First Quarterly Progress Report

September 27 through December 27, 1985

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Speech Processors for Auditory Prostheses

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CONTENTS

I. Introduction

The purpose of this project is to design and evaluate speech processors for auditory prostheses. Ideally, the processors will extract (or preserve) from speech those parameters that are essential for intelligibility and then appropriately encode these parameters for electrical stimulation of the auditory nerve. Work in the present quarter was a continuation of our previous efforts to develop speech processors for auditory prostheses under NIH contract N01-NS-2356. The major activities of this first quarter of the new contract (N01-NS-5-2396) included the following:

1. Tasks related to our first implant operation at Duke University Medical Center (DUMC) on October 24;

2. Subsequent testing of this implant patient, who was fitted with a percutaneous cable for extensive psychophysical and speech-perception studies;

3. Development and testing of new computer programs to support and extend such studies; and

4. Presentation of project results at the Neural Prosthesis Workshop, the 8th annual IEEE-EMBS Meeting, the annual ACEMB Meeting, and at Surgical Grand Rounds, DUMC.

In this report we will outline the psychophysical and speech-perception tests conducted with this first patient (SG) at DUMC and describe several of the major findings. In general, the results from the speech-perception tests were disappointing but consistent with SG's inability to (a) discriminate loudness-balanced stimuli delivered to different channels in her electrode array or (b) discriminate large changes in the frequencies of sinusoidal and pulsatile stimuli delivered to single channels in her electrode array. Although SG's onset of deafness at age 5 was postlingual, she had no recollection of sound and could not rank her electrically-evoked percepts along normal auditory dimensions of pitch or sharpness/dullness. We now believe, for reasons stated elsewhere in this report, that even though SG's onset of deafness was postlingual, the normal maturation of her central auditory system may have been interrupted at a critical stage. A compromised central system is certainly consistent with SG's inability to perceive well-represented peripheral stimuli as sound. If this idea of central dysfunction is correct, then results from SG's case may provide important new insights on the timing of cochlear implants in children and on the likely significance of the "critical period" for auditory input.

How

II. Description of the Patient

As mentioned in the Introduction, SG lost her hearing at age 5. The cause of deafness was meningitis. She was 19 at the time of the first implant operation to install her electrode array and percutaneous cable. When the round window and basal portion of the scala tympani were opened for insertion of the electrode array, many fine bony spiculae were observed, consistent with SG's etiology of meningitis. These structures did not impede insertion of the electrode array; nor did they appear on CT scans performed prior to the surgery. The electrode array was fully inserted to a depth of approximately 25 mm, and the electrical impedances of each electrode pair and each single electrode were measured. All impedances fell within the normal ranges for bipolar and monopolar configurations. Recovery from the surgery was uneventful, and testing with the percutaneous cable was begun after a two-week period for healing.

III. Outline of Psychophysical Tests and Brief Summary of Principal Findings

An intensive series of tests was conducted with patient SG during the months of November and December, 1985. A complete log of these tests is presented in Appendix 2. Briefly, the psychophysical tests included measures of the following:

1. Thresholds for a wide range of stimuli and electrode coupling configurations. The stimuli were (a) charge-balanced biphasic and "monophasic" pulses (where charge balancing for "monophasic" pulses was achieved by high-pass filtering) with durations of 0.1, 0.2, 0.3, 0.5, 1.0, 2.0, 4.0 and 8.0 msec/phase; (b) bursts of sinusoids with a burst duration of 200 msec, rise/fall times of 5.0 msec, and sinusoidal frequencies of 62.5, 125, 250, 500, 1000, 2000 and 4000 Hz; (c) continuous sinusoids at 100, 400, 1000 and 4000 Hz; and (d) single-cycle sinusoids, 300 microsec/phase, presented at the rate of 10 Hz. Thresholds were obtained for all these stimuli for all radial bipolar pairs of the electrode array, and for selected stimuli for all monopolar electrodes (referenced to a remote electrode distant from the cochlea);

2. Maximum comfortable loudnesses (MCLs) for most of the stimuli and coupling configurations listed in point 1 above;

3. Channel interactions, as determined by a loudness-summation paradigm in which loudnesses were measured and compared for in-phase and out-of-phase pulses delivered simultaneously to the two channels under study (a large difference in loudnesses between phase-reversal conditions indicated substantial interaction between channels). The pulses used were charge-balanced biphasic and "monophasic" pulses, 300 microsec/phase. All channel combinations were studied;

4. Temporal channel interactions, as determined by a modification of the procedure indicated in point 3 above, in which loudnesses were compared for in-phase biphasic and "monophasic" pulses delivered simultaneously and with relative delays to the two channels under study ("temporal release from channel interactions" was indicated by the relative delay at which stable loudness reports were obtained for that delay and greater delays). The pulses used were charge-balanced biphasic and "monophasic" pulses, 300 microsec/ phase. All channel combinations were studied;

5. Temporal integration, both within and across channels, as determined by the thresholds of pairs of pulses, where the time between pulses was vaired between 0.0 and 8.0 msec (the time between pulses at which threshold began to decline most-likely reflected the time constants of temporal integration at the membranes of stimulated neurons in the excitation field). The pulses used were charge-balanced biphasic and "monophasic" pulses, 300 microsec/phase. Alternate bipolar pairs of the electrode array were included in the studies of temporal integration;

6. Compensation of temporal integration and temporal channel interactions, as determined with the procedure outlined in point 5 above, except that the stimuli were high-pass filtered at 100, 200, 400 and 800 Hz, to "compensate" for the low-pass characteristic of passive integration at neural membranes (decreases in the differences between thresholds at long pulse separations and at short pulse separations indicated the degree of compensation of temporal integration). The apical-most bipolar pair of the electrode array was used for the studies of compensation of temporal integration;

7. Channel discrimination, as determined by differences in percepts evoked by loudness-balanced stimuli delivered to different bipolar pairs in the electrode array. The stimuli included (a) single sine-wave cycles, 300 microsec/phase, presented at the rate of 10 Hz; (b) charge-balanced biphasic pulses, 0.3, 1.0 and 2.0 msec/phase; (c) bursts of charge-balanced biphasic pulses, where the burst duration was 200 msec and rise/fall times were 5.0 msec, and where the pulse duration was 1.0 msec/phase; and (d) bursts of sinusoids, where the burst duration was 200 msec and rise/fall times were 5.0 msec, and where the sinusoidal frequencies were 50, 100, 200 and 2000 Hz. Most radial bipolar pairs were used in the studies of channel discrimination;

8. Frequency discrimination, as determined by differences in percepts evoked by loudness-balanced stimuli delivered to single bipolar pairs in the electrode array. The stimuli included 200 msec bursts of pulses and sinusoids, as in point 7 above. Most radial bipolar pairs were used in the studies of frequency discrimination;

9. Temporal discrimination, as determined by the temporal separation at which two pulses were perceived as two sounds instead of one sound;

10. Loudness and loudness matching for balanced biphasic pulses of various intensities and durations, where the pulses were delivered to alternate bipolar pairs in the electrode array; and

11. Extinction times, as determined by the time at which a suprathreshold, continuous stimulus would fade into inaudibility after initial application of the stimulus. The stimulus used was a 100 Hz train of single-cycle sinusoids, 300 microsec/phase. Several radial bipolar pairs were included in the studies of extinction times.

In addition to the psychophysical studies outlined above, measurements of the electric field patterns produced by SG's intracochlear electrode array were made for all monopolar and radial bipolar configurations, at the frequencies of 100, 400, 1000 and 4000 Hz. Finally, initial attempts at recording "intracochlear evoked potentials" were made, mainly to refine artifact-rejection circuitry built into our recording apparatus. Several problems were identified in this last effort, and design changes in the artifact-rejection circuitry are being implemented for studies with the next implant patient.

Because analysis of the huge volume of data from the psychophysical and field-mapping studies indicated above is not complete at this time, we will restrict our presentation in this report only to major findings from the psychophysical studies that bear directly on interpretation of SG's poor results in tests of speech recognition. Complete expositions of the psychophysical and field-mapping studies will appear in future quarterly reports.

First, threshold and dynamic range measurements indicated good peripheral nerve survival for SG over two pairs of electrodes (pairs 1-2 and 11-12) and poor or patchy survival over the remaining six pairs. Channel interactions between pairs of electrodes were generally severe, with only several combinations of channels exhibiting good isolation. Next, and perhaps most importantly, SG's percepts for simple stimuli such as sinusoids and pulse trains were not described as tone-like in character. Manipulations of sinusoidal or pulse-train frequency would change the quality of the percept from "bumpy" to "rough" to "smooth", depending on the direction and extent of frequency change, but the difference limens (DLs) for these changes were <u>much</u> larger than the DLs for frequency in "normal" cochlear-implant patients.

Also, alternation of channels to which loudness-balanced stimuli were delivered did not usually produce a difference in percepts that would allow SG to rank her electrodes. That is, even though some electrode pairs were well isolated from others (as demonstrated in the measurements of channel interactions), the percepts evoked by stimulation of any combination of pairs in the array were essentially indistinguishable. "Normal" implant patients (and even highly abnormal ones like patient LP, described in our 7th QPR for NIH project N01-NS-2356) would have had no difficulty in distinguishing percepts produced by stimulation of such well-isolated pairs.

Finally, the measures of temporal discrimination, as described in point 9 above, indicated thresholds of approximately 100 msec. These thresholds are much greater than the thresholds we have measured for other implant patients with good speech-perception results (e.g., the thresholds for patient EHT in the UCSF series were around 5-10 msec). The Hochmairs have indicated that performance on a similar test of temporal gap detection is highly correlated with subsequent performance in tests of speech understanding (with the Vienna speech processor, see Hochmair and Hochmair-Desoyer, 1985). These investigators further suggest that abnormally-high thresholds of temporal gap detection may indicate central dysfunction.

IV. Tests of Speech Perception

The results of studies using processed speech tokens were most disappointing. None of the percepts elicited by such stimuli was described as speechlike by SG; instead, the percepts were like those of indistinct "bumps" that coincided with stimulus presentations. Moreover, changes in the processing strategy did not produce the huge changes in speech perception results we saw with patient LP (QPR 7, NIH project N01-NS-2356). Specifically, SG would report only small qualitative changes in the "bumps" when fundamental changes were made in the processing strategy.

With these disappointing results in mind, we decided to give SG a take-home processor that implemented the "compressed analog outputs" strategy of the present UCSF/Storz cochlear-implant system. Our thought was that vigorous stimulation of her central auditory system through daily use of the speech processor might "reaffirm" long-dormant neural pathways and connections. Also, we thought that practice with the device, and correlation of electrically-evoked percepts with the visual cues of lipreading, might help SG enter the "speech mode" of auditory perception.

Unfortunately, use of the speech processor during the percutaneous-cable phase did not produce any measurable benefit. Subsequent use of the UCSF/Storz processor with the transcutaneous transmission system (after SG's second operation to remove the percutaneous cable and install the transcutaneous transmission system) <u>did</u> produce small improvements in aided lipreading scores over a one-month period. In addition, there were sporadic occasions where SG could make channel discriminations. These "sparkles" of improvement may reflect learning effects obtained with consistent use of the implant, possible changes in the "connectivity" of central auditory pathways, or both. We hope, of course, that substantial improvements will occur with continued use of the device. We will be monitoring SG's performance at approximately 1-month intervals for the next 6 months to chart her progress (monitoring will be primarily accomplished with selected tests of the minimal auditory capabilities battery). If clear improvements are found, we will evaluate alternative speech-processing strategies with the aim of realizing further improvements.

V. Concluding Remarks

The picture that emerges from our findings with patient SG is one of central dysfunction. Specifically, we know we can represent speech and other information at her largely-intact periphery, but, unlike other implant patients we have studied, she can't use this information to perceive the auditory environment. One possibility is that normal development of her central auditory system was arrested at a critical stage. Indeed, results from animal studies suggest that the central auditory system may not be fully developed until adolescence. If these results hold for humans, and if we are correct in ascribing SG's difficulties to an undeveloped central auditory system, then patients with histories like SG are unlikely to receive significant benefits from cochlear implants (at least in the early post-op period). Therefore patients who lost their hearing at an early age, and have had a long (several years or more) intervening period prior to a proposed implant, should probably be viewed as poor candidates for the procedure. On the other hand, a cochlear implant may "save" the hearing of an early-onset victim if the device can be applied in time to provide a reasonable continuity of auditory input. This, of course, is a prime argument for cochlear implants in children.

In summary, our findings in tests with patient SG suggest the following:

1. A largely-intact periphery does not guarantee success even with the best of cochlear implant devices;

2. Early loss of hearing combined with a long intervening period prior to implant may constitute an unfavorable history for good use of an auditory prosthesis; and

3. Implantation shortly after detection of an early loss of hearing may be absolutely critical to the maintenance of normal functions in the developing central auditory system of a child.

VI. Plans for the Next Quarter

Our second implant patient at DUMC is scheduled for her first surgery (to install the electrode array and percutaneous cable) on Feb. 22, 1986. The major activities of the next quarter will be (a) preparation for this event and (b) subsequent testing of the patient. In addition, Wilson will be presenting a paper, "Latency Fields in Electrically-Evoked Hearing," at the 9th Annual ARO Meeting in early February.

VII. References

Hochmair, E. S. and Hochmair-Desoyer, I. J., Aspects of sound signal processing using the Vienna intra- and extracochlear implants. In R. A. Schindler and M. M. Merzenich (Eds.), Cochlear Implants, Raven Press, New York, 1985, pp. 101-110.

Appendix 1

Summary of Reporting Activity for the Period of
September 27 through December 27, 1985,
NIH contract N01-NS-5-2396

The following major presentations were made in the present reporting period:

Wilson, BS: Speech processors for auditory prostheses. Presented at the Neural Prosthesis Workshop, November, 1985.

Finley, CC and BS Wilson: Models of neural stimulation for electrically evoked hearing. Invited paper presented in the special session on neurostimulation, ACEMB Meeting, Sept. 3--Oct. 2, 1985.

Wilson, BS and CC Finley: Speech processors for auditory prostheses. Invited paper presented in the special session on signal processing for the hearing impaired, IEEE Bioengineering Conf., Sept. 27-30, 1985.

Finley, CC and Wilson, BS: A simple finite-difference model of field patterns produced by bipolar electrodes of the UCSF array. Invited paper presented in the special session on cochlear implants, IEEE Bioengineering Conf., Sept. 27-30, 1985.

Farmer, JC, Jr., Kenan, PD and Wilson, BS: Cochlear implants. Presented at Surgical Grand Rounds, Duke University Medical Center, November, 1985.

Selected abstracts from these presentations are included in the remainder of this Appendix.

MODELS OF NEURAL STIMULATION FOR ELECTRICALLY EVOKED HEARING

C.C. Finley and B.S. Wilson

Design of advanced speech processors for multichannel auditory prostheses requires detailed knowledge of the mechanisms of electrical neural stimulation occurring within the cochlea. Such knowledge will afford the opportunity to optimize the coding of speech information by gaining precise control over the firing patterns evoked among fibers of the VIIIth nerve. Many factors are known to contribute to the overall characteristics of the "electrical-to-neural transformer" that links stimuli delivered to intracochlear electrodes to discharge patterns in the auditory nerve. These factors include the physical locations, dimensions and electrical characteristics of the electrodes, as well as, the physiological integrity and survival patterns of the remaining neural elements. In addition, to achieve successful encoding of speech on the VIIIth nerve by electrical stimulation, it is probably necessary to control the temporal and spatial profiles of neural discharge around each electrode or electrode pair while avoiding and/or exploiting field interactions between the electrodes. To evaluate the relative significance of each of these contributing factors, we have developed a series of computer-based models which describe the events of neurostimulation within the cochlea.

The first model describes the electrical field patterns within the cochlea, as a consequence of stimulation by electrodes within scala tympani. The objective of this model is to provide estimates of the profiles of potentials along the loci of surviving neural elements. These potentials are calculated by an iterative, two-dimensional, finite difference model of a cochlear cross section, which includes a pair of electrodes in the scala tympani. Grid points in the model are 20 microns apart and resistivities linking the grid points are defined according to published values for resistivities of tissues and fluids appearing in the cross section. The bipolar electrodes are defined as equipotential conductors mounted in an insulating carrier medium. Fixed voltages are assigned to each electrode and the resultant field patterns are computed by iteration for the entire cross section. Potential levels at points along the loci of the VIIIth nerve elements are extracted from the final field calculation. A second version of this model, containing the spiral of the cochlea, compressed into two dimensions, is used to estimate the interaction and crosstalk at a single neural element for stimulation of two or more electrode channels. Both models provide field pattern estimates that correlate well with published data from animal experiments.

A second model is a lumped-element description of an electrically—stimulated, myelinated neuron. Stimulus inputs for the model are the potential profiles calculated in the field potential models described above. This model is a modification of McNeal's axon model (IEEE Trans. BME 23: 329-337, 1976) of resistively-linked Frankenhauser-Huxley nodes. The modified model includes myelinated axon cable properties and uses mammalian node of Ranvier characteristics instead of the characteristics for Frankenhauser-Huxley frog nodes. Eighteen active nodes are included, each separated by ten myelinated segments. One section includes characteristics of a cell body, resembling the bipolar cells of the cochlea. A system of simultaneous, nonlinear differential equations is solved iteratively to calculate the model's response to any arbitrary stimulus waveform, applied as a voltage profile along the entire length of the axon. The neuron model, in conjunction with the field potential models, constitutes an integrated model of single fiber behavior in the electrically-stimulated cochlea.

However, speech encoding in the cochlea requires successful temporal control of an ensemble of neurons spatially distributed along the cochlear partition. A third model has been developed to describe ensemble responses of multiple neurons to stimulation by multiple electrodes. This model allows manipulation of field patterns and channel configurations of the electrodes, along with response characteristics and survival patterns of the neural elements. "Latency profiles" are calculated which show timing of neural firing as a function of both the location of the element along the cochlea and the stimuli and electrode channel configurations used. Initial studies indicate the occurrence of abrupt discontinuities in response fields as a consequence of electrode polarity and neural survival. These response discontinuities complicate the design of speech processors which seek to replicate firing patterns of the normal cochlea in response to speech stimulation.

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(KEMAR). Using a pseudorandom noise input, the phase characteristic of the transfer function at a high volume control setting was obtained with an FFT-based spectrum analyzer as well as the group delay through the hearing aid *in situ*. These measurements were repeated with a phase compensator designed to improve the linearity of the phase characteristic. The effect of the phase compensator was to flatten the group delay through the hearing aid which may result in improved speech perception for hearing impaired persons.

Speech Processors for Auditory Prostheses

BLAKE S. WILSON AND CHARLES C. FINLEY

In this presentation, we will describe strategies for coding speech signals for auditory prostheses. Two problems will be considered: 1) the "classic" problem of extracting parameters from speech that are essential for intelligibility, and 2) the problem of transforming these parameters into electrical stimuli that will produce patterns of neural activity that are perceived as intelligible speech.

Speech Processing for Cochlear Implants

E. L. V. WALLENBERG, I. J. HOCHMAIR-DESOYER, AND E. S. HOCHMAIR

Open-list sentence understanding without lipreading can be achieved by single-channel analog broad-band electrical stimulation by 75 percent of the postlingually deafened patients equipped with a Vienna cochlear prosthesis. Based on the analysis of vowel identification tasks, alternate speech coding strategies using feature extraction have been developed and evaluated.

Session C13—Cochlear Prostheses

Implications of Speech Encoding in the Normal Cochlea for a Cochlear Prosthesis

BEN M. CLOPTON

In addition to established analysis and transduction processes in the normal cochlea, significant nonlinearities and, possibly, extracellular electrical currents influence activity in the auditory nerve. Enhancement of specific features in sound waveforms, especially speech, occurs. These processes imply strategies for electrically encoding speech information for a cochlear prosthesis.

Current Spreading and Current Deconvolution in Scala Tympani Prostheses

DIRK VON COMPERNOLLE AND ROBERT L. WHITE

Multielectrode cochlear arrays have the potential to create, at the neurons, a current density representing the complex energy density spectrum of speech. For present devices, the desired current pattern is placed on the electrodes and degraded by current spreading to a blurred representation at the neurons. If the current spreading function is known, however, so is its inverse; and it is possible, using "current deconvolution," to compute the current pattern required at the electrodes to produce the desired pattern at the neurons. This paper will deal with the potential and the limitations of current deconvolution techniques as applied to scala tympani cochlear prostheses.

A Simple Finite-Difference Model of Field Patterns Produced by Bipolar Electrodes of the UCSF Array

C. C. FINLEY AND B. S. WILSON

A finite-difference model of field patterns produced by the UCSF multichannel bipolar electrode array is presented. The model describes the array as a spiral, compressed into a two-dimensional, homogeneous plane. Potential levels along radially directed dendrites of spiral ganglion cells are described. Comparisons between of measured animal data and model predictions are presented. Implications for channel interaction phenomena are discussed.

Mixed Boundary Value Problems in the Implanted Cochlea

J. T. RUBINSTEIN, M. SOMA, AND F. A. SPELMAN

This paper describes a three-dimensional analytical model of a finite-sized, cylindrical, segmented electrode array. The assumptions made are that the fields will be calculated at a distance much smaller than the length of the carrier, the carrier is a perfect insulator, and the electrodes are perfect conductors.

A Peak-Counting Model for Single-Electrode Cochlear Stimulation

LES ATLAS

A peak-counting model has been developed to predict the pitch and timbre perception for complex single-electrode electrical stimulation waveforms. This model has been studied for the case of amplitude modulated biphasic pulses and for more complex analog waveforms. The model has also been utilized to suggest speech processing algorithms.

Behavioral Responses to Intracochlear Electrical Stimulation: Possible Peripheral-Nerve Mechanisms

MARK WHITE

The rate at which neural activity increases with stimulus intensity is a function of fiber diameter. In addition, the intraaxonal resistance between nodes of Ranvier is a strong function of fiber diameter. These two primary factors may cause cochlear fibers of different diameter to be selectively excited depending on the stimulus level, the stimulus waveform, and distribution of surviving nerve.

Session D12—Electrical Control of Arrhythmias

Computer Algorithms for Tachycardia Detection in Antitachycardia Pacing

THOMAS E. BUMP

We have developed sensitive and specific decision rules for tachycardia identification which use rate in both chambers and degree of atrioventricular association to diagnose among tachycardias which do not have 1:1 atrioventricular relationship. Our technique delivers an atrial extrastimulus to differentiate sinus tachycardia from other

Appendix 2

Log of Studies Conducted with Patient SG,
November and December, 1985

Studies Conducted with Patient SG

date	tests
11/6	electrode impedances; thresholds and MCLs to continuous sinusoids at 100, 400, 1000 and 4000 Hz, obtained with manual manipulation of the analog oscillator. Also, reports of percepts were obtained; e.g, "high" sensations, "dizzy" sensations, etc. Frequency sweeps were made to probe changes in percepts with changes in frequency. Only bipolar pairs were used in the tests.
11/7	electrode impedances; further exploration of stimulus space with biphasic single sine-wave cycles, 300 usec/phase. SG was asked to rank percepts along loud/soft and sharp/dull dimensions. This lead to reports of "choppy," "smooth," "dull," "sharp," "weak," and/or "strong," which bore complex relationships to the rep rates and intensities of the stimuli. Finally, threshold data were obtained for single monophasic, triple monophasic and single-cycle biphasic sine pulses at the rep rate of 10 Hz. These data were obtained by manual adjustment of the oscillator.
11/8	electrode impedances; continued studies with single-cycle analog sinusoids. Continued training in psychophysical tests and reporting. Measured extinction times to 100 Hz trains of single-cycle sinusoids. Thresholds to single-cycle sinusoids, presented at the rate of 10 Hz, were measured with single ascending runs. All bipolar and monopolar stimulus combinations were tested for one leading phase of the single-cycle sinusoids, and the reverse phase was used in additional tests of all bipolar stimulus combinations.
11/12	electrode impedances; continued exploration of stimulus space with single sine waves, 300 usec/phase. In one set of tests, loudness was balanced for two channels and SG was asked "which channel?--A or B?" Also, loudness scales were established for selected channels. Finally, preliminary tests of channel summation were performed. In addition to these psychophysical tests, field mapping of the electrode array was performed with the single-cycle sinusoidal stimulus. Amplitude values were recorded from the scope for various configurations of the electrode array.
11/13	electrode impedances; continued exploration of stimulus space with single-cycle sinusoids. Preliminary measurements of channel interactions were made in which the polarities of stimuli delivered to the two selected channels were manipulated. Changes in loudness reports indicated the approximate magnitude of interactions. Also, several tests of channel discrimination were performed. Finally, unprocessed speech sounds were applied in an attempt to provide "speech-mode" percepts.

11/14 electrode impedances; began tests with computer-generated stimuli. Thresholds and "1" levels were obtained in single ascending runs for all bipolar channels in the array. Then thresholds were obtained for the 4 types of pulses, for selected channels in the array, using the automated procedures of program NEWTWO.

11/15 electrode impedances; continued measurements of pulse thresholds with NEWTWO. Loudness levels for 300 usec/phase biphasic pulses were obtained for pair 1-2. Multipulse excitation files were then prepared for delivery to pair 1-2, using the loudness and threshold data from the above tests. Multipulse files were made for tokens BOUGHT, BIT and BOAT. They were balanced for loudness and a discrimination test was conducted. The tokens were well discriminated, but not recognized as speech (instead, they sounded like "bumps"). Finally, thresholds for monopolar pulses were obtained for the four types of pulses for selected electrodes.

11/19 electrode impedances; MCLs were obtained for .3, 1.0 and 4.0 msec biphasic pulses, for bipolar pairs. Repeat of threshold measurements for biphasic pulses was performed to confirm reliability of tests (1) after a weekend away and (2) after changing the maximum output current (1 mA) and voltage compliance (35 v) of the stimulus-isolation unit. Also, MCLs for the other types of pulses were obtained for pair 1-2. Finally, brief channel discrimination studies were conducted for pairs 11-12, 15-16 and 1-2.

11/20 electrode impedances; initial attempts at EP measurements; measurements of <u>monopolar</u> thresholds to pulses for electrodes 3 and 4. Finally, a repeat measurement of thresholds to biphasic pulses was made for bipolar pair 3-4 after the max current level had been readjusted to 500 uA.

11/21 electrode impedances; artifact-reject trials; checked threshold for biphasic pulses delivered to pairs 5-6, 11-12, 13-14, 15-16. Measured thresholds for all pulse types, for pulses delivered to the monopolar electrodes of the above pairs. Probed channel discrimination with 2 msec/phase, loudness-balanced, biphasic pulses.

11/22 electrode impedances; continued channel discrimination tests, as above, at loudness levels of "1" and "3." Also, conducted channel-discrimination tests in which the stimulus delivered to one channel was a 300 usec/phase pulse and the stimulus delivered to the other channel was a 1.0 msec/phase pulse.

11/23 electrode impedances; frequency- and channel-discrimination tests, using 200 msec trains of 1.0 msec/phase biphasic pulses. Stimuli were balanced for loudness prior to comparison in the discrimination tests.

11/25 electrode impedances; MCLs for 200 msec tone bursts, at the tone frequencies of 125, 250, 500, 1000, 2000 and 4000 Hz, for bipolar pairs.

11/26 electrode impedances; channel discrimination experiments using 2 KHz tone bursts. Comparisons were made at loudness levels of "1" and "3." Tests of frequency discrimination, using 50, 100, 200 and 2000 Hz tone bursts were also conducted. Preliminary tests of temporal integration, both within and across channels, were performed using the "channel-interaction paradigm" described in the notes for 11/26.

11/27 electrode impedances; thresholds to 62.5, 125, 250, 500, 1000, 2000 and 4000 Hz tone bursts, for all bipolar pairs. Also, fit portable speech processor for use over the weekend.

12/2 electrode impedances; channel- and frequency-discrimination tests using 200 msec tone bursts, separated by an ISI of 400 msec. Measured thresholds again for biphasic pulses delivered to pairs 1-2, 5-6, 11-12 and 15-16. Obtained equiloudness contours for biphasic pulses of various durations, for pairs 1-2, 5-6, 11-12 and 15-16. In addition, obtained equiloudness contours for the other types of pulses for pair 1-2.

12/3 electrode impedances; multipulse, temporal-integration experiments for .3 msec pulses of all 4 types, for channels 1-2, 5-6, 11-12 and 15-16.

12/4 electrode impedances; two-pulse temporal discrimination tests were conducted, for channels 1-2 and 5-6. Various pulse waveforms were used, and discrimination measures were made at the loudness levels of "1" and "3."

12/5 electrode impedances; continued temporal discrimination tests, for channels 5-6, 11-12 and 15-16. Resumed multipulse, temporal-integration experiments for 2, 3 and 4 pulse sequences for channel 1-2; and for high-resolution measures with 4-pulse sequences for channels 1-2, 5-6, 11-12 and 15-16.

12/6 electrode impedances; multipulse, temporal-integration experiments for .1 msec pulses and .3 msec pulses delivered to channel 1-2. Also, measured channel interactions using a manual procedure with NEWTWO, to reverse the polarities of pulses delivered to the test channels and to introduce small time offsets between the pulses. Finally, the speech processor was readjusted to provide more compression, etc.

12/9 electrode impedances; field-mapping measurements.

12/10 electrode impedances; field-mapping measurements; channel-interaction studies in which loudnesses were measured for in-phase and out-of-phase .3 msec biphasic and monophasic pulses, for all bipolar pairs in the array. In addition, loudnesses were measured for various delay offsets of pulses delivered to the two channels, to gauge "release from temporal summation."

12/11 electrode impedances; continued field-mapping measurements and channel-interaction studies.

12/12 electrode impedances; continued field-mapping measurements and channel-interaction studies.

12/13 electrode impedances; multipulse, temporal-integration experiments, with pulses high-pass filtered at 100 Hz, channel 1-2.

12/14 electrode impedances; evoked potential studies using first version of Eclipse sampling and averaging software; multipulse, temporal-integration experiments with pulses high-pass filtered at 200, 400 and 800 Hz, channel 1-2.