A Computational Model of the Cochlea used with Cochlear Prosthesis Patients

Dirk Van Compernolle

Department of Electrical Engineering
Stanford University
Stanford, CA 94035

ABSTRACT

This paper describes a computational model of the cochlea with the appropriate interface for a cochlear prosthesis. This interface consists of two parts. First neural activity is mapped to electrical current density at the neurons. Then the optimal stimulation pattern is derived through a least squares fit, given the current spread function between the electrodes and the nerve endings. Practical limitations which limit the usage of this scheme, as well as the first tests performed with our patients, are described.

INTRODUCTION

The work in this paper was stimulated by a long existing dilemma in the signal processing for a multichannel cochlear prosthesis. Most multichannel patients are able to distinguish between the different electrodes on a frequency basis to some extent. This should give a multichannel speech processor a degreesoffreedom edge over a single channel device. In spite of this advantage, multichannel devices have proven only marginally superior for speech purposes. We believe that the two major reasons for this are an oversimplified cochlear model and the underestimation of the electrical field interactions between electrodes[1]. The model which is presented in this paper tries to overcome both failures.

ELECTRODE ARRAY

At Stanford we currently use an 8 channel electrode array which is in design very similar to the electrodes used by the Vienna group[2]. The electrode array is inserted in the scala tympani; the most basal electrode is located about 5mm from the round window and electrode spacings are 2.54mm, covering in total a little more than the first turn of the cochlea(see Fig. I). An extra ground electrode is placed in the mastoid and serves as return path for the net sum of currents delivered to the electrodes.

Based on physiological data we estimate that this electrode array spans characteristic frequencies from 800Hz to 9kHz[3]. In this frequency range the cochlear layout is close to logarithmic and the spacing of 2.54mm corresponds to 1/2 octave. These numbers are supported by anecdotal reports of our patients. The electrodes E7 and E8 give unpleasant scratchy, high frequency sounds outside the common speech range, and are hence of little practical use. This leaves us with 6 usable electrodes at locations with best frequencies ranging from 800 to 4500 Hz.

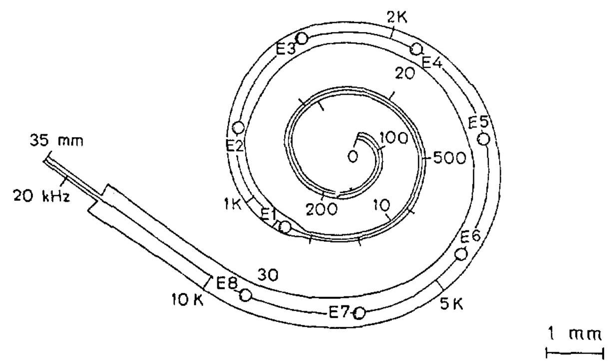

Fig. I : 8 channel electrode array

SIGNAL PROCESSING MODEL (Fig. II)

1. Computational Model of the Ear

The desired output of our model of the normal ear is a function which we will call "neural activity". It is an intensity function, representing the average firing rate for neurons at a given location. We do not attempt to model the output as the response of a single fiber, but this activity function is rather a spatial average of probability of firing of single fibers. Neural activity is the first common point in the processing chains of the normal and the implanted ear.

Speech is passed through a slow levelsetting AGC at the input in order to limit the operating dynamic range. The cochlear model consists of a 12-channel filterbank, instantaneous compressors with halfwave rectification, and neural adaptation. The filters are wide compared to high resolution models such as [4], but the small number of electrodes combined with the dramatic current spread, which will be described further, does not allow any better resolution and hence a higher number of channels would make little sense.

The filterbank has a cascade-parallel structure yielding high frequency slopes of 100dB/oct and much shallower low frequency slopes. A log compressor in combination with a halfwave rectification is used to model the hair cell nonlinearity and is followed by a low-pass function representing the haircell membrane time constant($\approx$1msec). These nonlinearities are parametrized to fit the Russell and Sellick measurements of hair cell potentials[5]. The broadband filters and the limited initial dynamic range eliminates the need for coupling the adaptation between channels, and single channel AGC's are considered a fair approximation.

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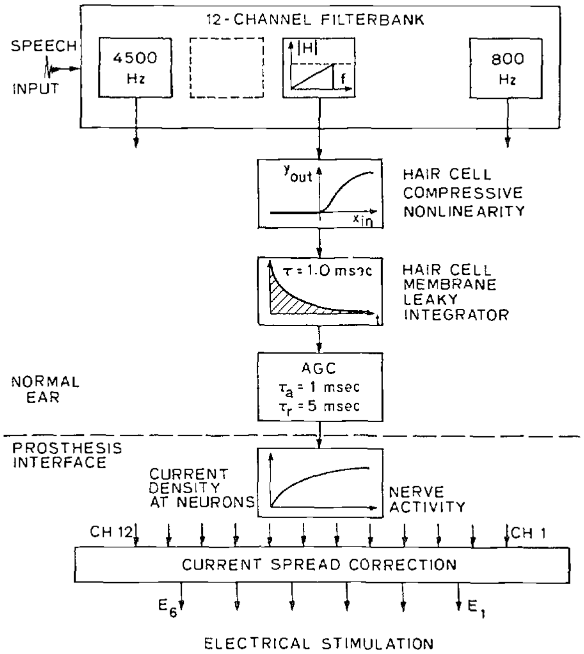

Fig. II : 12-channel Cochlear Model

2. From Neural Activity to Current Density

Single nerve fiber behavior in response to electrical stimulation is exponential above threshold[6]. We are, however, always stimulating large populations of fibers at once with a wide range of thresholds. For small currents the total activity will reflect the number of active fibers stimulated rather than the response function of a single fiber. As electrical current density increases most of the fibers in a given area are active and the activity function will start following the exponential behavior of the single nerve fibers.

It is assumed that the neural activity function is closely related to the loudness perceived by the implant patient, which exhibits an exponential behavior as well. The precise shape of this function is patient dependent, so it must be derived with the help of psychoacoustical experiments.

3. The Current Spread Problem in the Cochlea

The biggest problem with a multichannel cochlear prosthesis may well be the current spread in the cochlea.
The electrodes are resident in the low resistivity fluid of the scala tympani and are separated from the nerve endings by the bony wall of the cochlea which has a much higher resistivity. Hence the current patterns due to the stimulation of any electrode, seen on the neural side of the bone, are very broad and as much a function of the variable resistivity of the bone as of the place of the electrode (Fig. III.)

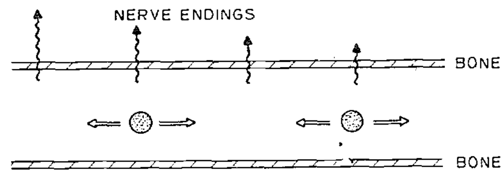

Fig. III : Current Spread in Cochlea

Psychoacoustical experiments were used by Neil Cotter to estimate these current spread functions for our patients[7]. He proposes a "vector sum model " in which the medium through which the currents pass is assumed homogeneous but in which the electrodes appear to be at a much larger distance from the nerve endings than they physically are. This model does not attempt exact physical modeling of the current spread but fits most of the available psychoacoustical data to within measurement accuracy.

The vector sum model has an easy mathematical formulation. If 'u' is a stimulation vector (currents at the electrodes) and then :

$$\begin{bmatrix} v_x \\ v_y \end{bmatrix} = \begin{bmatrix} xcnv \\ ycnv \end{bmatrix} \times u \qquad \text{(Ia)}$$

$$v(i) = \sqrt{v_x(i)^2 + v_y(i)^2} \qquad \text{(Ib)}$$

In our application the dimensions will be 6x1 for u (6 electrodes), and 12x1 for v (interfacing with the 12 channel model). In this formulation it is implicit that only the magnitudes of the current densities matter for neural stimulation and not their vectorial components.

The high relative value of resistance of the bone compared to the intracochlear fluid is bound to give large variances in current spread depending on the patient and the implanted hardware. For example, an implant technique or different hardware which replaces a higher proportion of the intracochlear fluid with a non conductive harmless medium would change the relative resistivities and hence would modify the current spread patterns drastically.

4. Coding and Current Spread Correction

As coding of the output of the cochlear model on the electrodes we propose to match the desired current densities at the neural excitation sites as close as possible at any given moment, using a least squares approximation. This is a purely spatial coding scheme, in which the time dependency does not enter past the last step of the computational model of the normal cochlea.

Given a desired set of magnitudes of current densities 'w', we must find $\hat{u}$, yielding $\hat{v}$ such that $|\hat{v}-w|$ is minimal. There is no simple mathematical solution because of the square roots in (Ib). However we know that the y-component in v is strongly dominating, hence we can replace the original problem with an alternative and much simpler one : instead of minimizing $|\hat{v}-w|$, minimize $\left| \begin{bmatrix} \hat{v}_x \\ \hat{v}_y \end{bmatrix} - \begin{bmatrix} 0 \\ w \end{bmatrix} \right|$, which reduces the problem to a linear least squares problem with solution :

$$\hat{u} = \begin{bmatrix} xcnv \\ ycnv \end{bmatrix}^+ \times \begin{bmatrix} 0 \\ w \end{bmatrix} \qquad \text{(II)}$$

in which $\begin{bmatrix} xcnv \\ ycnv \end{bmatrix}^+$ denotes the pseudo inverse of $\begin{bmatrix} xcnv \\ ycnv \end{bmatrix}$.

The current spread matrices obtained from the psychoacoustic tests have very high condition numbers, as a result of the wide current spread patterns (on the order of 10^3). This results in very large currents of opposite signs as solution of (II) and equally implies very high sensitivity to measurement error in the current spread functions. Hence in order to be able to apply this method, one must fine tune it to each patient's characteristics with a lot of care.

For some of the patients the measurement error and day to day variability can be too large to implement this method in a sensible way. Because of the ill-conditioning of the problem it is imperative for patient safety to use substantial underestimates of the measured current spread functions, as a slight overestimation might cause local peaks in the current densities which exceed expectations by far[1].

5. Modulation

In neural stimulation, charge balance, even over short periods of time, is necessary in order to avoid bone growth. Hence charge balance has to be imposed on the waveform. We use a modulation scheme such as the one proposed in [8], in which the values given by (II) are used as amplitudes of biphasic rectangular pulses.

PRACTICAL LIMITATIONS

The presented coding scheme runs into substantial practical problems as stated above, or results in conflicts with the goals of the cochlear model.

The sampling rate at the output of the cochlear model should not be lower than 4kHz if one wants to preserve a basis for neural synchrony up to 1.5-2kHz, as measured in a normal ear. This limits the total pulse duration to 250μsec (or 125μsec per phase). The maximum amplitudes, as protection against electrolysis, that we are using is 250μA.

The currents asked for by the least squares solution are often an order of magnitude larger than this and even larger.

EXPERIMENTS

We were unable to implement this scheme with the current spread corrections described in (II) because of the hardware constraints mentioned above. One of the most straightforward compromises, lowering the sampling rate and widening the pulses, gives only little help as lowering the sampling rate by half only reduces the current requirements by roughly 1.2. So extremely optimistic current spread functions as further compromise were used. It was assumed that this would yield results of similar quality as implementing the full least squares problem with inequality constraints on the solutions.

With the first patient(AS) it was necessary to underestimate greatly the current spreading (about 50%), even at a sampling rate of 1kHz, in order to obtain sounds at comfortable loudness. For the second patient(LJ) less compromises were necessary. We settled for a sampling rate of 2kHz and a 30% underestimate of the current spread functions. Even at this amount of underestimation multiple peaks in the current density patterns at the neurons will rarely be obtained. With both patients we were unable to improve significantly over previous speech processors, using these particular compromises.

CONCLUSIONS

The current spread observed with our type of electrodes makes the coding for a cochlear prosthesis based on a computational model from the cochlea a numerically ill-conditioned problem. Highly accurate measurements of the current spread function are required if one wants to apply the presented coding scheme to a cochlear prosthesis, an accuracy which we may not have attained yet. It is, however, likely that the current requirements asked for by this method will largely exceed the hardware capabilities and safety limitations.
A number of compromises will be necessary for any coding scheme using the output of a fairly complex cochlear model as input data. Lowering the sampling rate and underestimating the real current spread pattern which were tried as a first order approximation did not yield positive results.

We are convinced however that the model presented will serve as basis for future work in which an optimal coding scheme based on other compromises will be derived. It is likely that the time dependency will have to be included as the approximation at any given instance of intensity patterns is too poor. For example, one could attempt to recreate multiple peaks in the electrical current density patterns at the neurons by using a multiplexing coding scheme in which at any given moment one attempts to present only a single frequency peak. But this is only one of the many possible coding options which could be tried.

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