

How cochlear implants encode speech

Jay T. Rubinstein

Purpose of review

This review summarizes the history of cochlear implant signal processing and provides the rationale underlying current approaches. Present strategies are explained and recent research findings are summarized. It is suggested how these results may drive future advancements in signal processing.

Recent findings

Substantial advances have been made in our understanding of the spectral and temporal cues necessary for cochlear implant recipients to perceive music, speech in noise, and interaural timing. It is clear that higher levels of both spectral and temporal resolution, as well as better loudness and pitch coding are necessary for higher levels of performance. These factors are highly interrelated, however, and are beneficial for differing aspects of hearing. Signal processing algorithms incorporating these findings are under active development and some are currently undergoing clinical investigation.

Summary

Current implant devices, and those soon to be available, have substantial untapped potential to improve the auditory experience of their recipients. It is likely that in the near future, recent findings on pitch and loudness perception, as well as techniques to better emulate the normal functions of the cochlea will result in much higher levels of prosthetic hearing fidelity than are possible today. As the performance of these remarkable devices continues to improve, the population of hearing-impaired individuals who can benefit from implantation is likely to increase significantly.

Keywords

cochlear implant, signal processing, speech perception, music perception, binaural hearing

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Departments of Otolaryngology and Bioengineering, The University of Iowa, Iowa City, Iowa, USA, and Department of Otolaryngology, Leiden University Medical Center, The Netherlands

Correspondence to Jay T. Rubinstein, MD, PhD, The University of Iowa, 200 Hawkins Drive, Iowa City, IA 52242, USA
Tel: 319 353 7042; fax: 319 353 6767; e-mail: jay-rubinstein@uiowa.edu

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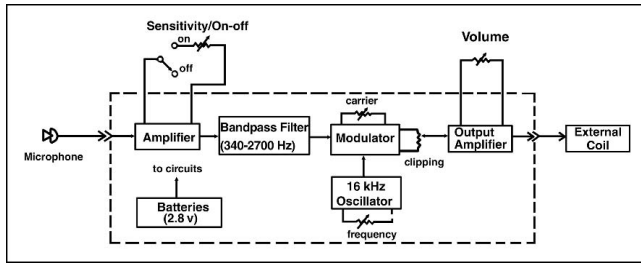
Introduction

Cochlear implant signal processors convert acoustic vibrations into electrical stimuli to be transmitted to the auditory nerve. This role is normally served by the external and middle ears, as well as the cochlea, organ of Corti, and inner hair cells. The processor must take a wideband signal with a dynamic range of greater than 100 dB and convert it into a set of parallel signals each with a maximum “data-transfer” rate of a few thousand spikes per second, the upper limit of auditory nerve phase-locking with electrical stimulation [1], and a dynamic range of ~2 dB, the typical dynamic range for electrical stimulation of an auditory nerve fiber [2••]. This challenging conversion from wideband, wide dynamic range acoustic signal to multiple parallel narrow-band, narrow dynamic range electric signals presented to the auditory nerve represents the performance limiting interface between the implant and the patient’s nervous system. In a normal cochlea, approximately 1000 inner hair cells transmit the details of basilar membrane motion via independent synapses to approximately 30,000 auditory neurons and subsequently to a normal central nervous system [3]. An implanted ear has between 1 and 22 intracochlear electrodes with poor spatial selectivity [4] activating a potentially reduced complement of auditory neurons [5], which convey signals to a potentially abnormal central auditory pathway [6]. How best to do this is a topic of considerable basic and applied interest that is under active investigation in many laboratories around the world. The task is made no easier by the fact that many basic aspects of normal hearing, for example the precise bases for loudness and pitch coding, are incompletely understood [7••].

Single-channel processor

Figure 1 illustrates the design of the first widely used cochlear implant signal processor, the House-3M single-channel device. The strategy was to amplitude modulate a 16kHz sinusoid by the bandpass filtered audio input signal. It was presumably chosen to modulate a high-frequency carrier rather than simply present the input signal due to the need for inductive coupling of the output signal across the skin [8]. As will be discussed, the high carrier rate may in fact have contributed to the success of this approach and to the more “natural” sound reported by the human subjects [9] due to the introduction of stochastic, or random, patterns of neural discharge. While single-channel devices provide valuable sound awareness, cues that aid lipreading, and in some

Figure 1. Diagram demonstrating the signal processor in the 3M House, single-channel cochlear implant



The input signal is filtered into a range relevant for speech. The filter output modulates a 16 kHz sinusoidal carrier. Figure redrawn from [8].

cases, remarkably high levels of speech perception, their use has essentially been replaced by multichannel devices [10]. The higher levels of speech perception produced with multichannel devices are due to the fact that periodicity, or temporal pitch perception, is limited, both behaviorally [11•,12•] and physiologically [1]. Exploitation of “place-pitch” with multiple electrodes is necessary to support higher levels of speech perception. Monosyllabic word scores with multichannel devices have been reported as high as 100% in quiet, although speech perception in noise is substantially lower [13,14].

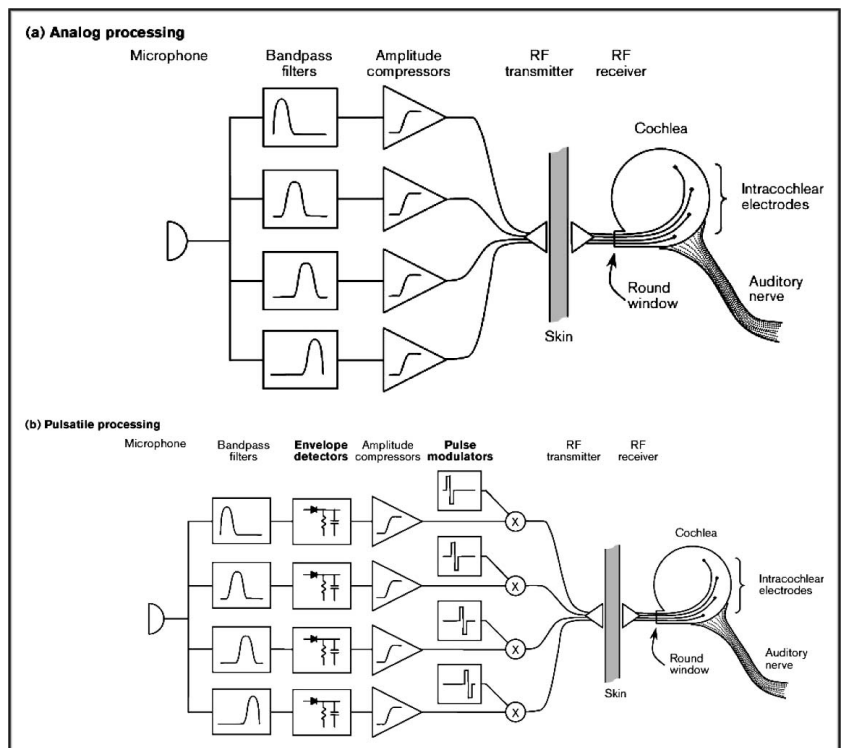
Frequency selectivity and multichannel implants

Figure 2A illustrates a generic multichannel analog cochlear implant processor. The Ineraid compressed analog

strategy used this approach [15]. Bandpass filters separate sound into different frequency regions and compression reduces the dynamic range in each band to the narrow dynamic range of electrical hearing, approximately 20 dB. While this approach attempts to mimic the tonotopicity of the normal cochlea, the similarity stops there. The normal cochlea performs a detailed frequency analysis of incoming sound via the micromechanics of basilar membrane motion under the influence of outer hair cell motility. Highly filtered signals are represented in the membrane potential of each inner hair cell and synapses convey a further transformation of these signals to between 10 and 30 individual auditory neurons [16,17•]. This sharply tuned system can be contrasted with an implanted cochlea where precise filtering can be performed using signal processing, but current spread [18•] and possibly other less well-understood central phenomena [19] limit the spatial selectivity of electrical stimulation [20]. This subsequently limits the spectral resolution of implant listeners [21]. The perceptual impact of poor spectral resolution has been studied in detail in hearing aid users [22] as well as for implant hearing [23] and in implant simulations [24]. It appears that “spectral mismatch” between the population of fibers excited by a given electrode, and the frequency range of sound presented to that electrode pose greater limitations to speech perception, at least in quiet, than does poor spectral resolution *per se* [25]. This mismatch may be related to depth of electrode insertion [26] but may also be due to our ignorance of, and relative inability to control, the precise

Figure 2. Generic multichannel analog cochlear implant processor

(A) A generic multichannel analog signal processor. The compressed outputs of a bandpass filterbank directly drive the intracochlear electrodes. Figure adapted from [29]. (B) A generic multichannel pulsatile processor. The bandpass filter outputs have their envelopes extracted and the envelopes are then compressed. The compressed envelopes modulate interleaved biphasic current pulses. Figure adapted from [29].



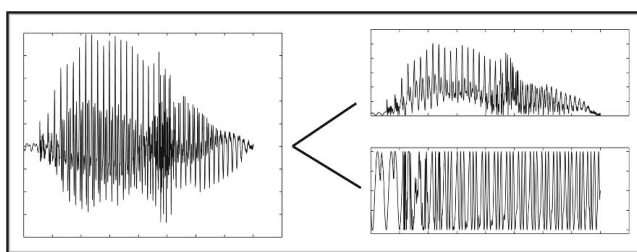
site of electrical excitation of auditory neurons [27–29]. Attempts to improve frequency selectivity include increasing the modiolar proximity of the electrode array [30], development of penetrating electrode arrays [31], possible use of neurotrophic factors to induce growth of neural processes toward electrodes, and temporal coding to reduce channel interaction.

Temporal coding

Figure 2B illustrates a generic multichannel pulsatile processor. The Continuous interleaved sampling, or CIS strategy [32], Spectral Peak, or SPeak and Advanced Combination Encoders, or ACE (Cochlear Corp., Sydney, Australia), as well as the High Resolution (Advanced Bionics Corp, Sylmar, CA) strategies are all variations on this theme. The primary differences among them are rate of stimulation, number of channels, the relation between the number of filters and the number of electrodes activated, as well as the details of how each channel's envelope is extracted and how much temporal information is preserved. These strategies all have in common the use of temporally interleaved pulses to avoid electrical field interactions that reduce frequency selectivity. SPeak and ACE attempt to emphasize spectral peaks in the signal by selecting a subset of filter outputs with higher energy levels. High Resolution augments the envelope with temporal information by allowing higher frequency components through the envelope detector. Although some patients clearly perform better with one strategy than another, average speech scores across populations of patients do not clearly demonstrate superiority of a particular strategy. Variability across subjects is still substantially greater than variability across hardware or software although there is a clear and substantial trend since the early 1980s for speech reception with implants to improve. Some of this improvement is due to advances in speech processing, some due to implanting patients with more normal auditory systems, and some is due to unknown causes.

Figure 3 illustrates the results of envelope extraction for the word “there”. All sound can be divided into envelope and temporal fine structure. Pulsatile signal processors typically do this because speech perception in quiet is effectively conveyed by envelope cues alone on as few as four channels [33]. In addition, neural membranes tend to respond to the envelope of a signal and discard the fine structure due to refractory effects. Since the neural system does not respond well to fine structure, there is no loss of usable information if the signal processor extracts it first. Hence there is usually, but not always, an increase in performance in shifting a patient from an analog to CIS processing presumably due to the decreased channel interaction. In quiet, there does not appear to be a counteracting detrimental effect of delivering only the signal envelope, which can be presented at substantially lower stimulus rates than the fine structure.

Figure 3. The Hilbert transform, and other techniques not shown, can extract both envelope and temporal fine structure from any signal



The signal for the word “boy” from a HINT sentence is shown on the left. It has been transformed into envelope (right upper panel) and temporal fine structure (right lower panel). If the envelope is multiplied by the fine structure, the original signal results. The envelope effectively codes speech in quiet. The fine structure contains temporal pitch and timbre information and is the primary source of interaural timing information. This process is similar to the Fourier transform, which decomposes a signal into amplitude and phase components from which the original signal can be reconstructed.

While four channels of envelope are effective for coding speech in quiet, perception of speech in noise, particularly with interfering speech [34•,35•], requires many more channels or better coding of fine structure and pitch. Music perception [36] as well as tonal language perception [37,38] should also benefit greatly from an increased number of channels but should be particularly susceptible to better representation of fine structure [39•]. Unfortunately, current technologies appear to provide a maximum of 6 to 8 channels of information regardless of the number of electrodes [21] and representation of fine-structure is quite limited [40]. Increasing the number of channels to 12 to 16, for example, would provide a substantial gain in speech perception in steady-state noise but would only have limited effects on music perception or speech perception with interfering talkers [14,34•].

Binaural processing

Binaural hearing depends on the cues of interaural amplitude differences (IAD) and interaural timing differences (ITD) between the two ears. Based on results from a few subjects, it was initially believed that implant recipients could not perceive ITD cues and that the benefits of binaural implantation would be limited to those provided by IAD cues. It is now clear that there is a range of ITD sensitivity in bilateral implant recipients [11•, 41•,42•] and that it will likely be important to match cochleotopic locations bilaterally if these cues are to be exploited by a binaural processor. Whether or not that processor needs the two implants to be temporally synchronized is unclear [11•]. While the central auditory system is sensitive to ITDs of envelope cues [43], synchronized processors could potentially deliver binaural fine-structure ITD, a critical cue not only for lateralization [39•] but also for binaural unmasking [17•]. Restoration of binaural unmasking would be manifested clinically as a robust squelch effect, which is currently limited

in bilaterally implanted patients [44]. If this can be accomplished, substantial improvements in the benefits and cost-efficacy of bilateral implantation are likely.

Promising research directions

It was discussed previously that both the number of information channels provided by implants, as well as the temporal resolution within each channel are significant limitations of current devices. Efforts to increase the number of channels via signal processing include both the use of “current focusing” to achieve “virtual channels” [45] as well as multiphasic waveforms to decrease temporal interactions on cell membranes [46,47]. Efforts to improve the representation of temporal fine structure include the use of “conditioning” or “desynchronizing” pulse trains, or high carrier rates to mimic the stochastic response properties of the inner hair cell synapse [48–52]. If temporal fine structure can be better represented using such techniques, it may be even more critical to deliver such stimuli to the tonotopically “correct” location in the cochlea [53]. Rapid adaptation is another normal property of the inner hair cell synapse and is the focus of strategies that enhance envelope transients [54,55].

Other promising efforts to simulate more closely the normal auditory system include attempts to better understand and enhance coding of pitch [56,57,58•] as well as loudness [59•]. Optimization of dynamic range, both of the acoustic input signal [60,61] and of the electroneural interface [50,51,62] are under active investigation. Lastly, optimization of the individual patient through manipulation of rate and number of electrodes [63], as well as choice of electrodes [18•,20] can offer substantial benefit. Such optimizations may be greatly enhanced through better imaging techniques, which may allow tonotopic localization of individual electrodes within the cochlea [64].

Conclusion

Cochlear implant signal processing is an active area of basic and applied research in multiple laboratories around the world. High levels of speech perception in quiet are currently possible due to the robust nature of speech but significant limitations of prosthetic hearing persist. Bilateral implants currently provide significant binaural benefit but primarily via interaural amplitude difference cues. Future advances in signal processing may soon result in higher levels of speech perception in noise, music perception, and the potential for binaural benefit from interaural timing cues. This would restore true spatial hearing and make the implant a bona fide auditory prosthesis rather than a “mere” speech recognition prosthesis. Such progress is likely dependent on continuing improvement in implanted hardware. Current implants, however, and those soon to become available, have significant untapped potential to provide

higher levels of auditory experience for their recipients. As the performance of these remarkable devices continues to improve, the population of hearing-impaired individuals who can benefit from implantation is likely to increase substantially.

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