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Cochlear Implants

Contributors: Fan-Gang Zeng
Edited by: Jack S. Damico & Martin J. Ball
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This entry describes what a cochlear implant is, how it works, and who may benefit from it. The chapter then discusses limitations of cochlear implants and solutions to overcome these limitations in the future. Finally, the chapter discusses issues related to cochlear implant simulation, accessibility, and impact on Deaf culture. The cochlear implant is a medical device used to restore functional hearing by electrically stimulating the auditory nerve in the damaged inner ear (the cochlea). It is an invasive device, requiring surgery to drill a hole in the skull behind the ear to access the cochlea and an additional hole in the cochlea to place the implant electrodes.

Since the Federal Drug Administration's (FDA's) approval of the first commercial device in 1984, more than half a million people have received cochlear implants worldwide, with roughly one half being adult users and the other half being pediatric users. Except for the first FDA-approved 3M/House cochlear implant system that had a single electrode contact in the cochlea and had about 1,000 users, all contemporary devices have multiple electrode contacts ranging from 12 to 24. In 2017, there were five cochlear implant manufacturers, with three having their products available globally (Cochlear, MED-EL, and Advanced Bionics) and two available regionally (Neurelec in Europe and Nurotron in Asia, Europe, and South America).

The contemporary cochlear implant has allowed its average user to carry on a conversation on the telephone. But more important, it has allowed prelingually deafened users to develop relatively normal language, with many of them attending regular schools. The cochlear implant has been considered a medical and technological wonder and is without question the most successful neural prosthesis that has been developed to restore a major human function.

The Cochlear Implant System

A typical multielectrode system consists of an external part and an internal part. The external part includes at least a microphone, a hearing-aid-like sound processor with batteries, and a coil that transmits radio frequency signals to the internal part. The internal part includes at least a coil that receives the radio frequency signals, a receiver and stimulator case, and an electrode array. The transmitting and receiving coils are aligned by magnets across the skin to form an efficient transcutaneous radio frequency transmission link. The receiver and stimulator case is hermetically sealed so that the electronics inside the case are protected from erosion by body fluids and moisture.

Except for earlier versions of the cochlear implant systems that used a ceramic case, all available commercial systems use a titanium case that improves mechanical protection of the enclosed electronics. The intracochlear electrode array consists of 12 to 24 small contacts typically made of 90% platinum and 10% iridium alloy for its low reactive electrical property and high resistance to corrosion. The electrode array may be straight or curved, similar to the snail-like cochlear shape. A return or reference electrode made of the same alloy is placed outside the cochlea as either a peripheral part of the intracochlear array, as a plate attached to the surface of the case, or as an independent array.

To ensure that the cochlear implant functions properly, audiologists use a programming unit to set the electric stimulation parameters customized to each user. Although the programming unit is not available to an actual user, the user has some controls of the electric stimulation parameters, such as sound sensitivity, stimulation volume, or one set of customized parameters or "maps." The user controls these either by buttons attached to the external sound processor or by a handheld unit that communicates with the sound processor wirelessly. The programming unit typically includes a back-telemetry function embedded in the cochlear implant system that monitors to some extent the status of the internal part but, more important, the electrode impedance, electric field distribution, and electrically evoked neural responses.

Working Principles

The most important working principle for a cochlear implant is to replace some of the cochlear functions that are damaged in a hearing-impaired individual. Frequency filtering and amplitude compression are two essential normal cochlear functions. A normal human cochlea handles an audio frequency from 20 to 20,000 Hz and can resolve < 0.1% difference in frequency. To replace frequency filtering, the cochlear implant takes advantage of the tonotopic organization of the auditory system by introducing multiple electrodes in the cochlea, with the basal electrode representing high frequency and more apical electrodes representing lower frequencies.

In a normal human cochlea, the tonotopic relationship between sound frequency (f) and cochlear position is well established as the so-called Greenwood map, developed by Donald Greenwood:

$$f = 165.4\left[10^{(2.1x/35)} - 0.88\right]$$

where x represents the cochlear length in mm from the most apical position, with the entire cochlear length being 35 mm. Most contemporary cochlear implants have used the Greenwood map or a more compressive spiral ganglion cell map to assign a sound frequency to a corresponding electrode.

The physical filtering is actually performed by a digital signal processing chip in the external sound processor. The implant frequency range is typically between several hundreds of Hz to roughly 8,000 Hz. This is narrower than the full 20 to 20,000 Hz audio range, partly because the cochlear implant does not have enough electrodes to cover the entire length of the cochlea and it deals mostly with speech sounds.

Cochlear compression allows a normal listener to handle a 100-dB acoustic dynamic range. To replace the normal compression function, the cochlear implant typically uses a logarithmic or power function to compress a 50-dB acoustic dynamic range into a 10 to 20-dB electric dynamic range. The 50-dB acoustic dynamic range is narrower than the 100-dB normal dynamic range but adequate for most daily conversations. In addition, a sensitivity control may shift the 50-dB acoustic range up or down to accommodate a desirable input signal, depending on the signal and background noise levels. Most important, the compression function is not directly applied to the instantaneous amplitude of a sound signal but instead to the instantaneous envelope of the sound signal.

First, the sound signal is divided into multiple narrow-band signals corresponding to the number of electrodes. For each narrow band, the envelope is extracted either by rectification with low-pass filtering or by the Hilbert transform, and the fine structure is typically discarded. Then, the band-specific envelope is compressed to amplitude modulate a constant-rate biphasic pulse carrier. The band-specific, compressed, envelope-modulated pulse carrier is delivered to its corresponding electrode. The constant-rate pulse carrier on a single electrode allows easy implementation of interleaved delivery of electric stimulation on multiple electrodes—namely, from either 1, 2, . . . to n, vice versa, or any predefined sequential order.

The interleaved stimulation ensures that no two electrodes are stimulated at once, avoiding complicated simultaneous electrical field interactions between two or more electrodes that would not only compromise the frequency-to-electrode map but also produce unacceptable loudness fluctuations. For example, simultaneous addition of two identical electric fields produces a 6-dB increase in current, which could be a third of or even the entire electric dynamic range. Except for earlier devices (e.g., Ineraid in 1970s–1990s and Clarion by Advanced Bionics in 1990s–2000s) that used simultaneous stimulation, all cochlear implants have adopted interleaved electric stimulation. Both the use of sound envelopes and the interleaved stimulation have contributed greatly to the high level of speech performance achieved by contemporary cochlear implants.

Several general working principles have also been developed to ensure safety of the cochlear implant. First, all contemporary devices have adopted current sources to deliver charge-balanced, biphasic, pulsatile stimu-

lation to minimize side effects of electric stimulation such as channel interference, bone growth, and neurotoxicity. Second, except for occasional recalls due to significant design- or manufacturing-related device failures, most contemporary devices have had an excellent safety record of 5% to 10% revision rate, with children having slightly higher revision rates than adults. The most common causes for revision include device failures from easily detectable implant or electrode migration to hardly identifiable soft failures; nonauditory side effects such as vestibular, taste, and facial nerve activations; and surgery-related complications from infection to cerebrospinal fluid leakage.

Third, contemporary devices are compatible with at least 1.5-Tesla magnetic resonance imaging (MRI) and may be compatible with higher Tesla imaging by temporarily removing the internal magnet or changing its polarity. MRI compatibility has become increasingly important because of the need of to MRI to evaluate neurological and metabolic status or diseases in the ever-increasing number of cochlear implant users who may have the device in their head throughout their lifespans and may have accompanying complications.

Candidacies and Indications

Candidacies and indications for cochlear implants are tightly coupled to their performance. When the single-channel 3M/House device was first approved by FDA in 1984, its performance was limited to improved detection of sounds, lipreading, and voice modulation. At that time, the only single-channel device candidates were adults with bilaterally profound hearing loss and no other accompanying sensory and neural disorders. With an additional benefit in open-set speech recognition (about 20% correct), the Nucleus-22 multichannel device by Cochlear was approved for deaf children in 1990. When implant performance was improved with advanced signal processing to about 80% open-set speech recognition in quiet, the criteria for cochlear implantation were further relaxed.

For example, in 2005, Medicare, the U.S. federal health insurance program, started to cover cochlear implants for individuals who received a pre-implant score of less than 40% correct open-set speech recognition under the best-aided listening conditions. Except for those who have active middle ear infection or auditory nerve or central auditory pathway lesions, cochlear implant candidates include those who may have severe or more hearing loss (≥70 dB HL) with 60% or less open-set speech recognition under the best-aided condition. Several other labeled (approved by the FDA) or unlabeled (unapproved by the FDA) indications have expanded cochlear implant candidacy and benefits.

First, a natural indication is bilateral cochlear implantation that would replace two damaged ears. About 20% of users have bilateral cochlear implants, improving sound localization mostly due to interaural level differences and speech recognition in noise due to the “better ear effect.” Bilateral implants provide limited access to interaural timing differences and other complex central processing responsible for the true binaural benefit in a normal hearing listener. There is evidence for additional benefits of simultaneous and early bilateral cochlear implantation, but bilateral implantation benefits are still limited, considering the doubling in cost. As a result, most national and private insurance programs do not cover bilateral cochlear implants.

Second, electroacoustic stimulation, or the hybrid cochlear implant, expands the criterion to patients with severe to profound middle- to high-frequency hearing loss but normal to moderate hearing loss in the low frequency (≤ 500 Hz). The hybrid implant is similar to a conventional device except for a shorter electrode array, which in combination with a soft surgical approach to minimize trauma can preserve low-frequency hearing and provide significant improvement in speech recognition, especially in noise.

In cases of low-frequency hearing loss, an accompanying hearing aid is used simultaneously with the hybrid cochlear implant. Similar benefits can be achieved by combining a conventional cochlear implant in one ear and a hearing aid in the other ear. Most insurance programs cover hybrid implants for their comparable cost to the conventional implants.

Third, cochlear implantation has been expanded to patients with single-sided deafness, especially when debilitating tinnitus is present as a comorbid symptom. Although there has been significant improvement in sound localization, speech recognition in noise, and tinnitus reduction in preliminary clinical trials, cochlear implantation in single-sided deafness or tinnitus treatment is still for off-label use.

Fourth, cochlear implantation has been widely used to manage auditory neuropathy as both labeled and unlabeled indications. The implantation is considered labeled if the auditory neuropathy patient has significant hearing loss that meets the standard criteria. The implantation is considered unlabeled if the patient's hearing loss, especially in terms of pure-tone audiogram, does not meet the standard criteria.

The reason for the relatively widespread unlabeled use is that many auditory neuropathy patients have mild-to-moderate hearing loss or even normal hearing but disproportionally low speech recognition, especially in noise. The outcome of cochlear implantation in auditory neuropathy has more to do with etiology (cause or origin) than with the degree of hearing loss as measured by audiogram. If the etiology is located in the inner hair cell or synapse (e.g., otoferlin deficiency), then cochlear implantation usually results in excellent performance. On the other hand, if an etiology gives rise to significant auditory nerve damage (e.g., Mondini malformation), implantation usually produces poor performance.

So far, cochlear implant indications have been expanded to individuals with multiple disabilities, including hearing-impaired children with autism and elderly with Alzheimer's disease. Cochlear implants are also used simultaneously with other medical devices such as pacemakers and deep brain stimulators to manage a wide range of diseases, from sensory and motor dysfunctions to cardiovascular and neurological disorders. Except for cautious notes to avoid electric interference between devices, no formal tests, standards, or approvals are available to ensure safe and effective operation of these coexisting implantable devices.

Limitations and Solutions

Despite the advances and successes made in cochlear implant technology and performance, there is still a huge gap between prosthetic and natural hearing. Pitch represents such a gap: A normal-hearing listener can discriminate a < 0.1% difference in frequency, whereas a cochlear-implant user's performance is about two orders of magnitude worse. This huge gap in pitch discrimination is deeply rooted in a technological limitation in electrode design, which has not fundamentally changed since its inception 40 years ago.

The two orders of magnitude difference reflect the difference in numbers (namely, 3,000 inner hair cells vs. 24—the max—electrodes) and also in size (namely, 1–10 μm auditory neurons vs. ~1 mm electrodes). Poor electric pitch discrimination is the root cause for the difficulty experienced by cochlear implant users in music appreciation, environmental sound recognition, and speech recognition in noise and reverberation. A fundamentally different interface is needed to overcome the electrode-to-neuron limitation in cochlear implants.

One incremental interface is to use penetrating electrodes that can be inserted in the auditory nerve bundle. Penetrating electrodes not only lower the stimulation thresholds and increase spatial selectivity; they can also access low-frequency neurons located inside the auditory nerve bundle. Acute animal studies have demonstrated technical feasibility and improvement in outcomes of using penetrating electrodes over cochlear electrodes, but chronic human application remains unclear due to stabilization of the stiff penetrating electrode in the soft human tissue. Laser or optogenetic stimulation of the auditory nerve can potentially close the gap between prosthetic and natural pitch discrimination, but their human application is still years if not decades away due to safety and implementation issues.

Power consumption is another huge gap between prosthetic and natural hearing. The normal cochlea has an estimated power consumption of 14μW, whereas a typical cochlear implant consumes three orders of magnitude more (10–50mW). Although there is some room for lowering power consumption in the external sound

processor and the transcutaneous radio frequency link, the total cochlear implant power consumption would not be significantly lowered if the inefficient electrode-to-neuron interface was not modified. Augmentative technologies such as piezoelectric MEMS transducers and coating electrodes with nerve growth factors may allow an order-of-magnitude reduction in power consumption. A totally implantable cochlear implant, which removes the need for the transcutaneous link, can further close the gap in power consumption between prosthetic and natural hearing.

Other Issues

One interesting question is what a cochlear implant sounds like to a normal-hearing individual. Several versions of cochlear-implant simulations are available on the Internet. A key feature underlying the simulations is the extraction of temporal envelope cues, ensuring that the same kind of acoustic information is available to both actual and simulated implant listeners at the same amount. This aspect of cochlear-implant simulations is accurate and appropriate because the same signal processing applies to actual implant users. The inaccurate and inappropriate aspect of simulations is the carrier used to deliver the envelopes to a normal-hearing listener.

From Robert Shannon's original noise bands to sinusoids or band-passed click trains, none can capture the use of interleaved biphasic pulses, electrode configurations, and individually dependent electrode-to-neuron interface in an actual implant user. To more accurately simulate interleaved electric stimulation and channel interactions, Gaussian noise and Gabor sinusoids, or clicks, have been used to improve simulation quality.

In a few unilateral cochlear implant users who had normal hearing in the contralateral ear, such simulations produced not only similar overall percentage speech recognition scores but also speech error patterns and, more important, a high similarity ranking in sound quality between the actual implant in one ear and the simulated stimulation in the other. In prelingually deafened children who receive cochlear implants, comparing their perceived sound quality with normal hearing is meaningless because these children's imprinted sound is via electric stimulation.

An important issue is the role of cochlear implantation in Deaf culture. In the early days, many people including some mainstream scientists viewed cochlear implantation as primitive and unethical. Not surprisingly, most in the Deaf community condemned cochlear implants and even compared them with technological genocide that threatens to wipe out the Deaf culture.

In 2000, the Position Statement on Cochlear Implants by the National Association of the Deaf had a relatively neutral view that cochlear implants provide sensitive hearing that is not a cure for deafness and certainly not appropriate for all deaf and hard-of-hearing children and adults. The National Association of the Deaf has not reissued any position statement on cochlear implants since; its most recent "Vision 2000" strategic plan emphasizes American Sign Language without any mention of the cochlear implant.

Cochlear implant accessibility is still an issue. Approximately 60,000 cochlear implants were sold worldwide in 2016, and the global market is expected to grow ~15% annually. Considering the 20% bilateral implantation rate, 50,000 individuals, including roughly 25,000 children, were implanted in 2016. Given 130 million births per year and 0.1% deaf birthrate, the total market for deaf children is 130,000. The market penetration rate for children is therefore only 19%, not counting backlog cases.

The rate for adults is about the same but will likely be much lower with the aging population, relaxing criteria, and expanding indications. Accessibility is presently unbalanced for the majority of cochlear implant users residing in developed countries, and the lack of awareness, affordability, and infrastructural support severely limits cochlear implant accessibility in developing countries.

See also Auditory Brainstem Implant; Auditory Neuropathy Spectrum Disorder; Deaf Culture; Psychoacoustics; Speech Perception, Theories of; Tinnitus

Fan-Gang Zeng
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