Final Progress Report

January 1, 2002, through June 30, 2005

Speech Processors for Auditory Prostheses

NIH Contract N01-DC-2-1001

Submitted by

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1.0 Introduction

Work performed with the support of this contract is directed at the design, development, and evaluation of sound-processing strategies for auditory prostheses implanted in deaf humans. The investigators, engineers, audiologists and students conducting this work are from four collaborating institutions: the Massachusetts Institute of Technology (MIT), the Massachusetts Eye and Ear Infirmary (MEEI), Boston University (BU) and the University of North Carolina at Chapel Hill (UNC-CH). Major research efforts are proceeding in four areas: (1) developing and maintaining a laboratory-based, software-controlled, real-time stimulation facility for making psychophysical measurements, recording field and evoked potentials and implementing/testing a wide range of monolateral and bilateral sound-processing strategies, (2) exploring new sound-processing strategies for implanted subjects, and (3) understanding factors contributing to the wide range of performance seen in the population of implantees through psychophysical and evoked-response measures.

This final report addresses two goals: (1) to report on the work accomplished since the 12th Quarterly Progress Report (QPR-12) and (2) to summarize the body of work accomplished over the course of this Contract.

2.0 Progress Since the Originally-Scheduled End of the Contract (12/31/05)

The original scheduled end of the Contract was December 31, 2004. A supplemental quarter of funding (1/1/05-3/31/05) and a no-cost extension (4/1/05-6/31/05) were awarded to facilitate the completion/extension of two general areas of the initially-funded work: (1) the development and delivery of a software/hardware system for the measurement of a wide range of biological and field potentials elicited in the course of intracochlear electric stimulation (e.g., intracochlear-recorded field (artifact) potentials, intracochlear-recorded evoked potentials, scalp-recorded field potentials and scalp-recorded evoked potentials) and (2) experiments with bilaterally-implanted cochlear implant subjects (bCIs) designed to guide future work directed at improving binaural performance in such implantees.

2.1 System for the Measurement of Biological and Field Potentials

This software/hardware system is described in QPR-11 that is submitted (along with a prototype system) at the same time as this Final Report. We refer readers to that QPR for a complete description of that system.

2.2 Preliminary Studies Related to Bilaterally-Implanted Cochlear Implant Subjects (bCIs)

The benefits of binaural over monaural hearing include more accurate sound-source localization and better reception of a target signal that is spatially separated from competing signals (e.g., Wightman and Kistler 1997; Arsenault and Punch 1999). On tests of sound-source localization where wide-band sources are constrained to the front 180^{o} of the horizontal plane, the mean total root-mean-square (RMS) error measured for 12 normal-hearing subjects in the presence of interferers was approximately 4^{o} (Hawley et al. 1999). In quiet, the total RMS error measured for most normal-hearing subjects is 0^o, even when the level is roved by 20 dB (QPR-6, (Poon 2004).

RMS error measured for bilaterally-implanted cochlear implant (bCI) users ranges from approximately 9^o to greater than 50^o depending on the subject, the range of azimuth tested, listening level and degree of level roving {e.g., QPR-6(e.g., Section C-1 of this application, van Hoesel et al. 2002; Eddington et al. 2003; Poon et al. 2003; van Hoesel and Tyler 2003; Litovsky et al. 2004; Nopp et al. 2004). The degree to which localization performance benefits from binaural over monolateral listening also varies by subject from none to highly significant (e.g., Nopp et al. 2004).

We also see this large range of performance in our subjects. For instance, the RMS error measured in a mildly reverberant room with seven speakers evenly distributed from -90^o to $+90^o$ in the front horizontal plane using wide-band noise (20 dB level roving) was 23^o for C109 (best performer) and 40^o for C105 (worst performer).

One unique feature of our approach is that the subjects wear a monolateral cochlear implant system for at least one year before the second ear is implanted. This means we are able to measure the monolateral performance using the monolateral listening strategy they use after at least 12 months of monolateral listening. The results in Table I for one of the three subjects for whom we have significant longitudinal data demonstrate why this approach is important. During the year of monolateral use of the first implant, the subject developed a relatively good monolateral strategy for localization with the RMS error (40^o) significantly better ($p<.01$) than chance (86^o). Her bilateral error measured before beginning to wear two sound processors (36^o) is not significantly better than the "practiced" monolateral score (40^o). After using both implants for 7 months, the monolateral error increased and the bilateral error decreased. One interpretation of these results is that over 7 months of bilateral use, the subject developed a new listening strategy for bilateral input that lowered the bilateral error from 36^o to 27^o but increased the monolateral error from 40^o to 70^o. She was unable to switch back to the earlier strategy for the monolateral listening within the two-hour testing session. Because other investigators compare the equivalent of our 7-month bilateral and monolateral listening errors (70^o vs. 27^o), the degree of bilateral benefit is greatly exaggerated compared to the appropriate 40^o vs. 27^o comparison.

Table I
RMS Error; Subject C092

Listening Condition	Before Bilateral Listening	7 Months Bilateral Listening
Monolateral	40^o	70^o
Bilateral	36^o	27^o

Binaural advantages to speech reception for spatially-separated sources measured with normal-hearing listeners are typically reported as signal-to-noise ratio (SNR) gains in the measurement of the speech reception threshold (SRT) (e.g., Bronkhorst and Plomp 1988; Bronkhorst and Plomp 1989; e.g., Arsenault and Punch 1999). Unfortunately, these advantages are often reported as gains in speech-reception scores measured at a specific SNR in the cochlear implant literature (van Hoesel and Clark 1999; Lawson et al. 2001; Gantz et al. 2002; Müller et al. 2002; Schön et al. 2002; Tyler et al. 2002; van Hoesel et al. 2002). Because of the different measure and the variation across studies of (1) the SNR at which the measure was made, (2) the nature of the noise/competitor signal and (3) the source positions, it is difficult to compare the results for bCI users with those for normal-hearing listeners.

Three studies (37 bCI subjects) report SNR gains in SRT in sufficient detail for computation of specific measures of binaural benefit that can be compared to normal-hearing results: 4 users of the Nucleus CI-24M device (van Hoesel and Tyler 2003), 14 users of the Nucleus CI-24R device (Litovsky et al. 2004) and 19 users of the Med-El Combi 40/40+ device. We computed three measures of the benefit to speech reception for these studies when the (Schleich et al. 2004) sources of the target signal and competing or noise signal are spatially separate[1]: binaural benefit (BB), binaural squelch measuring the positive impact of adding the ear ipsilateral to the noise to the monolateral listening condition (iSQ) and binaural squelch measuring the benefit of adding the ear contralateral to the noise (cSQ). We use these definitions when comparing results for normal listeners and bCI users.

Bronkhorst and Plomp measured a mean BB of -10 dB for 17 normal-hearing listeners using speech-shaped noise and KEMAR-derived signals (Bronkhorst and Plomp 1988). The magnitude of this benefit is significantly greater ($p<0.01$) than the mean BB of -1.8 dB computed from the results reported for 19 Combi users (Schleich et al. 2004), and the -4.3 dB for the CI24M subjects (van Hoesel and Tyler 2003)[2]. The average iSQ measured for the same 17 normal-hearing listeners was 2.5 dB which is significantly greater ($p<0.01$) than the 0.9 dB, 1.0 dB and 1.25 dB measured for the 19 Combi, 14 C24R (Litovsky et al. 2004), and the 4 CI24M users respectively. The reported mean cSQ for the normal-hearing subjects is 13.5 dB which is significantly greater ($p<.01$) than the 7.7 dB, 4.0 dB and 7.4 dB measured for the Combi, C24R and C24M users.

The functional performance of our bilaterally-implanted subjects is consistent with the current literature in showing a wide range of binaural advantage (speech reception in noise and localization) with even the best scores being significantly poorer than normal. For instance, Table II shows measures of BB and iSQ (see Section B, footnote 1 for definitions) with separated sources for two subjects after 7 months using bilateral cochlear implant systems. The speech signals were 16 American consonants in an /aCa/ context spoken by three

Table II
Binaural Advantage to Consonant Reception in Noise

Subject	Mean BB§	Mean iSQ§
C109	19 ± 4.3	7 ± 4.5
C105	2 ± 4.8	0 ± 4.8

§ Gain in percentange points ± standard error

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[1] We define binaural benefit as: $BB = SRT_{S_0N_0} - SRT_{S_0N_{\pm 90}}$, where $SRT_{S_0N_0}$ is the speech reception threshold (SRT) measured binaurally for the signal and noise sources at 0° in the horizontal plane and $SRT_{S_0N_{\pm 90}}$ is the SRT measured binaurally for the signal at 0° azimuth and the noise at +90° (right side) or -90° (left side). We define binaural squelch where the noise in the monolateral listening condition is contralateral to the listening ear/implant as: $iSQ = m_R SRT_{S_0N_{-90}} - SRT_{S_0N_{-90}}$ or $m_L SRT_{S_0N_{+90}} - SRT_{S_0N_{+90}}$, where $m_R SRT_{S_0N_{-90}}$ is the $SRT_{S_0N_{-90}}$ measured monolaterally using the right ear or implant and $m_L SRT_{S_0N_{+90}}$ is the $SRT_{S_0N_{+90}}$ measured monolaterally using the left ear or implant. Similarly, binaural squelch where the noise in the monolateral listening condition is ipsilateral to the listening ear/implant is defined as: $cSQ = m_L SRT_{S_0N_{-90}} - SRT_{S_0N_{-90}}$.

[2] The conditions reported in Litovsky, et al. are not sufficient to compute BB.

talkers {Fu, 2000 #1399}} and the competing signal was speech-shaped noise (Nilsson et al. 1994). Subject C109 enjoys a considerable advantage using two implants while C105 does not.

Sensitivity to interaural time and level differences mediate normal binaural advantages

When localizing wideband sounds in the azimuthal plane that include substantial components below 2 kHz, normal-hearing subjects use ITD and ILD but give stronger weight to ITD (Wightman and Kistler 1992; Macpherson and Middlebrooks 2002). For the wideband, flat-spectrum noise used by Macpherson and Middleboooks, ILD bias weights ranged across subjects from approximately 0.2 to 0.6 while the ITD weights ranged from 0.6 to 1. In the case of speech reception for separated signal and noise sources, Zurek's model (Zurek 1992), which incorporates the impact of head shadow and the binaural interactions associated with ITD and ILD [as implemented in a model of binaural masking-level difference (MLD) (Colburn 1977)], successfully predicts the general pattern of directional advantages measured by Bronkhort and Plomp (1988). Because the MLD decreases from 12 dB to less than 4 dB from 500 to 1500 Hz, one assumes that Zurek's model is influenced mainly by ITD for signals like speech with significant low-frequency energy and mainly by head shadow and ILD for signals with components above 1.5 kHz.

The poor functional performance of bCI subjects is likely due to abnormal ITD and ILD sensitivity

We hypothesize that the poor ITD and poor ILD sensitivity associated with bCI users is responsible for their abnormal functional performance.

ILD sensitivity

The just noticeable difference (JND) in ILD measured in normal-hearing listeners for tones (at 60 dB SPL) varies from approximately 0.6 to 1.1 dB depending on frequency (Yost and Dye 1988) and remains constant across a wide range of comfortable listening levels (Hershkowitz and Durlach 1969). While a number of studies report ILD JNDs for bCI subjects using single interaural-electrode pair stimulation (see below), the one study reporting ILD-JND measures made through the subject's sound processor used broadband sounds like click trains, speech and noise (Pok et al. 2003). Their results are similar to ours with subjects scoring poorer than normal (approximately 2.0 dB vs. 0.6 dB). Our measures of ILD JNDs using tone stimuli delivered through the subject's sound processors (see Table III below) also show abnormal performance for both our best-performing subject (C109: 1.4 dB through processor vs. 0.6 dB for normal listeners) and our poor performing subject (C105: 10 dB vs. 0.6 dB). It should be noted that while C105's performance on tasks related to bilateral stimulation is relatively poor, her monolateral speech-reception for NU6 words in quiet (38%) is only slightly below the median seen at the MEEI and is not consistent with severe CNS dysfunction.

For many bCI subjects, ILD JNDs measured using single, pitch-matched interaural electrode pairs at most-comfortable listening levels fall between 0.07 and 0.7 dB and is sometimes limited by an implant's smallest level step (Tabel III Lawson et al. 1996; Lawson et al. 1998; Lawson et al. 1999; Lawson et al. 2000; Lawson et al. 2002; van Hoesel et al. 2002; Tabel III van Hoesel and Tyler 2003). It is not known how ILD sensitivity varies with level in bCI users. The best ILD JNDs for bCI subjects represent 0.7% of a typical 10 dB dynamic range (DR) for electric stimulation; a value similar to that for normal-hearing subjects. For poorer performing subjects, the ILD JNDs can be greater than normal at more than 4% of dynamic range. In cases where ILD JNDs are measured in both the "through-processor" and "single-pair" conditions, the results of both good- and poor-performing subjects are consistent with the sound-processing strategy limiting ILD sensitivity (see text associated with Tabel III).

ITD sensitivity

In normal-hearing subjects, ITD JND for gated tones decreases from about 75 μs near 100 Hz to 11 μs at 1kHz where it steeply increases to 1.5 kHz beyond which humans show essentially no sensitivity to interaural phase differences. The normal sensitivity for broadband noise and 15 pulses/sec (pps) clicks is approximately 11 μs (Klumpp and Eady 1956). Only the group at Research Triangle Institute (RTI) has reported ITD sensitivity measures made through a subject's sound processor (Lawson et al. 2001). The measures were made in two subjects (Med-El Combi 40) using "click" (approximately 2.5 ms duration) trains at 50 pps with 25-μs ITD JNDs reported for both subjects. It is not clear whether these signals were presented acoustically or through auxiliary inputs, but compared to all other ITD-sensitivity measures (see below), they are the closest to normal. The ITD JNDs measured in our subjects for through-processor testing (see Table IV below) range from 187 μs for one subject listening to 50-pps click (0.2 ms duration) trains to >2 ms for our worst-performing subject listening to 455 Hz tone bursts. In general, the ITD sensitivity for bCI users as measured through their sound processor is substantially poorer than for normal-hearing listeners.

The abnormal ITD sensitivities measured using signals presented to the subject's sound processor are generally consistent with the poor sensitivities measured using a single interaural-electrode pair. ITD JNDs measured using unmodulated pulse trains have been reported for several, interaural-electrode pairs in each of 22 bCI subjects (Table IV this report, Lawson et al. 1996; van Hoesel and Clark 1997; Lawson et al. 1998; Lawson et al. 2001; van Hoesel et al. 2002; van Hoesel and Tyler 2003; Wolford et al. 2003). Of the measures for which the electrode pair and stimulus waveform parameters are provided, 75% are distributed uniformly between 100 μs and 500 μs and 23% are greater than 1 ms. From these well-documented measures, the group at RTI report the shortest ITD JNDs with five interaural pairs below 100 μs (Lawson et al. 1998; Wolford et al. 2003). The RTI group also reports 93 measures where the stimulus conditions are not completely specified (Lawson et al. 2001). Fifteen of these ITD-JND measures are less than 50 μs in 8 of 13 subjects tested with the other 77 measures distributed between 50 μs and 2 ms much like the measures reported by others. It is not known why the smallest ITD JNDs reported by the group at RTI are so much smaller than those reported by other groups (e.g., Table IV this report, van Hoesel and Tyler 2003). It does not seem to be a difference in devices since RTI reports ITD JNDs substantially shorter than 100 μs for both Nucleus and Med-El users.

Taken together, the reported ITD JNDs lead one to conclude that the sensitivity to ITD for bCI users is abnormal and is one important factor responsible for their poor binaural performance.

Stages at which ITD and ILD sensitivity may be limited

The schematic in Figure 1 illustrates stages associated with bilateral intracochlear stimulation that may contribute to poor binaural sensitivity. Depending on the implantee's etiology, the brain structures associated with binaural hearing may be compromised and limit the individual's access to binaural cues; even if the implants were to elicit normal patterns of AN spike activity. Our approach is to focus on the most peripheral and accessible aspects of this processing cascade initially, deferring emphasis on the central processing aspects of the problem. We assume that central processing will be far more successful if ITD and ILD cues are accurately delivered in the periphery.

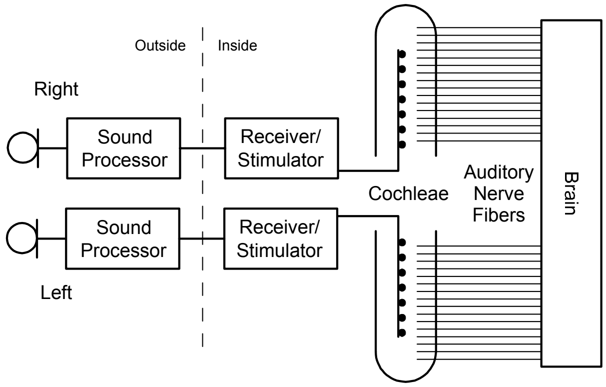

Figure 1. Schematic overview of bilateral, intracochlear stimulation. Beginning from the left: microphones, sound processors, implanted receiver/stimulators (rRSs), cochleae and fibers of the auditory nerves (ANs). The spike activity elicited on the AN fibers of both ears is interpreted by the brain.

Motivating this approach is the fact that the response properties of AN fibers to electric stimulation are far from normal (e.g., Kiang and Moxon 1972; Javel et al. 1987; van den Honert and Stypulkowski 1987; Parkins 1989; Dynes and Delgutte 1992; Litvak et al. 2001; Miller et al. 2001). When compared to acoustic tones of frequency less than 1 kHz for instance, single-electrode monopolar intracochlear stimulation using biphasic pulses (repetition rate < 1000 pps) elicit patterns of spike activity that are (1) broader for the same sensation level, (2) more synchronized to the stimulus period, (3) more coherent across fibers, (4) more likely entrained to the stimulus, and (5) are not delayed as a function of fiber place[3]. We hypothesize that the abnormal spike patterns elicited by bilateral intracochlear stimulation are factors that limit a cochlear implant user's sensitivity to ILD and ITD.

It is also possible that the implanted receiver/stimulator (iRS) impacts bilateral sensitivity by limiting the smallest ILD and ITD that can be represented in the electric stimulus. Our preliminary results suggest this is not the case for the Clarion implant system used by our subjects (see the text associated with Tables III and IV below).

Even when the implanted electronics do not limit binaural sensitivity, it is possible that the sound-processing strategy does. For instance, van Hoesel et al. (van Hoesel et al. 2002; van Hoesel and Tyler 2003) present results that suggest the automatic gain control (AGC) used by Nucleus sound-processor inappropriately limits or distorts ILD information conveyed to the user. Our preliminary results suggest that the AGC associated with Clarion CIS processing also limits ILD sensitivity (see the text associated with Table III and IV below).

ILD sensitivity and the Clarion CII implant system

Preliminary data from our best (C109) and worst (C105) performing bCI subjects shown in Table III indicate that ILD sensitivity as measured through the sound processor is abnormal and that it is unlikely the implanted receiver/stimulator (iRS) limits patient ILD sensitivity. ILD JNDs were measured through the sound processor using wideband noise (activating multiple interaural-electrode pairs) and single tones centered in an analysis channel's band (stimulating a single interaural-electrode pair). See Section D-9 for methods.

C109's ILD JND for noise delivered to the auxiliary (AUX) input was close to the normal JND using wide-band stimuli (0.8 dB vs. 0.5 dB). For tone signals, C109's JNDs

[3] In this document we use the phrase "fiber place" as a short hand to refer to the longitudinal cochlear position of the hair cell a (radial) fiber would normally innervate.

were about twice normal (1.4 dB vs. 0.6 dB) while C105's sensitivity was an order of magnitude worse than normal (10 dB vs. 0.6 dB).

Table III
ILD Sensitivity

Waveform	Subject	ILD JND	
		Through Processor	Single Pair
Wide-Band Noise	C109	0.8 dB	
Tone: 540 Hz	C109	1.4 dB	
Tone: 3590 Hz	C109	1.5 dB	
Tone: 455 Hz	C105	10.0 dB	
Tone: 762 Hz	C105	11.3 dB	
Train: 200 pps	C109		0.07 dB
Train: 1450 pps	C109		0.08 dB
Train: 200 pps	C105		0.27 dB
Train: 200 pps	C105		0.48 dB

ILD JNDs in interaural current level were also measured using single interaural-electrode pairs at comfortable listening levels with our custom laboratory system to directly control the implanted receiver/stimulator (iRS). These "single-pair" ILD JNDs range from 0.07 to 0.48 dB depending on the subject. ILD-JND values for 200 pps (0.07 dB) and for 1450 pps (0.08 dB) indicate ILD JND is not a strong function of pulse repetition rate. The minimum ILD step size for the levels at which these measurements were made is approximately 0.02 dB; close to limiting our ability to measure ILD JND in C109 but not in C105.

The measurements reported in Table III show that ILD JNDs measured at the sound-processor input are limited by the sound-processing strategy. There are at least two stages of the sound-processing strategy that are likely to impact ILD sensitivity: the front-end automatic gain control (AGC) and the nonlinear function that maps the instantaneous output level of each analysis channel to current level (level map). When one accounts for the Clarion CIS sound-processing strategy's level maps, a 0.1-dB ILD JND measured at the iRS is roughly equivalent to a 0.2 dB JND at the mapping function's input. To a first approximation, the stimulus *waveform* delivered to each channel's electrode during each of the tone bursts tested will be a 1450 pps pulse train, because the envelope is a DC increment modulating the 1450-pps carrier for both analysis channels. This means for a 540-Hz tone burst, the ILD JND at the iRS should be comparable to that of a 1450-pps pulse train or 0.08 dB (Table III). When one takes into account the level map, the ILD JND for "through-processor" testing should be 0.2 dB, rather than the 1.4 dB measured (Table III). We hypothesize that this discrepancy is due to a mismatch in the interaural AGC gain.

ITD sensitivity and the Clarion CII implant system

Preliminary data from our best (C109) and worst (C105) performing bCI subjects (Table IV) shows: (1) ITD sensitivity as measured through the sound processor is abnormal and (2) it is unlikely the implanted receiver/stimulator (iRS) limits patient ITD sensitivity. ITD JNDs were measured (Section D-9) through the sound processor using click trains (activating multiple interaural-electrode pairs) and tones centered in an analysis channel's band (stimulating a single interaural-electrode pair). Note that the in these measures, the two sound processors are running asynchronously.

Table IV
ITD Sensitivity

Waveform	Subject	ITD JND Through Processor	ITD JND Single Pair
Click: 50 pps	C109	187 µs	125 µs
Click: 50 pps	C105	1958 µs	539 µs
Tone: 350 Hz	C109	390 µs	
Tone: 540 Hz	C109	750 µs	
Tone: 3590 Hz	C109	700 µs	
Tone: 455 Hz	C105	>2000 µs	
Train: 200 pps	C109		125 µs
Train: 1450 pps	C109		535 µs
Train: 200 pps	C105		288 µs

The lowest ITD JND for 300-ms trains of 50 pps, 0.2-ms clicks delivered to the auxiliary (AUX) input of the sound processors is substantially greater than normal for both C109 (187 µs vs. 11 µs) and C105 (1958 µs vs. 11 µs). The same is true for low-frequency (<600 Hz) tones with the ITD JNDs ranging from 390 µs to greater than 2 ms as compared to the 30 µs to 15 µs for normal hearing listeners (Klumpp and Eady 1956).

Measures of ITD sensitivity for unmodulated pulse trains made with the Clarion iRS using a single pair of interaural electrodes are, for the most part, similar to those reported by other investigators (see Section B). We do not, however, see the below 100-µs ITD JNDs reported by the Research Triangle Group (e.g., Lawson et al. 2001). For interaural pairs that produce fused sound images and are paired in the subject's sound-processing strategy, ITD JNDs measured in C109 (our best performing subject) range from 125 µs to 194 µs (not shown in Table IV) using 50 pps pulse trains and from 539 µs to >2 ms (not in Table V) for C105. As repetition rate increases, the ITD JND increases (e.g., 125 µs at 200 pps vs. 535 µs at 1450 pps for C109).

It is unlikely that the minimum ITD step size used during these experiments (13.5 µs) limited the ITD sensitivities measured. We hypothesize that the abnormal ITD JNDs measured through the iRS are due to poor encoding of ITD in the AN responses to electric stimulation.

User ITD sensitivity is probably also limited by the standard Clarion CIS sound-processing system. The ITD JND measured in subject C109 was approximately 700 µs for both the 540 Hz and 3590 Hz tone-burst stimuli delivered through the AUX input of the sound processor. This is larger than the 535-µs ITD JND for the 1450-pps carrier used by the sound-processing strategy. We hypothesize this is due to the rise time of the analysis channel's filters as described in the next paragraph.

Preliminary measures of ITD sensitivity using unmodulated pulse trains delivered to a single interaural-pair of electrodes indicate that the ITD JND increases significantly when the onset cue is limited by increasing the rise and fall times. For instance, Table V shows the ITD JND for a pulse train (1450 pps; the carrier repetition rate for the Clarion CIS sound-processor strategy), almost doubles when the rise/fall time increases from 0 ms to 20 ms.

Table V
Onset Vs. Ongoing Cues; Subject C109

Stimulus Condition	Waveform	Rise/Fall Time	ITD JND
Single Electrode Pair	1450 pps Pulse Train	0 ms	535 µs
		20 ms	948 µs
Processor AUX Input	540 Hz Tone Burst	0 ms	735 µs
		5 ms	CNT§
		20 ms	CNT§

§ Could Not Test means the subject was unable to complete the task using a maximum ITD of 2 ms.

For the CIS sound-processing strategy, the 540-Hz tone burst centered in an analysis channel's band will produce a stimulus at its interaural-electrode pair that approximates an unmodulated pulse train at 1450 pps. Even when the tone-burst's onset is not limited, the ITD sensitivity through the sound processor does not reach the sensitivity available at the iRS (probably because the envelope's rise time is approximately 10 ms because of the filtering associated with the analysis channel). When a rise/fall time of 5 ms is applied to the tone, the ITD JND becomes longer than 2 ms.

3.0 Summary of Results for the Entire Contract Period

The major accomplishments/findings associated with the work supported by this contract are listed below and grouped by the major goals of the Contract's work scope.

3.1 Develop and maintain a laboratory-based, software-controlled, real-time stimulation facility for making psychophysical measurements, recording field and evoked potentials and implementing/testing a wide range of monolateral and bilateral sound-processing strategies.

- Only our laboratory has been able to synchronize bilaterally-delivered stimuli to within 90 ns using the Clarion CII and 90K class of implants. Because these implant systems are the most flexible devices available, this ability places us in a unique position to move forward developing new sound-processing strategies for bilateral cochlear implants.

- A potential recording hardware/software system was designed, fabricated and tested that enables users to measure a wide range of biological and field potentials elicited in the course of intracochlear electric stimulation (e.g., intracochlear-recorded field (artifact) potentials, intracochlear-recorded evoked potentials, scalp-recorded field potentials and scalp-recorded evoked potentials). Biological events as close as 20 μs to the generating stimulus can be captured with surface recording electrodes at sampling rates up to 400 kHz. This system was delivered to the Project Officer for loan to other investigators (QPR-11).

3.2 Develop and test new ideas for improving performance of cochlear implant users.

- Psychophysical measures demonstrated that triphasic waveforms result in substantially less nonsimultaneous electrode interaction than biphasic waveforms (QPR-1 and QPR-3).

- Psychophysical measures of nonsimultaneous electrode interaction demonstrated considerable variance across subjects for both biphasic and triphasic waveforms (QPR-9).

- A single-fiber model of nonsimultaneous electrode interaction was developed and tested. It predicts the patterns of biphasic and triphasic interaction measured in a group of cochlear implant subjects (QPR-9) and provides an unparalleled understanding of the physiological mechanisms underlying nonsimultaneous electrode interaction.

- Chronic testing of a CIS sound-processing strategy based on triphasic carrier waveforms in a group of cochlear implant subjects showed that only 4 out of 10 subjects benefited significantly (QPR-9).

- A measurement suite was developed and tested that identifies electrodes that should be paired interaurally when implementing *binaural* sound-processing strategies (QPR-3, QPR-5 and QPR6).

- The testing methodologies currently used to compare the benefit of monolateral vs. bilateral implantation were shown to over estimate the benefit of bilateral implantation. An alternative approach was demonstrated to overcome this fundamental flaw (QPR-6 and this Final Report).

- Preliminary data were acquired that identified several characteristics of today's bilateral sound-processing strategies that limit ITD and ILD sensitivity (see Section 2.2 of this report). Given these results, we (and others) will focus effort on the development of sound-processing strategies designed to overcome these specific limitation and improve the performance of bilateral cochlear implants.

3.3 Develop a better understanding of factors contributing to the wide range of performance seen in the population of implantees

- The new biological recording system described in Section 3.1 can be used with scalp electrodes to record the pulse-by-pulse stimulus artifact generated a cochlear implant system to determine whether the system is operating correctly. This is important when assessing factors contributing to performance. If the variance in across-subject performance associated with device (hardware/software) malfunction is not eliminated, the likelihood of identifying anatomical/physiological factors that influence performance is low. See QPR-2, QPR-7 and QPR-11.

- The new biological recording system has enabled us to make detailed and highly repeatable IEP measures that can be used in a number of ways. For instance, measures of electrode interaction based on IEPs recorded using this system have identified an (until now unknown) interaction phenomenon: hyperinteraction (QPR-4, QPR-8, QPR-10, QPR-12). IEP measures such as these are an essential component of a new approach to identifying the peripheral mechanisms accounting for the across-subject variance in patient performance.

- Subject-specific electroanatomical models (EAMs) of the mammalian cochlea have been developed that predicts both the IEP waveforms and the IEP-based measures of electrode interaction phenomena (e.g., spatially asymmetric distribution, hyperinteraction and inverse interaction) measured in human subjects. See QPR-12.

- Combining the ability to measure IEPs for a wide range a stimulating conditions and the ability of EAMs to predict those measures for specific subjects now provides the ability identify the degree to which specific anatomical and physiological factors in the periphery account for the variation in performance seen across subjects. In the case of electrode interaction, for instance, properties of the EAM can be manipulated (e.g., number and distribution of nerve fibers, electrode position and new bone formation) to determine the factors that account for the across-subject variance in the magnitude of interaction measured. This general approach provides the basis for identifying mechanisms that influence performance and will enable designers of new sound-processing strategies to focus on specific factors that limit performance.

4.0 References

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