Fourth Quarterly Progress Report

May 1 through July 31, 1996

NIH Project N01-DC-5-2103

Speech Processors for Auditory Prostheses

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I. Introduction

One of the principal objectives of this project is to design, develop, and evaluate speech processors for implantable auditory prostheses. Ideally, the processors will represent the information content of speech in a way that can be perceived by implant patients. Another principal objective is to develop new test materials for the evaluation of speech processors, given the growing number of cochlear implant subjects enjoying levels of performance too high to be sensitively measured by existing tests.

Work in the present quarter included:

- Studies with Ineraid subject SR2 (July 22-26), primarily to obtain measures of neural responses (with recordings of intracochlear evoked potentials) to pulses presented at high rates, up to 10000/s. The stimuli included trains of identical (unmodulated) pulses, sinusoidally amplitude modulated trains of pulses, and the modulated pulse trains produced by a single-channel speech processor.
- Speech reception and evoked potential studies with Nucleus percutaneous subjects NP5 (weeks beginning May 13 and May 20), NP4 (weeks beginning June 3 and June 10), and NP2 (July 8-10).
- Presentation of project results in two invited lectures at the *Third European Symposium on Paediatric Cochlear Implantation,* held in Hannover, Germany, June 6-8.
- Preparation for and conduct of a site visit for the project by Drs. Terry Hambrecht and William Heetderks (July 23).
- Recording of tokens for new tests designed to provide high sensitivity even for implant patients at the upper end

of the performance spectrum.

- Completion of new current sources, principally for use in studies to evaluate very high rates of stimulation (*e.g.*, 10000 pulses/s on each channel) in multichannel CIS processors.
- Continued development of the Evoked Potentials Laboratory, including incorporation of a 22-bit A/D converter, in part to allow recording of both stimulus pulse artifact and evoked potentials in the linear range of the recording system.
- Continued development of a new type of compression function for use in CIS processors, designed to mimic principal features of the noninstantaneous compression found in normal hearing at the interface between sensory hair cells and adjacent neurons.
- Further development of software and fitting procedures for the Geneva/MEEI/RTI portable processor, in anticipation of studies to evaluate possible learning effects with extended use of the CIS strategy, using the Geneva/MEEI/RTI processor.
- Continued analysis of speech reception and evoked potential data from prior studies.
- Continued preparation of manuscripts for publication.

In this report we provide a description of the new current sources, to be used in upcoming studies to evaluate very high rates of stimulation in conjunction with CIS processors. As indicated in section II, the new current sources also provide (a) greater flexibility in specifying patterns of stimuli that overlap in time across channels and (b) additional safety features and several enhancements to operator feedback and convenience compared to the prior system.

Results from other studies and activities indicated above will be presented in future reports.

II. New Laboratory Stimulator System

Special Note Regarding the Web Version

The figures in this report, mostly schematic circuit diagrams, have too much detail for an entire figure to be viewed at once on most video screens. Thus, in place of inline figures, we have provided links to high resolution versions that can be viewed a part at a time using browser scroll bars, or printed out at 150 dpi.

A new laboratory system has been designed and fabricated for direct stimulation of implanted electrodes in subjects with percutaneous connectors. This effort was undertaken because experimental manipulations of stimulus parameters had become constrained by the limits of the existing equipment. In particular the old equipment could not support the high pulse repetition rates (up to 10 kHz) and corresponding short pulses widths (5-10 µs) to be explored in the current contract period. The old equipment also limited studies with overlapping stimulation to 6 channels which was sufficient for Ineraid devices, but not for the 22-electrode Cochlear Corp. percutaneous implants. Overlapping stimulation is required for paradigms which utilize two or more intracochlear electrodes simultaneously (e.g. bipolar, current steering, virtual channel). In addition the new system provides additional safety features and several enhancements to operator feedback and convenience.

New System Design Objectives

Increased Current Source Speed

The speed of the old system is limited in three ways.

1) Isolator Bandwidth - Pulse waveform signals in the old system are coupled through analog isolators (Burr Brown ISO120) with approximately 60 kHz bandwidth. In addition to limiting bandwidth these couplers are also relatively noisy. Their noise floor is determined by sampling noise introduced by discrete-time transmission of the analog signal via pulse- width modulation. The analog isolators have been eliminated in the new system. Digital data and address outputs from the microprocessor are coupled through digital isolators to an array of 24 D/A converters on the isolated side of the circuitry.

2) Compliance voltage - The use of shorter pulses requires a concomitant increase in the current and hence, compliance

voltage. The old equipment regulates current by placing the load impedance in the feedback path of a conventional operational amplifier with ±12V supplies. Thus the compliance voltage across the any two electrodes (and coupling capacitors) is limited to about ±11 volts. The output stages of the new system use high voltage operational amplifiers with supply rails of ±30V. In addition the new system uses a dedicated ground-referenced (not floating) current source for every electrode (including the monopolar return electrode). Thus a maximum of ±58 volts can be developed across any electrode pair.

3) Parasitic shunting - With the old equipment high frequency components of the stimulating current pulse are shunted across parasitic capacitance in the cabling between the current sources and the subject. This limits the rise time of current actually delivered to the electrodes, and correspondingly limits the minimum effective pulse width. The new system uses guarding to reduce the effects of parasitic coupling between cables. Each current source output is carried in a coaxial cable whose shield is driven by a guard amplifier to be equipotential with the center conductor (plus the low-frequency voltage across the output capacitor). Thus parasitic current spikes flowing from shield to shield are supplied by the guard amplifiers and do not reduce the current flowing to the electrodes.

Simultaneous stimulation of 24 channels

The old equipment utilized different output stages for Ineriad and Cochlear subjects. The Ineraid interface provides for overlapping or simultaneous stimulation of six electrodes. However the output stage used for the Cochlear Corp. percutaneous electrodes utilizes a single current source multiplexed across the electrode array. It cannot stimulate more than one intracochlear electrode at a time. Because the new system provides a dedicated current source for every electrode, any combination of electrodes can be stimulated simultaneously.

Reduced battery count

The old equipment utilizes 12 batteries to power 6 current sources. The old current source design requires that the load be floating (i.e. neither side may be connected to the circuit ground). As a consequence each of the old current sources requires a separate pair of batteries. Extension to a 24 electrode system would have required 48 batteries (or more to achieve increased compliance voltage). In contrast all 24 of the new current sources share a common power supply and ground.

Another advantage of this configuration is that it is less susceptible to current "leakage" through parasitic pathways. With the old equipment only current returning through the "virtual-ground" electrode of each channel is regulated (*i.e.* current flowing to the inverting input of the op-amp). Current through the "active" electrode (connected to the op-amp output) is regulated indirectly by providing no other return path to ground. However any parasitic capacitance between the subject and the circuit ground provides an alternate unintended path to ground such that additional "leakage" current may flow through the active electrode, circumventing the regulated return path. The new current sources provide active regulation of current through every electrode which prevents such leakage.

Additional Safety Features

1) Charge Limit - The new system automatically disconnects all electrodes from the current sources if the input signal to any current source would result in passage of more than 100 nanocoulombs without a current reversal. It should be noted that this function is not intended to limit charge densities at the electrode surfaces to electrochemically safe levels. Operation with electrochemically safe chronic charge densities is achieved by selecting appropriate current and pulse width ranges in processor design. Instead this function protects against uncomfortably loud stimulation in the event of a hardware or software error which would otherwise produce erroneous current waveforms. For example a processor halt in the midst of generating a biphasic pulse might result in an indefinitely long first or second phase (limited only by charging of the output capacitor). Or a timing error might result in erroneously long current pulses. In either case the current source would detect the error, and all electrodes would be immediately disconnected.

This facility is also a sensitive indicator of small pulse width asymmetries which would otherwise reduce available compliance voltage in the current sources by causing the DC operating point of the output amplifiers to drift up or down.

2) Error Current Monitor - By definition the sum of all electrode currents (including the monopolar return) must be zero. That is, the sum of currents flowing into the tissue must equal the sum of currents flowing out. The processor code is responsible for ensuring that the sum of all regulated electrode currents is zero. However in practice, due to circuit imperfections, a small "error current" must exist even when the electrode outputs nominally sum to zero. A return path for this error current is provided through a surface electrode attached to the subject's wrist. The error current is monitored continuously. If it exceeds 10% of the full scale output level for more than 8 µs, all electrodes are automatically disconnected. This protects against any hardware or software error which results in significant imbalance among electrode currents.

This protection is particularly important, for example, during bipolar stimulation. In this case relatively high currents are sourced and sunk by adjacent electrodes. Consider the case where current through one electrode of a bipolar pair is interrupted (e.g. by a connector problem or wire breakage). The opposite current flowing through the remaining electrode becomes instantly monopolar, returning through the error path. This would very likely produce an uncomfortably loud sensation. The error current monitor prevents this by immediately disconnecting all electrodes when the currents are out of balance.

3) Connection sequence - The new system imposes a controlled sequence by which the subject's electrodes are connected to the current sources through relays. Connection cannot begin unless the volume slide control is at zero and any previously detected error conditions (charge errors, balance errors, or manual disconnect) have been eliminated. Electrodes are connected or disconnected sequentially, one-at-a-time, by turning a rotary encoder. This avoids any transients which might arise if 24 current sources were connected simultaneously. The volume control remains disabled (DAC outputs forced to zero) until the final current source has been connected. If the volume slider is raised before this time the encoder is disabled (connections are frozen), and the DAC outputs remain forced to zero.

Both the subject and the operator are provided with switches that can interrupt the supply current to the relay coils. If either switch is opened all of the electrodes are disconnected immediately. The electrodes can be reconnected only by repeating the sequence described above. A reconnection sequence also is required if the electrodes are disconnected due to a charge error or balance error.

Increased operator feedback/convenience

The new system accommodates both Ineraid and Cochlear Corp. percutaneous systems with no change in interface equipment. It also provides a variety of new operator conveniences and annunciators. Power to the entire system is controlled by a single switch. When this switch is off, the batteries are automatically connected to a charging circuit. A low battery indicator is provided on the supply module for each of the five battery-powered supply voltages. The operator panel includes error and alarm indicators which notify the operator of any of the following conditions: Charge error detected, balance error detected, manual disconnect switch open, one or more current sources saturated, or any low battery indicator activated. In addition it provides annunciators which indicate that all error conditions have been eliminated, that the rotary encoder is enabled, that the volume control is enabled, and which electrodes are connected.

Additional annunciators are provided on the current sources themselves. Each current source has three individual indicators. A red LED indicates that the current source has detected a charge error. A second red LED flashes for 50 ms each time the current source saturates. These two indicators allow the operator to determine which channel is the source of a global saturation or charge error indication on the operator panel. The third indicator is a green LED which illuminates for 50 ms each time the input signal to the current source exceeds 100 mV (corresponding to 20.6 µA of output current). This allows the operator to determine at a glance which channels are active, providing quick confirmation of normal processor operation. These signal indicators can be disabled to avoid providing visual cues to the subject during testing.

For purposes of developing and debugging processor code, the charge error and balance error protections can be defeated. However this is only possible when a dummy load (resistor array) is connected to the operator panel in place of the Ineraid subject cable. The dummy load assembly includes two switches which permit either or both of these protections to be defeated such that the relays do not open when the error(s) is(are) detected. When the dummy load is disconnected, the protections are automatically re-enabled.

System Components

Figure 1 shows a block diagram of the system. It comprises seven 12-volt/10A-hr lead-acid batteries and four physical enclosures: 1) the power supply; 2) the operator panel; 3) the digital rack housing the digital isolators and four hex-DAC modules; and 4) the current source rack housing eight triple current source modules. The enclosures are connected by a system bus which distributes power and global logic signals. Each of the components is described below.

Power Supply

The system uses 6 separate supply voltages. One of these (the +5e logic/relay supply) is line powered. It is generated by a medical grade linear supply module which feeds the operator panel. This supply provides power for relays in the operator panel and power supply module. It also powers the logic in the operator panel. Although it is isolated from earth, this supply does not power any circuitry to which the subject is connected. All such circuitry is powered by the remaining 5 system supply voltages (+5V, ±15V, and ±30V) which are generated from batteries.

Each of the six supplies appears on the front panel banana jacks. The isolated analog and digital grounds are connected together within the supply module, but are maintained as separate conductors throughout the rest of the system. The +5V digital supply is generated by a switching regulator from one of the 12-volt batteries. The ±15V and ±30V supplies are generated by linear regulators from a stack of six other 12-volt batteries. Each battery assembly includes a 2 ampere fuse, and is connected to the rear of the supply module through a shielded cable. The shield is connected to analog ground at the power supply module, and to a foil shroud which surrounds the battery.

The line powered logic/relay supply (+5e) is brought to the supply module from the **DC-Out** jack of the operator panel to a connector at the rear of the supply module. This +5e supply is controlled by the main power switch on the operator panel. When it is turned on, seven relays in the supply module close, connecting the batteries to the regulators to produce the isolated supplies. Five Low-Battery LEDs on the front panel of the supply module indicate when any of the unregulated input voltages to the regulators drops too low. The low battery thresholds are set to 11 volts per battery (±33 V for the ±30V regulators, ±22 V for the ±15V regulators, and 11V for the 5V regulator). The thresholds are well above the minimum input needed to maintain appropriately regulated supply rails. When any of these LEDs illuminates, a low battery LED is also illuminated on the operator panel. The following table shows which batteries feed each supply voltage:

SupplyBatteries

Supply	Batteries
+5V	#1
-30V	#2, #3, #4
-15V	#3, #4
+15V	#5, #6
+30V	#5, #6, #7

Charging current for the batteries is brought into the supply module through two banana jacks (red and black) on the rear panel. When the +5e logic/relay supply is turned off at the operator panel, the seven relays switch to connect each battery to the charging inputs through a series diode and 3W resistor. The batteries are charged in parallel, but current cannot flow from one to the other due to the series diodes which prevent outward current flow from each battery. The charging inputs are fed by a 13.7 V DC bench supply with a 2-amp current limit. (Plans call for this bench supply to be replaced with a dedicated medical grade linear supply.)

The supply module is connected to the operator panel and analog/digital racks through a 25- conductor ribbon cable terminated in DB-25 connectors. This cable also contains several system logic signals and the Reference Voltage used to set the global DAC output amplitude (volume control). The pin assignments for this connector are shown below.

System Supply / Signal Bus

+5 V	1	
+5 V		14
Digital Ground	2	
Digital Ground		15
+15 V	3	
+15 V		16
-15 V	4	
-15 V		17
Analog Ground	5	
Analog Ground		18
+30 V	6	
-30 V		19
Vref	7	
CLIP*		20
QERR*	8	
RS*		21
SL_ENAB*	9	
LOW_BATT*		22
nc	10	
nc		23
nc	11	
nc		24
nc	12	
nc		25
nc	13	

Operator Panel

The Operator Panel contains all operator controls for the system. Power is brought to the panel from the outboard medical grade linear 5V supply through the DC-In jack on the right side. This power is controlled by the main power switch on the right side of the panel. When this is turned on, relays in the power supply module close to connect the batteries and turn on the isolated supplies. A schematic diagram of the operator panel is shown in Figure 2a, and Figure 2b.

1) Subject Connections - Driven Ground

A surface electrode (e.g. ECG electrode) on the subject's wrist is connected to the rear panel DRIVEN GROUND BNC jack through a coaxial cable. The shield of this BNC is analog ground, and is insulated at both ends of the cable. It does not contact the subject. (Note: subject contact with analog ground poses no hazard, but might interfere with detection of any imbalance in electrode currents - see below). The center conductor of this cable is the driven ground. It is equipotential with analog ground, but has a very limited current capacity (about ±1 mA max). This connection serves two purposes. First it serves as a reference potential which maintains the subject at the same potential as the isolated analog ground and provides a return path for residual error currents. It also serves to detect any balance errors in the stimulating currents as follows.

The algebraic sum of all electrode currents should always be zero. In other words the amount of current flowing in to the tissue through source electrodes must be equal to the current flowing out of the tissue through sink electrodes. If the current sources are correctly balanced this way, there will be no residual current flowing to or from the driven ground electrode. However if the sum of the electrode currents is not zero the residual "error" current will return through the surface electrode to the driven ground. Small error currents are inevitable due to circuit imperfections. However a significant imbalance could occur due to a hardware problem (bad cable or saturated current source) or software problem (sum of programmed currents not zero). The current in the driven ground is monitored. If it exceeds a threshold for 8 µs or more a Balance Error occurs, and the subject is automatically disconnected from the current sources. The error threshold is ±10% of the full scale current output (as set by the Volume control) or ±20 µA, whichever is larger. The 8 µs window allows the circuit to ignore transient error current spikes which occur at pulse edges due to slight mismatches among the speeds of the current sources (see below).

2) Subject Connections - Electrode Cable

The cable from the subject's percutaneous plug is connected to one of two connectors on the top front surface of the panel. Ineraid subject cables connect to a DB-25 connector, and Cochlear Corp. subject cables connect to a Micro-D connector. All 24 of the current source outputs appear on each of these connectors.

The DB-25 connector is also used to connect a dummy load resistor network for system testing. The dummy load should not be connected to the system at the same time that a subject is connected for two reasons. First, if both the load and the electrodes are connected, an unknown portion of the stimulating current will be shunted through the resistor network, resulting in reduced (and uncalibrated) currents actually flowing to the implanted electrodes. More importantly, line powered measurement equipment (e.g. an oscilloscope) may be connected to the dummy load for test purposes, which would defeat the isolation of the outputs. In fact it is impossible to connect both an Ineraid subject and the dummy load, because both utilize the same DB-25 connector. However it is physically possible to leave the dummy load on the DB-25 connector while a Cochlear Corp. subject is connected through the Micro-D connector. Care must be exercised to ensure that this does not happen.

3) Subject Connections - Insulated BNC Jacks

The DB-25, Micro-D, and insulated BNC connectors are wired in parallel. Each pin of the DB-25 and Micro-D connectors is also connected to the center conductor of the corresponding insulated BNC jack on the top of the panel. All of these connectors remain connected to the subject cable when the disconnect relays are opened. Thus the BNC jacks can be used for impedance testing with the current sources disconnected.

When the disconnect relays are closed the center conductor of each BNC jack (and the corresponding subject connector pins) are connected to the center conductor of a coaxial cable which carries the stimulating current from the corresponding current source. In addition, the outer conductor of each BNC jack is connected to the shield of the coaxial cable. These shields are not connected to ground. Each shield is a separate driven guard signal. The guards prevent leakage of high frequency stimulating current components through parasitic capacitance to the system ground. Insulated BNCs have been installed to prevent inadvertent contact with these guard signals. The shield conductors are insulated and recessed such that neither the subject nor the operator can contact them.

4) Subject Connections - Disconnect Relays

A relay is connected in series with each pin of the subject connectors as well as the driven ground. The relays are normally open such that all electrodes would be disconnected in the event of a failure of the line powered +5e supply which powers the relays and associated logic. The 120VAC line is itself provided from a battery-backed uninterruptable supply. The operator and the subject are each provided with a switch which can immediately open all subject connections by interrupting current to the relay coils. The operator's disconnect switch is on the panel. The subject's switch is housed in a hand-held box and connected through two banana jacks at the rear of the panel. When either of these switches is opened the 25 relays open simultaneously, breaking the connection between the coaxial cables and the panel-top connectors (BNC, DB-25, and Micro-D), and the connection to the driven ground jack. If either of these switches is opened a red indicator LED at the lower left of the panel is illuminated and the connection sequence cannot be initiated (see below).

The driven ground is immediately reconnected to the surface electrode on the subject's wrist as soon as both disconnect switches are closed again, but the implanted electrodes are not. In order to reconnect the implanted electrodes to the current sources the usual connection sequence must be executed (see below).

5) Error/Alarm Signals

Five red LEDs on the lower left of the panel indicate error or alarm conditions. Three of these indicate error conditions which automatically cause the disconnect relays to open. In order to reconnect the subject after an error, a reconnection sequence must be conducted after the error has been eliminated (see below). The remaining two LEDs indicate alarm conditions which require operator attention but do not cause a disconnect.

Error Condition LEDs

- Disconnect Switch - Either the Operator or the Subject disconnect switch is open
- Charge Error - One or more of the current sources has received an input signal which would produced more than 100 nC of charge transfer without a reversal of current. The current source which generated the error can be identified by examining individual error LEDs on the modules.
- Balance Error - Current in the driven ground electrode exceeded threshold, indicating that the electrode currents were not balanced.

Alarm LEDs

- Clip - One or more of the current sources is saturated, such that it is not generating the intended current magnitude. This LED is illuminated for 50 ms each time a saturation is detected by any current source. The saturated current source can be identified by examining individual error LEDs on the modules. The most common cause of saturation is an open circuit between the current source and the electrode (e.g. relays open, bad or missing cable, etc.). A clipping error alone does not cause the relays to open. However clipping may result in a balance error, which will cause the relays to open.
- Low Battery - This LED indicates that one or more battery supplies to the 5 isolated regulators is low. The particular supply can be identified by examining individual low-battery LEDs on the power supply module. Low battery thresholds are adjusted such that the alarm condition arises while the regulator input is still adequate to maintain proper supply voltages. Therefore a low battery condition does not cause the relays to open. However extended operation with low batteries may result in loss of proper supply voltages.

6) Signal LED Disable

The front panel of each current source module has three green signal LEDs. Each of these illuminates for 50 ms each time the corresponding input signal exceeds ±100 mV (corresponding to an output of ±20 μA). These LEDs provide quick confirmation that appropriate channels are receiving stimulation during processor setup. However these LEDs could provide visual cues to the subject during testing. Therefore all of these LEDs can be disabled by the Signal LEDs toggle switch immediately to the right of the Error/Alarm LEDs at the bottom of the operator panel.

7) Volume Control

VOLUME (output current level) is controlled by a combination of a slide potentiometer and a calibrated rotary potentiometer. These potentiometers set the reference voltage used by all of the DACs. The rotary potentiometer is used to set the operating volume level. The slider is provided to permit a convenient method of smoothly raising the volume to the calibrated level, or lowering it back to zero. Testing is conducted with the slider at the maximum level so that the volume level is calibrated. Under some circumstances the volume level is forced to zero regardless of the pot settings (see below). Illumination of the green Volume Enabled LED under the rotary knob indicates that the VOLUME controls are active, and volume is not forced to zero.

8) Connection Sequence

When the system is turned on (or after an error has been detected) all of the disconnect relays are opened and the volume is forced to zero. In order to prevent perceptible transients when connecting the subject's electrodes to the current sources the relays are closed one-at-a-time. The connection sequence is as follows:

- System Reset - The connection sequence cannot be initiated until all error conditions have been eliminated and the VOLUME Control slide control has been brought to zero. When these conditions have been met the green Push Reset LED to the left of the connect knob is illuminated. This indicates that the system can be reset by pressing down on the knob, which extinguishes the charge error LEDs. The green Turn to Connect LED to the right of the connect knob is then illuminated, indicating that electrode connections can be made by rotating the knob.
- Connection - Successive electrodes are connected to corresponding current sources by rotating the knob clockwise. Rotation through each detent causes one additional relay to close as indicated by illumination of the corresponding LED in the array above the knob. Electrodes can be disconnected in sequence by rotating the knob counterclockwise. All electrodes can be disconnected simultaneously by pressing the knob (reset). The volume remains forced to zero (Vref to DACs is 0V) until all of the electrodes have been connected. If the VOLUME slider is raised before all of the electrodes have been connected, the connect knob is disabled (Push Reset and Turn to Connect LEDs turn off), and the volume remains forced to zero.
- Volume Adjustment - When the last electrode (#24) has been connected, the green Volume Enabled LED under the calibrated pot is illuminated. The slider can then be raised to bring the volume gradually to the level set by the calibrated knob. When the slider is raised the connect knob is disabled (Push Reset and Turn to Connect LEDs turn off). Turning or pressing the knob has no effect unless the volume slider is all the way down, so electrodes cannot be connected or disconnected (except by the DISCONNECT switches) with the volume raised.

Note: The restriction on raising the volume until all electrodes are connected protects against stimulation when some, but not all, of the electrodes-in-use have been connected. This would generally produce a current balance error. Neither of the percutaneous implants currently being studied actually utilizes all 24 current sources. Tests with Ineraid subjects may use anywhere from 2 to 7 sources, and tests with Cochlear subjects may used from 2 to 23. Since the hardware cannot know which of the electrodes are to be used for a particular subject and test, it requires that all 24 relays be closed before the volume is raised. Unused current sources with no load are programmed to generate zero current. If a non-zero current were programmed for an unloaded current source (e.g. current source 9 while testing an Ineraid subject), it would saturate and produce a clipping alarm. But there would be no effect on stimulation delivered to the electrodes.

· Disconnection - When testing is complete the VOLUME slider is lowered to zero. The Push Reset and Turn to Connect LEDs below the connect knob are then illuminated, indicating that electrodes can be disconnected one-at-a-time by turning the knob counterclockwise, or simultaneously by pressing the knob. When any electrode is disconnected the Volume Enabled LED goes out indicating that the volume has been forced to zero.

Isolator and DAC Modules

All DAC circuitry and associated logic is isolated and powered by the batteries. The four identical DAC modules and a digital isolator module are connected by a backplane which distributes the isolated +5V and ±15V supplies, digital control lines, and the DAC reference voltage. The digital control signals comprise 12 data lines, 5 address lines, and a strobe. They are generated by the (non-isolated) microprocessor and brought to the isolator module through two front panel ribbon cables. This module couples these control signals to the backplane through high speed digital isolators (Burr Brown ISO150). Unlike optical couplers these parts use faradic coupling (tiny capacitors) to transmit logic level transitions across the isolation barrier. They have higher barrier capacitance (5 pF) than optical couplers, but are faster and require less power.

Figure 3, parts a, b, c, and d shows a schematic diagram of one DAC module. It contains three Analog Devices DAC-8222 dual 12-bit multiplying DACs (U7,11,14). The DACs and output amplifiers are configured for two's complement bipolar operation. Each DAC has an individual address determined by jumpers JP1-4. Address decoder U2 gates the write strobe to the appropriate chip and pin. On-chip double buffering allows new data to be preloaded into the input latch for each DAC without changing its analog output level. When all of the input registers have been loaded (by sequential writes to their addresses), all of the DACs are triggered simultaneously to transfer the new data into their output latches. This is accomplished by a dummy write to address FFE1 which causes decoder U5 to assert the ALLLOAD* line. Thus all of the analog outputs change simultaneously.

A common reference voltage is distributed to all of the DAC modules to set the full scale (7FF) output for all of the DACs. It is buffered on board by U9:A,B and U16:A. The reference voltage varies between 0 and +10V, as determined by the operator panel volume controls. (Under some circumstances during the connection sequence the reference voltage is forced to zero regardless of the volume control settings.) The analog output from each DAC is connected by a BNC cable to one of the 24 current sources.

Current Sources Modules

Eight identical current source modules are connected by a second backplane which distributes the isolated +5 V, ±15V, and ±30V supplies as well as system error and control logic signals. Figure 4, parts a, b, c, d, e, f, g, and h shows a schematic diagram of one module. It contains 3 single-ended current sources each of which receives its input from a DAC and generates a proportional output current (206 μA/V). The outputs are not floating - each is referred to analog ground, and is connected through the operator panel to a single electrode. Bipolar stimulation is achieved by producing equal but opposite output currents simultaneously from two sources. Similarly for monopolar stimulation the current source which drives the extracochlear return electrode must generate the opposite of the algebraic sum of the currents generated by all of the intracochlear electrodes. Thus the sum of all output currents must always be zero, regardless of mode, and the return current carried by the driven ground (applied to the skin) should always be zero.

The following circuit description refers to components of the first current source on the board (shown in figures 4a, 4b, and 4g). High voltage op-amp U2 (OPA445) is connected in a bridge configuration to regulate output current through resistor R16 to ground. The output impedance of an ideal current source is infinite. However due to circuit imperfections practical current sources cannot maintain a very high output impedance at DC if the load is coupled through a series capacitor as in our case (C17). In this circuit DC stability is provided in two ways: 1) 5 Mohm resistor R21 shunts the current source output, providing a path for small DC output offset currents; and 2) 10 Kohm resistor R5 unbalances the bridge slightly at DC, providing additional negative feedback around the op-amp. For high frequency signals R5 is effectively short- circuited by 0.2 μF capacitor C2. The Norton equivalent output impedance of this stage is shown in Figure 5. The circuit behaves like an ideal current source shunted by a very large (293 Henry) inductor in series with a 154 Kohm resistor, and by the 5 Mohm of R21. This inductor in conjunction with the series 0.1 μF output capacitor C17 forms a second-order system with a natural frequency of about 30 Hz. However the system is overdamped (non-oscillatory) regardless of electrode impedance due to the heavy damping provided by the effective 154 Kohm series resistor.

Trim capacitor C1 is provided to adjust for equal speeds across all current sources. This is useful for minimizing transient error currents (balance errors) at pulse edges. Such errors arise if source and sink do not turn on and off at the same speed. Source 24 is trimmed for minimum rise time without overshoot into a 10 Kohm load. The speed of each of sources 1-23 is adjusted to match that of source 24. This is accomplished by trimming each source for minimum measured error current spike while generating a 30 μs/phase bipolar/biphasic pulse between the test source and source 24. Small residual speed differences are inevitable, however, as the output speed depends slightly on the load impedance which will vary from electrode to electrode.

A second high voltage op-amp (U7, Figure 4b) is configured as a follower to drive the guard shield of the output coaxial cable. In order to guarantee the integrity of the AC coupling, this op-amp actually tracks the voltage on the circuit side of coupling capacitor C17, rather than the electrode voltage itself. The error introduced, however, is a low frequency one which does not compromise the effective guarding of the output conductor at high frequencies.

Current gain is fixed at 206 μA per volt of input by resistor R9 and the ratio of R12/R11. Trimmer R6 is adjusted for critical balancing of the bridge such that the amplifier just enters saturation with 0V in, R5 shorted, and R21 open. Actual output impedances were determined by measuring output variations with loads of 1 Kohm and 100 Kohm with a 10 kHz square wave input. They varied from 2.7 Mohm to 3.5 Mohm, including the 5 Mohm shunt resistor.

Comparators U1:A and U1:B detect saturation of the current source when the output of U2 approaches either supply rail. The comparators trigger a one shot which illuminates the front panel clipping indicator, and also asserts the system bus CLIP* signal to activate the global clipping indicator on the operator panel. A clipping error does not itself cause the output relays to open. However in some cases clipping will result in a current imbalance condition because the algebraic sum of output currents will no longer be zero. The resulting error current will cause a balance error which will cause the relays to open. The relays will not open if the clipping does not result in an actual current imbalance. For example a current source not connected to the subject may receive an erroneous input signal and saturate; or all of the current sources may saturate simultaneously if the subject cable is disconnected. Neither of these conditions would cause a current imbalance condition.

Amplifier U4:B integrates the input signal to detect charge errors. A charge error occurs when a net charge of ±100 nC is delivered without reversing the output current. Since charge errors are detected by integrating the input waveform (not the actual current) a charge error may be detected even if no current is actually flowing (*e.g.* if the electrode is open-circuited) or if the delivered charge is less than the amount inferred from the input (*e.g.* if the output stage saturates). When U4:B integrates to ±2.5V comparator U1:C or U1:D sets flip-flop U8:A which activates the front panel charge error LED. The flip-flop is cleared by a system reset (bus signal RESETI*). The comparator also asserts the system bus signal QERR* in order to activate the charge error circuit in the operator panel which results in opening of all of the output relays. For DC stability U4:B is not actually an integrator (pole at 0 Hz) but rather a low pass filter with a time constant of 5 ms (pole at 32 Hz). Therefore it will not detect a charge error caused by a DC offset in the input of 100 mV (20.6 µA out) or less. Instead protection against DC output current is provided by the output capacitor.

Comparator U23:A triggers a one-shot to illuminate a front panel signal LED any time the input signal exceeds 100 mV. This provides quick confirmation to the operator that the current source is receiving an input. The one-shot can be inhibited by the system bus signal ENABLE-SIGLED* to defeat this indicator during testing.

Personnel

The conceptual design of the new system was developed primarily by Charles Finley and Marian Zerbi. Hardware designs were developed jointly by Marian Zerbi, and Chris van den Honert, and Charles Finley. Assistance in construction of the equipment was provided by Curtis Moore.

III. Plans for the Next Quarter

Our plans for the next quarter include the following:

- Presentation of project results in invited lectures at the annual *Neural Prosthesis Workshop* (October 16-18) and at the *International Workshop on Cochlear Implants*, to be held in Vienna, Austria, October 24 and 25.
- Completion of speech reception and evoked potential studies with Nucleus percutaneous subject NP5 (weeks beginning September 16 and 23).
- Continued studies with a local subject having a standard Nucleus device implanted on both sides (several half days over the course of the next quarter).
- Completion of recordings for new speech tests.
- Continued development of the Evoked Potentials Laboratory.
- Initial evaluation of CIS processors using very high rates of stimulation, in studies with Ineraid subject SR2 (presently scheduled for the week beginning September 30).
- Completion of preparations for use of the Geneva/MEEI/RTI portable processor in studies to evaluate possible learning effects with the CIS strategy.
- Continued analysis of speech reception and evoked potential data from prior studies.
- Continued preparation of manuscripts for publication.

Appendix 1: 1996 Discover Award

We are pleased to announce that the CIS processing strategy was recognized by a *1996 Discover Award for Technological Innovation*, in the category of "sound." Descriptions of this award and the awards made in other categories are included in the July, 1996, issue of Discover Magazine.

Appendix 2: Summary of Reporting Activity for this Quarter

Reporting activity for the last quarter, covering the period from May 1 to July 31, 1996, included the following:

Wilson BS, Finley CC, Lawson DT, Zerbi M: Temporal representations with cochlear implants. Invited lecture, *Third European Symposium on Paediatric Cochlear Implantation*, Hannover, Germany, June 6-8, 1996.

Wilson BS: Suggestions for the future development of cochlear implants. Invited lecture, *Third European Symposium on Paediatric Cochlear Implantation*, Hannover, Germany, June 6-8, 1996.

Wilson BS: Chair, Session on "Basic Science and Technical Aspects." *Third European Symposium on Paediatric Cochlear Implantation*, Hannover, Germany, June 6-8, 1996.

Wilson BS: Progress in the development of speech processing strategies for cochlear implants. Invited lecture, University of Iowa, Department of Otolaryngology – Head and Neck Surgery, Iowa City, IA, July 29, 1996.