could not demonstrate any open-set improvement; indeed, some of the subjects were unable to perform any open-set tests with either processing strategy. These results suggest that even for partial insertions it may be advantageous to use less than the total number of intracochlear electrodes, and to try to match the frequency allocation to the correct tonotopic position.

These results are less conclusive than those obtained by Whitford et al. in a small study of 4 patients. Notwithstanding the lack of statistical evidence that the charf mapping strategy is superior to Multipeak in these patients, none of the 11 wanted to return to the latter for the second phase of the study.

REFERENCES

NEW TEMPORAL CODING SCHEME FOR AUDITORY NERVE STIMULATION

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INTRODUCTION

Electrical stimulation of the auditory nerve provides deaf people with a perception of sound inferior to that of normal-hearing people. Since any acoustic perception is derived entirely from the output of the auditory nerve, effective cochlear implant stimulation strategies should evoke an output of the electrically stimulated auditory nerve that is in some sense a good approximation to that of an acoustically stimulated auditory nerve. A typical cochlear implant may have around 20 independent electrodes, each of which transmits data in the form of discrete electrical pulses at a practical limit of a few thousand pulses per second. In contrast, the normal-hearing ear has approximately 30,000 nerve fibers, each of which can achieve action potential independently of the others at any time during the stimulus. Clearly, it is beyond the scope of current technology to achieve an electrically evoked auditory nerve output identical to that evoked via the normal hearing situation (NHS), so cochlear stimulation strategies must be limited to approximating certain features of neural firing patterns. How should such features be chosen?

Emerging knowledge of methods of sound-coding within the auditory nerve, and within the brain (also Clark et al., this suppl, section 5), is helping to shed light on which features of neural response are important from a perceptual standpoint, and which can be discarded. However, certain questions remain unanswered. For instance, it is not yet clear whether the firing statistics of individual fibers, or the overall firing statistics of an ensemble of fibers, is more important for the coding of frequency (also Clark et al., this suppl, section 5). In this paper we explore methods for making the electrically evoked firing statistics of an ensemble of fibers approximate those of the NHS. This general idea has also been investigated by Parkins et al. The specific statistic we approximate here is the summed period histogram of all the fibers (the population period histogram [PPH]). Modifications to single fiber statistics are analyzed in a companion paper (Irlicht and Clark, this suppl, section 16).

In order to determine the neural firing patterns evoked from both acoustic and electrical stimulation, we have developed simulation programs that can calculate the neural firing pat-
required proportion of neurons to achieve action potential period histogram of the neural population to better approximate which the amplitude of the pulses is chosen so as to cause the within each section of the period of the waveform. produce neural firing patterns that are very different from bipolar electrode oriented at right angles to the fiber. We extend the model to calculate the response of 200 nerves Colombo and Parkins (CP) model. The CP model calculates these response differences are caused both by individual

SIMULATION ALGORITHMS

The model of electrical stimulation we use is based on the Colombo and Parks (CP) model. The CP model calculates the neural response for a single nerve fiber stimulated by a bipolar electrode oriented at right angles to the fiber. We extend the model to calculate the response of 200 nerves spread out along 1 cm of the basilar membrane. The pulse shape selected is a charge-balanced biphasic pulse with an initial positive phase lasting 100 milliseconds (ms), a 45-ms rest, a negative phase of 100 ms, and a 5-ms rest. All simulation parameters are identical to those of Colombo and Parkins.

The model of the normal-hearing ear is a concatenation of the basilar membrane model of Au et al. (this suppl, section 16), and the computational model of the inner hair cell–auditory nerve synapse. The basilar membrane model provides a time-varying basilar membrane displacement as a function of incident sound. These values, evaluated at 64 points along the basilar membrane and suitably normalized, are fed into the Meddis et al. model. The result is the fluctuating probability of neural output of 64 auditory nerve fibers evenly spaced along the basilar membrane.

SIMULATED POPULATIONAL PERIOD HISTOGRAMS

The period histogram (PH) of single fibers has been observed to phase-lock closely to the positive half cycle of low-frequency tones. Since each fiber responds differently, the PH is not necessarily a scaled version of a single fiber’s PH. These response differences are caused both by individual differences between the fibers, such as fiber diameter, and by phase delays due to the timing of the traveling wave in the cochlea reaching their positions along the basilar membrane. If the individual differences were correlated to position along the basilar membrane, then the traveling wave phase delay would cause their effects to also be correlated to the phase of the PH, resulting in a distorted PH. However, when measured within an area equivalent to the stimulating region of a single electrode of a multielectrode device, such individual differences are not strongly a function of the fiber’s position along the basilar membrane. Thus, fiber response differences are not a function of the phase delay of the traveling wave, but are averaged along the entire phase of the PH.

This averaging of single fiber response differences across the entire phase of the PH ensures that a reasonable approximation to the PH can be obtained via the simulation of a cohort of identical individual fibers spread along the basilar membrane, in which each fiber exhibits an “average” response. This is simulated via the acoustic simulation routines described above. For the simulation shown here (Fig 1), the acoustic stimulus used as input to the cochlear model is a 1-kHz tone of 100-ms duration, with the last 10 ms of data used in the calculation. The result is a phase-locked signal, resembling a scaled version of the PH of a single unit. (Other simulations not described here demonstrate that the amount of phase locking is a function of the spatial distribution of nerves included in the PH.)

STIMULATION STRATEGY

The MP3 stimulation strategy proposed in this paper is designed to be implementable on a speech processor for cochlear implants that codes signals in terms of biphasic pulses. Planned refinements to the MP3 strategy involve modifications to the pulse parameters, but in this initial study we investigated the applicability of MP3 with a fixed-width biphasic bipolar pulse, with an overall pulse width of 250 microseconds.

The MP3 approach involves causing the electrically stimulated neural PH to approximate a target PH, possibly the NHS PH. Once chosen, the target PH will be approximated via a rectangular approximation, with the approximation...
The electrical stimulation simulations described above were applied to investigate the PPH of a typical electrical stimulation routine. Figure 3 shows simulation of the PPH for a stimulation of a current level of 3 mA, at 200 Hz. It is clear that such fixed-rate stimulation techniques cannot form a close approximation to the target PPH at any frequency other than that of stimulation.

A multiple-pulse electrical stimulation model was iterated in a trial and error fashion until a set of MP^3 current levels was found that provided the required PPH. The results can be seen in Fig 4. It is clear that for each period, the actual histogram closely resembles the desired approximation of Fig 2. Clearly, when compared to Fig 3, the MP^3 system yields a vastly improved PPH.

Figure 5 shows the current levels required to produce the PPH of Fig 2. Note that the current levels required are smaller at first, and then become larger when more nerves are in a refractory phase. After a short time (approximately 3 ms), the relative number of nerves in each quadrant within a refractory period has reached a "steady state," and close to periodic (1-kHz) current values provide the required output. However, since the relative current levels across each period are affected by refractory effects across the stimulating period, they are not linearly dependent on the required PPH level. Thus, forcing the current levels to follow a simple sinusoid will not result in the required PPH; the shape must be modified to take refractory effects into account.

Figure 6 shows the spread of action potentials as a function of nerve position and time segment. The nerves excited by the simulation were grouped close to each electrode, and the spread of excitation was limited to 0.25 cm on either side of the electrode midpoint. As expected, it can be seen that after achieving an action potential, it took most nerves about 1 ms or so before the next action potential. Also, after the initial stages, the firing pattern took on a very predictable pattern, with each nerve predominantly firing within one of the four stages of the stimulation.

**CONCLUSION**

Advanced computer simulation techniques can be used to compare the temporal-spatial characteristics of neural responses for electrical stimulation, and the NHS. These results can be used to propose new stimulation strategies. Here, we propose a Multipeak per period strategy that results in improvements to the electrically evoked PPH.

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