

# Comparison of Electrically Evoked Whole-Nerve Action Potential and Electrically Evoked Auditory Brainstem Response Thresholds in Nucleus CI24R Cochlear Implant Recipients

Marcia J. Hay-McCutcheon\*  
Carolyn J. Brown\*<sup>†</sup>  
Kelly Schmidt Clay\*  
Keely Seyle<sup>†</sup>

## Abstract

In this study, differences between electrically evoked whole-nerve action potential (EAP) and electrically evoked auditory brainstem response (EABR) measurements within Nucleus CI24R cochlear implant recipients were evaluated. Precurved modiolus-hugging internal electrode arrays, such as the CI24R, are designed to provide more direct stimulation of neural elements of the modiolus. If the electrode array is closer to the modiolus, electrically evoked and behavioral levels might be lower than were previously recorded for the straight electrode array, the CI24M. EAP and EABR growth functions and behavioral levels were obtained for 10 postlingually deafened adults. Results revealed no significant differences between EAP and EABR threshold levels, and these levels were not significantly lower than those obtained using the CI24M.

**Key Words:** Cochlear implants, electrical action potential, electrical auditory brainstem response, electrically evoked auditory potentials, neural response telemetry

**Abbreviations:** C-level = upper comfort level; EABR = electrically evoked auditory brainstem response; EAP = electrically evoked whole-nerve action potential; MAP = speech encoding program downloaded to speech processor; MP1 = monopolar ball electrode; MP2 = monopolar plate electrode; NRT = neural response telemetry; PCI = Programming Computer Interface; T-level = threshold level

## Sumario

En este estudio se evalúan las diferencias entre los potenciales auditivos del nervio evocados eléctricamente (EAP) y las respuestas auditivas del tallo cerebral evocadas eléctricamente (EABR) medidos en pacientes con un implante coclear Nucleus CI24R. Los dispositivos de electrodos internos, precurveados para abrazar el modiolus (HD1), tales como el CI24R, están diseñados para aportar una estimulación más directa de los elementos neurales del modiolus. Si los electrodos están más cercanos al modiolus, el nivel de los estímulos conductuales y el de los eléctricamente evocados puede ser menor que los previamente registrados para electrodos rectos (CI24M). Se obtuvieron funciones de crecimiento para EAP y para EABR y los niveles conductuales en 10 adultos sordos post-lingüísticos. Los resultados no revelaron diferencias significativas entre los niveles umbrales de las EAP y las EABR, y estos niveles no fueron significativamente menores que los obtenidos utilizando el CI24M.

**Palabras Clave:** Implantes cocleares, potencial eléctrico de acción, potencial auditivo eléctricamente evocado, respuesta auditiva eléctrica del tallo cerebral, telemetría de respuesta neural

**Abreviaturas:** C-level = nivel superior de comodidad; EABR = respuesta auditiva del tallo cerebral eléctricamente evocada; EAP = potencial auditivo del nervio eléctricamente evocado; MAP (HD2) = programa codificador de lenguaje grabado en el procesador del lenguaje; MP1 = electrodo monopolar de bola; MP2 = electrodo monopolar de placa; NRT = telemetría de respuesta neural; PCI = interfase de computadora para programación; T-level = nivel umbral

---

\*Department of Speech Pathology and Audiology and <sup>†</sup>Department of Otolaryngology—Head and Neck Surgery, University of Iowa, Iowa City, Iowa

Reprint requests: Marcia J. Hay-McCutcheon, Department of Speech Pathology and Audiology, 119 Wendell Johnson Speech and Hearing Center, Iowa City, IA 52242

Cochlear implant technology has made a number of advances during the past 25 years. The original cochlear implant provided only a single channel of stimulation and limited flexibility in speech processing capabilities. Current cochlear implants allow for the simultaneous stimulation of multiple channels and are designed to provide a selection of high-rate speech encoding strategies to the recipient (Wilson et al, 1995; Skinner et al, 2000). The design of the internal array electrode has also evolved such that current electrode arrays sit closer to the modiolus than was possible with previous technology (Gstoettner et al, 2001).

Research conducted using temporal bones from Nucleus 22-channel cochlear implant donors, along with animal studies, has suggested that as the internal array is moved closer to the modiolus, more direct stimulation of the spiral ganglion cells occurs. This occurrence allows for more focused or discrete electric current fields within the cochlea (Shepherd et al, 1993; Kawano et al, 1998). It is hoped that the closer proximity of the point of stimulation to the neural fibers will result in lower stimulation thresholds, increase individual dynamic ranges, reduce the amount of channel interaction associated with closely spaced electrodes, and reduce power consumption, thereby prolonging battery life (Fayad et al, 2000; Roland et al, 2000; Richter et al, 2001). To achieve these goals, Cochlear Corporation introduced their version of the perimodiolus-hugging array (CI24R) in October 2000.

There are several differences between the electrode array of the CI24R device and Cochlear Corporation's previous generation of implant, the CI24M. The electrode array of the CI24R contains 22 electrodes and is precurved. This array tapers in size from 0.8 to 0.5 mm. The electrode array of the CI24R is held in a straight position during insertion by a metal spine or stylet. During the insertion process, the stylet is removed, allowing the array to curve around the modiolus. The electrode contacts of the Contour device are "half bands" that are arranged logarithmically along the array. Electrodes at the base are spaced further apart than electrodes at the apex. In addition, the receiver/stimulator is smaller in volume than the CI24M. This design represents a significant change from the previous design for the CI24M device that contains 22 full-banded electrodes that were equally spaced along the array and positioned along the lateral wall of the scala tympani.

Both the CI24M and CI24R devices are equipped with a neural response telemetry (NRT) system that is capable of assessing the

integrity of the electrodes in addition to measuring electrically evoked whole-nerve action potentials (EAPs). The EAP is recorded as a negative peak (N1) with a latency of approximately 0.2 to 0.4 msec, followed by a positive peak (P2). The amplitude and morphology of these action potentials show large variability across subjects, reflecting, at least in part, differences in the underlying properties of individual neural populations (Brown and Abbas, 1990; Brown et al, 1996, 1998; Abbas et al, 1999).

One reason this technology was incorporated into the device was to assist with the programming of speech processors for pediatric patients. Brown and colleagues, (2000) and Hughes and colleagues, (2000) demonstrated that individual EAP responses can be used in combination with a small amount of behavioral information to program cochlear implants for small children and difficult-to-test patients. Prior to the introduction of NRT, the electrically evoked auditory brainstem response (EABR) was used in much the same manner to facilitate programming of the speech processor (Shallop et al, 1990; Mason et al, 1993; Brown et al, 1994, 1999; Hodges et al, 1994; Firszt et al, 1999). Research has shown that EABR testing can be used in the same way as EAP testing for programming the implant (Brown et al, 2000). EABR testing may be preferable to EAP testing for individuals who have malformed cochleas, otosclerotic bone growth, or significant ossification of the cochlea. In these patients, the electrical stimulus artifact that is generated can be large and atypical. These atypical electrical fields can distort the EAP response, making this measurement impossible.

The major practical limitation of EABR is that it requires a very quiet subject. During EABR testing, adults are required to remain fairly motionless and small children are sedated. In that regard, EAP testing does have advantages over EABR testing. Specifically, EAP measurements are near-field recordings requiring no surface recording electrodes and are not contaminated by muscle artifact, as is frequently the case during EABR recordings for which surface recording electrodes are required. Individuals undergoing EAP testing are free to move around, get involved in a quiet activity, read, or watch a video. The number of sweeps required for EAP testing is between 100 and 200, making EAP data collection much more time efficient than EABR testing, which typically requires 1000 to 2000 sweeps (Gantz et al, 1994; Brown et al, 1996, 2000; Hughes et al, 2000).

Previous research using patients implanted with the CI24M device has shown that EAP and EABR thresholds are highly correlated (Luk, 1999; Brown et al, 2000). Whether that relationship is still strong in the CI24R device remains to be seen. The CI24R electrode array is designed to sit closer to the modiolus than the CI24M electrode array does. The closer proximity of the electrode contacts to neural tissue may lead to more selective neural stimulation than was possible with the CI24M. Theoretically, moving the electrode array closer to the modiolus should result in lower threshold (T) and comfort (C) levels. However, the size and shape of the electrode contacts have also been changed in the CI24R device. How these changes in the internal electrode array will affect EAP and EABR thresholds and their relationship to behavioral T- and C-levels is unclear.

The primary purpose of this study, therefore, was to evaluate how the physical changes in the design of the CI24R device impact the physiologic and behavioral measures. This goal is accomplished by evaluating the differences between EAP and EABR measurements using a within-subject design for patients using the Nucleus CI24R Contour implant. The data obtained from this study will be compared with similar, previously published data collected from our laboratory for CI24M cochlear implant users. If the intracochlear electrode array of the CI24R is closer to the modiolus, one might expect behavioral T- and C-levels as well as EAP and EABR thresholds to be lower than previously was observed for the CI24M (Brown et al, 2000).

## METHOD

### Subjects

Ten postlingually deafened adults, six males and four females ranging in age from 33 to 87 years, who received the Nucleus CI24R Contour device at the University of Iowa Hospitals and Clinics between March 2000 and January 2001, participated in this study. All subjects experienced an uneventful surgery and each had a fully inserted electrode array. These individuals received the Sprint Speech Processor approximately 1 month postsurgery. Impedance telemetry testing, using the Cochlear Corporation WinDPS software version 116.02, revealed that all electrodes for each subject had normal electrode impedance levels. At the time of testing, individual implant experience varied from 3 to 13 months, and T-levels and maximum

C-levels had stabilized. Five subjects were programmed with the Advanced Combination Encoder (ACE) speech encoding strategy, one was programmed with the Continuous Interleaved Sampling (CIS) strategy, and four were programmed using the spectral peak (SPEAK) strategy.

### General Procedures

Prior to beginning physiologic testing, behavioral measures of T- and C-level for a 250 pulses/sec stimulus and a 49 pulses/sec stimulus were obtained. Psychophysical measures of T- and C-levels were collected on electrodes 5, 10, 12, and 20 using Cochlear Corporation's WinDPS software and the Programming Computer Interface (PCI). The stimuli were presented in a monopolar fashion using MP1, the monopolar ball electrode, as a reference. A pulse width of 25  $\mu$ sec/phase was used for both stimuli. The subjects were required to estimate the softest level of detection (T-level) and to judge the upper C-level for each stimulus presentation. The stimulus duration and the interstimulus interval were both set at 500 msec for the 250 pulses/sec stimulus. T- and C-levels were also obtained for the stimulus used to record the EABR.

Once behavioral measures were completed, surface recording electrodes were applied and subjects were seated in a reclining chair and encouraged to sleep while the EABR was measured. EABR testing lasted approximately 1 hour. After the completion of EABR testing, the surface electrodes were removed and a short break was provided to the subject. EAP testing followed, which took approximately 30 minutes to complete. During this testing, subjects chose to read or watch a video while sitting in a reclining chair. The entire data collection session took between 2 and 2½ hours to complete.

### EABR Recording Procedures

The stimuli used to record the EABR consisted of a series of 25  $\mu$ sec/phase biphasic current pulses, which were presented to the subject at a rate of 49 pulses/sec via the Sprint speech processor. The PCI provided an external trigger, which was used to initiate averaging. Auditory brainstem responses were recorded on a separate computer equipped with a 16-bit analog-to-digital converter and an RF-shielded, ground-isolated differential amplifier with a gain of 10,000. The electroencephalographic signal was recorded from surface electrodes positioned on

the vertex (positive), contralateral mastoid (reference), and ipsilateral mastoid (ground). A sampling rate of 50 kHz was used. Artifact rejection was used to minimize contamination by myogenic activity. These stimulating and recording parameters were the same as those used by Brown and colleagues (2000) so that direct comparisons of results from both studies could be made.

Owing to the constraints associated with the current Cochlear Corporation EABR software, it is not possible to alternate the polarity of the presented current pulse, and, subsequently, this limitation can lead to considerable stimulus artifact. In an attempt to minimize the stimulus artifact present in some recordings, the first 0.8 msec of the averaged waveform was digitally removed and linear regression techniques were used to determine the equation of a line that best fit the waveform. The slope of this line was primarily determined by stimulus artifact. This regression line was then used to correct the raw EABR waveform to minimize artifact contamination. This is a technique used to record EABR data from subjects with the CI24M device and is one that we have used widely to obtain EABR measures in Nucleus cochlear implant users (Abbas and Brown, 1991; Brown et al, 1994, 1999, 2000). EABR threshold was estimated based on these corrected waveforms.

Growth functions were obtained for electrodes 5, 10, and 20. If time permitted, growth functions also were obtained on electrode 12. Stimulus levels were initially set 5 to 10 levels higher than the threshold levels indicated for the 49 pulses/sec pulse train and thereafter were raised by five programming units until a wave V response was clearly identified, which was typically within the subject's dynamic range for the stimulus used to evoke the EABR. Step sizes of two to three units were used near the threshold. Actual EABR threshold was defined as the lowest level at which two replications of the stimulus level resulted in a detectable wave V as judged by two raters experienced with EABR measures. When different judgments resulted (within 3 units), the more conservative threshold value was chosen.

### **NRT Recording Procedures**

The EAP is recorded using the NRT software, which measures the voltage across a pair of electrodes after the presentation of electrical stimulation. Data were collected using version

2.01 of the NRT software. In an attempt to reduce the stimulus artifact present in the EAP recordings, an on-line subtraction procedure is used after the data have been collected. Data collection occurs in four different conditions: (1) a probe-alone condition, (2) a masker-plus-probe condition, (3) a masker-alone condition, and (4) a control condition in which the switching artifact associated with the presentation of a zero-amplitude probe pulse is recorded. Each of these four recording conditions consists of 16 samples obtained at a rate of 10 kHz using a time window of 1.5 msec. Sampling begins after presentation of the probe pulse and a short user-controlled sampling delay. A second set of the four recordings is then made using the same stimulation parameters but with the onset of the sampling window slightly delayed in comparison with the first recording. These two sets of samples are interleaved with one another, resulting in a 1.6-msec sampling time window and an overall sampling rate of 20 kHz (see Brown et al, 2000, for further detail).

To record the EAP, subtraction of stimulus artifact is required. The stimulus artifact is removed through a process that takes advantage of the refractory properties of the auditory nerve and involves the presentation of individual probe and masker conditions and a masker-plus-probe condition. In the probe-alone condition, the resultant response contains both the stimulus artifact and the neural response, and in the masker-plus-probe condition, the auditory nerve will fire in response to the masker and be unable to fire in response to the probe. The EAP response without the stimulus artifact can then be obtained by subtracting the recording made in the masker-plus-probe condition from the probe-alone condition.

Successful recording of an EAP requires systematically changing a series of specific stimulating and recording parameters. In this study, we used the guidelines set by Abbas and colleagues (1999) for this purpose. These guidelines suggest using the following sequential steps to obtain an appropriate EAP response: (1) lengthen sampling delay, (2) change recording electrode, (3) adjust probe level, and (4) change amplifier gain from 60 to 40 dB. The masker advance, or the time between masker and probe pulses, was always set at 500  $\mu$ sec. When beginning each session, the sampling delay was set at 60  $\mu$ sec, and the amplifier gain was set at 60 dB. The stimulating electrode pair was chosen so that one intracochlear electrode was referenced to the MP1 ball electrode, and the recording pair was

chosen with a different electrode referenced to MP2, the plate electrode. Responses were collected using 100 to 200 sweeps and the high-resolution option was always activated, resulting in a sampling rate of 20 kHz. Initially, stimulation began at levels near the top of the subject's dynamic range.

Changes in the parameters were adaptively varied from subject to subject so that representative growth functions could be obtained. If little improvement of the response was indicated with two changes of delay, the recording electrode was then changed. In some instances, the recording electrode was changed from two to five electrodes away from the stimulating electrode. In addition, changing the recording electrode in a basal direction sometimes provided a more robust response than changing the electrode in an apical direction. Table 1 shows the parameters that were controlled by the experimenter and the settings used for each subject. These data are provided for electrodes 5, 10, 12, and 20.

Recording parameters indicated for each subject are the stimulating electrode, recording electrode, sampling delay, amplifier gain, and number of sweeps. The mean and range values of the parameters are indicated at the bottom of each electrode grouping.

Once stimulating and recording parameters were adjusted to yield an optimal response, EAP growth functions were obtained on electrodes 5, 10, 20, and, when time permitted, 12. EAP growth functions were obtained by decreasing the probe level in steps of five programming units while keeping the level of the masker constant, which was typically near the upper comfort level. Step sizes of approximately two programming units were used for stimulus levels near threshold.

EAP threshold was determined by identifying a single high-level response waveform to serve as a template. Recordings obtained at lower levels were then scaled and cross-correlated with the template response. The

**Table 1 Recording and Stimulating Parameters for the CI24R Device**

<i>Subject</i>	<i>SE</i>	<i>RE</i>	<i>DL</i>	<i>GN</i>	<i>SWPS</i>	<i>SE</i>	<i>RE</i>	<i>DL</i>	<i>GN</i>	<i>SWPS</i>
24R-1	5	2a	90	60	100	10	5a	110	60	100
24R-12	5	2a	100	60	200	10	3b	100	60	200
24R-3	5	2a	70	60	100-200	10	3a	130	60	100-200
24R-14	5	4a	110	60	200	10	5a	100	60	200
24R-16	5	2a	100	60	100-200	10	3a	150	60	100-200
24R-2	5	3a	90	60	200	10	3a	120	60	200
24R-13	5	2a	90	60	100	10	3b	90	60	100
24R-4	5	2b	70	60	100	10	4b	60	60	100
24R-9	5	2a	100	60	100	10	2a	90	60	100
24R-7	5	2a	90	60	100-200	10	3b	90	60	100-200
Mean	NA	NA	91	60	NA	NA	NA	104	60	NA
Range	NA	2b-4a	70-110	0	100-200	NA	4b-5a	60-150	60	100-200
24R-1	12	3b	90	60	100	20	5b	90	60	100
24R-12	NA					20	5b	100	60	200
24R-3	12	3a	90	60	100-200	20	2a	90	60	100-200
24R-14	NA					20	3b	130	60	200
24R-16	12	3a	100	60	100-200	20	2a	130	60	100-200
24R-2	12	3a	160	60	200	20	5b	150	60	200
24R-13	12	4b	60	60	100	20	2a	90	60	100
24R-4	12	4b	70	60	100-200	20	3b	80	60	100
24R-9	12	2b	90	60	100	20	3b	70	60	100
24R-7	12	3b	90	60	100-200	20	5b	60	60	100-200
Mean	NA	NA	93.8	60	NA	NA	NA	99	60	NA
Range	NA	4b-3a	60-160	60	100-200	NA	5b-a	70-150	60	100-200

Mean and range values are provided for recording electrode (RE), sampling delay (DL), and gain (GN). Mean values are not provided for the number of sweeps (SWPS) because a range of 100 to 200 sweeps was used within single recordings of electrodes for specific subjects. The values provided for the RE indicate the number electrodes apically (a) distant from and basally (b) distant from the stimulating electrode (SE). For example, if the RE is two away from the SE in a basal direction, then "2b" represents the recording electrode location. Similar nomenclature is used for recording electrodes located away from the stimulating electrode in an apical ("a") direction.

EAP threshold was defined as the largest level that resulted in a correlation coefficient of .8 or greater. Previous results in our laboratory from children indicate that visual and correlation methods of threshold detection are highly correlated (Hughes et al, 2000). In two instances, the visual detection thresholds differed significantly from the correlational thresholds. In those two instances, the visual detection thresholds were used rather than the correlational thresholds. To ensure that these visual detection thresholds were appropriate, a second audiologist experienced in using both the template method and the visual detection method also examined the EAP data.

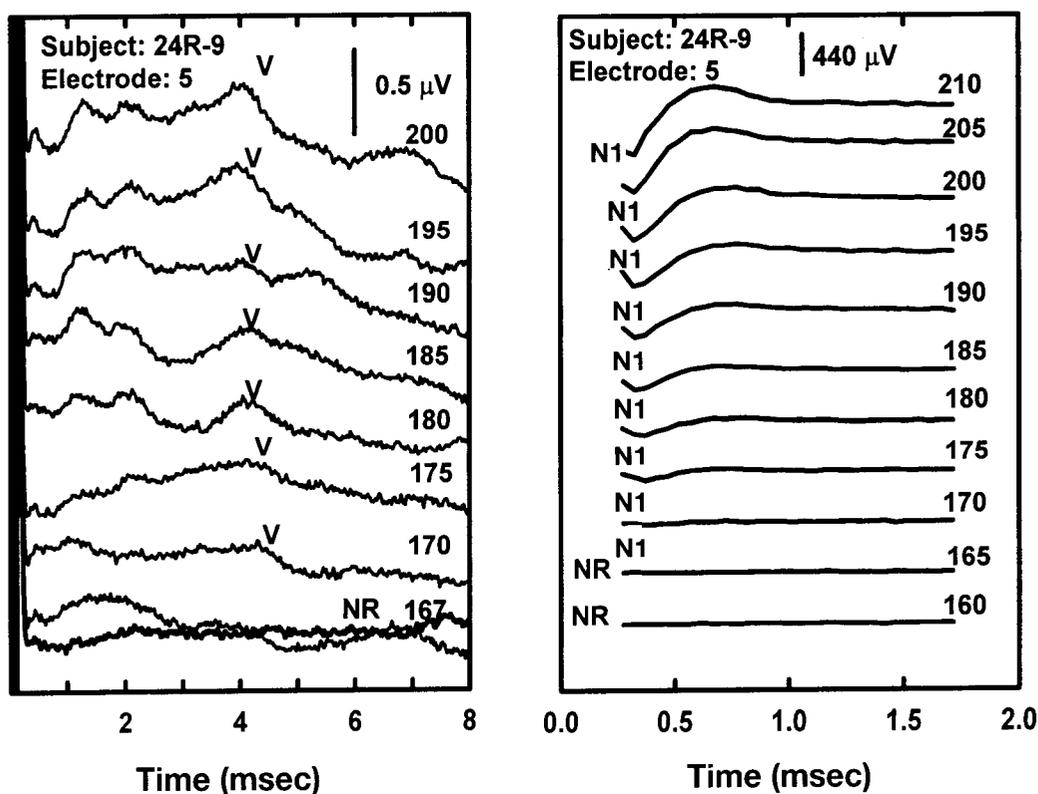
## RESULTS

Figure 1 displays examples of fairly robust EABR and EAP waveforms obtained from a Nucleus CI24R cochlear implant subject. The left panel shows a series of EABR responses obtained for electrode 5 as the stimulus was progressively increased from 167 to 200 programming units. Wave V of the EABR is indi-

cated for each waveform. The two waveforms recorded at 167 units clearly indicate a non-response. The visual detection threshold for this series of waveforms was 170 programming units.

The right panel of Figure 1 shows the EAP responses recorded for the same electrode and subject. A large negative peak, N1, with a latency of approximately 0.38 msec measured at high levels, can be seen followed by a positive component, P2. The amplitude of the response is determined from the negative trough to the positive peak. In this example, the amplitude gradually decreases from 797  $\mu\text{V}$  at 210 programming units to 0  $\mu\text{V}$  at 165 programming units. The very small response at 170 programming units, 47  $\mu\text{V}$ , was clearly discernable on a larger scale and was highly correlated ( $r = .88$ ) with the high level response at 210 programming units. EAP threshold was determined to be 170 programming units both by correlation analysis and visual detection.

The main objective of this study was to compare EABR and EAP thresholds within subjects. Figure 2 provides these individual subject data. The stimulus threshold levels for EABR,

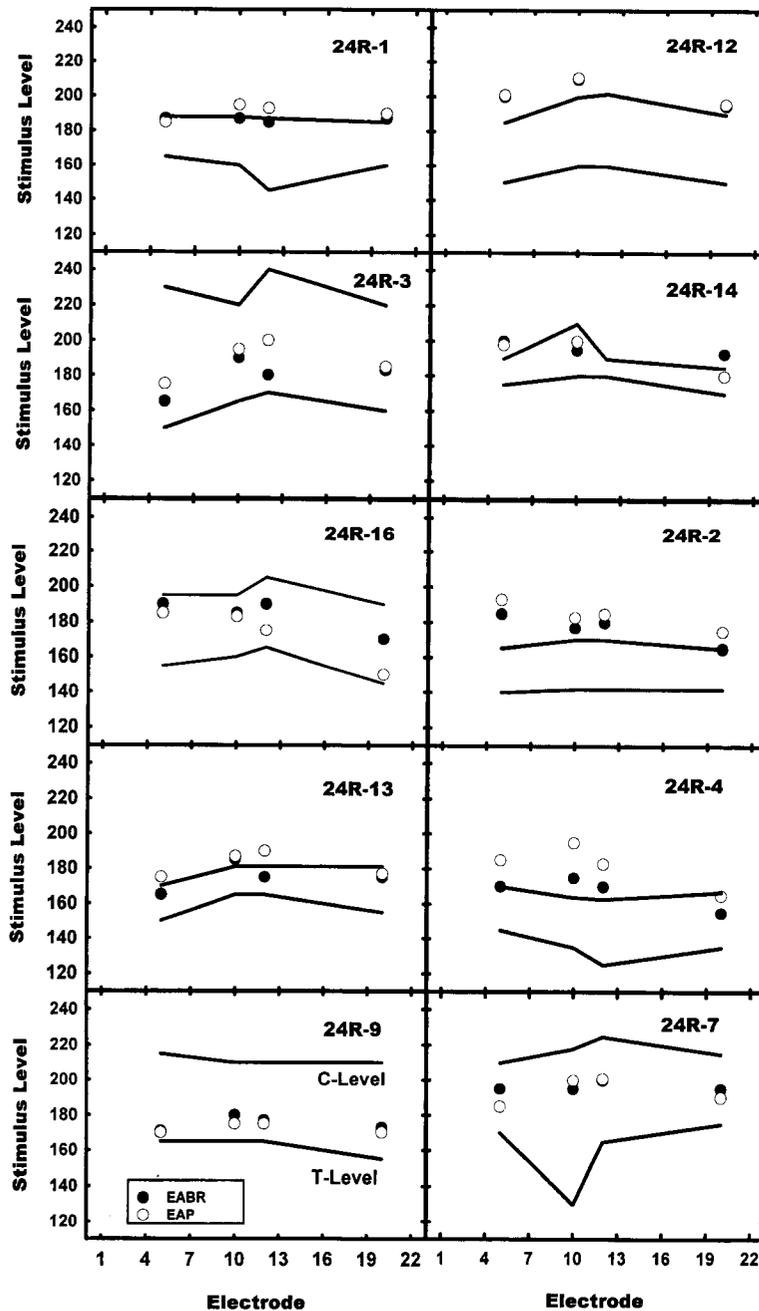


**Figure 1** Electrically evoked auditory brainstem response (EABR) and electrical action potential (EAP) growth functions for one Nucleus CI24R subject. The left panel shows the EABR waveforms and the right panel displays the EAP responses. Stimulus levels (in programming units) used to elicit each response are indicated at the right of each waveform. Waveform amplitudes are indicated in  $\mu\text{V}$ . No-response waveforms are indicated by NR.

EAP, and behavioral T- and C-levels are indicated for electrodes 5, 10, and 20 for each subject. Additionally, these data are provided for electrode 12 for eight subjects. The solid lines show the behavioral T- and C-levels, and the EABR and EAP thresholds are indicated by the filled and open circles, respectively. Generally, EABR and EAP thresholds are similar to each other. In 31.6 percent of cases, EAP thresholds exceed EABR thresholds by more than five programming units. In 7.9 percent of cases, EABR thresholds exceed EAP thresholds by more than five programming units. This figure also illustrates

the variability of EAP and EABR thresholds relative to the MAP dynamic range. In some subjects, EAP and EABR thresholds are recorded near behavioral T-level (e.g., 24R-9), whereas in other subjects, both EAP and EABR thresholds exceed C-levels (e.g., 24R-12 and 24R-2).

Figure 3 shows the relationship between EAP and EABR thresholds, pooled across electrodes, for the 10 Nucleus CI24R CI users participating in this study. For comparison purposes, similar data obtained previously from CI24M CI users (Brown et al, 2000) are shown. Both graphs show EAP thresholds plotted as a function of

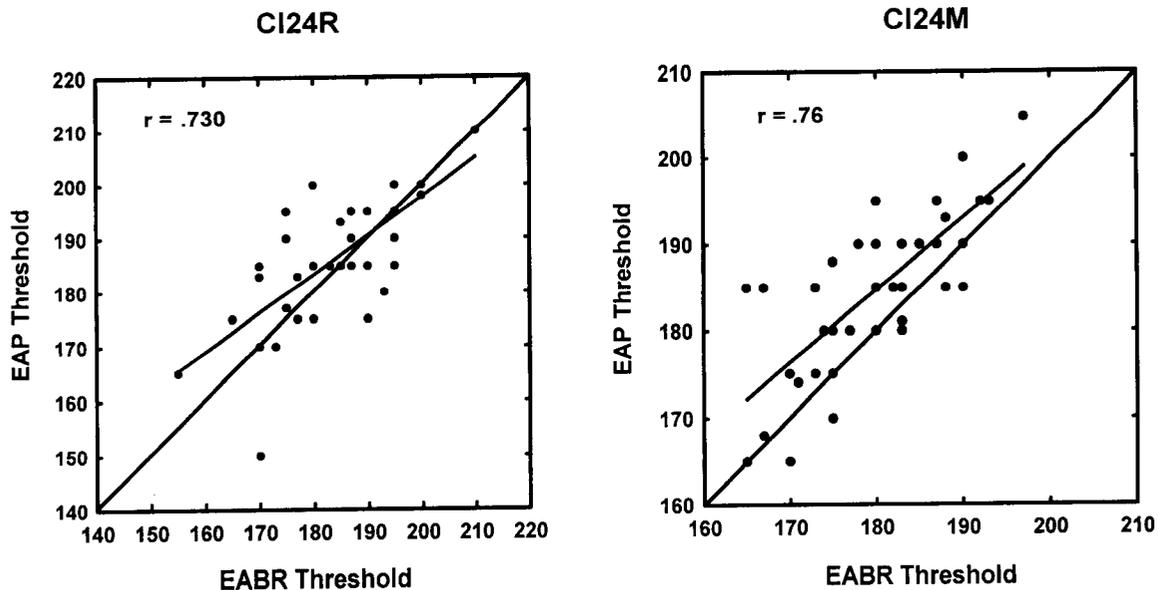


**Figure 2** Individual electrophysiologic and behavioral threshold data (electrodes 5, 10, 12, and 20) for 10 Nucleus CI24R subjects. The filled and open circles represent electrically evoked auditory brainstem response (EABR) and electrical action potential (EAP) values, respectively. The lower solid line represents individual behavioral threshold level. The upper solid line represents individual behavioral upper comfort level.

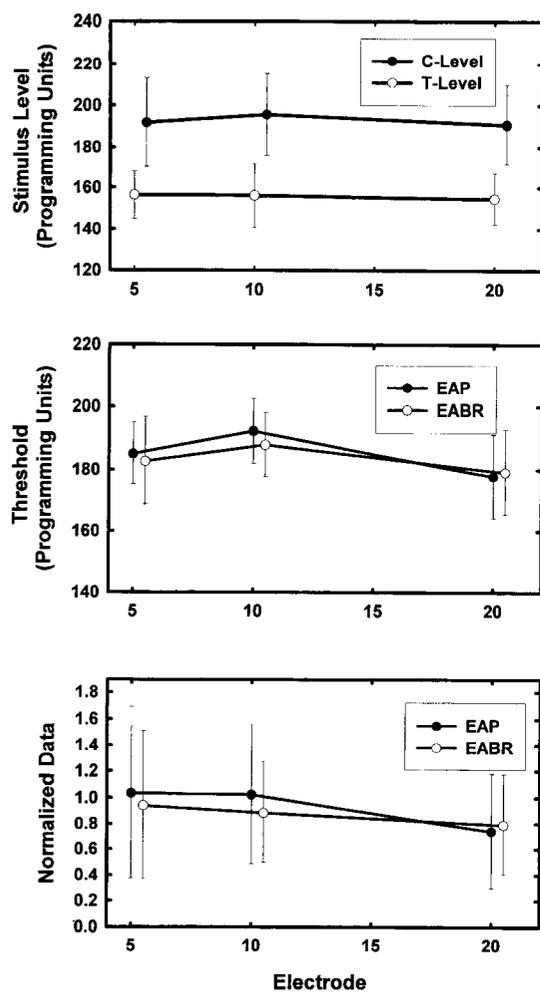
EABR thresholds. In both cases, EAP and EABR thresholds were recorded using monopolar stimulation with 25  $\mu$ sec/phase biphasic pulses. The solid diagonal line in both panels represents the values at which the EAP and EABR thresholds are equal. The shorter line is the best fitting linear regression line. The correlation between the EAP and the EABR threshold for the CI24R implant is .73 ( $p < .001$ ) and for the CI24M implant is .76 ( $p < .001$ ). For both array types, the mean EABR threshold is slightly lower than the EAP threshold, but this difference is smaller for the CI24R than the CI24M device (2.58 units vs 4.73 units). Paired  $t$ -tests revealed that the difference between the EAP and EABR threshold for the CI24R was not statistically significant ( $t = 1.795$ ,  $df = 37$ ,  $p = .081$ ) but was statistically significant for the CI24M ( $t = 5.644$ ,  $df = 47$ ,  $p < .0001$ ). Additionally,  $t$ -tests performed between the CI24R and CI24M devices revealed no significant differences for EABR or EAP thresholds (EABR  $t = 1.36$ ,  $df = 84$ ,  $p = .177$ ; EAP  $t = 0.362$ ,  $df = 84$ ,  $p = .718$ ).

Whereas Figure 3 shows pooled electrode data, Figure 4 shows the mean behavioral and physiologic thresholds for the individual electrodes tested. These mean data from electrodes 5, 10, and 20 are shown. The mean data points ( $\pm 1$  SD) are slightly offset at each electrode

interval to more clearly display the results. Examination of the mean T- and C-levels in the top panel reveals that the T-levels for all three electrodes center around 156 programming units, and the C-levels center around 193 programming units. The EAP and EABR thresholds shown in the middle panel indicate that the mean threshold values are similar for each electrode. In addition, these values are somewhat reduced for the apically placed electrode 20 in comparison with the mean thresholds revealed for electrodes 5 and 10. When comparing evoked potential thresholds for electrodes 5, 10, and 20, a repeated measures one-way analysis of variance (ANOVA) revealed a significant difference for EAP thresholds ( $F = 8.138$ ,  $df = 2$ ,  $p = .003$ ). Pairwise comparisons indicated that EAP thresholds were significantly different for electrodes 10 and 20 ( $p = .002$ ) but not for electrodes 5 and 10 or 5 and 20. Similar statistical analyses for EABR thresholds ( $F = 3.623$ ,  $df = 2$ ,  $p = .048$ ) revealed that thresholds also were significantly different for electrodes 10 and 20 ( $p = .003$ ) but not for electrodes 5 and 10 or 5 and 20. In the bottom panel, the EAP and EABR thresholds have been normalized so that the mean EAP and EABR thresholds for the 10 subjects are expressed as a percentage of each subject's dynamic range (i.e., the 250 pulses/sec



**Figure 3** The relationship between electrically evoked auditory brainstem response (EABR) and electrical action potential (EAP) thresholds for CI24R and CI24M cochlear implant recipients. The *left panel* provides threshold values in programming units for the 10 CI24R subjects evaluated in this study. The *right panel* displays data from Brown et al (2000) for 14 (16 ears) CI24M subjects. Mean EAP and EABR CI24R thresholds for the 38 electrodes tested were 185.58 ( $\pm 11.95$ ) programming units and 183.00 ( $\pm 12.14$ ), respectively. Mean EAP and EABR CI24M thresholds for the 48 electrodes evaluated in the Brown et al (2000) study were 184.77 ( $\pm 8.75$ ) and 180.04 ( $\pm 7.94$ ), respectively.



**Figure 4** Mean and normalized behavioral and physiologic data for electrodes 5, 10, and 20. The data points ( $\pm 1$  SD) are offset at each electrode to more clearly display the results. The *filled circles* in the top panel represent mean upper comfort level for each electrode, and the *open circles* represent mean threshold level. The middle panel shows mean EAP and EABR thresholds using *filled* and *open circles*, respectively. The bottom panel displays EAP (*filled circles*) and EABR (*open circles*) normalized data, where a value of 1 means that the threshold was recorded near upper comfort level and 0 means the threshold was recorded near threshold level.

T-level is subtracted from the physiologic threshold and then divided by the difference between the C-level and the T-level for the 250 pulses/sec stimulus). On this graph, 0 means that the EAP or EABR threshold was recorded at a level equal to the T-level; 1 means that the EAP or EABR threshold was recorded at a level equal to C-level. The normalized data suggest that the mean EAP and EABR thresholds fall closer to C-level than to T-level, and the EAP thresholds are slightly higher than EABR thresholds for elec-

trodes 5 and 10 and slightly lower for electrode 20. These values range from approximately 80 to 100 percent of the subject's dynamic range, but a large amount of variability, as suggested by the standard deviation bars and previously suggested in Figure 2, is demonstrated.

## DISCUSSION

Our findings have indicated that the EAP and EABR thresholds obtained with the CI24R device are strongly correlated, which is consistent with previously published data obtained for CI24M patients (Brown et al, 2000). Previous results using the CI24M device showed that, on average, EABR thresholds were significantly lower than EAP thresholds. In the present study, however, when the CI24R device was used no significant difference was found between mean EAP and EABR thresholds (see Fig. 3). Additionally, the mean EAP and EABR behavioral T- and C-levels recorded in this study with the CI24R users are very similar to those recorded previously for CI24M users. If closer modiolar location results in physiologic and behavioral responses that are more place specific, we would have expected both the EAP and EABR thresholds to be decreased compared with those observed with the CI24M device. Although the sample size in this study was small, little difference between EAP and EABR thresholds recorded with the CI24R and those recorded with the CI24M device was observed. It is possible that this trend may have become apparent with larger sample sizes. The lack of a difference observed in this study suggests that the difference would be small at best.

Even though no significant differences between the EAP thresholds were indicated for the CI24R and CI24M devices, the manner in which the CI24R EAP responses were obtained was altered from that used to obtain responses with the CI24M device. EAP responses with the CI24M in the Brown and colleagues' (2000) study were frequently obtained using the initial default setting provided by the software. In the present study, however, when recording parameters were kept at initial settings, typically the responses still had significant levels of artifact contamination. Consequently, a number of manipulations to the sampling delay and recording electrode were required to obtain an optimal response. The data shown in Table 1 illustrate the range of different recording parameters needed for individual subjects and electrodes when obtaining EAP growth functions with the patients who

wear the Nucleus CI24R device. We followed a procedure for optimizing the EAP outlined in Abbas and colleagues (1999) and found that it was necessary to change the recording parameters from default settings (excluding the delay) for 71 percent of the 38 electrodes evaluated. Alternatively, results from Abbas and colleagues (1999) using the CI24M indicated that in 92 percent of cases (excluding the delay). EAP responses were successfully recorded using the default parameters outlined in their study. Results from Table 1 indicate that for the Nucleus CI24R, a different set of default parameters might be required. For example, using a default setting of 70 or 80  $\mu$ sec for the sampling delay may be more appropriate than the currently used default setting of 50  $\mu$ sec. Additionally, for the more apically placed electrodes, perhaps recording electrodes should be set at three electrodes away from the stimulating electrode in a basal direction rather than two away in an apical direction, as is currently the software default. In part, the differences between the recording parameters needed to obtain an optimal response for the CI24M and CI24R devices may be attributable to the fact that with the CI24R, the distance varies logarithmically along the array. The distance between electrodes is smaller for the CI24R device than the distance between electrodes of the CI24M device, which are equally spaced across the electrode array.

Two different methods for optimizing EAP recordings have been suggested. The protocol suggested by Abbas and colleagues, (1999) involving the systematic manipulation of the sampling delay, recording electrode, probe level, and amplifier gain was followed in this study. Cochlear Corporation's protocol, outlined by Lai (1999), suggests manipulating the sampling delay, amplifier gain, or stimulation rate. For this protocol, moving the recording electrode more than two away from the stimulating electrode was not specifically mentioned. Results from the present study suggest that moving only two electrodes away will typically not result in optimal response amplitudes.

Generally, findings from this study, compared with previously published results, suggest that one of the hoped for advantages of the modiolus-hugging internal array, the reduction of behavioral and electrically evoked auditory thresholds, has not been realized. Contrary to the work of Kawano and colleagues (1998) and Shepherd and colleagues (1993) indicating that behavioral and electrically evoked thresholds are reduced as the electrode array is positioned closer to the modiolus, findings from this study

do not suggest that electrically evoked thresholds or behavioral thresholds are significantly reduced relative to the respective threshold recorded from patients who use the straight array. Luk (1999), using the CI24M, revealed that T- and C-levels were 152 and 192 programming units, respectively, which are very similar to the overall mean T- and C-levels observed in this study (mean T-level 156 units, mean C-level 193 units). Additionally, electrically evoked physiologic thresholds recorded for the CI24R users in this study are not significantly different from those revealed by Brown and colleagues (2000). The expected decrease in behavioral and electrically evoked thresholds for the CI24R device may not have been realized owing to sampling error. Possibly, with a larger subject pool, reduced thresholds may have been present. In addition, it is possible that the CI24R array was not tightly hugging the modiolus in the small set of subjects used in this study. Follow-up radiologic studies would be needed to determine if array placement within the cochlea is significantly different between the CI24R and CI24M users.

Another consideration for the lack of reduced electrically evoked thresholds revealed in this study compared with previously reported data could be the result of differences in methodologies between studies. As previously mentioned, work done by Kawano and colleagues (1998) was conducted using temporal bone donations, and Shepherd and colleagues (1993) used cats, the majority of which had been deafened less than 6 months prior to implantation. It is possible that results obtained using live human data could be dissimilar to either the results obtained with temporal bone donations or from recently deafened animals. In fact, data from Shepherd and colleagues (1993) indicated that EABR thresholds were not reduced when a modiolus-hugging array was used with a cat deafened for 3 years. Perhaps the length of deafness prior to implantation may play a role in determining the effectiveness of a curved electrode array for individual cochlear implant recipients. With greater durations of deafness, auditory neural stimulation is vastly reduced, encouraging neural degeneration. Without stimulation, neural elements extending to the basilar membrane from the modiolus degenerate over time (Spoendlin, 1984; Fayad et al, 1991; Xu et al, 1993b), which could hinder the effectiveness of a precurved array designed to reduce behavioral and electrically evoked thresholds. In these cases, close proximity to the modiolus

may be irrelevant as there could potentially be a limited number of dendrites present for stimulation. Possibly, then, degeneration of neural fibers would prevent noticeable differences of threshold levels between the modiolus-oriented and straight electrode arrays. Evidence for this position may be found in the work of Xu and colleagues (1993a). These researchers, using an electrode array that was both full-banded and half-banded, revealed that no differences were noted in EABR thresholds between current delivered through the full-banded array and the half-banded modiolus-oriented array. The animals used in this study were cats deafened for 2½ years.

In the present study, the length of deafness prior to implantation could have been a factor contributing to the lack of reduced electrically evoked thresholds. All subjects, with the exception of two, had been severely deafened for 3 or more years before receiving the CI24R implant. The two alternate subjects had been deafened for 8 and 12 months, respectively. Perhaps the subjects used in this study, owing to their duration of deafness, had experienced sufficient neural atrophy such that the position of the internal array did not significantly change their electrophysiologic or behavioral thresholds relative to a group of CI24M users.

In summary, data from this study suggest that EAP and EABR thresholds for the CI24R device are highly correlated and the mean threshold values are not significantly different. This finding is in contrast to previously published data obtained using the CI24M device suggesting that EABR thresholds are lower than EAP thresholds. Additionally, comparison with previously published data suggests that significant reductions in EAP and EABR thresholds for the CI24R precurved modiolus-hugging array were not realized. Data from this study suggest that recording EAPs in CI24R cochlear implant recipients may require a different set of defaults.

**Acknowledgment.** This work was supported in part by NIH/NIDCD Grant P50 DC00242, NIH/NCRR Grant RR00059, and the Iowa Sight and Hearing Foundation. We would also like to thank Michelle Hughes for her assistance with figure preparation and Greg Flamme for statistical analyses support. In addition, the comments of three anonymous reviewers on an earlier version of this document were very much appreciated. Portions of this article were previously presented at the 2002 Mid-Winter Meeting of the Association for Research in Otolaryngology, St. Petersburg Beach, FL.

## REFERENCES

- Abbas PJ, Brown CJ. (1991). Electrically evoked auditory brainstem response: growth of response with current level. *Hear Res* 51:123–138.
- Abbas PJ, Brown CJ, Shalloo JK, et al. (1999). Summary of results using the Nucleus CI24M implant to record the electrically evoked compound action potential. *Ear Hear* 20:45–59.
- Brown CJ, Abbas PJ. (1990). Electrically evoked whole-nerve action potentials: parametric data from the cat. *J Acoust Soc Am* 88:2205–2210.
- Brown CJ, Abbas PJ, Fryauf-Bertschy H, et al. (1994). Intraoperative and postoperative electrically evoked auditory brain stem responses in Nucleus cochlear implant users: implications for the fitting process. *Ear Hear* 15:168–176.
- Brown CJ, Abbas PJ, Borland J, Bertschy MR. (1996). Electrically evoked whole nerve action potentials in Ineraid cochlear implant users: responses to different stimulating electrode configurations and comparison to psychophysical responses. *J Speech Hear Res* 39:453–467.
- Brown CJ, Abbas PJ, Gantz BJ. (1998). Preliminary experience with neural response telemetry in the Nucleus CI24M cochlear implant. *Am J Otol* 19:320–327.
- Brown CJ, Hughes ML, Lopez SM, Abbas PJ. (1999). Relationship between EABR thresholds and levels used to program the Clarion speech processor. *Ann Otol Rhinol Laryngol* 108:50–57.
- Brown CJ, Hughes ML, Luk B, et al. (2000). The relationship between EAP and EABR thresholds and levels used to program the Nucleus 24 speech processor: data from adults. *Ear Hear* 21:151–163.
- Fayad JN, Linthicum FH, Otto SR, et al. (1991). Cochlear implants: histopathologic findings related to performance in 16 human temporal bones. *Ann Otol Rhinol Laryngol* 100:807–11.
- Fayad JN, Luxford W, Linthicum FH. (2000). The Clarion electrode positioner: temporal bone studies. *Am J Otol* 21:226–229.
- Firszt JB, Rotz LA, Chambers RD, Novak MA. (1999). Electrically evoked potentials recorded in adult and pediatric Clarion implant users. *Ann Otol Rhinol Laryngol* 108:58–63.
- Gantz BJ, Brown CJ, Abbas PJ. (1994). Intraoperative measures of electrically evoked auditory nerve compound action potential. *Am J Otol* 15:137–144.
- Gstoettner WK, Adunkd O, Franz P, et al. (2001). Perimodiolar electrodes in cochlear implant surgery. *Acta Otolaryngol* 121:216–219.
- Hodges AV, Ruth RA, Lambert PR, Balkany TJ. (1994). Electric auditory brain-stem responses in Nucleus multichannel cochlear implant users. *Arch Otolaryngol Head Neck Surg* 120:1093–1099.
- Hughes ML, Brown CJ, Abbas PJ, et al. (2000). Comparison of EAP thresholds with MAP levels in the Nucleus 24 cochlear implant: data from children. *Ear Hear* 21:164–174.

- Kawano A, Seldon HL, Clark GM, et al. (1998). Intracochlear factors contributing to psychophysical percepts following cochlear implantation. *Acta Otolaryngol (Stockh)* 118:313–326.
- Lai W. (1999). *An NRT Cookbook*. Basel, Switzerland: Cochlear AG.
- Luk B. (1999). *Within Subject Comparison between Electrically Evoked Auditory Brainstem Response Thresholds and Electrically Evoked Compound Action Potential Thresholds in Nucleus CI24M Cochlear Implant Users*. University of Iowa, Iowa City, IA.
- Mason SM, Sheppard S, Garnham CW, et al. (1993). Application of intraoperative recordings of electrically evoked ABRs in a paediatric cochlear implant programme. *Advan Otorhinolaryngol* 48:136–141.
- Richter B, Aschendorff A, Lohnstein P, et al. (2001). The Nucleus Contour Electrode Array: a radiological and histological study. *Laryngoscope* 111:508–514.
- Roland JT Jr, Fishman AJ, Alexiades G, Cohen N. (2000). Electrode to modiolus proximity: a fluoroscopic and histologic analysis. *Am J Otol* 21:218–225.
- Shallop JK, Beiter AL, Goin DW, Mischke RE. (1990). Electrically evoked auditory brain stem responses (EABR) and middle latency responses (EMLR) obtained from patients with the Nucleus Multichannel Cochlear Implant. *Ear Hear* 11:5–15.
- Shepherd RK, Hatsushika S, Clark GM. (1993). Electrical stimulation of the auditory nerve: the effect of electrode position on neural excitation. *Hear Res* 66:108–120.
- Skinner MW, Holden LK, Holden TA, Demorest ME. (2000). Effect of stimulation rate on cochlear implant recipients' thresholds and maximum acceptable loudness levels. *J Am Acad Audiol* 11:203–213.
- Spoendlin H. (1984). Factors inducing retrograde degeneration of the cochlear nerve. *Ann Otol Rhinol Laryngol* 112(Suppl):76–82.
- Wilson BS, Lawson M, Zerbi M, et al. (1995). New processing strategies in cochlear implantation. *Am J Otolaryngol* 16:669–675.
- Xu S, McAnally KI, Xu J, Clark GM. (1993a). Comparison of half-band and full-band electrodes for intracochlear electrical stimulation. *Ann Otol Rhinol Laryngol* 102:363–367.
- Xu SA, Shepherd RK, Chen Y, Clark GM. (1993b). Profound hearing loss in the cat following the single co-administration of kanamycin and ethacrynic acid. *Hear Res* 70:205–215.