Chapter 6

Restoration of Hearing by Electrical Stimulation of the Human Cochlea, Brainstem, and Midbrain

Robert V. Shannon, Ph.D.[1]

[1] House Ear Institute, 2100 W. Third St., Los Angeles, CA 90057
PH: 213-353-7020; FX: 213-413-0950; EM: shannon@hei.org

6.1 Introduction

When cochlear implants were in an early stage of development in the 1970s most auditory neuroscientists were highly skeptical that electrical stimulation of the cochlea could ever produce useful auditory sensations. The scientists predicted that patients might hear "beeps and boops" but would never be able to understand complex auditory patterns like speech. It was unthinkable at the time that a small number of electrodes stimulating broad regions of neurons could provide functional replacement for the complex nonlinear hydrodynamics of the cochlea and the highly complex pattern of neural responses by more than 30,000 stochastically independent nerves. The initial devices did provide mostly sound awareness and rudimentary sound discrimination. However, improvements in electrode arrays and signal processing have resulted in steady improvements over time. Figure 6.1 shows a meta-analysis of average levels of word and sentence recognition by cochlear implants from published clinical studies. The initial bars on the left were from single channel cochlear implants in the early 1970s and the rightmost bars are recent results from the latest implant technology. As of 2008 the average postlingually deaf adult cochlear implant recipient can understand about 96% of sentences and 64% of isolated single words using only the sound from their implant (Spahr and Dorman, 2006). This level of performance is easily good enough to converse normally on the telephone. People with cochlear implants can now converse with people so effortlessly that many people cannot believe that they are really deaf. How could the auditory neuroscientists of the 1970s have been so wrong? How can a small number of electrodes so successfully replace the function of the cochlea and its 30,000 hair cells? In this chapter we will explore the technological and scientific discoveries that contributed to this amazing improvement in function. First let us review the history of this technology and its variations.

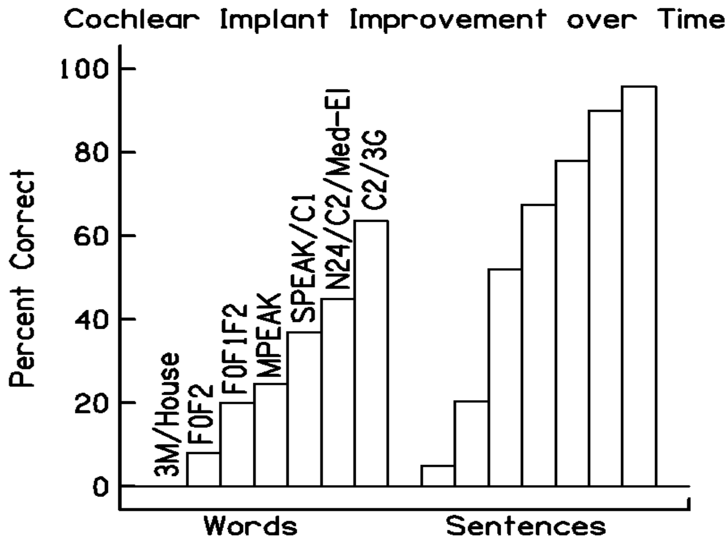

Figure 6.1. Cochlear implant outcomes over time. The results shown are derived from clinical studies reporting word and sentence recognition of postlingually deafened adults across evolving signal processing strategies. The earliest signal processing strategies (left-most bars) were single-channel and occurred in the early 1970s and the latest results are from the newest signal processing strategies (C2 and 3G) reported by Spahr and Dorman (2006).

6.2 Brief History of Cochlear Implants

Volta may have been one of the first people to experience electrical stimulation of hearing. It is reported that he charged up a glass rod and shoved it in his ear canal. He described a sound like "thick boiling soup". It's not clear if he was hearing the sounds of the dc current electrolysis of his inner ear fluids or some electrophonic activation of his inner ear, but he found the experience unpleasant and did not pursue it.

The first scientific study of electrical stimulation of hearing was performed by a French engineer Andre Djourno and the surgeon Charles Eyries. They implanted a simple coil in the auditory nerve of two deaf volunteers in 1957. While the devices provided rudimentary sound sensations, the devices failed in a short time and personal disagreements between the two men halted the project (Eisen, 2003).

Modern cochlear implants started with single-channel devices in the 1960's (House and Urban, 1973; Michelson et al., 1973; Simmons, 1966). While these devices did not provide recognizable speech they were highly useful as communication aids and patients were highly enthusiastic. Multichannel devices were developed simultaneously by several research groups (Eddington et al., 1978; House and Edgerton, 1982; Merzenich et al., 1979; Clark et al., 1990; Simmons et al., 1981; Chouard, 1980; Hochmair-Desoyer et al., 1983). Some of these early devices evolved into the modern implants while others were abandoned.

The auditory system consists of the external ear (pinna) and ear canal, the middle ear (ear drum and middle ear bones), and inner ear (cochlea). Sound vibration enters the outer ear and is transduced into the fluids of the inner ear by the vibration of the ear drum and bones of the middle ear (ossicles). The vibrations within the inner ear (cochlea) convert the sound from vibrational energy into nerve impulses to the brain. The cochlea contains some 30,000 tiny cells, called hair cells, which transduce the vibrations into nerve impulses. Most types of deafness are caused by the loss of these fragile hair cells. In this case the hydrodynamic vibration of the cochlea is still intact and the nerve is still intact, but the connection between the two is lost so sound energy does not reach the brain. Cochlear implants are intended to replace the hair cells and activate the nerves directly. An array of electrodes is inserted into the cochlea (Fig. 6.2) and electrical stimulation is delivered which stimulates the nerves. The normal cochlea separates sounds by frequency, called a tonotopic representation, with high frequency sounds represented near the base of the cochlea and low-frequency sounds represented near the apex. The multiple electrodes of the cochlear implant are designed to activate different pitch regions to reconstruct the frequency combinations of sound. While the hearing ear contains about 30,000 hair cells, implants have 16-22 electrodes distributed along a region that would normally respond to acoustic frequencies of 500 - 8000 Hz. Figure 6.3 shows a modern cochlear implant inserted into a clear acrylic model of a human cochlea. The electrode is designed to lie along the inner wall of the chamber, which is closest to the stimulable neurons.

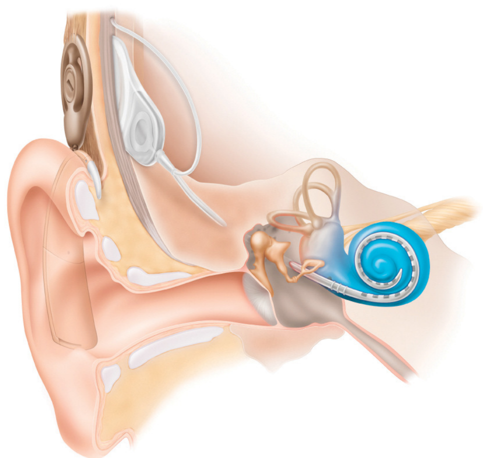

Figure 6.2. Illustration of a cochlear implant. The cochlear implant consists of an external portion, and an implantable portion. The external portion (brown) contains a microphone, signal processor, and rf transmitter. The implanted portion contains an rf antenna receiver, a signal decoder, and a stimulus generator. The cochlear implant electrode array, containing 12-22 stimulating sites, is inserted into the scala tymapni of the cochlea (blue). Acoustic signals are received at the microphone, processed and transmitted to the internal electronics. The received signal is then decoded and sent to the appropriate electrodes in the cochlea.

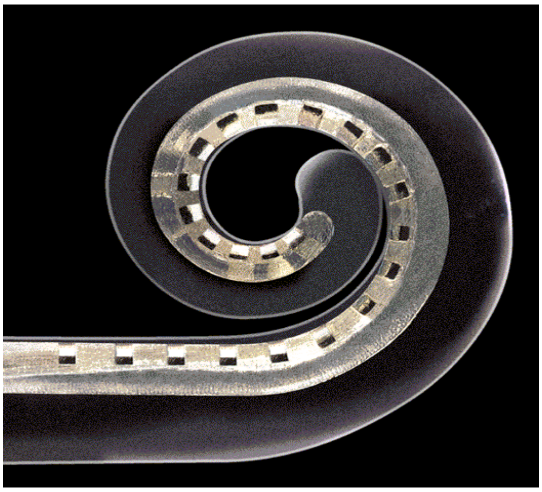

Figure 6.3. The cochlear implant electrode array is designed to be inserted more than one full turn into the scala tympani of the cochlea. This picture shows the ideal placement of the cochlear implant electrode array in an acrylic model of a human cochlea.

6.2.1 *Auditory brainstem implants*

Some people don't have an auditory nerve that can be stimulated by a cochlear implant. The auditory nerve can be damaged by head trauma, severe ossification, or tumors. The most common cause of VIII nerve damage is from neurofibromatosis type 2 (NF2), a genetic condition resulting from a mutation on chromosome 22 (Baser et al., 2003). NF2 produces schwannomas on the vestibular branch of the VIII nerve. Removal of these tumors usually damages the VIII nerve and results in complete deafness. A cochlear implant is of no use to these patients since there is no auditory nerve. An auditory brainstem implant (ABI) was developed for these patients (Fig. 6.4).

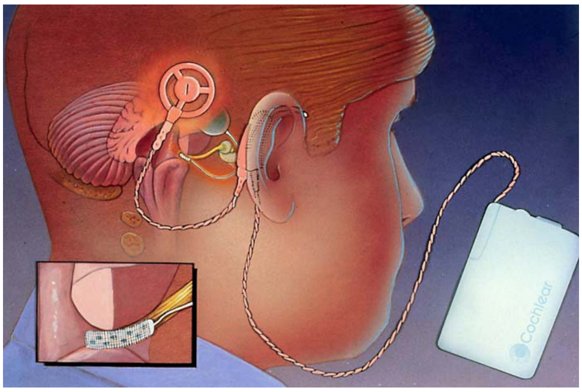

Figure 6.4. A schematic representation of the auditory brainstem implant (ABI). The exterior microphone and signal processing are similar to that of a cochlear implant. The implanted package is also similar except for the electrode array.

The first auditory brainstem imlant (ABI) was done in 1979 by William House and William Hitselberger in Los Angeles (Edgerton et al., 1982; Eisenberg et al., 1987). They placed a ball electrode into the region of the cochlear nucleus complex in the brainstem during a surgery to remove a vestibular schwannoma (VS) in a patient with NF2. That first ABI is still functioning in 2008 and the patient wears the device every waking hour. From 1979 to 1990 the ABI progressed slowly from a single electrode to a three electrode model (Fig. 6.5), with successive electrodes designed in collaboration with the Huntington Medical Research Institute in Pasadena. The array was designed with a mesh backing to encourage ingrowth of a fibrotic tissue reaction. Unlike a cochlear implant, where the electrode array is inserted into a bony chamber, the ABI is placed in the lateral recess of the IV ventricle and is not strongly attached to the brainstem. The mesh backing allows the natural foreign body fibrotic reaction to fix the electrode to the brainstem.

In 1992 Cochlear Corporation, the leading manufacturer of cochlear implants, collaborated with the House Ear Institute and HMRI to develop a commercial multichannel ABI (Brackmann et al., 1993; Shannon et al.,

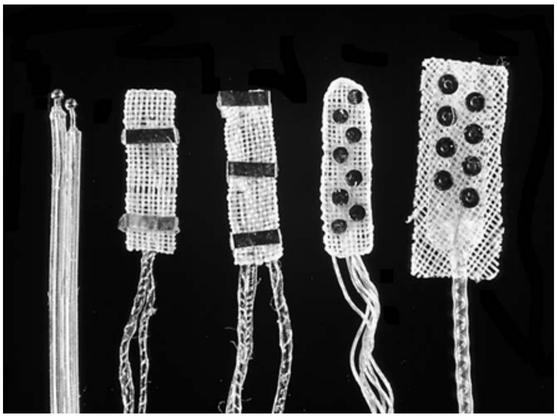

Figure 6.5. The evolution of ABI electrode design. The first ABI was a single pair of ball electrodes. Subsequent designs incorporated a woven mesh to encourage fibrous ingrowth to fix the array in place.

1993). The initial design contained 8 electrodes on a silicone substrate. Later the number of electrodes was increased to 21 to take advantage of the stimulation capabilities of their system. Figure 6.6 shows the present commercial ABI system. The implanted portion of the device contains

Figure 6.6. The ABI implantable system consists of a receiving antenna, a hermetically sealed electronics package that receives and decodes the signal, and a current source stimulator, and the electrode array.

the electrode array and the hermetically sealed receiver/stimulator package. This implanted unit received signals transmitted across the skin, decodes the signal, and presents a biphasic current-controlled pulse to the appropriate electrode. The external unit contains a microphone, digital sound processor, and rf transmitter circuitry. The external transmitter coil and the internal receiver coils are aligned by magnets.

The ABI was approved for clinical use by the Food and Drug Administration in 1996 and has now been implanted in more than 700 patients worldwide. The average level of performance is not as high as that observed in cochlear implant recipients. Most ABI listeners receive sound awareness, sound differentiation, some recognition/discrimination of environmental sounds, and help with lipreading (Otto et al., 2002). On average ABI listeners can understand 30% more speech when using the ABI in conjunction with lipreading than with lipreading alone. But overall, most ABI patients are not able to understand speech with only the sound from their implant. Average ABI performance is similar to the early single-channel cochlear implants, represented by the leftmost bars in each cluster in Fig. 6.1.

The reasons for the rather limited performance with the ABI are not clear. It was thought that the surface electrodes on the cochlear nucleus were not making good contact with the tonotopic axis of the auditory system. Cochlear implants are inserted into the scala tympani where they are nicely aligned with the normal tonotopic axis of the cochlea; electrodes at one end activate nerves that produce a high pitch sensation and electrodes at the other end produce a low pitch sensation. But the surface ABI electrode array is placed along the outside of the posteroventral cochlear nucleus and the dorsal cochlear nucleus in the brainstem. The cochlear nucleus has several independent representations of pitch and none of them are well represented on the surface. Anatomical studies suggest that the tonotopic axis of the PVCN runs deep to the surface, with low frequencies represented on the surface and high frequencies represented deep to the surface (Moore, 1987).

It was thought that it would be necessary to use penetrating microelectrodes to access the tonotopic organization of the auditory brainstem (McCreery et al., 1998). Under an NIH contract electrodes were developed and safety studies verified that such a device was feasible. Microelectrodes were 50 micron diameter iridium shafts insulated with parylene. The tip of each electrode was etched into a conical shape with a final tip diameter of 6 microns. Repeated insertion studies in cat spinal chord showed these electrodes to produce minimal tissue and vascular injury. The first human patients were implanted in 2004 (Fig. 6.7) and the evaluation continues (Otto et al., 2008). At the present time it is clear that the penetrating microelectrodes are implanted correctly in the cochlear nucleus and that they produce highly selective activation of the tonotopic dimension of the CN. However, it is not clear at present if this improved spectral selectivity will result in improved speech recognition over the surface electrode ABI.

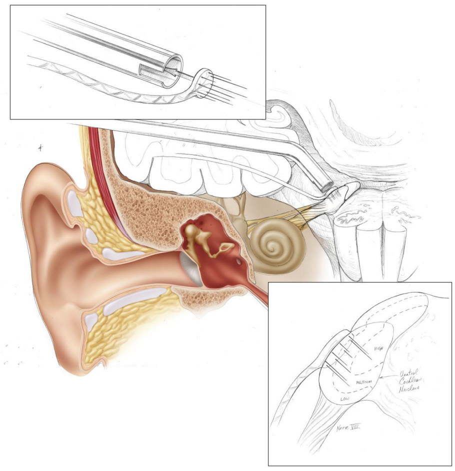

Figure 6.7. Illustration of the penetrating auditory brainstem implant. Top insert shows the electrode insertion tool and the lower insert shows the optimal placement of the penetrating microelectrodes into the ventral cochlear nucleus.

A new development in ABI results throws doubt on the original reason for developing penetrating microelectrodes. It was thought that surface electrodes were limiting performance because of their poor connection to the tonotopic dimension of the auditory brainstem.

However, some patients were observed who could obtain high levels of speech recognition with the normal surface electrode ABI - performance that was similar to that seen in cochlear implants (Colletti and Shannon, 2005). This result shows that excellent speech recognition is possible with surface electrode stimulation of the brainstem and suggests that the pathology associated with NF2 is responsible for limiting performance, not the device design or placement.

6.2.2 *Auditory midbrain implants*

If the ABI can produce excellent speech recognition, but NF2 patients cannot achieve this result due to localized damage to the brainstem, then a better outcome might be possible by stimulating a higher level of the auditory system. The inferior colliculus (IC) is a large midbrain nucleus with a well-studied tonotopic organization. Surgical access is possible using a traditional neurosurgical approach to the midbrain: the infratentorial supracerebellar approach (Fig. 6.8). At the present time there are two approaches to an implant for the auditory midbrains: the Auditory Midbrain Implant (AMI) using a penetrating electrode array (Lenarz et al, 2006; Lim et al, 2006) and the Inferior Colliculus Implant (ICI) using a surface electrode array placed on the dorsal surface of the IC (Colletti et al., 2007; see Fig. 6.6). The first human patient was

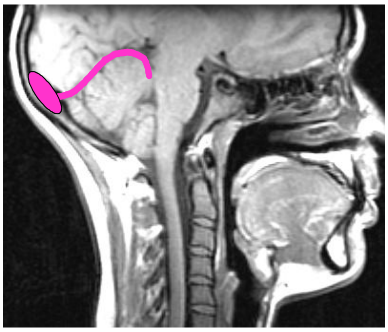

Figure 6.8. Schematic diagram of the placement of the ICI on the surface of the Inferior Colliculus. Surgical access was via the infratentorial supracerebellar approach.

implanted with the ICI in 2005 and the first patient with the AMI was implanted in 2006. It is still too early to know how well these approaches will work relative to cochlear implants and ABIs. The initial results of the ICI and AMI show auditory sensations and different pitch percepts across the electrode array but relatively poor speech recognition ability (Colletti et al., 2007; Lim et al., 2007, 2008a, b). For patients whose cochlea and brainstem are damaged from NF2 and other pathologies these implants may provide the only possibility for auditory function. But the higher in the system we stimulate the more intrinsic neural processing is bypassed. At some point we will likely reach the "point of diminishing returns", where stimulation is not useful in terms of speech. The excellent speech recognition in non-NF2 ABI patients shows that the CN is not too central. It is not yet clear whether the IC is still peripheral enough that prosthetic activation can still provide useful pattern recognition.

6.3 Safety of Electrical Stimulation

Useful hearing sensations would not be as desirable if the prosthetic devices caused damage to the stimulated neurons that eventually reduced their effectiveness. What do we know about the long-term safety and stability of electrical stimulation of neurons? Firstly, we know that cochlear implants have been in use for more than 35 years and ABIs for 30 years without any evidence of long-term decline in performance. Some of the original patients with these devices use them as much as 16 hours per day and performance and perceptual levels are stable.

Extensive animal studies (Agnew and McCreery, 1990; McCreery et al., 1990) have defined the limits of safe stimulation. Two factors contribute to neural damage from electrical stimulation: non-reversible ionic reaction products near the electrode-tissue interface, and excitotoxicity. Most reaction products that occur when charge is passed from an electrode are reversible if the charge is recovered by a reverse phase within a short time. Lilly et al. (1955) first proposed a brief biphasic charge-balanced waveform for limiting neural damage.

Neural damage can occur from excitotoxicity even if no damaging reaction products are generated near the electrode surface. In this case the neuron is driven to produce action potentials at a rate that cannot be sustained by the cell's metabolic machinery. With sustained stimulation the neuron's ionic balance is disrupted to the level that it is fatally damaged. Extensive chronic experiments (McCreery et al., 1990) have shown that such damage is determined both by the absolute level of charge and the charge density. A meta-analysis of this data resulted in a model of safe stimulation levels for electrical excitation of neurons (Shannon, 1992b). It appears that the primary factor determining toxicity of electrical stimulation is the charge density at the neuron. If the electrode is close to stimulable neurons, then the local charge density near the electrode surface is critical. Some electrode geometries can produce edge effects and "hot spots" where charge density is higher and it is in these areas where neurons are most likely to be damaged. If the neurons are more distant from the electrodes, these local concentrations of charge are dispersed by the distance and the relevant factor is simply the charge density at the neuron.

How is it possible to know the distance between the stimulating electrode and the neurons? A seminal paper by Ranck (1975) plotted the distance between the stimulating electrode and the neurons over hundreds of biophysical studies, including myelinated and unmyelinated neurons, in vivo and in vitro. Figure 6.9 replots the data from these studies. In general, a 200 μs/phase pulse of 1 mA can stimulate a neuron that is about 1 mm away. The consistency across studies was impressive, given the diversity of methods and preparations. We had an opportunity to confirm this result when we obtained the brain of an ABI patient who had died (Shannon et al., 1997). We were able to measure the actual distance between stimulating electrode and the cochlear nucleus in the brainstem. We also had years of measurements from this patient of the current levels necessary to produce auditory percepts. When we compared these results to the figure of Ranck the agreement was excellent. Our ABI patient had thresholds of 250-600 μA and the actual distance measured was 500 μm between electrode and brainstem. This consistency between a human implant and laboratory measures gives us confidence that we can estimate the distance between the stimulating electrode and the target neurons from the current levels needed to elicit a response.

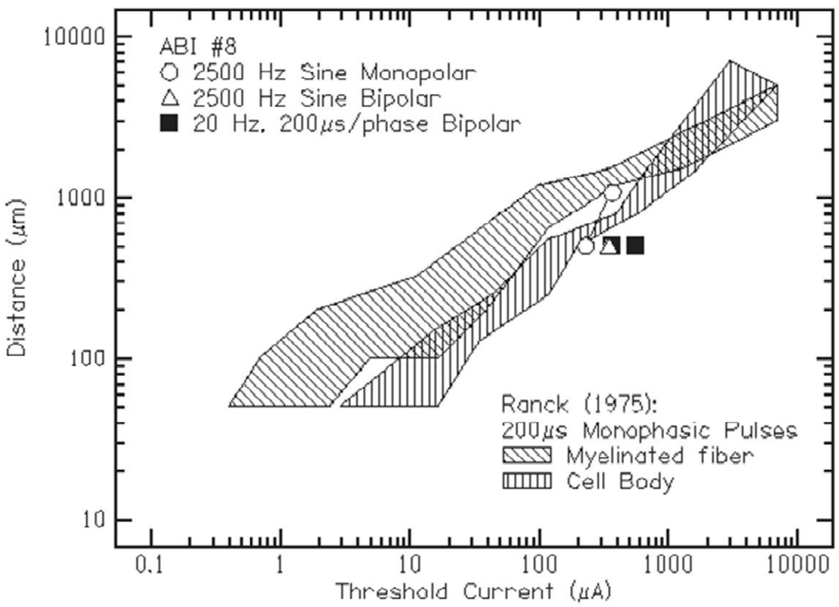

Figure 6.9. Composite plot of the meta-analysis of Ranck on the relation between stimulation amplitude and distance between electrode and neuron. Superimposed on Ranck's results are data from an ABI patient post-mortem. Note that the behavioral thresholds and the anatomical distances relate well to the animal experiments summarized by Ranck.

6.4 Signal Processing for Implants

The original single-channel implant simply placed a filtered version of the acoustic waveform on the electrodes. In some cases the acoustic waveform was compressed and used to modulate a high-frequency carrier for better coil-coupled transmission across the skin. Such a waveform is highly inappropriate for transmission of speech information because the raw acoustic waveform is not charge balanced and because the biological mechanisms of the deafened cochlea cannot extract frequency-specific information from an analog electrical waveform. All neurons near the electrode are stimulated whenever the electrical waveform delivers sufficient charge to trigger action potentials. The timing of such nerve spikes will be related to the gross timing of voice fundamental frequency, but will not reflect the spectral elements of speech.

To achieve frequency-specific stimulation, multichannel electrode arrays were developed to take advantage of the normal tonotopic organization of the cochlea. Sound was divided into multiple frequency bands and the frequency-specific signals were delivered to different electrodes, with low-frequency information presented to electrodes near the apex of the cochlea where low-frequency information is normally processed. The early multichannel processing either used frequency-specific analog waveforms, or attempted to extract the most important features from speech and stimulate with a simplified signal (so called feature extraction strategies). Neither of these processing methods worked very well. The multichannel analog strategies were complicated by electrical field summation across electrodes. The frequency-specific analog waveforms from two adjacent electrodes would add and subtract in the regions where their fields overlapped, producing uncontrollable and undesirable activation patterns. The feature extraction strategies were inconsistent in extracting speech cues, especially in noisy listening environments. When the background noise consisted of other voices the feature extraction strategies would select features from the background voices as well as the target voice. In contrast to single-channel analog implants, at least these multichannel strategies produced some limited open-set speech recognition (see second two bars in Fig. 6.1).

A breakthrough occurred in the 1990s when signal processing techniques were developed to present biphasic pulses that were interleaved in time across electrodes (Wilson et al., 1991). The use of non-simultaneous biphasic pulses eliminated the problem of electric field summation across electrodes. Brief biphasic pulses are sequentially presented to each electrode. The amplitude of each pulse represents the energy in that frequency band of speech at that moment in time. The overall rate of stimulation was originally about 250 pps/electrode, but newer hardware allows pulse rates of 1000 pps/electrode and higher.

The other breakthrough that occurred at the same time was to abandon feature extraction processing. When the hardware was limited in bandwidth it was felt that some reduction in information rate was necessary. Newer hardware has sufficient bandwidth that the raw information from each signal band is sent to the associated electrode with relatively little modification. The result has been a significant improvement in speech recognition (the 5^{th} and 6^{th} bars in Fig. 6.1). Research has demonstrated that the post-lingual human brain is better at extracting relevant information about speech than even the most reliable pre-processing strategy (a fact that has been widely appreciated in the automatic speech recognition community for many years). Humans are trained at speech pattern recognition over a lifetime, with more than 30 million words received just in the first 4 years of life (Hart and Risley, 1995). The pattern recognition ability of the average human brain for speech is amazing - we can extract the correct words from running speech of talkers with high or low pitch voices, heavy accents, spoken fast or slow, in high noise levels, and even heavily distorted by a bad communication channel. No machine or processing algorithm is even close to human performance. Cochlear implants experienced a dramatic improvement in performance when the processing algorithms simply presented the signals in tonotopic region of the cochlea and "got out of the way" and let the brain do the rest.

The next level of improvements in cochlear implant signal processing depend on achieving a better understanding of the trade-off between signal processing and allowing the over-trained brain pattern recognition system to recognize the patterns of speech. What information is not getting through to the patient in a cochlear implant? Is the limitation in the signal processing or in the damaged nervous system or is it simply an inescapable limitation of prosthetic stimulation? Let us look at the basic perceptual capabilities of cochlear implants and how they are affected by signal processing. In the next section we will review the basic perception of amplitude, timing, and frequency with a cochlear implant.

6.5 Perceptual Issues in Implant Design

6.5.1 *Amplitude cues*

All sensory systems incorporate some degree of amplitude compression at the periphery because the range of physical amplitudes far exceeds the range of neural firing. In hearing the normal ear can code sound levels over a range of 120 dB, which is a range in amplitudes of one million to one. Each auditory neuron can only code sound amplitudes over a range of about 30 dB. The cochlea normally provides a logarithmic compression of sound amplitudes to match the large range of sound amplitude to the smaller range of neural outputs. However, in electrical stimulation we must control the level of neural activity to mimic the normal range of perceptual loudness. If one acoustic sound is twice as loud as another then we need to code those two sounds into electrical amplitudes that will mimic this loudness ratio in the electrically stimulated ear. Studies have shown that loudness is an exponential function of the electric current level, whereas it is a power function of acoustic amplitude (Zeng and Shannon, 1992, 1994, 1999). Most commercial implant devices use a logarithmic compression of acoustic amplitude prior to conversion into electric current amplitude. While a logarithmic compression is not theoretically ideal, it is close enough mathematically to the ideal transformation that it is a reasonable compromise in a prosthetic device.

6.5.2 *Temporal cues*

Temporal cues in hearing are traditionally divided into three classes: envelope cues, periodicity cues, and fine structure (Rosen, 1992; Plomp, 1983). Envelope cues are low-frequency temporal fluctuations (<50 Hz) that are perceived as temporal fluctuations rather than as a distinct pitch. Periodicity cues are temporal fluctuations in the 50-500Hz range that produce a sensation of pitch that is related to the periodicity. Temporal fine structure cues are temporal fluctuations faster than 500 Hz. In the normal acoustic cochlea fine structure temporal information is conveyed by a complex combination of the cochlear place of activation and the temporal structure. Cochlear implants convey envelope cues quite well and CI listeners can generally detect periodicity cues up to 300-500 Hz. However, CI listeners cannot detect temporal fine structure cues. CI devices convey fine structure information by stimulating different electrodes in different cochlear locations. This allows a crude discrimination of pitch by utilizing the type of pitch information that is coded by cochlear location. However there is no indication that CI listeners can make use of temporal cues above 500 Hz strictly in the time domain and CI listeners do not appear to be able to combine timing and cochlear place information for complex pitch as occurs in acoustic hearing.

CI listeners appear to have relatively normal processing of global temporal cues. Global temporal processing is measured by several psychophysical techniques: gap detection measures the smallest detectable silent interval in an ongoing sound, forward masking measures the recovery from adaptation following a sound, and modulation detection measures the ability to detect fluctuations in an ongoing sound. On all of these measures implant listeners are similar in their capabilities to normal hearing, when compared at equivalent loudness levels (Shannon, 1989, 1990, 1992a). Implant listeners can detect gaps of 1-3 ms, have time constants of recovery of 60-100 ms following a stimulus, and can detect modulation of 1-3% up to modulation frequencies of 100-300 Hz. The implication is that these kinds of temporal processing utilize central processing mechanisms that are not adversely affected by deafness or by the lack of fine temporal structure delivered by a cochlear implant.

6.5.3 *Spectral cues: The number of spectral channels*

One of the most important factors in the success of cochlear implants is in the replication of spectral information presented to the correct tonotopic location along the cochlea. Many studies have shown that speech recognition performance improves as the number of spectral channels is increased (Dorman and Loizou, 1997; Dorman et al., 1997; Fishman et al., 1997; Friesen et al., 2001; Fu and Shannon, 1999;

Shannon et al., 2004). In cochlear implant listeners and in simulations of cochlear implants speech recognition improves dramatically as the number of spectral channels is increased from 1 to 8. Adding spectral channels beyond eight continues to improve performance in normal hearing but does not appear to improve performance in implants. It is not clear why implant performance seems to be limited in the number of effective spectral channels, but it is likely to be limited by interactions across electrodes. More spectral channels are needed for speech recognition in noise (Fu et al., 1998; Friesen et al., 2001; Fu and Nogaki, 2005) and for complex pitch and music (Smith et al., 2002; Shannon et al., 2004).

6.5.4 *Spectral cues: Warping and shifting the tonotopic map*

One of the potential problems in cochlear implants is that of presenting spectral information to the wrong tonotopic location in the cochlea. The normal cochlea in humans is 35 mm in length and frequency information is distributed approximately logarithmically along the cochlear length, with high frequencies near the base and low frequencies near the apex (Greenwood, 1990). However, cochlear implant electrode arrays are about 25 mm in length and so cover a region in the cochlea that would normally respond to acoustic frequencies of 500-5000 Hz. A cochlear implant signal processor typically takes acoustic information from 200-10,000 Hz and presents it to the electrodes. This type of processing distorts the normal tonotopic distribution of information: the frequency to cochlear place mapping in the implant is shifted and compressed relative to the normal acoustic mapping. Studies have shown that distortion in the tonotopic map results in a reduction in speech recognition (Shannon et al., 1998; Baskent and Shannon, 2003, 2004). Presumably, the central mechanisms for speech pattern recognition are developed over a lifetime of listening experience and there is a tight association between the correct tonotopic pattern in cochlear place with correct identification of the phoneme. It is not clear how flexible this association is, i.e. how long it might take to relearn the association of new speech patterns produced by the implant. There is some evidence that these patterns can be quickly relearned (Rosen et al., 1999; Fu et al., 2005), but relearning may not be possible if the distortion is too great (Fu et al., 2002).

6.5.5 *Bilateral implants*

Localizing sounds in space is an important ability for locating the source of a sound and also for being able to recognize a single talker in a noisy environment like a restaurant or cocktail party. Normal acoustic listeners can follow conversations in a noisy room when the signal level is 10-20 dB less than the noise level. Implant listeners have generally only had a single implant and so are not able to use the binaural cues to locate sound sources and to understand speech in noisy conditions. Recently, many adults and children have received bilateral implants and the results are mixed (Litovky et al., 2004; Nopp et al., 2004; Schleich et al., 2004; Buss et al., 2008; Grantham et al., 2008). Some patients with bilateral implants can locate sound sources in space, although not as well as normal hearing listeners. Bilateral implants also allow some patients to understand speech in noisier conditions than with a single implant, but again the improvement is not as large as that observed for NH listeners. It appears that approximately one third of bilateral implant recipients can use the two implants better than either implant alone, one third benefit from improved performance by using the head shadow effect (the ear opposite a noise source will have a more favorable signal to noise ratio), and one third derive no additional benefit from two implants. Research into the causes of performance limitations and variations is ongoing.

6.5.6 *Combined acoustic and electric stimulation*

As CI performance improves the selection criteria have been relaxed to allow patients with more and more residual hearing. Some patients have a pattern of hearing loss in which they have good residual hearing at low acoustic frequencies (below 1000 Hz) and essentially no hearing at high frequencies. Cochlear implants are being used to supplement the residual hearing of such patients so that they use acoustic hearing and implant hearing in the same ear (Gantz et al., 2005, 2006; Vermeire et al., 2008). Although the residual hearing may not be sufficient to allow speech recognition by itself, the acoustic hearing often preserves fine spectral resolution and temporal resolution within the region of acoustic hearing. When this residual hearing is combined with electrical stimulation at higher frequency locations in the cochlea these patients are able to recognize speech at a high level, allowing telephone use in most. However, some patients have lost some or all of their residual hearing following implantation and others do no better than a full cochlear implant. It remains to be determined what the most appropriate approach is for such patients. Since hearing is usually present in the contralateral ear, preserving acoustic hearing in the implanted ear may not be necessary.

6.6 Future Directions in Device Design and Signal Processing

6.6.1 *Noise reduction pre-processing*

Speech recognition in real-world listening situations is often limited by noise, produced either by competing talkers or machines. Many companies making consumer electronics (cell phones, speech-to-text transcription, hearing aids) are actively working on techniques for improving the signal to noise ratio at the input to the device. Spectral enhancement methods have shown promise (Yang and Fu, 2005) as have active beamforming microphone arrays (Soede et al, 1993). Some of these methods show undesirable processing artifacts under some conditions, but have shown the potential to improve the real signal-to-noise ratio by about 3 dB.

6.6.2 *Virtual channels*

One factor that appears to limit the capabilities of cochlear implants is the limited number of spectral channels. Modern cochlea implants have 12-22 stimulating electrodes spaced along the tonotopic axis of the cochlea. However, studies have shown that implant patients are actually performing as if they were using only 8-10 channels (Fishman et al., 1997, Friesen et al., 2001) or even as few as 4 (Fu and Nogaki, 2005) when listening in noise. Although there are more stimulating electrodes the effective number of information channels is probably limited by the interaction of electrical fields from adjacent electrodes. If the distance between neurons and electrodes is large relative to the distance between electrodes, then the current fields from adjacent electrodes may be nearly indistinguishable at the point of stimulation. There have been several techniques proposed that would sharpen the stimulating electrical field by controlling the relative phase of stimulation across the electrode array in a manner similar to beamforming or phased array processing. There is now some evidence that such field sharpening can result in an increased number of distinct pitch sensations between adjacent electrodes that could be used as "virtual" channels of stimulation (Donaldson et al., 2006). The initial results of such schemes in clinical speech processors are encouraging.

6.6.3 *Photolithographic electrode arrays*

If the electrode array is in close proximity to the stimulable neurons then the performance of the device may be limited by the number of electrodes. All implant manufacturers are actively investigating modern photolithographic techniques for increasing the number of electrodes in a cochlear implant array. Electrode stimulating surfaces can be printed on a flexible polyimide substrate and overcoated with an insulating layer of parylene. Electrodes have also been designed with distributed stress gauges along the length so the position and force could be monitored during surgical insertion (Wang and Wise, 2008). The resulting unit could be inserted into the cochlea directly or mounted on a silicone carrier, much like present implant technology. However the high electrode counts, up to hundreds of stimulation sites would require sophisticated multiplexing telemetry to send and receive signals from outside the skin.

6.6.4 *Optimizing parameters*

As hearing aids and cochlear implants add more electrodes and evolve more complex signal processing algorithms, the number of parameters increases dramatically. For example, each stimulated electrode in a cochlear implant has electrical pulse parameters (pulse phase duration, inter-phase gap duration, stimulation rate, amplitude, temporal interleave with adjacent electrodes), perceptual parameters (threshold, maximum loudness level, loudness growth function), as well as parameters for the signal processing channel that feeds that electrode (spectral bandwidth, cutoff frequencies, filter slopes, amplitude compression etc.). As the number of electrodes increases from 16 to 120 the number of parameters becomes unmanageable. Some of these parameters can be fixed across electrodes, but some must be determined for each individual patient. Genetic algorithms have shown promise for achieving rapid optimization in this large parameter space (Wakefield et al, 2005; Baskent et al., 2007) and will likely be implemented in the near future to reduce the time required to fit a complex device to an individual patient.

6.6.5 *Totally implantable CI*

Research is progressing on a totally implantable cochlear implant device that would not require the use of an external processor, microphone, or transmitter coil. This obviously would have tremendous practical appeal in that it would not have any outwardly visible parts and it would allow activities like swimming. Periodic charging of the device would occur transcutaneously, using a charging module (Cohen, 2004). This technology would also require the implantation of a microphone (Maniglia et al., 1999). All major implant companies are actively developing the technologies that would allow totally implantable devices.

6.7 Summary and Conclusions

Cochlear implants and brainstem implants are highly successful sensory prostheses, exceeding all of our expectations. People who lose their hearing after acquiring language have an excellent chance at regaining a high degree of functional hearing from electronic implants. Advances in engineering technology are making the devices smaller, faster and have higher signal processing capability. The limitations in device improvements are probably not on the engineering side but in terms of matching the device design to the neurobiology of the damaged sensory system. As performance improves we learn more about the potential benefits and limitations of artificial hearing and more about auditory neuroscience. Each advance in neuroscience leads to engineering improvements in the next generation of devices. Although the level of performance with auditory prostheses is much higher that originally imagined, there is reason to think that we will see continued improvement in hearing quality from these devices in the future.

6.8 References

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6.9 Review Questions

Q6.1 A cochlear implant restores perfect hearing. True or False.

Q6.2 People with a cochlear implant can understand speech in quiet but have difficulty in noisy rooms. True or False.

Q6.3 Cochlear implants and auditory brainstem implants provide similar levels of speech understanding. True or False.

Q6.4 Prosthetic electric stimulation of the auditory nervous system has been tried in humans with electrodes positioned in:

A. the cochlea

B. the cochlear nucleus in the brainstem

C. the inferior colliculus in the midbrain

D. the auditory cortex

E. all of the above

F. only A, B, and C

Q6.5 Because cochlear implants have a limited bandwidth it is necessary to extract only the most important features of speech and present a simplified version of the speech signal to the implant. True or False.

Q6.6 Cochlear implants are only for the totally deaf. Anyone with even the slightest amount of residual acoustic hearing should not get an implant. True or False.

Q6.7 Single channel implants present the full bandwidth acoustic signal in electric form to the electrode. Thus single channel implants can provide speech recognition performance that is as good as that of a multichannel implant. True or False.