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COCHLEAR PROSTHETICS

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THE STATE OF THE ART

Cochlear prostheses are being used clinically to restore functional hearing in patients suffering from profound sensorineural deafness. The devices include one or more electrodes implanted in or near the cochlea to provide electrical stimulation of the remaining auditory nerve fibers, thereby bypassing the defective sensory hair-cells (Figure 1). There are many different designs in various stages of development, testing, and availability, with widely differing therapeutic benefit both among devices and among individual patients receiving a given device.

This review focuses on those aspects of the current development and evaluation processes that involve researchers from the fundamental neuroscience community. In the past, such a review might have focused on issues of tissue damage and stability of evoked percepts; while still important, these problems have been largely resolved by improved materials and designs, functional testing in animals (Walsh & Leake-Jones 1982, Maslan & Miller 1987), mechanical testing in cadaver temporal bones (Kennedy 1987, Webb et al 1988), and clinical experience (Yin & Segerson 1986, Waltzman et al 1986, Clark et al 1988, Terr et al 1988). Current research is geared toward integrating the clinical results from these implants with theories of auditory perception, in the hope of developing efficient approaches to improving therapeutic benefit.

In addition to the specific literature citations and the list of recent book-length treatments of the field provided here (Table 1), the opinions and evaluations expressed here are based on a survey of 120 leading researchers active in this field, 54 of whom provided detailed responses to a seven page questionnaire that was circulated in January, 1989 (De Foa and Loeb, in preparation).

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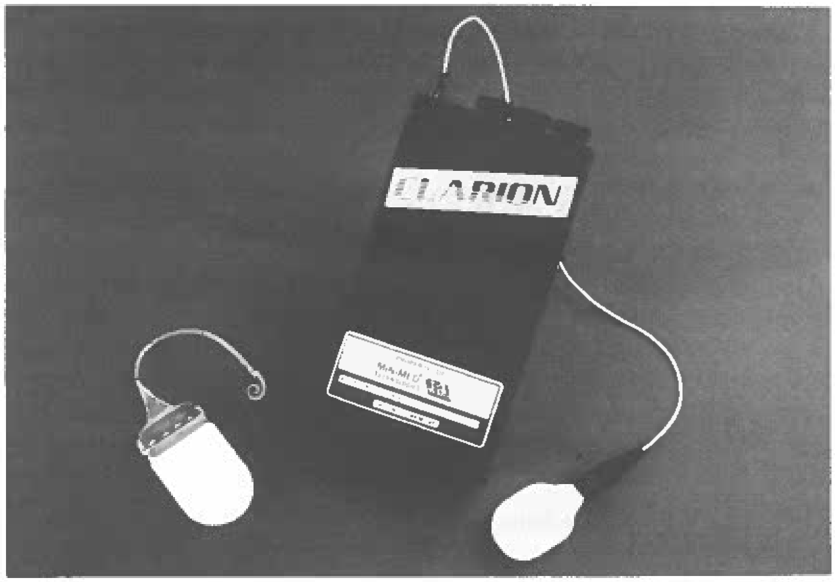

Figure 1 Clarion™ cochlear prosthesis produced by Minimed Corp. of Sylmar, California, showing spirally molded electrode and ceramic case (*lower left*) containing custom IC chip and related circuitry for driving the 16-contact electrode array with signals specified by the external speech processor unit (*center*) and transmitted along with power by an inductive coil (*bottom right*). (Photograph courtesy of Minimed Technologies.)

Table 1 Recent review volumes

Cochlear Prostheses, ed. C. W. Parkins, S. W. Anderson, *Annals of the New York Academy of Science*, Vol. 405, 1983
Cochlear Implants, ed. R. A. Schindler, M. M. Merzenich, New York: Raven Press, 1985
British Journal of Audiology, Vol. 20, number 1, February 1986
Otolaryngology Clinics of North America, Vol. 19, number 2, May, 1986
Cochlear Implant: Current Situation, ed. P. Banfai, Erkelenz, GDR: Bermann GmbH, 1988
Cochlear Implants in Young Deaf Children, ed. E. Owens, D. K. Kessler, Boston: College Hill Press, 1989
Models of the Electrically Stimulated Cochlea, ed. J. M. Miller, F. A. Spelman, New York: Springer-Verlag, 1989

Socio-Economic Factors

There are at least two million functionally deaf individuals in the USA alone (defined as being unable to understand speech or most ambient sounds even with the use of an acoustic-amplification hearing aid; DiPietro 1984). Most mechanical problems with middle ear conduction of sound are now treatable surgically; most of the remaining deafness represents sensorineural pathophysiology. Although there are many proven non-aural techniques for restoring functional communication (e.g. speech-reading and sign language), a large proportion of these individuals are excluded, for various reasons, from most of the social and economic interactions that virtually define human society. Yet many of these patients have lesions confined primarily to the cochlear hair cells, with largely intact auditory nerves and central pathways. In such patients, direct electrical stimulation of the remaining auditory nerve fibers can restore this communication channel without interfering with other sensory and motor activities, in contradistinction to visual and vibrotactile displays of acoustic information, which have had very limited clinical acceptance (Rose et al 1988, Skinner et al 1988, Thornton 1988).

For the purposes of this discussion, it is useful to define three broad categories of deaf individuals:

1. POST-LINGUALLY DEAFENED ADULTS These have been the most successful and enthusiastic users of cochlear prostheses. Their communicative skills and social ties are strongly based on hearing and their acquired deafness is often a severe psychological blow. Even so, there is a wide range of awareness and attitudes toward prostheses among these patients and their therapists.

2. PRE-LINGUALLY DEAFENED ADULTS These have been the least successful or enthusiastic users because they have acclimated to life without hearing and because their mature nervous systems seem to lack the ability to learn to deal with auditory sensations.

3. PRE-LINGUALLY DEAFENED, YOUNG CHILDREN Only a limited number of implants have been performed in this group, which poses a host of ethical, technical, medical, and scientific quandries. In addition to the problems and potential of stimulating and evaluating the undeveloped auditory system (discussed below), there is much controversy on whether and how to integrate this new technology with the several different, fiercely competitive approaches to developing language skills in deaf children.

Technology

Table 2 provides an overview of the myriad devices that have evolved (and often become extinct) during 30 years of active research and development.

Table 2 Cochlear prostheses

Device Name	P.I./Institute/City	Manufacturer/City	First Desc.	Est. # Implants	Approx. Cost US $	Availability [a]
-	Djourno & Eyries/Paris	-	1957	1	-	never
-	Doyle/Univ. Southern California/Los Angeles	-	1964	1	-	never
-	Simmons/Stanford/Calif.	-	1966	4	-	superceded
Bioear	Simmons, White/Stanford	Biostim/Brooksville, FL	1984	7	4,500	defunct
UCSF	Michelson/Univ. California/San Francisco	-	1971	5	-	superceded
UCSF-Storz	Merzenich, Schindler/UCSF	Storz/St. Louis, MO	1984	18	12,000	withdrawn
Clarion	Schindler, Merzenich/UCSF + Wilson/Research Triangle/ Durham, NC	Minimed/Sylmar, CA	1988	0	13,000	R&D
3M-House	House/House Hearing Inst./Los Angeles	3M Corp./St.Paul, MN	1973	3000	6,500	withdrawn
Chorimac	Chouard/CHU Saint-Antoine/Paris	Bertin/Paris	1973	150	9,000	superseded
Monomac	Chouard/CHU Saint-Antoine/Paris	Bertin/Paris	1987	?	?	local
Minimac	Chouard/CHU Saint-Antoine/Paris	Bertin/Paris	1988	?	?	local
EPI	Douek, Fourcin/ Guy's Hosp./London	-	1978	9	?	local
UCH-RNID	Fraser/Univ. College Hosp./London	Fine Tech./London	1988	45	1,100	local
Vienna	Hochmair/Innsbruck	-	1978	70	?	local
3M-Vienna	Burian/Hochmair/Tech. Univ./Vienna	3M Corp./ St. Paul, MN	1978	80	9,500	withdrawn
Implex	Banfai, Hortmann/ Cologne-Duren Group	Hortmann, GmbH/ Cologne	1978	100	6,000	Europe superseded
Exco-16	Banfai, Hortmann/ Cologne-Duren Group	Hortmann, GmbH/ Cologne	1985	60	?	Europe
Ineraid	Eddington/Univ. Utah/ Salt Lake City	Symbion/ Salt Lake City, UT	1980	100+	11,000	IDE
Nucleus	Clark/Univ. Melbourne	Cochlear-Nucleus/ Sidney, Aust	1980	1500	12,200	PMA
-	Dillier, Spillman/ Universitatspital/Zurich	-	1982	10	?	local
MSR-UCL	Geradorff, Sneppe/Catholic Univ.Leuven/Brussels	Siemens	1985	9	?	local
Laura	Marquet Peeters/Univ. Instelling Antwerp	Forelec/Antwerp	1988	2	?	R&D
Prelco	Cazals/INSERM/Bordeaux	Racia/Bordeaux	1985	23	2,000	local
Medtronic	Fraysse/Toulouse	-	-	17	?	local
-	Gerhardt/Humboldt Univ./Berlin	-	1987	10	?	local
Omina	Bosch/Hosp.de la Cruz Roja/ Barcelona	-	-	12	?	local
ECME	Bochanek/Acad. Med./Warsaw	-	-	3	?	local
-	Valvoda, Tichy/Acad Science/Prague	-	1988	3	?	local

Compiled from ASHA (1986), Chouard et al (1988), market survey by Cochlear A.G. (1988) and references cited.

a U.S. Food and Drug Administration designations: IDE=investigational implants only at approved centers; PMA=approved for marketing to licensed practitioners

b Number of electrically separate, active contacts, excluding reference or ground.

c M=directly in auditory nerve in modiolus; X= extracochlear

d Number of channels commonly activated in parallel; Nucleus device sweeps 2 output channels among 21 bipolar sites.

Electrode Contacts [b]	Electrode Depth mm [c]	Stimulus Channels [d]	Stimulus Waveform [e]	Coupling [f]	Reference
1	M	1	pulse	perc.	Djourno & Eyries (1957)
1	?	1	AM	trans.	Doyle et al (1964)
8	M	8	pulse	trans.	Simmons (1983)
1	X-3	1	analog or pulse	trans.	Dent et al (1988)
2	15	1	analog	trans.	Michelson (1971)
16	24	4	analog	trans.	Schindler and Kessler (1987)
16	24	8	analog	trans.	Wilson et al (1988)
1	6	1	AM or pulse	trans.	Fretz and Fravel (1985)
8, 12	0-3[g]	8, 12	analog	trans.	Chouard (1978)
1	X-10	1	pulse	trans.	Chouard et al (1988)
15	?[h]	15	pulse	trans.	Chouard et al (1988)
1	X	1	pulse	ext.	Walliker et al (1985)
1	X	1	analog	ext.	Conway (1988)
8	14	1	analog	trans.	Burian et al (1986)
1	X	1	analog	trans.	Burian et al (1986)
1,4,8	X	1,4,8	analog	perc./trans.	Banfai et al (1979)
16	X	16	pulse	perc./trans.	Banfai et al (1986)
6	22	4	analog	perc.	Parkin and Stewart (1988)
22	25	2/21	pulse	trans.	Franz et al (1987)
1	X	1	pulse	trans.	Spillman et al (1982)
1,2	X-5	1,2	analog	trans.	Gersdorff et al (1988)
16	18	8, 16	analog or pulse	trans.	Marquet (1986)
1	X	1	analog	trans.	Negrevergne et al (1988)
1	X-2	1	pulse	trans.	-
2	X	2	analog	trans.	Gerhardt and Wagner (1987)
1,6,8	?	1,6,8	?	?	-
1	X	1	?	?	-
1	X	1	analog	trans.	Valvoda et al (1988)

e Analog=band-pass filtered acoustic signal; Pulse=brief stimuli, usually at F_0 rate, amplitude-modulated by acoustic envelope; AM=unrectified carrier modulated by acoustic signal

f perc.=percutaneous plug; trans.=transcutaneous inductive coupling; ext.=exteriorized, removable appliance.

g Electrodes inserted through individual fenestrations in the lateral wall of the cochlea.

h In Scala tympani via round window.

The landscape has changed significantly within the past six to eight years, with commercially designed and manufactured devices supplanting the "home-built" efforts of university shops but still undergoing attrition among themselves. For the moment, the commercial field is dominated by the multichannel device manufactured by Nucleus Corp. from Australia, which is the only device that is now approved by the Food and Drug Administration and marketed in the United States (as of July 1989).

The various designs are distinguished by the set of decisions that they embody regarding the following options:

SINGLE VS. MULTICHANNEL ELECTRODE Speech perception requires a signal with at least 3 kHz bandwidth, whereas the ability of neurons to follow each cycle of acoustic or electrical stimulation is limited to about 300–500 pps. (Curiously, the phase-locking that appears to contribute to acoustic pitch discrimination in the 500–5000 Hz band has no clear influence on electrically evoked percepts; Loeb et al 1983b). The intact auditory system solves this by spatial filtering and parallel processing along the basilar membrane. Multichannel prostheses attempt to replicate this function as well as simple transduction by having multiple, independently addressable electrode-contacts located near different subpopulations of auditory nerve fibers. In general, the percept elicited by each has a noisy timbre but a distinct pitch related to the tonotopic map of the auditory nerve fibers in the cochlea.

INTRACOCHLEAR VS. EXTRACOCHLEAR ELECTRODE Both single and multichannel devices have had their electrode contacts positioned outside of the cochlea (usually on the round window and bone overlying the cochlear turns; "X" in Table 2), within the cochlea (usually on a flexible, slender probe inserted along the scala tympani), and in the auditory nerve itself as it passes through the modiolar bone ("M" in Table 2). More recently, feasibility studies have been conducted on placing electrodes on and in the cochlear nucleus, thereby bypassing the auditory nerve entirely (McElveen et al 1985, 1987).

PERCUTANEOUS VS. TRANSCUTANEOUS COUPLING The acoustic signals are picked up and processed by a wearable unit similar to a hearing aid, which formats the stimulus waveforms and transmits them to the electrode. This can be done via a percutaneous connector affixed to the skull and passing through the scalp or via transcutaneous inductive-coupling of radio-frequency signals. For multichannel systems, such RF signals often include sophisticated digital encoding schemes and power transmission to drive custom integrated circuitry in a hermetically encapsulated implant (e.g. see Figure 1).

SPEECH PROCESSING Three general approaches have emerged for transforming the acoustic signal into stimulus waveforms. The simplest is to use the acoustic waveform itself (so-called "analog" stimulation), suitably band-filtered and compressed in its dynamic range to conform to the sensitivity of neurons to electrical stimulation (White 1986). Multichannel versions of such an approach employ a bank of filters, driving each electrode in parallel according to the output of its corresponding bandpass filter. The rationale is that the nervous system may be able to make some use of the information contained in the raw acoustic waveforms, although the biophysical events leading to spike initiation suggest that the resultant fine temporospatial patterning is highly unphysiological (Kiang et al 1979). More recently, improved channel isolation and overall speech performance have been achieved by converting the envelope of the signal from each filter into a set of narrow pulses that can be delivered in a basally to apically sequenced, non-overlapping pattern to each electrode at a repetition rate that is determined by the voiced-fundamental pitch (Wilson et al 1988a,b). These approaches, based on the frequency-channel vocoder, contrast with the formant-extraction method (e.g. Nucleus) in which an on-line microprocessor tracks the spectral location and relative amplitude of one or two vowel formants and selects one or two electrode contacts for stimulation based on a previously stored map of the pitch sensations that are elicited at each available site (Franz et al 1987).

ISSUES IN NEUROSCIENCE

Biophysics of Auditory Nerve Stimulation

Obviously, a larger number of functionally separate stimulation channels permits a more faithful representation of the speech signal in the evoked neural activity and, generally, provides better speech comprehension. The problem is in defining and obtaining "functional separation." In the typical scala tympani approach, the electrode contacts lie 0.5–1.0 mm from the spiral ganglion cells that are embedded in the medial wall of the scala. Stimulus current injected through one such contact tends to spread diffusely in the volume-conductive fluids and tissues that surround it. At threshold, the first auditory nerve fibers to be recruited will be those closest to the contact, but any attempt to provide a useful dynamic range of stimulus intensities produces rapid spread of activation to nearby neurons in the same and even adjacent turns of the cochlear spiral. These more distant neurons are intended to be under the separate, perhaps simultaneous control of other stimulating electrodes.

Theoretical considerations (Finley et al 1989) and empirical studies (Merzenich & White 1977) both indicate that the best spatial selectivity can be achieved by using bipolar pairs of contacts that are oriented radially (perpendicularly) to the axis of the cochlear spiral. Such a configuration produces a potential gradient that is oriented parallel to the long axis of the overlying spiral ganglion cells, whose apical dendrites (formerly innervating hair cells in the organ of Corti) and central axons are similarly radially oriented. Under ideal conditions, with the contacts oriented optimally against the medial wall of the canal, such an electrode provides a space-constant of about 0.87 mm (10 dB attenuation per millimeter distance away) for stimulus spread both apically and basally from the bipolar pair. This space constant would be compatible with the use of eight such pairs arranged at 2 mm intervals in the region 10–24 mm from the round window where the critical speech frequencies are normally transduced (Vivion et al 1981). In contrast monopolar electrodes have a space constant of 13.0 mm (0.67 dB/mm) and longitudinally oriented bipolar pairs (Black & Clark 1980) have a space constant of 2–4 mm (4.3–2.2 dB/mm); extracochlear contacts are even more severely compromised in their selectivity. Attempts to focus the electrical fields by introducing antiphasic waveforms on adjacent electrodes have produced relatively little improvement (Ifukube & White 1987), presumably because the large anodal currents that are required result in virtual cathodes and new sites of spike initiation at more distant nodes of Ranvier along the auditory nerve fibers (Ranck 1975).

Considerable circumstantial evidence suggests that much of the variability in the functional benefit realized by individual patients using a multichannel implant is related to the histopathology of the spiral ganglion cells, which is highly variable among patients with the same nominal etiology of their deafness and often heterogeneous in different regions of the same cochlea (Hinojosa et al 1987). Patients with poorer clinical results generally have higher current thresholds to achieve any auditory sensation, narrower dynamic ranges before reaching maximal loudness, and psychophysical test results suggestive of overlapping recruitment when adjacent channels are activated simultaneously (Shannon 1983, White 1984). Detailed modeling work using finite-element analysis is under way to determine the optimal electrode configuration(s) for different regions and conditions of the cochlea (Finley et al 1989).

Unfortunately, most of the design features that make an electrode array biophysically ideal also make it extremely tedious to fabricate and difficult to insert atraumatically. To get the electrode contacts to lie on the medial wall of the canal, either the electrode must fit snugly into the scala tympani or it must spring spontaneously into a tightly curved spiral. To get the electrode into the scala tympani without rupturing the overlying basilar membrane (which sequesters potassium-rich endolymph that is toxic to neurons), the surgeon must slide it in without pushing, while working through the long, narrow access afforded by the external auditory canal. To date, multichannel prostheses have opted either for electrodes that are known to be far from optimal (e.g. monopolar—Chorimac, Ineraid, Exco-16; longitudinal bipolar—Nucleus) or have paid the price of very slowly progressing research and development programs (e.g. near-radial bipolar—University of California at San Francisco; Loeb et al 1983a).

Now that reasonable numbers of patients have been implanted with various electrode designs, it should be possible to determine which electrode design features are most important for psychophysical function. This knowledge, it is hoped, will lead to design compromises that are based on informed trade-offs rather than historical accident.

Prognostic Testing

The above-noted patient variability in functional results now represents the single largest hindrance to the widespread clinical application and commercial success of cochlear prosthetics. The problem is that the number of post-lingually deafened adults with profound deafness (no detectable acoustic threshold) is quite small, whereas a much larger number have some residual hearing but are unable to make effective use of a well-fitted acoustic hearing aid. Implantation of a multichannel prosthesis generally results in the loss of any residual acoustic hearing, and therefore is not justifiable unless its function in that patient can be guaranteed to be better than what will be lost. At present, the results from such a prosthesis can vary from sufficient speech recognition for conversation over the telephone to only a general awareness of the rhythm of ambient sounds (Gantz et al 1988).

Obviously, a preoperative evaluation of the condition of the spiral ganglion cells and CNS auditory pathways is highly to be desired, particularly if, as expected, it can be shown to correlate with prosthetic function. Unfortunately, the few attempts to develop and validate such tests to date have encountered technological problems and paradoxical results.

AUDITORY SENSATIONS FROM PROMONTARY STIMULATION It is a relatively simple office procedure to pass a needle electrode through the tympanic membrane so that its tip rests against the boney promontory next to the round window. Electrical stimulation in deaf subjects usually produces auditory sensations, confirming the presence of a functional auditory nerve (Chabolle et al 1988). Paradoxically, such stimulation in subjects with normal hearing produces no auditory sensations (Eddington et al 1978, Liard et al 1988). In the absence of a plausible mechanism for this phenomenon, there seems to be little enthusiasm for developing detailed psychophysical tests based on this procedure.

ELECTRICALLY EVOKED AUDITORY BRAINSTEM RESPONSES (EABR) The highly organized, tonotopic projections throughout the auditory CNS result in coherent, remote field potentials in response to both acoustic and electrical stimulation. Because of their temporal coherence, even the very small signals recordable noninvasively as scalp potentials can be revealed by stimulus-triggered signal averaging (Chouard et al 1979, Dobie & Kimm 1980, Starr & Brackmann 1981, Waring et al 1985, Kileny & Kemink 1987). This has been used to evaluate prosthesis function in children who are too young to provide psychophysical data (Miyamoto & Brown 1987). The rate of growth of such potentials with increasing stimulus strength has been used to evaluate the spatial selectivity of intracochlear electrodes (Gardi 1985) and has been suggested as a potentially useful way to interpret promontory stimulation in terms of auditory nerve survival (Simmons & Smith 1983). Unfortunately, the earliest waves of the EABR are the most likely to correlate with nerve survival (particularly in view of the perceptual paradox noted above), and these are easily obscured by electrical artifact from the stimulation. Special techniques are needed to alternate stimulus phases to obtain cancellation of the artefact (Gardi 1985, Banfai et al 1986) and to protect high-gain amplifiers from saturation; these have not proven easy to integrate into the normal repertoire of audiometric procedures. In theory, it should be possible to obtain fairly detailed representations of the integrity of the auditory pathways by constructing three-dimensional vectors of the EABR (based on orthogonal lead placements) and using current-source-density analysis to relate this to the tonotopic representation in various projections (Gardi 1985). If these could be shown to relate well to the function of cochlear prostheses, they would easily justify substantial developmental costs (Stypulkowski et al 1986).

SCANNING TECHNIQUES Computed X-ray tomography is now routinely used to screen intracochlear prosthesis candidates for patency of the scala tympani. Resolution is now marginally adequate and likely to continue to improve. However, there have been and will probably continue to be reports of both unexpected obstructions to electrode insertion and successful insertions obtained by drilling through clearly ossified barriers (Balkany et al 1988, Gantz et al 1988). Atrophy of the cochlear nucleus has been noted in experimentally deafened animals (Webster & Webster 1979); it might be detectable with high-resolution magnetic resonance imaging (MRI). Magnetoencephalogram (MEG) scanning (Hari et al 1988) is also a distant but promising way to get around technical problems with the EABR.

ETIOLOGICAL FACTOR ANALYSIS Attempts to correlate prosthetic outcome with patient-history items such as cause and duration of deafness, age and rate of onset, audiometric pattern, etc have been discouraging (Chouard et al 1987, Fritze & Eisenwort 1988, Geers & Moog 1988). Many deaf patients have no defined etiology and many probably had gradual hearing loss that went undetected for many years. Furthermore, even when these factors are well known, there seems to be little correlation with the histopathology of spiral ganglion cells (Hinojosa et al 1987).

Development and Plasticity

The application of cochlear prostheses to prelingually deaf children is undoubtedly the greatest challenge and research opportunity for neuroscientists in this field (reviewed by Loeb 1989). Two hypotheses underlie the urgency and the promise:

CRITICAL PERIOD HYPOTHESIS The auditory and speech centers in the CNS (like the visual system) have critical periods in their natural development, at which times a sustained absence of organized sensory input will lead to a permanent and irreversible deficit in their ability to process information.

PLASTICITY HYPOTHESIS During these critical periods, the plasticity of the nervous system permits it to develop information processing strategies that are, in some way, optimized to extract the information that is embedded in complexly encoded and often noisy sensory channels.

If valid, these hypotheses suggest that cochlear prostheses should be implanted in deaf children at the earliest possible age and that such children will learn to make much better use of the distorted input thus provided than would an adult with or without previous hearing. Unfortunately, evidence supporting these hypotheses is still limited and difficult to obtain. Evidence from animals with experimentally induced auditory deprivation supports the notion of a post-natal critical period and the efficacy of chronic electrical stimulation in preventing disuse atrophy (Wong-Riley et al 1981, Lousteau 1987; although see Balkany et al 1986). Certainly the results of cochlear implants in adults who were deafened prelingually are quite poor (Tong et al 1988). However, in the absence of animal models for speech comprehension and production, questions about functional plasticity will probably have to be resolved through carefully designed psychophysical experiments on children who have received cochlear implants.

Long-term data from such children are just starting to be available but are very difficult to interpret. In addition to the obvious problems of designing and administering appropriate tests, the only two devices used to date in young children are the least likely to benefit from plasticity, at least in the precortical brain. The 3M-House implant provides a single-channel signal consisting of an unrectified 16 kHz carrier that is over-modulated by the acoustic envelope; this permits very simple implanted electronics but provides almost no neural signal beyond prosody. However, even this device seems to work much better in some children than it has in adults (Berliner & Eisenberg 1987). The multi-channel Nucleus system works much better than the 3M-House device in post-lingually deafened adults (Gantz et al 1988), but its formant-extraction approach is intended specifically to evoke percepts related to the speech-recognition strategies of individuals with previously normal hearing. None of the raw spectral information that might support alternative speech-recognition strategies is represented in the output stimuli. Even so, the few reports to date from young children suggest excellent results, with slower initial learning curves than in adults but continued improvement over longer periods of time, particularly regarding the high-level cognitive and speech-production skills that are particularly difficult to teach to deaf children (Luxford et al 1988, Berliner et al 1988).

CONCLUSIONS

In the early days, the clinical pioneers ignored and were usually ignored by the basic research community. It is now widely recognized that the resultant blind empiricism is no longer an effective way to advance this complex art. Cochlear prosthetics provides neuroscientists with an almost unprecedented opportunity to apply basic knowledge to an important and rapidly evolving area of clinical practice, while at the same time offering unique, high-technology tools to conduct psychophysical experiments on human CNS functions that have no animal models. For the individual investigator, the problem is to identify a suitable niche in the complex and shifting social order of commercial manufacturers, clinical trials, and engineering support.

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