How do cochlear prostheses work?

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The past two decades have witnessed a revolution in the treatment of sensorineural hearing loss. Cochlear prostheses have evolved from laboratory experiment to a commercial technology that has benefited over 20,000 people. Paralleling this phenomenal development has been a substantial increase in our understanding of the biophysical, physiological and psychophysical mechanisms underlying the function of these devices.

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Abbreviations
FDA Food and Drug Administration (USA)

Introduction

Cochlear prostheses are systems incorporating electrode arrays that are surgically implanted into the cochlea as a treatment for severe to profound sensorineural hearing loss. Most cases of sensorineural hearing loss involve the preferential loss of hair cells, with relative preservation of the spiral ganglion [1]. These neurons are the target for electrical stimulation, which attempts to replace the function of cochlear hair cells. Figure 1 demonstrates the position of the implanted electrode array in the scala tympani adjacent to the modiolus — the 'core' of the cochlea in which the residual auditory neurons are located. There are multiple manufacturers and designs for these devices, but all operate on similar principles. Figure 2 illustrates simplified schematics of two standard implant designs: analog and pulsatile.

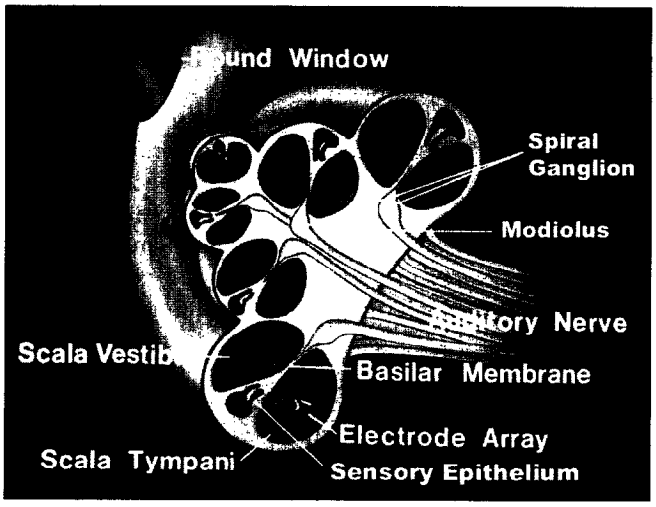
Speech comprehension is the primary determinant of candidacy for implantation and is the main outcome measure. Current FDA criteria for implantation of postlingually deafened adults (i.e. individuals defeaned after aquisition of spoken language) requires that speech reception be limited to less than 40% words in sentences in the 'best-aided condition'. Such patients will typically score less than 10% on monosyllabic word tests. Prelingually deafened adults receive little benefit from these devices and usually are not candidates. Pre- or postlingually deafened children are candidates if they receive no benefit from an adequate hearing aid trial and are over eighteen months of age. In adults, the 'gold standards' of outcome measures are monosyllabic word recognition and sentence recognition in standardized noise. Because of a limited number of word lists, learning effects can be problematic with either of these tests. When frequent and repeated speech reception measures are necessary, such as in development of speech-processing strategies, vowel and consonant testing can be repeated indefinitely without learning and are preferred. Outcome measures in young children are complex and evolving, and they will not be discussed here.

Implantation of cochlear prostheses has progressed from a highly experimental procedure to a standardized clinical intervention performed worldwide with substantial auditory benefit. Over 20,000 people have received these devices and much is known about their expected performance. Wide variability has been documented but average performance has been steadily improving. In the early 1980s, average monosyllabic word recognition was approximately 10%; it is now over 45%. Approximately one-third of this improvement can be attributed to improved speech-processing strategies, one-third to patient factors such as duration of deafness and degree of residual hearing, and one-third to unknown causes [2•].

It is now possible to predict the results of cochlear implantation with reasonable confidence intervals based on duration of deafness and residual speech comprehension. Assuming straightforward surgery and no unusual auditory nerve pathology, the correlation coefficient for the prediction is 0.77 [2•]. The sources of the residual variance are unknown but may include electrode location, pattern of neural survival, surgical trauma, and integrity of the central auditory pathways. Greater understanding of the interface between the cochlear prosthesis and the auditory system is needed before we can completely characterize the technology's limitations. Such understanding promises to further enhance hearing for those currently implanted with these devices, as well as for the millions of individuals who do not meet FDA criteria for the procedure because they have substantial, albeit socially inadequate, speech reception.

In a classic temporal bone study, Otte *et al.* [3] determined that 90% speech discrimination was possible with only one-third the normal complement of auditory neurons. They also demonstrated that 45% of temporal bones from people with profound deafness had at least this many auditory neurons. A 'perfect' cochlear implant technology should therefore provide at least 90% discrimination to at least 45% of profoundly deaf individuals. While such levels of speech reception occasionally occur with current devices, they are rare. Higher levels of performance might be possible in people with less severe hearing loss, potentially offering substantial assistance to those who currently derive only limited benefit from hearing aids. Cochlear

Figure 1



Cut-away of a cochlea showing the electrode array of a cochlear prosthesis in the scala tympani. The array is inserted through the round window and typically extends into the second turn of the cochlea. The sensory epithelium of implantees is generally severely damaged, with few remaining hair cells and afferent terminals. Cell bodies of human spiral ganglion neurons persist long after hair cell loss, although there is typically some loss of neurons as well. Current from the electrodes stimulates the spiral ganglion, although the precise sites of excitation are unknown. Modified from a drawing by Kate Sweeney.

prostheses are the only form of hearing rehabilitation that improves speech intelligibility; basic research into biophysical and psychophysical mechanisms of electrical stimulation has the potential to substantially expand the population assisted by this technology.

As in the study of normal hearing, three fundamentally different experimental paradigms for the analysis of cochlear implant function have evolved: the biophysical/physiological, psychophysical and clinicopathological approaches. These are the focus of this review. The biophysical/physiological paradigm was initially confined to animal work [4,5] but now includes human studies as well [6•,7]. This approach attempts to elucidate neural patterns in response to electrical stimulation. As much is known about patterns of auditory nerve activation in normal hearing, it is reasonable to assume that by understanding the mechanisms underlying electrical stimulation, it will become possible to engineer devices that more closely mimic normal physiology. The psychophysical paradigm initially focused on human studies [8,9] but has also been expanded to experimental animals [10–12]. This approach models the electro-neural interface as a 'black box' and analyzes perceptual measures with varying stimulus configurations. The clinicopathological paradigm attempts to correlate clinical measures of hearing ability with histopathological data [11,13–16]. These three approaches have helped define the spatiotemporal limitations of current cochlear prostheses and direct us to possible means to overcome them.

Biophysical/physiological studies

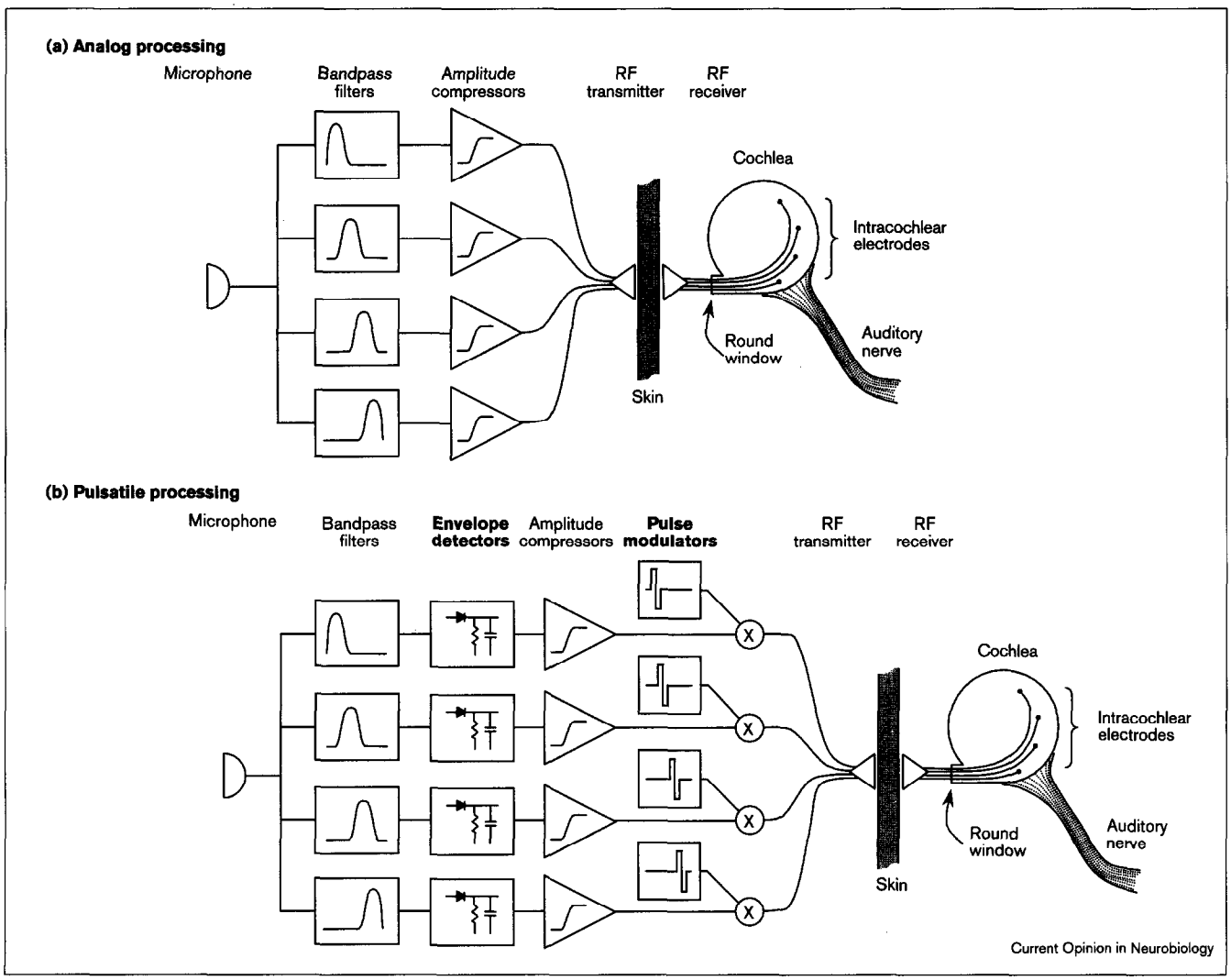
Intracochlear current spread

Because multi-electrode cochlear implants produce much higher levels of speech perception than do single-channel devices [17], there is great interest in minimizing channel interaction. Until recently, it was generally believed that only bipolar electrical stimulation could selectively activate spatially specific regions of the spiral ganglion. This belief is attributable to early single-unit recordings in cats suggesting that monopolar stimulation widely activated the auditory nerve, whereas spatially selective excitation occurred with closely spaced bipolar electrode pairs [18]. Although earlier work demonstrated that pitch-ranking could be performed with monopolar electrodes in humans [8], it was widely concluded that bipolar stimulation would be more spatially and spectrally specific and provide a greater number of independent information channels. For analog stimulation (see Figure 2a), indirect clinical data suggest greater speech reception in the bipolar than in the monopolar mode [19]. In contrast, when pulsatile stimuli are used (see Figure 2b), direct within-subject comparisons have failed to demonstrate any speech reception advantage of bipolar over monopolar [20] or of closely spaced over widely spaced bipolar configurations [21].

Our recent work [22], as well as that of Liang *et al.* [23•], clearly demonstrates at the single-fiber level that place tuning is possible with monopolar stimulation. Both studies employed intracochlear electrode arrays and showed that thresholds and input–output functions of auditory nerve fibers differed at different sites of intracochlear stimulation. This finding explains the clinical data mentioned above, but does not rule out the possibility that future multipolar electrode designs could produce more focused stimulation with resultant higher levels of speech perception. Using both intracochlear potential measurement and single-fiber recordings, Kral *et al.* [24•] demonstrated that multipolar stimulation can produce more restricted excitation fields in cats. They found that bipolar stimulation was more restricted than monopolar, and tripolar more restricted than bipolar. Tripolar thresholds were intermediate between monopolar and bipolar.

Given these findings, it is still unclear why bipolar stimulation does not produce better speech perception than monopolar. One possibility is the failure to produce sharpened psychophysical tuning with closely spaced bipolar stimulation (see discussion below and [25]); another is the complex neural activation patterns that may result from the different polarities of stimulation that occur adjacent to the two electrodes of a dipole. Miller *et al.* [26,27•] and Shepherd and Javel [28•] have demonstrated several different excitation modes, accommodation, latencies, and stochastic temporal properties that can result from reversing the polarity of excitation as occurs near the return electrode of a bipolar pair. By consistently exciting the same location on a population of auditory neurons, monopolar stimulation may have a temporal coding advantage over bipolar that offsets bipolar's putative spatial superiority. A third possibility

Figure 2



Simplified schematic diagrams of two standard multichannel cochlear implant designs illustrating basic components found in most cochlear prostheses. Four-channel devices are shown to simplify illustration; typically, 6 to 22 channels may be present. In both examples, the acoustic signal is first bandpass filtered so that each channel receives energy from a different portion of the sound spectrum. The band-limited signals are then compressed to accommodate the limited dynamic range available with electrical hearing. These signals are transmitted to the cochlea in tonotopic order by means of a transcutaneous radio-frequency (RF) transmitter or directly via a percutaneous connector. **(a)** The system shown transmits compressed analog signals to each of the four intracochlear electrodes. **(b)** In this device, the four analog signals have their envelope extracted and compressed followed by multiplication by trains of biphasic pulses. This results in the presentation of amplitude-modulated pulsatile stimuli to each electrode. Because the pulses generated for each of the four signals are staggered (or interleaved) in time, no two electrodes are active simultaneously, thus limiting channel interaction.

is that monopolar activates a larger population of 'fringe' neurons at lower levels, with a more stochastic and therefore 'natural' response. It remains to be seen whether a modiolar-hugging radial bipolar electrode configuration, coupled with high-charge-density tolerant electrodes and better current sources, could improve performance. Likewise, electrode and waveform manipulations such as multipolar electrodes [29] and multiphasic waveforms [28•,30] are promising.

Temporal properties of neural activation

Since the first physiological studies of the cochlear prosthesis by Kiang and Moxon [4], it has been well known that short-duration electrical stimulation of a deaf ear results in very different temporal patterns of auditory nerve activation than does acoustic stimulation of a hearing ear. With no inner hair cell synapses in the deaf cochlea, there is a relative lack of spontaneous activity [31] and substantially decreased within- and across-fiber temporal jitter. In the early 1970s, it was suggested that this lack of across-fiber independence would make speech recognition impossible with a cochlear prosthesis. It is still reasonable to ask why implants work as well as they do given this significant limitation. Recent studies in our laboratories [32,33•] and others [34] suggest that the higher stimulation

rates typical of present generation speech processors may partially reproduce the stochastic properties of the inner hair cell synapse. This may result in spike timing resembling that seen with acoustic stimulation rather than the highly deterministic patterns observed in the early physiological experiments. If this is so, substantial further improvements in temporal coding may be possible [35].

Psychophysical studies

Various studies have demonstrated that high levels of speech recognition in both implant patients and normal hearing listeners are possible with as few as four channels of speech information. This holds even when the information is coded by noise bands or sinusoids modulated by a low-pass filtered speech envelope [36–39]. Acoustic simulations of prosthetic hearing appear useful for setting an upper bound on possible performance with the device; in the absence of background noise (i.e. in quiet), the best-performing implantees perceive speech as well as do normal hearing listeners with an acoustic simulation using the same number of channels [40•]. These studies have been extended to perception of speech with background noise (i.e. in noise) and attempts to identify the critical components of speech analysis and stimulus presentation in an implant.

Shannon *et al.* [41••] used four-channel noise-band acoustic simulations of an implant speech processor presented to eight normal hearing subjects. Vowel, consonant and sentence scores were obtained in quiet. Speech processing was parsed into signal analysis and presentation. For the simulations and for implant patients, analysis is represented by the input filters of the speech processor. For the simulations, presentation is represented by the spectral content of the noise-band carriers. For implant patients, electrode location, neural survival and intracochlear current spread determine the spectral characteristics of presentation. Two major conclusions were obtained from the simulations that are pertinent to implant function. First, the precise spectral location of filter and noise-band carrier cutoff frequencies did not significantly affect speech reception as long as they were matched. Decreased speech reception occurred when filter and carrier bands did not match in spectral width or when carrier bands were shifted to higher frequencies mimicking the basal location of cochlear implant electrodes. Second, spectral overlap, or smearing, of filter and carrier bands did not have a negative impact on speech reception in quiet.

These measures were subsequently obtained in noise by Fu *et al.* [42•], who studied three implantees as well as four normal hearing listeners with 3, 4, 8 and 16 channel acoustic simulations of an implant processor. They concluded that, much like hearing-aid users [43,44], implant listeners (and normal hearing listeners with acoustic simulations) require an increasing number of channels to obtain good speech reception as the signal-to-noise ratio is decreased. Similar speech reception results were also obtained in implantees [45•] and in acoustic simulations by Dorman *et al.* [46]. The same implant listeners studied by Fu *et al.* [42•] had their psychophysical channel interaction quantified using both gap detection [25] and forward masking [47] measures. Despite the fact that the small number of widely spaced electrodes used resulted in essentially no measurable channel interaction, information transmission analysis revealed that for spectral place, performance in noise was worse than in the normal hearing listeners. Thus, low channel interaction does not guarantee high spectral fidelity. This counterintuitive finding may limit any performance gains obtained using modiolar-hugging electrodes [48–50], which seek to improve performance by decreasing channel interaction.

Clinicopathological studies

Attempts to correlate cochlear implant function with spiral ganglion cell populations have been hamperead by the limited amount of postmortem material available from patients with multichannel cochlear prostheses, and by the great degree of variability inherent both in speech reception and in histopathological measures. To date, no such correlation has been reported [16,51], despite the fact that duration of deafness is the best predictor of both spiral ganglion cell populations [52] and speech reception with an implant [2•]. Our clinical research center has recently demonstrated a smaller but significant correlation between residual speech reception preoperatively with a hearing aid and such measures postoperatively with an implant [2•]. This finding suggests that the better hearing ear should be chosen for implantation, a clinically controversial conclusion also supported by Incesulu and Nadol [53•]. They avoided the problems of across-subject variability by comparing ears within subjects, and demonstrated a strong correlation between interaural differences in spiral ganglion cell counts and pure tone thresholds. The better-hearing ear during life consistently had higher spiral ganglion cell counts postmortem.

Kawano *et al.* [54] studied postmortem the temporal bones of five patients who had a cochlear prosthesis and demonstrated three significant correlations: first, psychophysical threshold of a given electrode correlated with distance of that electrode from the spiral ganglion; second, psychophysical threshold increased as pathological change (i.e. new bone and fibrous tissue secondary to implantation) increased; and third, psychophysical dynamic range correlated with spiral ganglion cell counts.

These correlations are consistent with those seen in animal studies [11,13–15]. Thus, while many factors, both peripheral and central, may play a role in speech reception with an implant, the morphology and pathology of the electroneural interface clearly contributes to some aspects of performance. It is possible that anatomical and geometrical study of this interface during life will soon become possible with new applications of computerized tomography [55]. Such anatomical data, coupled with intracochlear electrophysiological measures [6•,7] may allow device programming to be tailored to a specific patient without the use of behavioral testing.

Conclusions

In the past year, substantial progress has been made in defining the critical features of speech coding for cochlear prostheses. The required number of channels has been documented and the expected upper bounds on possible performance have been defined. Physiological information has been obtained regarding temporal representation of electrical stimuli in both acutely and chronically deafened ears. A variety of novel stimulus waveforms and speech-processing strategies have been suggested on the basis of these studies and are undergoing testing in both experimental animals and human subjects. Human histopathological data are accumulating, and when combined with psychophysical, speech reception, electrophysiological data and continued animal and computer modeling, they should lead to the clinicopathological correlations necessary to answer precisely the question of "How do cochlear prostheses work?".

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