RESEARCH TRIANGLE INSTITUTE

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PART I

TECHNICAL PROPOSAL B

Speech Processors for Auditory Prostheses

Submitted in response to NIH RFP No. NIH-NINCDS-85-09.

POST OFFICE BOX 12194 RESEARCH TRIANGLE PARK, NORTH CAROLINA 27709

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*Denotes that this section and its subsections are identical in technical proposals A and B.

I. Introduction

The project outlined in NIH RFP No. NIH-NINCDS-85-09 is directed at the improved design of speech processors for auditory prostheses. Briefly, the project requires collaboration with a team or teams of investigators to evaluate processing strategies, and further requires the selected contractor(s) to (1) "design and develop a computer-based, multichannel waveform generator," which when coupled with the collaborating investigator's multichannel neural stimulator will permit basic psychophysical studies on improved methods of stimulus coding; (2) "design and develop a computer-based, multichannel auditory signal processor for use in evaluating promising speech extraction and stimulus encoding schemes;" (3) "design and fabricate wearable speech processors based on the results obtained with the computer-based simulated designs;" (4) "supply at least two of these wearable speech processors to the Project Officer by at least three years after the start of the contract;" and (5) "assist the collaborating human subject evaluation team in implementing the above mentioned waveform generator, computer-based signal processor and the wearable speech processors."

In this proposal we will first review our progress in the first 18 months of effort for NIH project N01-NS-2356, "Speech Processors for Auditory Prostheses." This project, of course, is the predecessor to the one described in the present RFP. In the current contract we at the Research Triangle Institute (RTI) have had the good fortune to establish a most productive collaboration with the excellent cochlear-implant group at the University of California at San Francisco (UCSF). The main activities of our combined team during the initial period of project effort were the following:

1. Design, build and test a hardware interface to provide a high-bandwidth communications link between an Eclipse computer and implanted electrodes;

2. Develop and evaluate an integrated field-neuron model of electrical stimulation by intracochlear electrodes;

3. Identify and contrast promising approaches to the design of speech processors for auditory prostheses, the product of which is a detailed set of plans for experiments to evaluate "stimulus primitives," single-channel coding strategies, and multichannel coding strategies;

4. Build a computer-based simulator that is capable of rapid and practical ~~evaluation~~ emulation of most of these approaches in software;

5. Conduct tests with an implant patient at UCSF to confirm proper operation of the equipment and software indicated in points 1 and 4 above, and to obtain measures of basic psychophysical performance with the UCSF transcutaneous transmission system;

6. Design and build a portable, real-time speech processor for single-channel auditory prostheses, based on use of the "AMDF" algorithm for extraction and presentation of voice pitch and voice/unvoice boundary information;

7. Help to establish a strong collaboration between UCSF, Duke University Medical Center (DUMC) and RTI, so that parallel series of tests with implant patients can be conducted in the immediate future at both UCSF and DUMC.

In all, we are proud of the progress we have made in the first 18 months of project effort. In addition to meeting fully all requirements of the contract work statement with the completion of tasks 1, 3, 4 and 6, we have been able to build a powerful tool for understanding and defining the "electrical-to-neural transformer" linking the outputs of the speech processor to the inputs of the central nervous system (task 2) and we have been able to help initiate a parallel testing effort at Duke (task 7).

Our primary goal for the work described in this proposal is to define the classes and parameters of processor design that will allow full recognition of speech without lipreading for recipients of multichannel implants in whom survival of peripheral dendrites is good. Other important objectives of the work we propose include (1) development of improved strategies for implant recipients in whom survival of peripheral dendrites and/or ganglion cells is patchy or poor; (2) development of objective tests to determine the pattern of nerve survival in implanted patients; (3) further development of portable, real-time processors to implement particularly-promising strategies in "take home" units for implant patients; (4) investigation of learning effects (mainly by the collaborating psychophysical teams) with these take-home units; (5) development of improved strategies for coding speech information with single-channel, extracochlear auditory prostheses, primarily for the safe (and hopefully efficacious) use by infants and young children; and (6) development of improved understanding of the encoding of electrical and acoustic stimuli at the auditory nerve, through the studies we will describe on "stimulus primitives."

This is the second of two proposals we are submitting in response to this RFP. The first proposal outlines a project that will meet the requirements of the RFP at approximately our present level of effort, and the second proposal outlines an "expanded-scope" project in which the RTI team will (1) increase its effort in supporting the teams at UCSF and Duke, to evaluate processing strategies and stimulus primitives; (2) help initiate and continue to support a new evaluation effort at Washington University and Central Institute for the Deaf; (3) thoroughly evaluate more alternative designs for portable, real-time processors than the necessarily limited number described in the "present-scope" proposal; and (4) provide additional take-home processors for patient use and for further evaluation of possible learning effects.

To save time for readers of these proposals, we want to mention that both proposals have the same organization and that many sections are identical in the two documents. The identical sections are marked with asterisks on the CONTENTS pages of each proposal. In the background sections of both proposals we describe tools we have developed in the first period of our present contract. These include the hardware interface for communications between the Eclipse computer and implanted electrodes; the models of field patterns and neural discharges produced by stimulation with intracochlear electrodes; and the computer-based simulator of speech processors for auditory prostheses. In the background section we also present preliminary results obtained from the field-neuron model and from the tests with the implant patient at UCSF. Finally, we describe the collaborations we have established with UCSF, DUMC and Storz Instrument Company, and we describe our work on development of portable, real-time hardware for a single-channel auditory prosthesis.

The next major section of both proposals is the "plan of proposed effort." Here, all subsections are the same with the exception of the one on further development of portable, real-time processors. The identical subsections describe (1) the design of stimulus primitives, single-channel coding strategies and multichannel coding strategies; (2) our suggested experimental plan to assist the collaborating psychophysical teams in evaluating these stimulus primitives and coding strategies; and (3) our view of the prospects for the proposed projects.

As might be expected, there are large differences in the sections on "Project Organization and Management" and "Statement of Work, Schedule and Budget." These sections present in detail the distinctions between the two projects.

Finally, the last sections, on protection of human subjects, literature references and RTI experience, are the same in both proposals.

II. Background

A. Patient Stimulator and Computing Hardware

The RTI Patient Stimulator is a flexible research tool, designed to allow great flexibility of stimulus control for speech processor experiments in cochlear implant research. The stimulator functions as an extension of the RTI Block-Diagram Compiler software, allowing full functional realization of the speech processors designed with the compiler. Its large dynamic range and bandwidth capabilities make the Patient Stimulator a transparent component in the simulation of a wide range of speech processors. Patient safety has been heavily emphasized in the design. An added feature is the ability to measure intracochlear potentials via nonstimulated electrodes within the implanted electrode array. This option has been included to allow more direct assessment of the correlations between applied electrical stimuli and elicited neural activity. A functional overview of the interface's capabilities, a review of patient safety features, and a discussion of the intracochlear potential measurement system follow below, along with a brief description of the computer hardware and software which control the patient interface.

1. Functional Overview

Designed as a transparent element in the testing of speech processor systems with patients, the RTI Patient Stimulator is a research tool for the laboratory environment. No attempt has been made to address the issues of the portability that will ultimately be required in useable patient devices. Rather, the stimulator design has focused upon maximizing the opportunity afforded by the percutaneous cable to deliver highly controlled stimuli, as well as to assess the physiological events underlying electrical stimulation within the cochlea. The stimulator also can simulate the poorer performance characteristics of hardware drivers used in current prosthesis designs, thus allowing full emulation of presently realizable systems.

Figure II-A-1 shows the three hardware units that comprise the complete

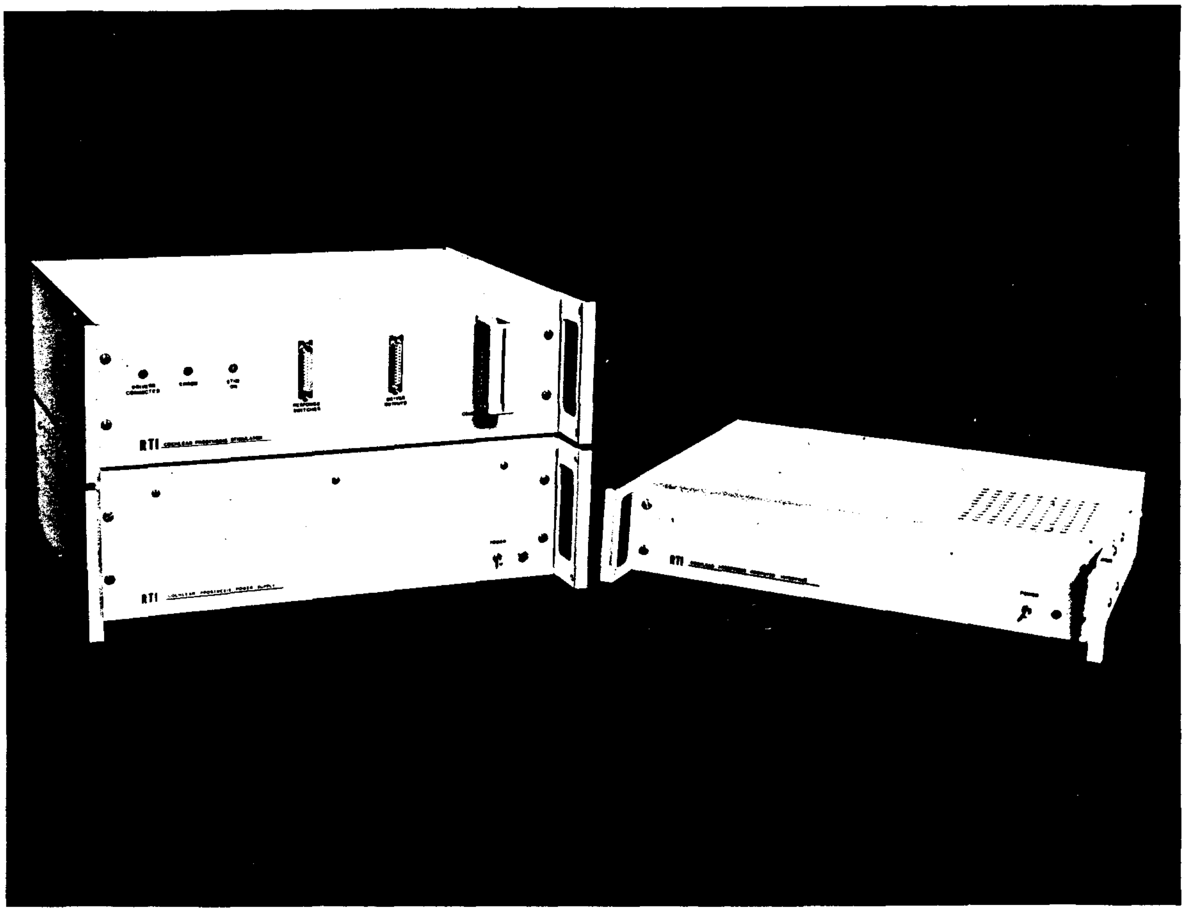


Fig. II.A.1. RTI Patient Stimulator System - On the left is the stimulator unit sitting atop its power supply unit. On the right is the logic interface unit which couples the stimulator system to its controlling computer.

stimulator system. On the left is the main stimulator unit, atop its associated power supply. These two units reside in the testing location with the patient. Three front panel connectors can be seen on the stimulator itself. On the left is the connector for a patient response panel. In the center is the connector for the cable leading to the patient's electrodes. Finally, on the right is a configuration jumper plug for establishing connections between stimulator channels and implanted electrodes. Three lights on the left end of the stimulator indicate various status conditions to the experimenter.

Up to 100 feet of parallel bi-directional logic line connect the stimulator with a logic interface unit, shown on the right in Figure II-A-1. This unit couples the stimulator's communications logic bus to the digital control unit (DCU) of a Data General Eclipse computer system. The logic interface unit is physically located with the Eclipse computer. The DCU is a separate programmable processor, capable of sharing memory with the Eclipse via data channel communications. It is connected both to the Eclipse data bus and to its own data bus. A standard parallel-bus interface card connects the DCU data bus to the logic interface unit of the prosthesis stimulator.

Figure II-A-2 is a block diagram of the computer facilities at UCSF, RTI, and DUMC that run our patient testing software. The lower portion of the diagram details the patient interface connections to each of the Eclipse computers. All three laboratory systems have been similarly configured to insure transportability of software. Horizontal dotted lines indicate communication links between the systems. Bulk data and software transfers are handled by magnetic tape, whereas daily maintenance, communication and coordination activities are handled by modem/telephone links. A more complete description of the computing facilities may be found in section IV.B., Facilities.

Figure II-A-3 is a functional block diagram of the stimulator system. The stimulator consists of six major functional subsections.

First are eight latched 12-bit multiplying digital-to-analog converters (DAC) which may be set independently under program control. Each DAC reference input may be connected to either an internal 10 V reference or an external signal source. DAC outputs are routed to the back panel of the stimulator where they are available as sources to external equipment

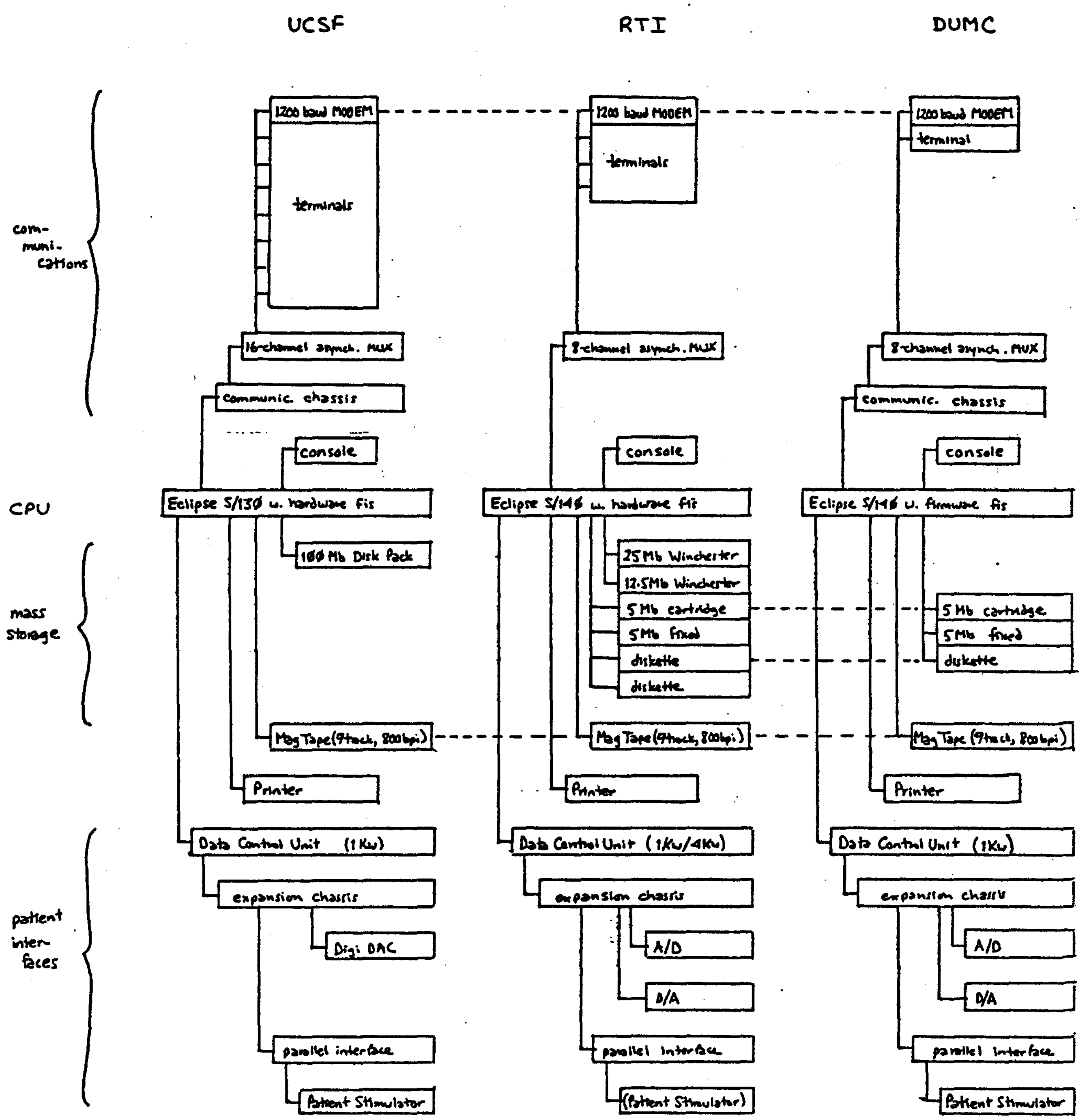


Figure II-A-2. Computing facilities.

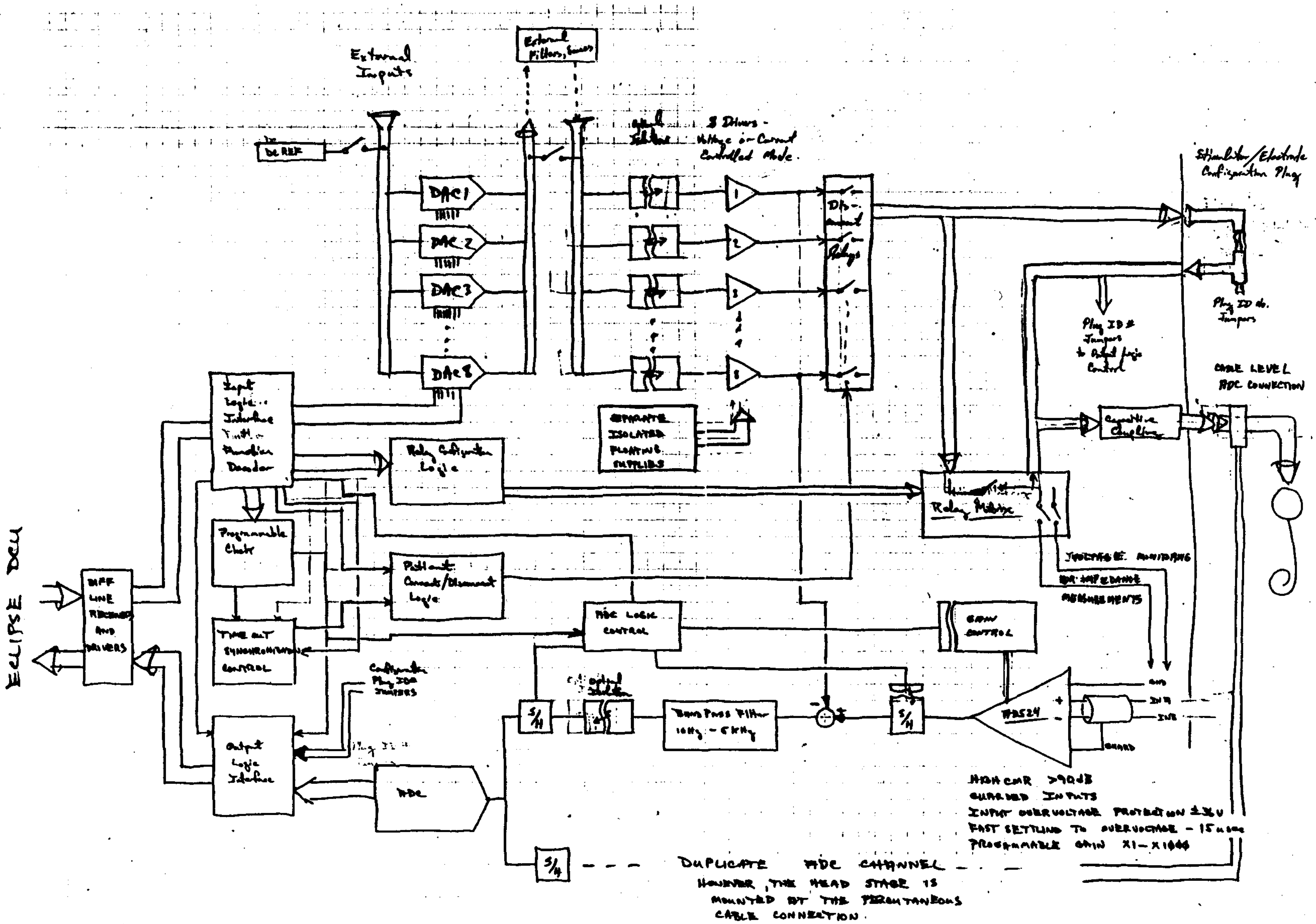


Figure II-A-3 Functional block diagram of Patient Stimulator (acoustic monitors, anti-aliasing filters, stimulus drivers for animal experiments) or are redirected as inputs to the stimulus drivers.

Second are eight high-voltage, wide-bandwidth op-amp driver stages that receive input from either the internal DAC's or external sources (anti-aliasing filters, etc.). Each driver stage is electrically isolated by a wide-bandwidth (60KHz) isolation stage. Each driver may be operated in either a voltage-controlled mode or in a current-controlled mode. Output voltage compliance is + or - 70 volts. Peak output current levels are limited to a preset level (in the range of approximately 1 to 2 mA).

Third are separate, isolated power supplies (+ and - 70 volts) for each driver stage. Grounds for each stage are maintained separately.

Fourth are two methods of connecting the driver outputs to the patient electrodes. In both cases all driver outputs are routed through a series of patient connect/disconnect relays, allowing rapid interruption of all electrical connections to the patient. These relays may be operated under program control, manually by the patient, or automatically by the interface itself when valid data ~~has~~ have failed to arrive from the computer within a specified period (typically 50 usec.). Following the connect/disconnect relays are a jumper plug and a relay matrix for configuring driver connections to the electrodes. The jumper plug provides manual selection of connections of drivers (8 total) to electrodes (17, including ground) in various configurations (such as monopolar, bipolar, pseudobipolar). The relay matrix allows configuration of electrode connections for two driver stages under program control. The latter feature facilitates electrode impedance measurements and allows speed and flexibility in channel interaction studies.

Fifth is a two-channel analog-to-digital converter system for electrode potential measurement. This analog sampling system is used in impedance measurements and intracochlear evoked potential studies. Further discussion of this subsystem is in the section below on Intracochlear Potential Measurements.

Sixth is the control logic interfacing the stimulator subsystems (described briefly above) to the DG Eclipse computer. Communications with the Eclipse computer occur via two separate unidirectional parallel buses. One carries 16 bits, plus handshaking, to the interface from the Eclipse DCU; and, the other returns 16 bits plus a clock interrupt to the DCU. Both buses are provided with line drivers, allowing the stimulator to be located up to 100 feet from the computer installation. Within the stimulator logic system is a programmable clock that synchronizes all data transfers to the stimulator system. Each data transfer is made in response to an interrupt generated by the stimulator clock. In the event that the stimulator fails to receive a valid data transfer from the computer in one interrupt clock period, the stimulator automatically disables itself, disconnects the patient, and signals the DCU that an error has occurred. Programming provisions for the stimulator are described in Appendix 1 for the interested reader.

Figure II.A.4. is a view of the main stimulator unit showing its physical construction. Six circuit boards are shown, each constructed using insulation displacement wiring techniques. Board interconnections are via ribbon cables connected at the ends of each board. The boards, as viewed from back to front, include (1) two logic circuit boards, (2) one eight-channel DAC board, (3) one eight-channel driver board with optical isolator~~s~~ stages (shown with only two channels populated), (4) one relay matrix and (5) one interconnect board with coupling capacitors to the patient electrodes. Not shown is the analog-to-digital converter board which inserts between the driver board and the relay matrix. The configuration jumper plug is on the right end of the front panel. The patient electrode cable connects to the front panel connector ~~converter~~ marked "DRIVER OUTPUTS."

A summary of the electrical and functional features of the Patient Stimulator follows:

- a total of eight stimulus channels, each consisting of a computer-controlled stimulus driver;
- each channel may function in either a voltage-controlled or current-controlled mode with + or - 70 volts output voltage compliance;
- each channel features a large-signal bandwidth of at least 60Hz to 60kHz;
- each channel functions independently, with electrically floating grounds and isolated supplies;
- each channel uses a latched digital-to-analog converter (DAC) allowing updating of stimulus channels only when stimulus magnitude changes are desired;

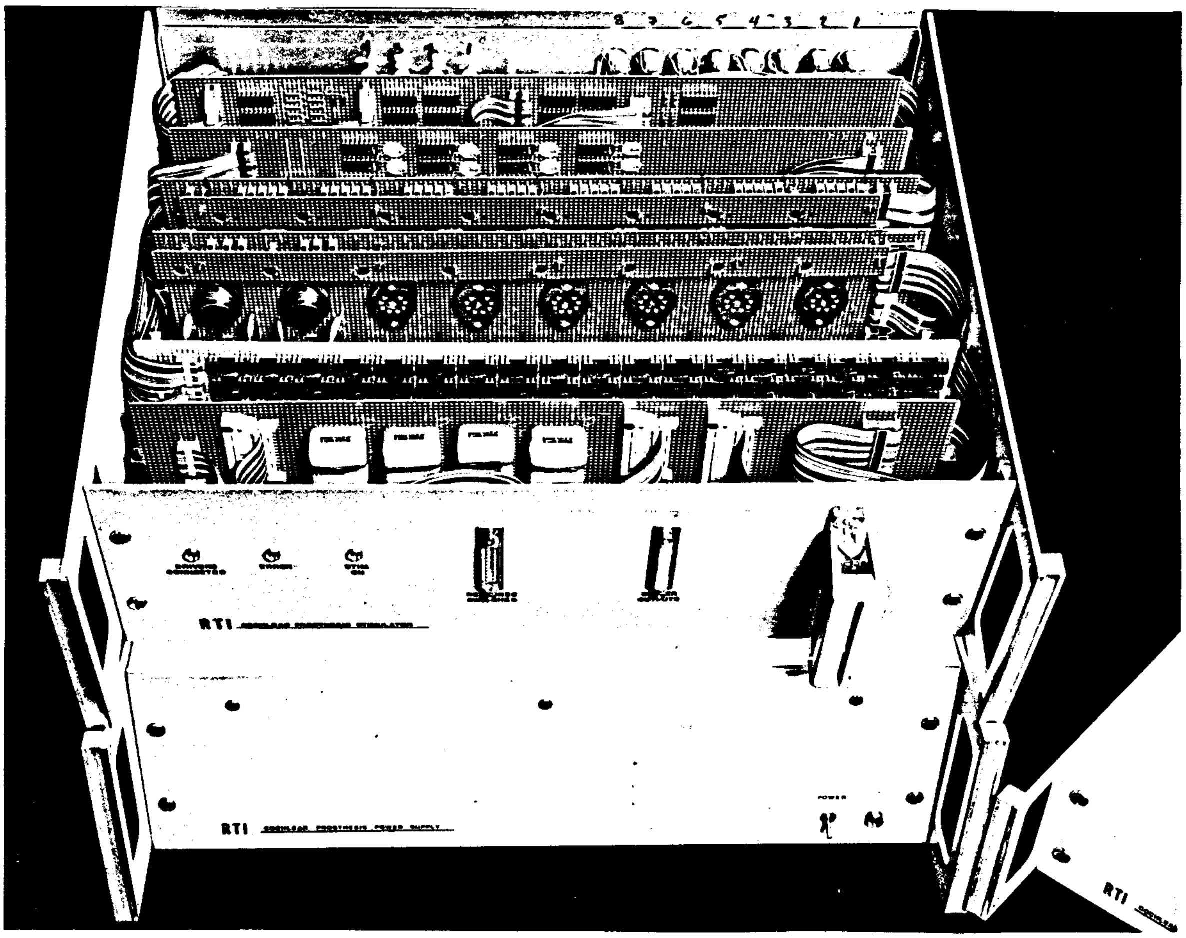


Fig. II.A.4. Patient stimulator with cover removed.

- all stimulus channel magnitude transitions are synchronized across all channels with a programmable timer built into the stimulator itself;
- electrode impedance can be measured between any two patient electrodes (when connected via a percutaneous cable directly to the electrodes);
- the patient and interface may be connected or disconnected under program control;
- connections between stimulus drivers and the electrodes may be achieved in two ways:

    First, for complete flexibilitiy under program control, two stimulus channels may be switched via a relay matrix to any combination of seventeen implanted electrodes, including a remote ground. This option facilitates characterization of all electrode impedances, as well as enabling rapid configuration of two stimulus channels for channel interaction studies.

    Second, for rapid change of stimulator/electrode configurations, connections among up to eight stimulators and up to seventeen electrodes, including remote ground, are made by a front panel jumper plug. Each configuration plug is prewired for the desired stimulator/electrode configuration. Each configuration plug is also prewired with an identification number (0-32) which may be read and verified under program control during testing. This aids in preventing errors in changing stimulator/electrode configuration during evaluation of various speech processor designs.
- patient safety features are included (these are summarized in the next section);
- a two channel analog-to-digital converter (ADC) system is included for measurement of potentials appearing at the electrodes. This system allows both electrode impedance measurements and measurements of signals from unstimulated electrodes. The latter function is intended to be used in the measurement of intracochlear evoked potentials and stimulus artifacts. (A more detailed discussion of this measurement system follows in section

3 below.)

2. Patient Safety Design Features

Patient safety design features are listed here for review:

- optical isolation of analog circuitry;
- dual output blocking capacitors of low capacitance and low leakage;
- peak limiting of voltage and current stimuli are available at preset levels;
- patient disconnect relays on each channel operate under patient control or after a preset timeout period if communications with the controlling computer are interrupted or delayed;
- the high voltage supply for each patient channel consists of a standard modular supply, driven with an isolation transformer;
- the entire unit is further isolated with a medical-grade isolation transformer on the primary power support;
- standard maximum leakage current criteria for Subject Instrumentation Equipment are met or exceeded (lead leakage to ground <50uA).

3. Intracochlear Potential Measurements

A two-channel analog-to-digital conversion system has been included within the Patient Stimulator. This subsystem serves two functions. One is the measurement of electrode impedances. The other is the measurement of potentials from nonstimulated electrode pairs during active stimulation. This latter measurement technique will be used to assess the field patterns and evoked physiological responses occurring within the cochlea in response to stimulation. Below are brief descriptions of the hardware and software used for the impedance and intracochlear potential measurements. Techniques designed to eliminate the stimulus artifact component also are discussed.

Impedance measurements are conducted in a straightforward manner by injecting a small stimulus current and monitoring the consequent electrode voltage. One channel of the analog sampling system is connected by the programmable relay matrix across any two electrodes. Simultaneously, an electrode channel driver is connected across the same two electrodes. Current is injected and the resultant potential is measured. Since all connections are under program control, measurement across many electrode combinations may be made rapidly. This process has been fully automated in software. By measuring the interelectrode impedances of at least three separate electrodes in all combinations, the impedance of each separate electrode may be calculated. Our impedance measurement software does this and presents characterizations of each individual electrode at various test frequencies and current injection levels.

Measurement of intracochlear potentials via nonstimulated electrode pairs is a more difficult task, but one that promises great rewards in understanding the physiological basis of intracochlear stimulation in an individual patient. Factors complicating these measurements include the small signal amplitude and the presence of stimulus artifact components. A variety of measures are available for use in circumventing these problems.

When making intracochlear potential measurements via nonstimulated electrodes a small op-amp head stage and switch matrix may be inserted between the patient's percutaneous cable and the cable from the patient stimulator. (Refer to the functional block diagram of the stimulator, Figure II.A.3.) The switch matrix allows the disconnection of the intracochlear electrode leads from the stimulator (thus significantly reducing capacitive loading and extraneous noise sources) and connects them directly to a differential op-amp buffer. The buffer head stage provides a high common-mode rejection ratio of at least 90 dB, input overvoltage protection up to + or - 36 volts and fast settling times (15 usec) after input overvoltages. Additionally, a programmable gain feature is included, providing preset gain settings (X1 - X1000) under program control.

Following the head stage is a sample and hold unit (S/H) that rapidly disconnects the signal input to subsequent filter stages, preventing latchup and slow recovery problems with high level stimuli. This S/H operates under program control and is typically used to disable the input signal path during high level stimulus periods. Immediately after the stimulus, the signal path is enabled and sampling is resumed. Subsequent to the S/H is a summing junction, allowing the injection of signals from an unused stimulus driver. This feature may be used as a cancellation tool in reducing unwanted artifact components during low to moderate level, continuous stimulation (for example, a sine wave stimulus).

A two pole bandpass filter (10 Hz to 5 KHz) follows for biological signal filtering. This filter also functions as an anti-aliasing filter in that the upper frequency cutoff of 5 KHz is a factor of four below the standard 20 KHz sampling rate of the ADC stage. All previous stages are powered by isolated supplies and all digital and analog signal lines are optically coupled.

A two channel 12-bit ADC completes the sampling system. All sampling and data transfer functions are controlled by the stimulator unit logic circuits. Sampling occurs synchronously with updating of stimulus driver channels, based upon the onboard programmable clock frequency.

Tests are presently being conducted with this unit to evaluate its performance in controlling artifact signals. Good performance is expected in that these techniques have been used with success in other recording situations. McGill, Cummins, Dorfman, Berlizot, Leutkemeyer, Nishimura and Widrow (1982) have recently reviewed the problem of neural recording with stimulus artifact suppression. In an experiment directly related to the intracochlear recording problem, Stypulkoski and van den Honert (1984) have recently succeeded in recording compound action potentials from monopolar electrodes in the cat modiolus during electrical stimulation.

B. Models of Electric Fields and Neural Discharge Patterns Produced by Intracochlear Electrical Stimulation

1. Overview

It is our view that the design of advanced speech processors for single and multichannel auditory prostheses is a two-part problem. First is the "classic" problem of extracting (or preserving) from speech those parameters that are essential for intelligibility. Second is the problem of transforming those parameters into electrical stimuli that will produce patterns of neural activity that are perceived as intelligible speech. Speech extraction is discussed in detail in Section III.B, Design of Single-Channel Coding Strategies, and in Section III.C, Design of Multichannel Coding Strategies. The present section describes our work on encoding the speech information into neural firing patterns.

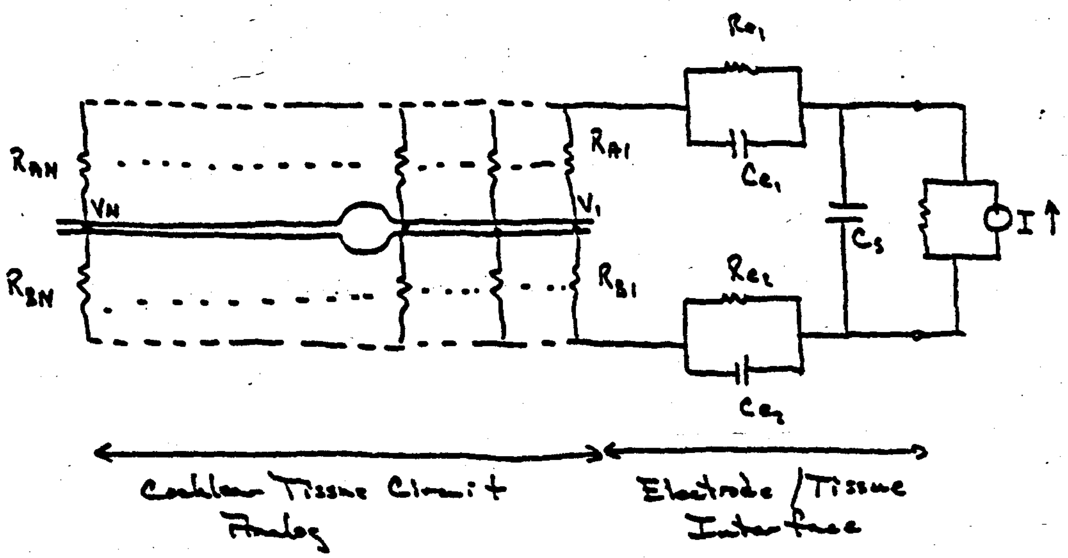
Somewhat intermediate to the the extraction and encoding tasks described above is the determination of the neural firing patterns that best mimic those patterns elicited in a normal cochlea for a similar stimulus. We consider this task part of the design of encoding strategies. Further discussion on this topic is found in Sections III.B and III.C.

Consequently, for purposes of the present discussion, we define the encoding problem to be that set of issues that limit or influence the control that can be exerted over the firing patterns of VIIIth nerve fibers by electrical stimulation. In this context, we consider the encoding task to be a weak link in the design of any speech processor for a cochlear prosthesis and one that must be dealt with in an informed manner to ensure success in speech encoding. The total sequence of mechanisms that contribute to the electrical stimulation of neural elements within the cochlea is referred to as the "electrical-to-neural transformer".

Solution of the encoding problem requires detailed knowledge of the mechanisms of electrical stimulation of neural elements within the cochlea. Many factors are known to contribute. 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, it is probably necessary to control both temporal and spatial profiles of neural discharge around each electrode or electrode pair. This task is further complicated by the presence of field interactions among the electrodes and the highly nonlinear response of the neural elements to electrical stimulation. To evaluate the relative significance of each of these contributing factors, we have developed a series of biologically-authentic computer models of the physical structures and biophysical mechanisms thought to be involved in the transduction of electrical stimulation to neural cochlear outflow.

Figure II.B.1 is a schematic diagram of the equivalent electrical circuit involved in the electrical stimulation of a single nerve fiber. This electrical analogue is the basis of our modeling efforts. More complicated arrangements which include multiple neurons and multiple electrodes are constructed upon this basic circuit. Re1 and Re2 are the series DC resistances of the electrodes 1 and 2, respectively. Ce1 and Ce2 are the electrode/tissue interface capacitances. Resistive dividers formed by the combination of Rax and Rbx (x ranges from 1 to N) are distributed along the course of the neuron producing potentials (V1 to VN). The ratios of Rax to Rbx, along with the electrode drive voltage, combine to produce the node voltages (VN). Cs is the electrode cable shunt capacitance. The current driver is described as a Norton equivalent circuit whose parameters may be changed to represent a voltage-controlled, transcutaneous, radio-frequency-coupled link or the high-voltage compliance current sources used with the percutaneous cable. Observation of the model reveals that the absolute magnitude of the potentials along the course of the neuron is a direct function of the magnitude of the stimulus current passing between the electrodes; whereas, the relative ratios of the potentials along the neuron are a function of the physical geometry and resistive characteristics of the electrodes and surrounding tissue.

Estimates of the node voltages are calculated by an iterative, two-dimensional, finite-difference model of a cochlear cross section. The model contains a scala tympani bipolar electrode pair representing the UCSF bipolar electrode design, compressed into two dimensions (Loeb, Byers, Rebscher, Casey, Fong, Schindler, Gray and Merzenich, 1983). Resistivities are included in the finite difference calculations that approximate



Equivalent circuit of field calculation results.

Figure II.B.1

appropriate tissue characteristics. Fixed voltages are assigned to the electrode pair, and the outer boundary of the plane is held constant at zero potential level. Field patterns for the entire cross section are calculated by iteration. Those familiar with electromagnetics will realize that the computation as outlined is essentially an electrostatics problem. This formulation of the problem is appropriate since the tissue is predominately resistive and linear in nature up to 20 kHz. (Spelman, Clopton, and Pfingst, 1982). Finally, the potential levels at points along the locus of VIIIth nerve elements are extracted from the final field calculation. These potentials include the node voltages (VN) described above. A more complete description of this modeling approach is presented in section II.B.2. below.

The motivating interest in the model is in predicting and studying the discharge characteristics of the neurons themselves. Consequently, the next step of modeling is the calculation of the neural responses to stimuli delivered by the implanted electrodes. This moves the modeling problem into the time domain and therefore links the modeling work directly to the issues of speech encoding strategies and the specification of stimulus current patterns. It is here that the modeling approach may prove especially valuable by providing insight into the factors controlling the temporal features of electrically-induced neural firing.

Temporal modeling is accomplished by feeding the calculated voltages along the neuron into a lumped-element model of a 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 (1976) axon model consisting 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. The stimulus is 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.

The final step in modeling is to explore the interactions and relationships between neurons both with single and multiple channel stimulation. Figure II.B.2 schematically outlines this approach by showing an integrated field-neuron model for a simple system composed of two neurons and two electrode pairs with current sources. Neurons A and B represent elements of an ensemble of neurons located at different positions within the spiral ganglion or along the basilar membrane. Each neuron has its own unique electrical relationship with each electrode pair as defined by the neuron's position in the field and the characteristics of the surrounding tissue. Furthermore, each neuron may have its own unique electrical characteristics, based upon is own biophysical features, such as diameter variation, loss or reduction of myelin, or even complete loss of peripheral segments. Stimulation with the current sources for channel Y and Z, either alone or simultaneously, allow study of the firing characteristics of both neurons A and B.

Full exploitation of these modeling tools will afford considerable insight into controlling the temporal and spatial profiles of neural discharge around an electrode or electrode pair. Additionally, insight may be gained into the complexity of field interactions occuring during multichannel stimulation, making it possible to avoid and/or exploit such interactions in speech processor design. Finally, appreciation of the deleterious effects of pathology may be factored into the speech processor design equation. This latter point may be the single most important contribution of the models, since interpatient variations in speech perception performance confound interpretation of test results as they apply to speech processor evaluations. Interpatient differences in pathology are prime candidates as sources of these variations. Clear insight into the details of electrical stimulation, specifically applied to intracochlear prostheses, can transform the stimulus encoding problem into a rational design task.