exp}}$ there was a cluster of experimental points with higher selectivity than predicted by the model. This may be due to a slight overestimation of activation level in a subset of the experiments. By analogy with the model, $|\boldsymbol{\tau}_{\text{max}}|$ is intended to estimate the sum of the forces exerted by each motor unit. If all of the motor units do not exert torques in the same direction, however, $|\boldsymbol{\tau}_{\text{max}}|$ underestimates this total

by not accounting for components of the torque vectors that cancel each other out. This has the effect of overestimating $AL_{exp}$, since $|\tau_{max}|$ is in the denominator. Therefore, there may be a systematic error in a subset of the experiments where the points have not been shifted up, but to the right. This cluster of points resulted from stimulation of the CP or Tib fascicles, which serve multiple muscles with torque vectors pointing in different directions. Because the angles between the torque vectors were small, as discussed above, however, this is a small effect causing a maximum error of approximately 10%. The fundamental conclusions regarding the selectivity of the FINE are, therefore, still valid.

## CONCLUSIONS

It was found in this study that it is possible to selectively activate portions of individual fascicles using the FINE. This selectivity is variable, however, depending on the orientation of the contacts with respect to the fascicles. These results suggest that by increasing the contact density the FINE may provide selectivity comparable to intrafascicular electrodes. Further work is required to test the chronic properties of this electrode as well as quantify its long-term stability.

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