NIH Public Access
Author Manuscript
IEEE Rev Biomed Eng. Author manuscript; available in PMC 2009 November 25.

Published in final edited form as:
IEEE Rev Biomed Eng. 2008 January 1; 1: 115–142. doi:10.1109/RBME.2008.2008250.

Cochlear Implants:System Design, Integration and Evaluation

Fan-Gang Zeng,
Departments of Anatomy and Neurobiology, Biomedical Engineering, Cognitive Sciences and Otolaryngology - Head and Neck Surgery, University of California, 364 Med Surg II, Irvine, CA 92697, USA (fzeng@uci.edu)

Stephen Rebscher,
Department of Otolaryngology - Head and Neck Surgery, University of California, San Francisco, CA 94143, USA

William V. Harrison,
Silere Medical Technology, Inc., Kirkland, WA 98034, USA

Xiaoan Sun, and
Nurotron Biotechnology Inc., Irvine, CA 92618, USA

Haihong Feng
Institute of Acoustics, Academia Sinica, Beijing 100190, China

Abstract

As the most successful neural prosthesis, cochlear implants have provided partial hearing to more than 120,000 persons worldwide; half of which being pediatric users who are able to develop nearly normal language. Biomedical engineers have played a central role in the design, integration and evaluation of the cochlear implant system, but the overall success is a result of collaborative work with physiologists, psychologists, physicians, educators, and entrepreneurs. This review presents broad yet in-depth academic and industrial perspectives on the underlying research and ongoing development of cochlear implants. The introduction accounts for major events and advances in cochlear implants, including dynamic interplays among engineers, scientists, physicians, and policy makers. The review takes a system approach to address critical issues from design and specifications to integration and evaluation. First, the cochlear implant system design and specifications are laid out. Second, the design goals, principles, and methods of the subsystem components are identified from the external speech processor and radio frequency transmission link to the internal receiver, stimulator and electrode arrays. Third, system integration and functional evaluation are presented with respect to safety, reliability, and challenges facing the present and future cochlear implant designers and users. Finally, issues beyond cochlear implants are discussed to address treatment options for the entire spectrum of hearing impairment as well as to use the cochlear implant as a model to design and evaluate other similar neural prostheses such as vestibular and retinal implants.

Index Terms

Auditory prosthesis; electric stimulation; auditory nerve; auditory brainstem; signal processing; radio frequency; current source; hermetic sealing; electrode; loudness; pitch; temporal resolution; fine structure; speech recognition; music perception; biocompatibility; biomaterials; safety

Correspondence to: Fan-Gang Zeng.

NIH-PA Author Manuscript
NIH-PA Author Manuscript
NIH-PA Author Manuscript

NIH-PA Author Manuscript
NIH-PA Author Manuscript
NIH-PA Author Manuscript

I. Introduction

Using electric currents to directly stimulate the auditory nerve in a totally deafened person, the cochlear implant is the only means available at present to penetrate the "inhuman silence which separates and estranges", as stated by the famous blind-deaf writer, lecturer, and social advocate, Helen Keller (1880-1968). The development of the cochlear implant has a long, distinguished, and interesting history, including dynamic interplays between engineers and physicians as well as an intricate balance between experimentation and ethics (Figure 1). The cochlear implant today has not only provided useful hearing to more than 120,000 deaf persons, but also become a multi-billion-dollar industry, attracting discussion in the MBA classroom, and attention in popular culture from the Oscar nominated documentary film *Sound and Fury* to TV shows such as *ER, The Young and The Restless*, and *Cold Case*.

The journey started with a "crackling and boiling" sensation, when the Italian scientist Alessandro Volta (1745-1827) placed the two ends of a 50-volt battery to his ears more than two centuries ago [1]. The unpleasant sensation was likely the first demonstration that electric stimulation, instead of sound and light, can induce auditory and visual sensations. In honor of Volta's work in electricity, an important electrical unit, the volt (V), was named after him. In addition, Napoleon made him a Count and established the annual Volta prize.

In 1880, the most notable Volta prize went to Alexander Graham Bell (1847-1922) who received 50,000 francs for his invention of the telephone. Bell used the prize to help hearing-impaired people including his deaf wife and Heller Keller, who devoted her first book "*The story of my life*" to him. Although it is hard to tell whether Bell had envisioned it, there is no question that Bell's enterprise, the Research Laboratories of the Bell Telephone System, conducted early comprehensive research in hearing and speech that forms the theoretical foundation needed for later success of the cochlear implant [2]. For example, Bell Labs' vocoders that describe how to break up and then reassemble speech [3,4] have been absolutely critical to the development of signal processing in modern cochlear implant systems [5-7].

Over the next 150 years, a few brave ones repeated Volta's heroic experiment and all obtained the same disagreeable sensation as a result of the direct current (DC) stimulation [8,9]. Recognizing the danger of DC stimulation and armed with vacuum-tube-based oscillators and amplifiers, Harvard researchers S.S. Stevens and his colleagues used AC (sinusoidal) electric stimulation to identify three mechanisms underlying the "electrophonic perception" in the 1930s [10-12]. The first mechanism was related to the eardrum's conversion of the electric signal into an acoustic signal, resulting in a tonal pitch perception but at the doubled signal frequency. The second mechanism was related to the "electromechanical effect," in which electric stimulation causes the hair cells in the cochlea to vibrate, resulting in a perceived tonal pitch at the signal frequency as if it were acoustically stimulated. Only the third mechanism was related to direct electric activation of the auditory nerve, as the subjects reported noise-like sensation in response to sinusoidal electrical stimulation, much steeper loudness growth with electric currents, and occasionally activation of non-auditory facial nerves.

In the middle of last century, physicians became the driving force to translate these early research efforts into clinical practice. In 1957, A French physician, Djourno and his colleagues reported successful hearing using electric stimulation in two totally deafened patients [13-15]. Their successful story crossed the Atlantic, spurred an intensive level of activity in attempts to restore hearing to deaf people, mostly on the U.S. west coast in the 1960s and 1970s [16-19]. By one historian's account [20], William House in Los Angeles in 1961 implanted a gold electrode insulated with silicone rubber in the scala tympani of two deaf patients. He used a low-frequency square wave (40-200 Hz) as the carrier of electric simulation, whose amplitude was modulated by the sound. House's initial device only lasted two weeks but both patients

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

reported useful hearing with electric stimulation. In 1964, Blair Simmons at Stanford implanted a cluster of six stainless-steel electrodes into the auditory nerve through the modiolus in a 60-year-old man with profound hearing loss. In 1971, Robin Michelson in San Francisco implanted a form-fitting (from human temporal bones) single-channel electrode pair in four deaf patients. In 1978, Graeme Clark in Australia developed a 20-electrode (Platinum rings) cochlear implant system and implanted in two deaf patients. Other similar experimental efforts included Chouard in France [21], Eddington in Utah [22], and Horchmair in Austria [23].

However, the efficacy of these early cochlear implants, particularly the single-electrode systems, met with strong suspicion and perhaps resistance from the mainstream scientific community in 1970s. Compared with the three thousand tuned inner hair cells in a normal cochlea, a single electrode cannot provide the level of tuning and timing that resembles the normal pattern of neural activity produced by acoustic stimulation. For example, Nelson Kiang, a prominent physiologist at Harvard and MIT, suggested that little or no discriminative hearing could be generated from a single-electrode cochlear implant [20,[24]]. Even the National Institutes of Health (NIH) condemned human implantation as being morally and scientifically unacceptable in the early 1970s [25]. At that time, any researchers who became involved in cochlear implants did so at their own professional risk. To settle the scientific issues, Michael Merzenich and colleagues organized the First International Conference on Electrical Stimulation of the Acoustic Nerve as a Treatment for Profound Sensorineural Deafness in Man in San Francisco in 1973. An outcome of this conference was, as shown in Figure 2, an intensified research effort in cochlear implants, particularly using animal experiments, in the mid 1970s to 1980s. To settle the safety and efficacy issues, NIH, in 1975, commissioned Bilger and colleagues at the University of Pittsburgh to evaluate objectively and independently the audiological performance in the world's first group of single-electrode cochlear implant recipients, including 11 implanted by House and 2 by Michelson [26-[31]]. Bilger's report confirmed that the single-electrode cochlear implants provided useful hearing in terms of aiding lipread and identifying common environmental sounds, but these devices could not provide open-set speech recognition. To a large extent, the San Francisco meeting and Bilger report legitimized the cochlear implant as an acceptable and valid clinical practice.

Next, the race to commercialize the cochlear implant had just begun because the technology for commercialization was ripe. The development of cardiac pacemakers helped identify biocompatible materials, design insulated electrodes and set safe electric stimulation limits. Thanks to the cold war, advances in the space industry provided crucial technology in integrated circuits and hermetical sealing, with titanium encapsulation still being the standard for today's cardiac pacemakers and cochlear implants. One industrial giant, 3M, became interested, initiating contact with the simpler and safer House single-electrode device over the more complicated Melbourne multi-electrode device in 1978. Indeed, the 3M/House single-electrode implant was the first to win FDA approval in 1984 and became the industrial leader with several hundred users in the mid 1980s (see Insert in Fig. 2).

On the other hand, supported by a grant from the Australian Department of Productivity, the University of Melbourne and Nucleus Limited (a medical device company focusing on pacemakers) entered a public and private cooperative agreement in 1979 to manufacture and market the 22-electrode cochlear implant. In middle 1980s, NIH also helped speed up the acceptance of multi-electrode cochlear implants by funding the UCSF and the University of Melbourne device development (1RO1-NS21027) and hosting the first consensus conference concluding that "multichannel implants may have some superior features in adults when compared with the single-channel type" [33]. Time proved that the multi-electrode devices indeed not only produced much superior performance over the single-electrode devices (Fig. 3), but also eventually phased out the single-electrode devices in the market (Fig. 2).

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

In addition to the Nucleus device, several other multi-electrode devices were also developed. The University of Utah developed a six-electrode implant with a percutaneous plug interface [22,34] and was called either the Ineraid or Symbion device in literature, which was uniquely suited for research purposes [35-38]. The University of Antwerp in Belgium developed the Laura device that could deliver either 8-channel bipolar or 15-channel monopolar stimulation. These devices were later bought out and are no longer available commercially. The MXM laboratories in France also developed a 15-channel monopolar device, the Digisonic MX20, which is marketed by Neurelec (www.neurelec.com).

At present, there are three major cochlear implant manufacturers including Advanced Bionics Corporation, USA (www.advancedbionics.com), Med-El Corporation, Austria (www.medel.com), and Cochlear Corporation, Australia (www.cochlear.com), with Cochlear being the dominating player controlling 70-80% of the cochlear implant market worldwide. Several startup companies are also developing advanced and low-cost multi-electrode cochlear implants, including Advanced Cochlear Systems (www.advcoch.com) in Seattle, Washington, Nurobiosys Corporation in Seoul, Korea [39], and Nurotron Biotechnology Inc. based in both Irvine, CA and Hangzhou, China (www.nurotron.com).

Figure 3 summarizes sentence recognition scores in quiet by 4 different devices over 27 years. The sentence recognition task is chosen because it can best measure the user's ability to communicate in daily life, e.g., a 70% sentence recognition score would support a telephone conversation [40]. Starting out with the single-electrode device providing little or no open-set speech recognition in 1980, it took 15 years and the collective effort from academia and industry to achieve this level of performance, allowing an average multi-electrode cochlear implant user to converse on the telephone. It appears that speech recognition in quiet has reached a plateau in the last 10 years, with much effort being focused on improving speech recognition in noise, music perception and tonal language understanding, all of which are currently difficult tasks for implant users (see Section IX for details).

In the remainder of this paper, system design and specifications will be laid out first in terms of the overall system architecture and functions (Section II). Then the subsystem components from speech processors to electrode arrays and fitting programs will be presented and analyzed in depth (Sections III-VII). System integration will be discussed in Section VIII and functional evaluation will be presented in Section IX. The paper will end by addressing issues beyond cochlear implants in Section X.

II. System Design and Specification

The over-arching goal of a cochlear implant is to use electric stimulation safely to provide or restore functional hearing. Figure 4 shows graphically a typical modern cochlear implant system. The behind-the-ear external processor with ear hook and a battery case (2) uses a microphone to pick up sound, converts the sound into a digital signal, processed and encodes the digital signal into a radio frequency (RF) signal, and send it to the antenna inside a headpiece (3). The headpiece is held in place by a magnet attracted to an internal receiver (4) placed under the skin behind the ear. A hermetically sealed stimulator (5) contains active electronic circuits that derive power from the RF signal, decode the signal, convert it into electric currents, and send them along wires (6) threaded into the cochlea. The electrodes (7) at the end of the wire stimulate the auditory nerve (8) connected to the central nervous system, where the electrical impulses are interpreted as sound.

All modern cochlear implant systems have the following architecture and functional blocks (Figure 5). An external unit, also known as the speech processor, consists of a digital signal processing (DSP) unit, a power amplifier, and an RF transmitter. The DSP is the brain of the

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

cochlear implant system that receives sound, extracts features in the sound, and converts the features into a stream of bits that can be transmitted by the RF link. The DSP also contains memory units or “maps” that store patient specific information. The maps and other speech processing parameters can be set or modified by a PC fitting program.

An internal unit consists of the RF receiver and a hermetically sealed stimulator. Because the internal unit has no battery, the stimulator must first derive power from the RF signal. The charged up stimulator will then decode the RF bit stream and convert it into electric currents to be delivered to appropriate electrodes. All modern systems also contained a feedback loop that can monitor critical electric and neural activities in the implants and transmit these activities back to the external unit.

Table I summarizes system and functional specifications of the latest cochlear implant from the three major manufacturers. Over the last 25 years, there seems to be a convergence of technology in terms of system specifications. For example, the input dynamic range (IDR) was set at 30 dB in the early Nucleus 22 device, but has now been increased to be 75-80 dB with the default value of 45-60 dB in the latest devices to match the range of amplitude variations of natural speech and environmental sounds [48-50]. Similarly, the frequency range has broadened to include components lower than 300 Hz to take advantage of the temporal pitch code in an attempt to improve pitch and tonal language perception [51-53]. The latest devices all contain the standard CIS plus other proprietary multiple speech processing strategies. The Nucleus device is slightly ahead in adopting sound field processing, a directional microphone and other cosmetic (e.g., water resistant) technologies.

The Nucleus device also has the longest history and the best reliability record, but it appears to lag behind in the internal unit design and technology. The latest Nucleus Freedom system is still the slowest in terms of RF transmission frequency and data rate, hence the lowest overall stimulation rate. It has kept to its original design from the 1980s with only one current source, unable to provide simultaneous stimulation and electrical field imaging. Detailed presentation and analysis of these subsystem components, functions, and specifications will be provided next.

III. Signal processing

Except for the paradigm shift from the single electrode device to the multi-electrode device in the early 1980s, advances in signal processing are largely responsible for the continuous and steady improvement by cochlear implant users. Because several review papers have been published to focus on this topic [40,54-56], this section provides a brief overall review and an update.

The theoretical basis for signal processing in cochlear implants can be traced to early research in the source-filter model in speech production [57] and the vocoders in telephone communication [3,4]. Briefly, speech sounds can be modeled as either a periodical (for voiced sounds) or noise (for unvoiced sounds) source whose frequency spectrum is filtered by the resonance properties in the vocal tract. Alternatively, the source can be modeled as a carrier while the vocal tract acts as a modulator, reflecting the opening and closing of the mouth or the nose. Generally speaking, the source varies rapidly whereas the filters vary more slowly. Recently, Zeng has argued for a general model in which the rapidly varying fine structure contributes mainly to auditory object formation whereas the slowly varying envelope contributes to speech intelligibility [58].

Figure 6 classifies signal processing used in modern cochlear implants. Most cochlear implants discard the fine structure and encode the coarse features only. The first generation multi-electrode Nucleus 22 device extracted the fundamental frequency (F0), which is a source

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

information reflecting voice pitch and the second resonance frequency in the spectral envelope (also known as the second formant, or F2) [59]. In later versions of the implant, the first formant was added [60], followed by additional three spectral peaks between 2000 and 8000 Hz [61, 62]. Consistent improvement in speech recognition was observed as more spectral details were added (see Fig. 3).

In the late 1980s and early 1990s, another paradigm shift occurred from spectral envelopes to temporal envelopes, showing that the temporal envelope from a very limited number of spectral channels can support a high level of speech recognition [5,63,64]. Parallel changes occurred in cochlear implant signal processing, from explicit encoding of spectral envelope cues to explicit encoding of temporal envelope cues [6,65].

Figure 7A shows the functional block diagram of the continuous-interleaved-sampling (CIS) strategy, which has been implemented by all major manufacturers and is still available in their latest devices (see Table 1). The sound is first subject to a number of bandpass filters with the number being as few as 5 in the original CIS implementation [6] and as many as 20 in the Nucleus Freedom device [46]. The temporal envelope from each band is extracted by either half-wave (shown in the figure) or full-wave rectification followed by a low-pass filter; or more recently by the Hilbert transform [47, 66]. The envelope is then logarithmically compressed to match the widely varying acoustic amplitudes to the narrow electric dynamic range [50, 67]. The compressed envelope amplitude modulates a fixed-rate biphasic carrier, whose rate can vary from several hundreds to several thousands per second. To avoid simultaneous electrical field interference, a problem that apparently bothered early devices such as the analog Ineraid implant [22], the biphasic carriers are time interleaved between the bands so that no simultaneous stimulation occurs between the bands at any time. In practice, a single current source is needed in a CIS strategy. The CIS strategy can avoid simultaneous channel interaction while preserving the temporal envelope samples for each band as long as the carrier rate is sufficiently high. Typically the cutoff-frequency of the low-pass envelope filter is set between 400 Hz or slightly lower, requiring at least 800 Hz carrier for faithful representation of these envelopes [6].

Figure 7B shows a functional block diagram of the “n-of-m” strategy, which was first described by Wilson and colleagues [68] and had been refined in subsequential development as the SPEAK and ACE strategies in the Nucleus devices [46, 65]. The pre-processing in the n-of-m strategy is similar to the CIS strategy, including the bandpass filters and the envelope extraction blocks. However, there are several major differences between the two strategies. One difference is that the n-of-m strategy has a greater number of bandpass filters, e.g., m=22 in the Nucleus implementation, than the CIS strategy. The number of bandpass filters is typically set to equal the number of intra-cochlear electrodes. The second difference is that the n-of-m strategy is based on temporal frames, typically lasting 2.5 to 4 msec, whereas the CIS strategy does not have any explicit processing frames. In each frame of the n-of-m strategy, an “n” number of bands with the largest envelope amplitude are selected (by definition, $n \leq m$). Envelopes from the selected bands are subject to the same amplitude compression and used to determine the current level of the biphasic pulse. The biphasic pulses are interleaved between the output channels, with the per channel stimulation rate being determined by the frame rate. Finally, only the corresponding “n” electrodes (dark bands in the figure) out of the “m” electrodes are stimulated in a particular frame. The SPEAK strategy selects 6-8 largest peaks and has a fixed 250 Hz per channel rate. The ACE strategy has a larger range of peak selection and higher rate than the SPEAK strategy. If n=m, then the SPEAK and ACE strategies are essentially same as the CIS strategy.

Figure 8 shows the spectrogram of a sample speech (i.e., acoustic input) and electrodograms of the same speech by two different processing strategies (i.e., electric output). The spectrogram

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

is a pseudo 3-dimensional plot, showing frequency (y-axis) and intensity (color scale) changes as a function of time (x-axis). The horizontal lines in the voiced portion of the speech represent harmonics, with the bottom line representing the fundamental frequency. The energy concentration and movements represent formant frequency and its transitions. Electrodograms are the electric parallel to spectrograms, displaying electric stimulation information such as electrode, amplitude and timing as a function of time [69]. Electrodograms are an effective and valuable tool to visualize and evaluate the effects of electric stimulation parameters and signal processing on cochlear implant performance [70-71].

Panel B shows a 6-channel CIS processor output, in which only the most apical 6 electrodes are used. Each electrode carries slowly varying envelope information, with the energy difference between electrodes carrying the spectral information. For example, the apical electrodes (4-6) are stimulated at a higher intensity than the basal electrodes (1-3) at 500 ms, when the sound is a vowel (/a/ as in large); whereas only the most 3 basal electrodes (1-3) are stimulated at 750 ms, when the sound is a consonant (/s/ as in size). The insert in panel B shows detailed timing information between 900 and 910 ms, during which the interleaved pulses between electrodes can be seen as staggered lines at 1.25 ms interval (or 800 Hz per electrode stimulation rate). Note that the electric stimulation pattern does not contain any spectral or temporal fine structure (e.g., fundamental frequency, harmonics, and formant transition) in the original sound.

Panel C shows the electrodogram of an 8 of 20 (n of m) SPEAK strategy, which has been implemented in the Nucleus 24 system [65]. In the SPEAK strategy, the number of bandpass filters can be 16 or 20 (m). Every 4 ms, 6 or more bands (n) with the highest energy are selected and the corresponding electrodes are stimulated. Compared with the 6-channel CIS strategy, the 8 of 20 SPEAK strategy produces more spectral details but less temporal details because of the much lower per channel stimulation rate at 250 Hz. With the faster Nucleus Freedom system, the per-channel stimulation rate can be increased to 2500 Hz to preserve greater temporal details. Cochlear has termed this “faster” SPEAK strategy as the advanced combination encoder or the ACE strategy [46].

At present, signal processing focuses on how to encode spectral and temporal fine structure cues in cochlear implants. To encode the spectral fine structure, more independent electrodes are needed. Given the current electrode manufacturing technology and the placement of these electrodes in the cochlea (see Section VI for details), it is difficult to increase the number of physical electrodes. Instead, several innovative signal processing techniques are being investigated to increase the spectral resolution using focused stimulation and the number of functional channels using virtual channels [72]. Recently, encoding of the temporal fine structure cue has received much attention [73-78]. One way to encode the fine structure is to increase the electric stimulation carrier rate so that the temporal fine structure cue can be represented in the waveform domain, e.g., Med El’s FSP processor [47]. A second way is to extract frequency modulation from the temporal fine structure and then use it to frequency modulate the carrier rate [79]. A third way is to use multiple carriers to encode the fine frequency structure [80]. The effectiveness of these new strategies has not been demonstrated in actual cochlear implant users.

IV. Radio Frequency (RF) Link

To ensure safety and to improve convenience, the internal unit in all modern devices is now connected to the external unit via a transcutaneous RF link. The RF link uses a pair of inductively coupled coils to transmit both power and data. The RF transmission has to address a host of challenging technical issues [81-83]. For example, the external unit needs to provide not only reliable communication protocols including a signal modulation method, bit coding,

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

frame coding, synchronization and back telemetry detection, but also high efficiency RF power amplifier and immunity to electromagnetic interference (EMI). The internal unit, on the other hand, needs to harvest power with high efficiency and retrieve data with high accuracy. In addition, the size of the transmitting and receiving coils needs to be minimized and cosmetically appealing. Table II describes RF characteristics of the 3 major cochlear implant manufacturers. The remainder of this section focuses on bit coding, frame coding, and power management.

A. Bit Coding

The output of the external unit is a digital stream of 1s and 0s. Prior to sending these bits to the RF power amplifier, bit coding is usually required for reliable and accurate wireless transmission and decoding. At present, the major cochlear implant manufacturers, at least in their forward transmission systems, all use the amplitude shift keying (ASK) modulation for RF transmission. Figure 9 shows a two-layered bit coding ASK scheme implemented in the Nucleus Freedom system [81]. In the first layer, 1 is coded by 5 ON cycles of the 5MHz carrier frequency, i.e., RF is present for 5 cycles; 0 is by 5 OFF cycles of the 5 MHz carrier frequency, i.e., RF not present for 5 cycles. In the present example, the amplitude modulated RF signal gives a 111011110111 bit pattern, displayed right underneath the waveform. In the second layer, the raw bit signal is grouped into 6 bits to encode a 3-bit data token. In this example, the raw bit signal “111011” represents a data token “110”, whereas “110111” represents a data token “010”. There are a total of 8 data tokens (2^3=8), plus 2 synchronization tokens and 2 error tokens that are defined by the same 6-bit patterns, for a total of 12 tokens. The 12 tokens are only a subset of the available 64 patterns (2^6=64), with the rest being unused. This redundancy is critical for the system to detect transmission errors. In case the decoder produces a pattern that is not part of the 10 data tokens, or the 2 synchronization tokens, or the 2 error tokens, then the system detects it as an error.

B. Frame Coding

The RF link uses a frame or packet coding scheme to transmit specific stimulus parameters to the internal stimulator. In the Nucleus system, the parameters include pulse amplitude, pulse duration, pulse gap, active electrode and return electrode that are used to define a biphasic pulse and the stimulation mode. Dependent on the timing relationship between a frame and the pulses it generates, frame coding schemes can be classified by either the expanded mode or the embedded mode.

The expanded mode was first used by Cochlear in the Nucleus 22 device with a carrier frequency of 2.5 MHz and later in the Nucleus 24 device with a carrier frequency of 5 MHz [81,[84]]. Figure 10 shows the expanded frame coding used in the Nucleus 24 system, in which a frame consists of a SYNC burst and an additional 5 bursts that specify a legal biphasic pulse [85]. The SYNC burst is short, containing no more than 7 RF clock cycles.

The number of RF cycles (n) within each burst is multiples of 8 cycles and conveys the information needed to specify the electric biphasic pulse. The active electrode number is determined by (n-4)/8, where 12≤n≤180, and ranges between 1 and 22. The stimulation mode is determined by (n-4)/8, where 12≤n≤244, and ranges between 1 and 30. When the stimulation mode returns a number between 1 and 22, it specifies bipolar (BP) configuration with the return electrode being in the cochlea (however, the active and return electrodes cannot be the same). On other hand, monopolar stimulation is specified by a stimulation mode number of 24 with the reference electrode being a ball electrode placed under the temporalis muscle (MP1), 25 being a plate electrode on the package (MP2), and 30 being both the plate and the ball electrodes (MP1+2). The pulse amplitude is coded by 271-n, where16≤n≤271, resulting in 256 discrete clinical units from 255 to 0. The pulse duration of phase 1 is determined by the duration of the Phase 1 burst, with the number of RF cycles from 18 to 3,00, or 3.6 to 600 μS. The phase delay

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

determines the interval between the negative-going phase and the positive-going phase and can range from 6 to 50,000 RF cycles, or 1.2 to 10,000 μS. The polarity of phase 2 is opposite to that of phase 1, but their durations have to match for balanced charge. The residual inter-pulse-interval (RIPI) or inter-frame gap is inserted to produce the designed stimulation rate. The number of the RF cycles for the RIPI can range from 6 to 1,250,000, equivalent to 1.2 μS to 250 mS. This 250 mS upper limit is determined by the requirement of at least a 4 Hz pulse rate to keep the internal circuit powered up. Note that critical parameters such as active electrode and mode are encoded by multiples of eight RF cycles, which allows theoretically a counting error up to 4 RF cycles for reliable transmission.

The Expanded mode requires a relatively simply decoder on the receiver side. However, it has several limitations. First, the maximal total stimulation rate is low because no stimulation is generated when parameters such as Sync, Electrode, Mode, Amplitude, and RIPI bursts are being transmitted. Furthermore, the RIPI is not a constant between frames even at a constant stimulation rate, because it is affected by other stimulation parameters such as electrode, mode, and amplitude. Finally, the amplitude and pulse duration parameters are prone to RF cycle detection errors, which may lead to an unbalanced charge.

To overcome the limitations of the expanded mode, the embedded mode frame coding scheme was developed and has become the de-facto standard for the current cochlear implants [86]. Figure 11 illustrates schematically the embedded mode frame coding scheme. The basic idea is to transmit the information regarding electrode, mode, and amplitude (E, M & A) for the next biphasic stimulus, shown as Stim(N+1), while the present stimulus, shown as Stim(N), is being delivered. Another advantage of the embedded mode is that there is a period of time between the end of the present stimulus and the start of the next stimulus for the internal circuitry to check the validity of stimulation parameters, E, M and A. In case of errors, the stimulus can be stopped before it is actually delivered.

C. Maximum Total Stimulation Rate

The maximum total stimulation rate depends upon the bit rate and the frame rate in the RF transmission link. Using 5 RF cycles to encode the raw bits and 6 raw bits to encode 3 actual data bits, the bit rate is 250 kBits/second in the Nucleus 22 system and 500 kBits/second in the Nucleus 24 system. The Nucleus 22 uses only the expanded mode to encode frames and has a theoretically maximal total stimulation rate of 5,900 Hz. In practice, this theoretical rate is not attainable. For realistic stimulation parameters such as pulse duration of 100 μS/phase, 30 μS phase delay, electrode 1 and 2 bipolar mode with the highest amplitude, the maximum rate is just above 3,000 Hz [84]. The Nucleus 24 system supports both the expanded mode and the embedded mode. For example, with pulse duration of 12 μS/phase, the Nucleus 24 (CI24M) system can produce the maximum pulse rate of 8,500 Hz using the expanded mode and 14,400 Hz using the embedded mode. To further increase the maximum total stimulation rate, the electrode mode and pulse duration information can be set initially with only the electrode number and amplitude information being transmitted on a pulse by pulse basis. The Nucleus Freedom system (CI24R and CI24RE) adopts this high rate mode and can attain a maximum total rate of 32,000 Hz [46].

Both the Advanced Bionics and Med El systems have used higher RF and multiple current sources to achieve higher maximum total stimulation rate than the Cochlear system. The higher RF leads to higher bit rate in the RF transmission (see Table 2). The multiple current sources allow one-to-one mapping between the current source and the electrode, eliminating the need to transmit the electrode information. The HiRes 90K device has 16 current sources corresponding to 16 electrodes and is capable of producing a maximum total stimulation of 83,000 Hz. Sonata has 12 current sources corresponding to 12 electrodes and is capable of producing a maximum total stimulation rate of 50,700 Hz pulses. Although there has been a

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

trend for all manufacturers to push for higher stimulation rates, there is little or no scientific evidence suggesting that higher rate produces better performance [87-89].

D. Power Transmission

To extend battery life, high power efficiency needs to be achieved in the RF transmission link. A highly efficient Class-E power amplifier is typically used in the present cochlear implants [90,91]. Properly designed transmitting and the receiving coils are another critical component to determine the power efficiency in the RF link. There are many conflicting requirements in the RF design. For example, the power transmission efficiency is maximized if the RF system works at its resonance frequency, i.e., a narrow bandwidth. However, the data transmission requires the RF system to have an unlimited bandwidth. Moreover the highly efficiency Class-E power amplifier is also highly non-linear, with its distorted waveform limiting the data transmission rate. Another example is the conflicting requirement in coil size: The large the coil size, the higher the transmission efficiency; on the other hand, the coil size has to be limited by the head size and cosmetic considerations. In general, the RF power amplifier and the coils need to be designed and integrated to take into account of the overall system power consumption, variations in skin thickness, and forward and backward data transmission [92, 93]. At present, the RF link has about 40% transmission efficiency, delivering 20-40 mW power to the internal unit over the skin thickness from 4 to 10 mm.

V. Receiver and stimulator

The internal unit consists of a receiver and a stimulator, and is sometimes referred as the “engine” of a cochlear implant. Figure 12 shows the block diagram of a typical implanted receiver and stimulator [81, 94-96]. The centerpiece is an ASIC (Application Specific Integrated Circuit) chip, shown as the dotted box, which performs critical function of ensuring safe and reliable electric stimulation. Inside the ASIC chip, there are a forward pathway, a backward pathway, and control units. The forward pathway usually includes a data decoder that recovers digital information from the RF signal, error and safety check that ensures proper decoding, a data distributor that sends the decoded electric stimulation parameters to the right place (i.e., the programmable current source) at the right time (i.e., by switching on and off multiplexers). The backward pathway usually includes a back telemetry voltage sampler that reads the voltage over a period of time on the recording electrode. The voltage is then amplified by the programmable gain amplifier (PGA), converted into digital form by an analog to digital converter (ADC), and stored in memory to be sent out to the external unit via back telemetry. The ASIC chip also includes many control units from the clock generated from the RF signal to the command decoder. There are several circuits and devices that cannot be easily integrated in the ASIC chip, including the voltage regulator, the power generator, the coil and RF tuning tank, and the back telemetry data modulator. The following highlights the design, implementation, and function of several important components.

A. Safety Measures

Stimulation safety is a top priority in the receiver and stimulator design. Under no circumstance should harmful electrical stimulation such as over stimulation or unbalanced stimulation be delivered to the cochlea. Additional considerations are needed to prevent erratic functions of the receiver stimulator in cases of unpredictable events, such as drop of the headpiece, strong electromagnetic interference, and malfunction of the external DSP unit. Several levels of safety check are commonly implemented in current cochlear implants, including:

- Parity check to detect bit error from either RF transmission or data decoding.
- Stimulation parameter check to ensure the validity of electrode number, mode, amplitude, pulse duration, and inter-pulse gap.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

- Maximum charge check to prevent over stimulation, with the charge density being typically less than 15 to 65 μC/cm2/phase dependent on the electrode material, size and shape [97,[98]].
- Charge balance check to prevent unbalanced stimulation and DC stimulation because they generate gases, toxic oxychlorides, corrosion products, and associated pH changes that can cause tissue damage [99].
- To prevent DC stimulation, capacitors are serially connected to the electrodes to block any unbalanced charge being delivered to electrodes. All modern devices have used this method.
- To prevent accumulated unbalanced charges, especially with high-rate stimulation, Nucleus cochlear implants short all stimulating electrodes between pulses [100].

B. Current Source

The current source generates a stimulating current according to the amplitude information from the data decoder. It usually consists of a digital to analog converter (DAC) and current mirrors. The design of an accurate current source is demanding. In the Nucleus 22 device, the amount of current in the drain of a MOSFET is controlled by the voltage difference between the gate and the source of the MOSFET. Because the relationship between the drain and the source is not constant due to the process variation of integrated circuit fabrication, a trimmer network was needed to fine tune the reference current in the Nucleus 22 device. Recent devices have abandoned this technique, instead they combine multiple DACs to obtain the desired amount of current.

Another factor is the impedance of a current source. An ideal current source has infinite impedance. In practice, the impedance of a current source should be high relative to the impedance of the load. Several techniques have been developed to design a high impedance current source. For example, Cascode current mirrors are commonly used to increase the current source output impedance, but the increased impedance usually comes at the expense of reduced voltage compliance and power dissipation [101].

For cochlear implants with multiple current sources, e.g., Advanced Bionics HiRes 90k and Med El Sonata devices, a switching network is no longer needed to connect one current source to multiple electrodes. Instead, multiple current sources are used sequentially or simultaneously, in which both the N-channel and the P-channel current sources are used to generate positive and negative phases of stimulation [102]. It is technically challenging to match between the N-channel current source and the P-channel current source to ensure that the positive and negative charges are balanced.

C. Low Power Design

Low power design and implementation of the internal ASIC chip are critical to relaxing the RF transmission efficiency and extending battery life [103,[104]]. A recent implantable neural recording system design only consumes 129μW power with the total chip consuming less than 1mW power [104]. Several design principles and methods should be considered to lower the IC chip power consumption.

For any circuits, assuming that the current is already at the low limit, high frequency and high voltage usually lead to high power consumption. For the receiver and stimulator circuitry in a cochlear implant, the data detector usually requires high frequency operation. One of the reasons that ASK modulation is preferred to FSK modulation is the relatively simple implementation and low power consumption of the ASK data detection circuitry, especially with the high frequency RF signal.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

To handle a wide range of electrode impedance, the current source in a cochlear implant typically requires high compliance voltage, thus leading to high power consumption. Minimizing voltage drop in the devices other than the electrode load is one way to achieve low power consumption. Using adaptive compliance voltages is another way to balance the need between the low power consumption and the wide impedance range [105,[106]].

D. Back Telemetry

One function of back telemetry is for the external unit to check the status of the internal unit, such as regulated voltage, compliance voltage, register values, and hand shaking status. This function of back telemetry is critical to ensure that the internal circuit works in the proper state and can correctly execute commands sent from the external unit. The other function of back telemetry is to measure and monitor critical information regarding the electrode-tissue interface, including electrode impedance, electrode field potential, and neural responses [107].

Electrode impedance is derived by measuring the voltage drop across an electrode for a given current. The current is delivered below the audible threshold, with a value in tens of μA or even lower. Extremely low electrode impedance suggests shortage while extremely high electrode impedance suggests open circuitry. Electrodes with both extreme values are typically eliminated in the fitting process (See Section VII). Electrode field potential can be obtained by stimulating one electrode while recording the potential in other non-stimulating electrodes. Electrical field imaging plots the potential distribution as a function of electrode position and can be a useful clinical tool to probe the interference and interaction between electrodes [108].

Neural response telemetry (NRT) measures the auditory neural response to electric stimulation. Because the neural response is tiny, usually buried in the artifact of electric stimulation, special techniques are required to remove the artifact. Figure 13 shows three techniques used to remove the electric artifacts [109, [110]]. First, alternating phase assumes the neural responses are the same to anodic and cathodic-leading stimuli so that simple averaging between the two responses would cancel the artifacts while preserving the neural response. However, this assumption is not true and additionally the nerve may respond to the second phase, limiting the alternating phase utility [111]. Second, forward masking takes advantage of neural refractoriness in that a probe following a masker will produce artifact but no neural response [109]. Third, template subtraction uses statistical properties to characterize electric artifact and use it to remove the artifact and recover the neural response [112, [113]]. At present, forward masking is the most widely used technique.

It is technically challenging to design a robust back telemetry system. Although all three back telemetry measurements sample the voltage on the electrode, the sampled voltage varied in a wide dynamic range. The voltage can range from several volts in electrode impedance measurement, mVs in electrical field imaging and μVs in NRT. A programmable gain amplifier is typically employed to match the sampled voltage signal to the accepted range by the analog-to-digital converter. There are two methods to transmit the information from the internal unit to the external unit. One method uses load modulation, as typically used in the RFID applications, to change the load of the internal coil so that the external unit can detect small changes in the amplitude of the RF signal applied to the external coil. The advantage for the RFID method is that only one set of coils is needed to transmit both forward and backward signals. The disadvantages are the relatively complicated RF reader in the external unit and the inability to perform both forward transmission and back telemetry simultaneously. The other method is to use a second set of coils for only backward signal transmission [114]. The advantage for the second method is the ability to perform forward and backward transmission

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

simultaneously and independently, but the disadvantage is the increased size due to the second set of coils and additional hardware.

VI. Electrode Arrays

The electrode array is the direct interfaces between the electrical output of the speech processor and the auditory neural tissue. In the last three decades, these electrode arrays have evolved from single channel to multiple channels with 12 to 22 active contacts, from an in situ position near the lateral wall of the scala tympani to a position closer to the modiolus, and from larger molded silicone elastomer "carriers" to smaller profiles. These changes are summarized here and reflect improved understanding of cochlear anatomy and electrophysiology and their relationship to cochlear implant performance.

A. Design goals for intracochlear electrodes

Three goals have guided the development of contemporary cochlear implant electrodes. The first goal is to insert the electrode array more deeply into the scala tympani to better match the assigned frequency band of electrical stimulation with the existing tonotopic organization of the cochlea and the auditory nerve. The second goal is to improve overall coupling efficiency between the electrode and the nerve. The third goal is to reduce the incidence and severity of insertion associated trauma and potential infection.

1). Insertion Depth—Recent anatomical and high resolution imaging studies [115-[118]] have allowed accurate prediction of the relationship between the frequency organization of the spiral ganglion and the position of electrode placement in an individual subject. To access lower frequency components of the speech spectra (200-1200 Hz), it is necessary to insert the electrode array to an approximate depth of 1.5 cochlear turns or 540° degrees measured from the round window. Recent studies have shown that in most cases the current generation of electrodes can be inserted to a depth of less than 400° and that deeper insertion results in higher rates of intracochlear damage and misplaced electrodes [119,[120]]. Thus, significant advances in electrode design will be required to reliably achieve optimum insertion depth with minimal trauma.

2). Coupling efficiency—Reducing the electrode-to-nerve distance decreases power consumption and interaction between channels. Early attempts to reduce this distance were made by molding electrodes to exactly match the volume of the scala tympani. Because of individual anatomical and dimensional variation this strategy was not successful. Shortly thereafter, electrodes were designed and built with a spiral shape to hold the array in a position closer to the modiolus. These electrodes were first applied in larger clinical trials by Advanced Bionics (Clarion™) and later used in conjunction with a separate elastomer positioner to further reduce the distance from the array to the spiral ganglion. In the late 1990s Cochlear Corporation, Advanced Bionics and Med-El all produced electrodes designed to be located near the modiolus and termed them as either perimodior or modiolus "hugging" electrodes.

3). Insertion Trauma—Intracochlear damage occurs when insertion forces at any point on an electrode exceed the strength of the tissue resisting that force. In many cases this damage results in decreased coupling efficiency and inconsistent channel-to-channel performance. Analysis of the forces that affect the occurrence of trauma and the angle at which an electrode contacts tissue have identified two primary mechanisms of injury [121-[126]]. First, damage most frequently occurs as a straight electrode, or curved electrode held on a stylet, contacts the outer wall of the scala tympani as it spirals away from the round window or cochleostomy through which the electrode was inserted. Second, insertion injury occurs when an electrode is inserted

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

beyond the depth at which it fully occupies the volume of the scala tympani [127]. Techniques to reduce the incidence and severity of trauma will be discussed.

B. Current Intracochlear Electrodes

Table III details specific design parameters for currently available clinical cochlear implant devices from 3 major manufacturers. Figure 14 illustrates three commonly applied cochlear implants, the Med-El Combi 40+™, Advanced Bionics Helix™ and Cochlear Contour™. Each of these devices is similar in construction with stimulating contacts fabricated from platinum-iridium (PtIr) alloy foil held on a silicone elastomer carrier. The connecting cable and lead wires in all cochlear implants are subject to breakage, particularly in active children. In addition, these wires dominate the mechanical properties of the electrode array and the increased stiffness they impart may be a cause of increased trauma. Figure 15 illustrates the crinkled wires used to reduce stiffness and increase reliability in the Med-El Combi 40+™ (Panel A) and Advanced Bionics Helix™ (Panel B) electrodes. Note that the wires in the Helix™ array are stacked along the inner radius of the spiral array to further reduce stiffness in the horizontal plane of the spiral and to increase stiffness in the vertical plane. The channel (arrow, B.) in the center of the Helix™ array is premolded to accept a wire stylet used to straighten the electrode during insertion. Panel C illustrates helical winding used to increase reliability in the cable section connecting the implanted stimulator and the electrode array in the Advanced Bionics Helix™ and 1J devices. The large cylindrical platinum iridium contact is a reference electrode used for either monopolar cochlear stimulation or measurement of intracochlear action potentials resulting from electrical stimulation. PtIr alloy wire leads are currently used for all devices with helical winding or fine zig zag patterns formed into each lead to increase flexibility and minimize the rate of failure from repeated bending.

1). Standard Arrays—The Advanced Bionics HiFocus Helix™ and Cochlear Contour Advance™ arrays are spiral shaped to match the diameter of the modiolus when fully inserted. Both devices are mounted on a straight stylet to permit initial insertion to the first cochlear turn then are advanced off of the stylet to complete the insertion. The premolded spiral shape of the Contour and Helix devices also ensures that the electrode contacts mounted on the inner radius of the electrode will be facing the spiral ganglion upon full insertion.

In contrast, the HiFocus 1J and Med-El Combi arrays follow a path close to the lateral wall during implantation. When fully inserted in temporal bones the HiFocus 1J electrode was positioned with a mean distance of 1.23 mm (electrode n=4) from the modiolus [120]. Although there have been no quantitative studies of the Combi array published to date it is reasonable to assume that the position of this electrode would be similar and published images of this electrode in place confirm their location near the spiral ligament. This position compares with a mean in situ distance from the electrode to the modiolus of 0.16 mm (electrode n=3) for the earlier HiFocus II electrode array with positioner [128], 0.65 mm (electrode n=4) for the HiFocus Helix array [120] and 0.33 mm (n=3) for the Cochlear Contour electrode [128]. Although the straighter electrodes are located laterally, it should be noted that the location of the perimodiolar arrays is highly variable and that this variability may affect threshold and channel interaction in highly idiosyncratic ways along the length of an array in an individual subject and across subjects throughout the patient population.

2). Electrodes for combined electrical acoustic stimulation (EAS)—As cochlear implant performance has steadily improved, the implantation of subjects with significant levels of residual hearing has increased dramatically [129-131]. For patients with high frequency hearing loss, it seems appropriate and possible to stimulate the basal cochlea electrically while maintaining acoustic sensitivity in the apical, low frequency, region of the cochlea. Several studies have shown that this group of subjects receives increased benefit from basal only

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

electric stimulation combined with amplified acoustic stimulation in the same ear, particularly in the presence of noise [129,[132],[133]].

Although recent temporal bone studies report that standard length arrays from each of the major cochlear implant manufacturers can be inserted without significant trauma to approximately 400°, several shortened electrode arrays (6 to 24 mm in length) have been developed to minimize the probability of injury in the basal cochlea which would disrupt the transmission of acoustic energy along the basilar membrane in the cochlea. To maximize the flexibility in programming, most of these EAS electrodes retain the same number of stimulating contacts found in the corresponding standard length electrodes in regular cochlear implants. Temporal bone studies using these shortened electrodes indicate that they can be introduced without damage and human trials have shown that subjects implanted with these electrodes experience a smaller average increase in acoustic threshold than is seen with full length electrodes in similar subject populations [134-[137]].

It is clear that a growing number of subjects with residual hearing will receive cochlear implants in the future. Whether acoustic information is used from the implanted ear or the contralateral ear, it will be increasingly important to minimize intracochlear damage in these subjects in which we presume that nerve survival is better than in subjects with greater hearing loss and in which we hope to maintain acoustic function. Thus, the continued development of strategies to reduce insertion related trauma should be a high priority for manufacturers. With the improvements outlined above, it appears that current standard length devices are close to meeting the goal of atraumatic insertion with a minimal increase in the acoustic threshold. Because most insertion trauma begins at or near the first cochlear turn, there is no logical reason that an electrode shortened to 16 –24 mm inserted to a depth of 250 – 300° should be less traumatic than a full length electrode with the same mechanical properties. Ideally, an electrode with a pre-molded spiral shape, such as the Cochlear Contour™ or Advanced Bionics Helix™, which is inserted using the AOS technique, will have no contact with the spiral ligament or basilar membrane. Such a “free floating” insertion would not only be without damage but would also minimize changes in the mechanical properties of the basilar membrane. Additionally, the majority of patients with severe high frequency hearing loss will eventually lose lower frequency hearing as well. Although this may be a very gradual process, the additional cost and medical risk associated with re-implanting these subjects with a full length cochlear implant must be considered and weighed against the perceived benefits of a shorter EAS electrode in the short term

C. Insertion Trauma, Depth and Electrode Design

Temporal bone studies of cochlear implant electrodes developed in the 1980’s and 1990’s demonstrated that severe insertion related trauma was prevalent with these first generation electrodes [127,[138]-[147]]. In contrast, recent reports with standard length electrodes indicate that the incidence of observed trauma has been significantly reduced when devices are inserted to an average depth of approximately 400° [119,[120],[124],[134],[147],[148]]. As a whole, these reports give an optimistic view that improved design of electrode arrays may well decrease the frequency and severity of injury. However, we believe that there are still significant challenges to be overcome before these electrode arrays can be reliably inserted to optimum depth without damage. First, these studies report that insertions deeper than 400° (measured from the round window) are likely to produce severe damage including fracture of the OSL and tearing of the basilar membrane or its connection to the spiral ligament. Optimum placement of these electrodes will require an additional 135°, or more, of insertion depth to match stimulation location with the tonotopic organization of the adjacent spiral ganglion. We anticipate that the design of electrodes capable of this deeper insertion will require significant invention. Second, these recent temporal bone studies were conducted by experienced cochlear implant surgeons

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

with extensive practice in the analysis of insertion trauma. Although we have not seen a correlation between the incidence of trauma observed and a surgeon's clinical cochlear implant experience, numerous surgeons have anecdotally remarked that repeated cycles of implanting and dissecting temporal bones in the laboratory is extremely beneficial to the development of their surgical expertise in this specialized area. Unfortunately, this level of feedback is impossible for most surgeons to achieve even as they gain substantial experience with cochlear implants. Thus, the small group of research oriented surgeons that are currently conducting temporal bone studies may become less and less well suited to evaluate the development and safety of future cochlear implant electrodes that will be implanted by surgeons without those insights.

To better understand the mechanics of intracochlear trauma and where it occurs we identified the initiation and extent of each injury site in a set of 13 temporal bones with severe trauma. These specimens were selected from a series of temporal bones implanted with Cochlear Contour™ electrodes evaluated at UCSF. Figure 16 illustrates the region of damage in each cochlear structure in this series of specimens. As predicted, initial damage in these specimens occurred at or near the first site of contact with the lateral wall of the ST. In most cases this injury continued for a distance of less than 90° after which the electrode was located in either the ST or scala vestibuli with minimal associated trauma. It is clear that if damage resulting from this initial high angle contact of the electrode tip with the lateral wall can be avoided the incidence of injury will be minimized.

Several techniques have been used to minimize this mode of trauma including curving the electrode tip to reduce the initial angle of contact, use of the advance off stylet (AOS) method to further reduce contact with the outer wall of the ST, changes in the shape and stiffness of the electrode tip and mechanical control of the vertical and horizontal stiffness of the electrode to reduce upward bending [121,[134],[147]-[149]]. Figure 17 illustrates the intended insertion path of a pre-molded spiral electrode using the AOS technique.

Two observations from current temporal bone studies are highly relevant to clinical practice. First, although manufacturers intend that electrodes will be inserted only until initial resistance is felt by the surgeon, data from temporal bone studies and surgeon descriptions indicate that all electrodes are most often implanted to their full depth. In our trials at UCSF surgeons have observed little resistance to insertion and were confident that the electrodes were implanted with little or no trauma. Particularly in cases with the Advanced Bionics HiFocus™ electrode with positioner, subsequent dimensional analysis [127,[149]] and temporal bone evaluations indicate that many, or most, of these insertions exceeded the depth at which the electrode and positioner could be located without trauma. This disparity indicates that it is very difficult to sense the level of resistance that signals the initiation of injury with a large tapered electrode carrier. Further, it is probable that sensation may be reduced or masked by the use of insertion tools. Due to concerns that the two part HiFocus™ electrode and positioner might contribute to the incidence of meningitis this system was voluntarily removed from the market in 2002 [150]. Looking toward future electrode designs it will be necessary to prevent this mechanism of trauma by insuring that perception of resistance by the surgeon is not required for injury free insertion.

Second, high levels of performance have been seen in subjects using all models of electrodes that were presumably traumatic in many cases. This dilemma highlights the tradeoffs between possible traumatic insertion and presumed optimum electrode position near the modiolus. As an example, Balkany [128] documented that the mean distance from the modiolus to the electrode surface was significantly less for the HiFocus™ system than for other clinical electrodes. A large body of animal based electrophysiology research and computer modeling indicates that subjects with electrodes located farther from the spiral ganglion and those with

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

greatly degenerated spiral ganglion resulting from injury should have higher thresholds and greater channel interaction than subjects with electrodes in a perimodiolar position adjacent to less damaged neurons [140,[151]-[155]]. In a similar tradeoff the current Med-El Combi 40+ electrode may be located nearer lower frequency regions of the spiral ganglion when it is deeply inserted. However, trauma was observed at deeper levels in the cochlea (> 360°) in 10 of 21 (47.6%) temporal bones implanted with the Med-El Combi 40+™ and Flex Soft™ electrodes [119]. It is not clear whether this trauma was the result of the electrode tip being too large to fit in the volume of the ST in these deepest insertions or if another mechanism of damage is responsible with these particular electrode designs. Recent studies evaluated the relative importance of insertion depth, electrode placement and trauma in clinical subjects and correlated these measures with speech recognition scores. The results indicate that trauma, as measured by the number of electrode contacts located in the scala vestibuli, has a strongly negative affect on performance even in subjects with very deeply inserted arrays [156,[157]].

D. Future Intracochlear Electrode Arrays

Development of cochlear implant electrodes will continue with improvements in safety, larger numbers of functional channels (either physical or virtual), deeper insertion and decreased manufacturing cost. In current devices several mechanical characteristics appear to reduce the incidence of insertion trauma. These factors include the relationship of vertical stiffness to horizontal stiffness in the array, the shape and stiffness of the electrode tip and the overall shape and size of the array [121,[137],[149]]. Each of these factors interacts in different ways with individual surgical techniques and surgical instrumentation. For this reason we feel that it will be necessary for manufacturers to approach the development of future electrodes with a very broad view that includes a critical examination of current electrodes, anatomy of the ST, the surgical approach, surgical instrumentation and device specific training for surgeons. We envision that advances in imaging will allow both improved speech processor fitting by documenting the unique position of each electrode within the cochlea and increased safety by giving the surgeon detailed feedback after each surgery [158-[160]]. With the application of successful design features from current electrodes and improved surgical training and instrumentation, it appears that relatively modest modifications of the current designs will ensure routine atraumatic electrode insertion to a depth of up to 400°.

At present, there is a large variation in distance from the array to the modiolus in current perimodiolar designs [120,[161]]. Refinement will be needed to consistently locate stimulating contacts close to the spiral ganglion. Because straighter electrodes are located against the spiral ligament, they appear to have less variable placement. This variability confounds comparison and analysis of performance between different perimodiolar designs and between perimodiolar electrodes and laterally located electrodes as a group.

Over the longer term advances in materials technology will be incorporated into electrode array design. Low cost thin film based electrodes were first proposed in the early 1980s [162,[163]], however, delamination and failure in both the insulation and conductive layers have prevented these electrodes from progressing to clinical trial.

One of the proposed advantages of thin film electrodes has been larger numbers of contact sites. Another approach to achieving greater spectral resolution with a CI is to use current focusing, or “steering”, to generate percepts between two physical electrode contacts. This approach has been documented in animals [164] and human subjects [165-[167]] and is applied clinically in the Advanced Bionic HiRes120 signal processing algorithm. Optimization of electrode site spacing and configuration for virtual channel stimulation will depend on further basic research and testing of clinical subjects.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

Additional features may be added to cochlear implant electrodes to further increase subject performance. This may include sensors to guide the insertion process by measuring the distance from the sensor to the margin of the surrounding ST or by measuring pressure when the electrode contacts a surface of the cochlea. To enable the surgeon to respond to this feedback the tip of a future electrode might also be steered by fluidic channels or mechanical controls. Anti-inflammatory or neurotrophic agents might also be administered via fine channels in a cochlear implant electrode to support neural regeneration or attract growth of dendrites to specific stimulating contact sites.

VII. Fitting Program

Because each patient is unique, the cochlear implant has to be customized to meet his or her conditions and needs. A fitting program is used to interface between a PC and the external speech processor. The interface provides two-way communication during fitting sessions, which is generally conducted one month after surgery and adjusted annually or as necessary. The fitting program collects critical patient-specific information from patient preference of speech processing strategies to the setting of threshold and comfortable loudness values on all electrodes. Once these critical electric stimulation parameters are set, the fitting program stores the information as a “map” on the PC, which is then downloaded to the memory (usually EPROM) of the speech processor.

Figure 18 shows a section of the fitting program interface used in the Nucleus Freedom device, called Custom Sound. The y-axis represents the clinical units for electric currents whereas the x-axis represents the number of electrodes. The electrodes can be flagged (electrode 12 and 3 in this case), hence not used, because they are malfunctioning electrically, e.g., short or open circuitry, or cause unpleasant or unwanted problems such as dizziness or other facial nerve stimulation. To create the map, on all other useful electrodes, both the lowest current level inducing threshold (T level) and the highest level inducing comfortable loudness (C level) are typically measured and recorded, with the difference being the dynamic range (DR). These electrical parameters usually change with different processing strategies and electrode configurations, and sometimes may change with time and experience, thus requiring separate measurements and monitoring. Four to nine different maps can be downloaded to the speech processor dependent upon the manufacturer (See Table 1). The fitting process can be time consuming, lasting from 20 minutes to 2 hours, depending on the device complexity, patient condition, and audiologists’ skill levels.

At present, efficient, integrated, and objective fitting needs to be developed. Significant progress has been made in efficient fitting of cochlear implants, includes measuring only the comfortable loudness level (Clarion), streamlining the mapping by interpolating the T and C levels in a subset of the electrodes [168], and optimizing the map by genetic algorithms [169]. Because structured training has been shown to benefit cochlear implant performance [170], a future fitting system may be integrated with a structured training system.

While electrode impedance has long been used to determine a short or open connection, other objective measures are slowly moving into cochlear implant fitting. A promising measure is the electrical compound action potential or ECAP that can be measured by the latest device from all three manufacturers. The ECAP threshold is typically between the behaviorally measured T and C levels, and can be used to create a map without any subjective responses from the patient. Such objectively created maps are especially valuable to the increasing pediatric implant populations as well as those who cannot give accurate behavioral responses, but their speech performance is still sub-optimal compared with the behaviorally created maps [171-173]. Another measure under investigation for objective mapping is acoustic reflex, which contracts the middle ear muscle to attenuate a loud sound, but its clinical acceptance appears

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

to be at least several years away [174,[175]]. Finally, high resolution imaging data before and after surgery will likely be integrated in the fitting system to only predict performance but also optimize the frequency map between the electrodes and the residual nerve [176,[177]].

VIII. System Integration

To provide safe, reliable and useful electric hearing, the cochlear implant imposes extreme physical, power, environmental and handling constraints on both system design and system integration. First, the implant must be safe, causing no harm, chronic pain or discomfort to the people who use it. Second, the implant must be reliable, because children who receive cochlear implants will depend on them to develop their speech and communication skills and will use their devices for a significant portion of their life before it is replaced if at all. This imposes a serious need to design the product to withstand the implant environment for up to, if not beyond, 30 years. Third, for the implant to reside safely and unobtrusively on the patient's head, severe restrictions are imposed in the physical size and shape of the package available to house electronics. Fourth, the implant is powered by batteries so that a low power consumption design is vital. Furthermore, the implant should be able to handle not only a very harsh operating environment inside the human body for a long time but also extreme physical abuse such as mechanical impact in case of a fall or crash. The implant users want to participate in the normal activities of daily living unshackled by the need to use and protect an overly delicate prosthesis. Finally, the implant must be designed in close collaboration with surgeons and patients to accommodate a broad range of anatomical variations and to ensure reliable, effective and satisfactory outcomes. To illustrate the importance of system integration, the following two sections illustrate some of the problems that occurred in earlier generations of the cochlear implants and had led to reduced overall system performance, cumbersome administrative overhead, and increased risk to the patients.

A. Design and Implementation issues

One example is the setting of input dynamic range in the speech processor. In the early cochlear implant systems [e.g., [59]], the input dynamic range was set to be around 30 dB because of the beliefs that speech dynamic range is also about 30 dB [178] and that the 30 dB range better matches the 10-20 dB electric dynamic range. Detailed acoustic analysis and cochlear implant speech evaluation have shown that greater than 30 dB input dynamic ranges are needed for better performance [50], leading to a setting of the 50-60 dB input dynamic range in current cochlear implants.

The second example is related to accuracy of current sources. In the first-generation Clarion device [95], the current source had little head room, resulting in saturating response at relatively low current values. This nonlinear saturation likely contributed to unusually wide electric dynamic range and low speech performance in some cochlear implant users [50] and possibly also produces artificially high perceptual thresholds of electric vision when the Clarion cochlear implant was adopted by Second Sight for its retinal implants [179]. A different current source problem also created an administrative burden in the Nucleus 22 implant users. Because of large variability in the chip fabrication process, the current source in each Nucleus 22 device produced different values (>10%) when the same current amplitude was specified. The manufacturer had to calibrate each individual device and create a patient-specific lookup table for fitting and research purposes [62]. The manufacturer has been able to reduce the current source variability to less than 10%, thus eliminating the cumbersome device-specific calibration and patient-specific lookup table in the Nucleus 24 devices.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

B. Cochlear implant and bacterial meningitis

In 2002, the U.S. Food and Drug Administration received a growing number of reports of bacterial meningitis in cochlear implant recipients. Although these cases occurred in subjects implanted with devices produced by all the major manufacturers the frequency of infection was clearly greatest in the device with a separate elastomer positioner, the HiFocus™ II system. At this time, the HiFocus™ II was withdrawn from the market voluntarily and reintroduced a short time later without the attached positioner. Cohen and colleagues [180] found that the number of meningitis cases was actually greater than previously thought and confirmed that devices from all manufacturers were associated with these infections. Concurrently, the FDA issued an advisory notice that all children receiving cochlear implants should also be vaccinated against streptococcus pneumoniae and other species common to the middle ear space [181, 182].

From a device design perspective it appears that two factors may contribute to the incidence of meningitis. First, a two-part electrode might allow increased bacterial biofilm formation between the individual components of the device, which would be resistant to normal immune response. This mechanism could also occur in a single part device with enclosed fluid channels or a stylet channel that opened into the ST due to damage. The second mechanism that may play a role in infection is intracochlear trauma. A severe fracture of the osseous spiral lamina or damage to the inner surface of the ST provides a direct path for bacteria to enter the central nervous system via the internal auditory meatus [183]. Future electrode arrays should be designed with these two factors in mind.

C. Safety Considerations

Absolute product safety is almost impossible to demonstrate in human implant applications, however, safety is a very important consideration in the design of any neural prosthetic device. Safety should occupy a high priority in any set of product design requirements. The safety of an implantable neural prosthetic device can be segmented in to four categories: 1) materials and their biocompatibility and toxicity, 2) sterilization to eliminate infection, 3) mechanical design with its potential to cause structural tissue damage, and 4) energy exposure limits and the resulting tissue and neural damage. Safety problems related to each of these categories can have both short and long term consequences.

1). *Biocompatibility* is the property of a material being biologically compatible by not producing a toxic, injurious, or immunological response in living tissue. The materials used to fabricate implantable system components need to be compatible with the tissue and structures in the vicinity of the device and need to be appropriately selected for their specific use.

There is a significant history of implantable materials demonstrating the successful biocompatibility in implantable applications. The most direct approach to selecting materials to achieve biocompatibility in system design would be to use materials that have established acceptable biocompatible safety records in implanted applications. By selecting materials that already meet an FDA recognized biocompatibility standard you can avoid the need for animal testing and may be able to submit a declaration of conformity in place of performance data [184]. In addition to saving time and expense, declaration of conformity allows design energy and focus to be placed elsewhere. Materials selections would then be based on factors related to their mechanical properties, electrical properties and abilities to achieve hermetic isolation. Table IV presents a partial list of frequently used biocompatible materials in cochlear implants.

These materials are currently used in cochlear implants that have received FDA approvals. Implanted devices fabricated with these materials have demonstrated significant histories of safe and effective performance in the field. While the materials listed above are considered to

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

be biocompatible, caution must be exercised in their specific use. For example, when selecting materials for the design of electrodes used to deliver electrical stimulation, it has been found that platinum-iridium electrodes are safer and less damaging to neural tissue then titanium electrodes used at the same stimulation exposure levels [185]. Therefore, even though titanium is biocompatible, its use should be limited to specific applications that do not include delivering electrical stimulation to neural tissue. The materials most commonly used for this application in cochlear implants are platinum-iridium alloys. A metal alloy with a ratio of 90% platinum to 10% iridium is a common selection and offers good electrical conductivity, good mechanical strength and ductility properties that have a reasonable resistance to fatigue failure. Changing these ratios can achieve different material properties while maintaining the biocompatibility of the final combination of materials. Caution should always be exercised and reviewed with experts and regulatory authorities before any final decisions are made.

Fabricating system components from biocompatible materials frequently requires the use of non-biocompatible and toxic materials. While these other materials are usually removed during the manufacturing process caution needs to be exercised to ensure there are no residual amounts left on the final component. Manufacturing processes must be designed and validated to ensure the end result is a component that contains only biocompatible materials.

2). *Sterilization*: A serious but often overlooked and underestimated need for the design and manufacturing of an implantable device is ensuring it can be sterilized and that the final result is that the sterile product reaches the customer. Several publications are available that deal with sterilization requirements for implanted devices. Ethylene oxide (EtO) is frequently used to sterilize cochlear implants and has been standardized (ANSI/AAMI/ISO 11135: “Sterilization of medical devices. Validation and routine control of ethylene oxide sterilization”; ANSI/AAMI/ISO 10993-7: “Biological evaluation of medical devices part 7, ethylene oxide residuals”). Other acceptable sterilization methods are available for cochlear implants. In fact, there are situations where a device may be exposed to multiple sterilization processes. These are covered by multiple standards and documents available from the FDA and other sources [186]. It is suggested that the designer review these standards before design requirements are finalized.

Even though the material used for a specific design may be sterilized, it must also be designed and manufactured to tolerate the process that will be used to achieve sterilization. Sterilization processes often expose materials to high temperatures and harsh chemicals. The EtO sterilization process applies both high temperatures and a chemical gas to the components to achieve sterilization. Materials exposed to the EtO process must be able to withstand exposure to both of these conditions without damage. Additionally, the structure of implanted components and housing must be designed to avoid pockets, crevices or other small spaces in the external surfaces where bacteria can collect, making it more difficult if not impossible for the sterilization processes to be effective.

Device testing to assess the effectiveness of the selected sterilization process shall be conducted and documented before regulatory approval is granted. Failure to pass this test is a serious problem and frequently occurs late in the design process. This problem would cause a significant delay in the delivery schedule. It is suggested that the implant designer conduct an early sterilization test on the package to ensure it is effective and a later test on the completed product to ensure that exposure to the sterilization process does not harm the product.

3). *Mechanical Safety*: Tissue trauma is frequently the results of mechanical stress or chronic force applied to tissue by the implantable package and electrode. Tissue trauma can also be induced by surgeries that result from a design that is difficult to put in place and stabilize. Relative to the package, smaller is generally better, but this is not the only consideration. The

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

cochlear implant package is normally placed behind the pinna of the ear embedded in bonny tissue around and adjacent to the mastoid cavity. A very small package that cannot be stabilized will probably cause more tissue damage then a well-designed package that can be stabilized in an effective bone bed. A new device implanted in a healthy subject will probably be encapsulated in tissue that will stabilize the device after several weeks; the design goal is to provide stabilization until this encapsulation occurs. The top surface of the package needs to be shaped to reduce internal tissue trauma that could result in long term problems. Softly rounded corners and soft silicone rubber encapsulation helps prevent these problems. There have been cases of severe tissue necrosis reported with some implants and this usually results in an explant and a subsequent re implant with a different device or a different placement method. Designers need to work carefully with skilled surgeons to ensure the final design meets the surgical needs and minimizes the potential for chronic tissue trauma.

4). *Energy exposure* must be constrained to safe levels in implanted products. The type of energy that is encountered in implants includes, electricity, heat, light and sound. The bio-effects of light and sound energy sources, particularly the interaction between tissue and laser or ultrasonic sound, are well documented. This section focuses on electrical and heat energy exposures that are most relevant to cochlear implants.

Stimulation of the cochlea requires exposure to adequate amounts of electrical energy to achieve sufficient neural recruitment to achieve loudness. Increasing requirements to improve tonotopic selectivity is forcing a reduction in contact surface area along with a higher density electrode pitch. These trends are placing additional burdens on keeping electrical energy below safe exposure levels. The standard parameter used to quantify energy delivery for neural activation is charge density. The maximum total charge, charge density and its delivery must be specified for safe operation of the stimulator. Most modern cochlear implants use current source stimulation drivers. The charge is the product of current and time of the signal applied to the contact. Driver currents in cochlear implants range from a few micro-amps to as high as two milliamps. Electrical contacts vary in range from 0.12 mm^2 up to over 1.5 mm^2. Typically the safe charge density limit is less than 15 to 65 μC/cm2/phase although higher values have been considered safe in electric stimulation of the nerve tissue [97,[98]].

External devices that maintain surface contact with the skin should not have chronic temperatures that exceed 41 degrees centigrade, although recent amendments to IEC standards have raised the chronic temperature to 43 degrees. It is suggested that a careful review of this standard be conducted during the design process. This criteria allows considerable rise if the environmental temperature is low but may be difficult to achieve in very hot climates. The temperature rise of implanted electronics must be minimized to safe levels. Implanted devices must not have surface temperatures that exceed 39 degrees centigrade under any condition in vivo. The implant environment must be considered during design and testing to ensure the final device meets requirements [187]. For example, implanting a device in tissue increases its thermal mass and will aid in limiting the rise in temperature. Different tissue groups and locations in the body will have different influences on rises in temperature. Highly perfused tissue will have a more pronounced affect on temperature than poorly perfused tissue. It is important to test the final design in an effective tissue environment. One approach is to measure the temperature rise while the device is implanted in the tissue of a live animal with tissue similar to the final human implant subjects. The temperature rise is a result of normal operation but is probably increased more as a result of exposures to outside energy sources such as MRI unless the device contains a rechargeable battery. Analytical treatment of this topic is difficult and early in vivo modeling and testing is suggested.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

D. Risk Management

It is critical to recognize the interactions of the device, the user, and the environment and their contributions to the safe and effective device use or unsafe or ineffective use. The FDA has adopted a human factors engineering approach to identify and isolate these factors and to address and manage their risks [188]. Figure 19 shows the dynamic interactions among the device, its user and environment and their contributions to the safe vs. unsafe and effective vs. ineffective usage. The goal of the FDA guideline is to minimize use-related hazards, assure that intended users are able to use medical devices safely and effectively throughout the product life cycle, and to facilitate review of new device submissions and design control documentation. Verification and validation are needed to ensure that the system specifications be met and the device be safely used under simulated or actual environments.

IX. System Evaluation

The cochlear implant has to be judged and accepted by its user. Detailed psychophysical, speech, music, language, and cognitive performance has been systematically measured and evaluated. Cost-utility benefits have also been analyzed and demonstrated, resulting in complete or partial coverage of cochlear implantation by major insurance companies and Medicare in the US [189]. Here basic psychophysical performance and challenges under complex listening situations are highlighted to illustrate both the limitation of current cochlear implant design and the opportunity for the future.

Basic psychophysical performance helps us understand what the cochlear implant limitations are and how to compensate for these limitations. For example, the psychophysical performance may be limited by the design, the hardware, the electrode-tissue interface, or the brain. Figure 20 shows cochlear implant performance in loudness, pitch, tuning, and temporal processing. Different from a power function in normal loudness growth and a wide 100-120 dB dynamic range, cochlear implant users typically have an exponential loudness growth function and a much narrower 10-20 dB dynamic range. The altered loudness growth and the reduced dynamic range are to large extent due to the damaged cochlea [190] and can be compensated effectively by the amplitude compression circuitry in the speech processor. The temporal pitch is limited to 300-500 Hz in a typical cochlear implant user, whereas pitch above this frequency may be provided by place pitch, as evidenced by the different saturation points between apical and basal electrodes at 1000-2000 Hz high stimulation rates. Current processors have not taken advantage of this dual pitch code. The spatial tuning curve in electric hearing can be 10 or more times wider than that in acoustic hearing. Although tripolar stimulation has been proposed, it is not clear how it and other similar proposals can effectively compensate for this huge difference in the spatial tuning curve [191, [192]]. The psychophysical measure that justifies the “bionic ear” nickname is the superior performance in detecting temporal variations by cochlear implant users. Apparently this superior performance also has to do with the lost cochlear compression [193] and significantly contributes to cochlear implant speech performance [194].

One key question in early cochlear implant research was to evaluate how many channels are needed to support open-set speech recognition. The answer was clearly more than one, but whether 3-4 channels or 32 or more are needed depends upon the difficulty of the task, particularly upon whether the task requires extraction and use of pitch of complex sounds [196,[197]]. At present, many modern cochlear implant users have approached normal performance in standard audiological tests, e.g., daily sentence recognition in quiet [198].

Another interesting question regarding system evaluation is: What does a cochlear implant sound like? This line of research is also called acoustic simulation of cochlear implants, which has not only allowed a normal-hearing listener to appreciate the sound of electric stimulation

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

but also helped identify the key parameters associated with cochlear implant performance. The acoustic simulation of the cochlear implant is based on vocoders that extract temporal envelopes from a number of spectral bands and use them to amplitude-modulate noise-band or sinusoidal carriers [4,[5]]. The acoustic simulation has captured the essence of a cochlear implant from the signal processing point of view but is severely limited by a lack of incorporation of the highly distorted electrode-to-neuron interface. This limitation produces several problems associated with the acoustic simulation of a cochlear implant. One problem is that the acoustic simulation is likely to predict the performance by the best cochlear implant users, therefore cannot account for the large individual variability among the cochlear implant users. The other problem is that the acoustic simulation cannot simulate the qualitative aspects of a cochlear implant. As cochlear implant patients with almost normal hearing on the contralateral side as well as detailed physiological data become available, realistic acoustic simulations are being developed to overcome not only the present simulation limitations but also provide an accurate account of the complicated interactions between electric stimulation and the nervous system.

Despite the high level performance, significant difficulty still remains, even for today's star cochlear implant users. Figure 21 highlights three challenges of aural and oral communication in the present cochlear implant users. Panel A shows that a normal-hearing listener requires -5 dB SNR to achieve 50% correct performance, whereas a cochlear implant user requires 10 dB, resulting in a 15 dB deficit in SNR. The SNR deficit is increased to 30 dB when the noise is a competing voice [70, [199]-[201]]. Panel B shows that while the cochlear implant users can use rhythmic cues normally, they cannot perform simple melody and timbre recognition [202]. It is amazing to observe that these implant users can converse relatively easily on the phone but cannot tell the difference between two familiar nursery songs, suggesting that speech recognition and music perception are two different processes [203]. Panel C shows that cochlear implant users cannot perceive or produce Mandarin tones normally [204, [205]]. There is also evidence that cochlear implant users have a great deal of difficulty in talker identification [206] and emotion detection [207]. Advances in signal processing (Section III) and electrode design (Section VI) are needed to bridge the performance gaps between normal-hearing and cochlear implant listeners.

X. Beyond Cochlear Implants

Intensive and interesting investigation is being conducted to extend the utility of cochlear implants and beyond. One active area of research is to demonstrate the effectiveness of bilateral cochlear implants and justify the cost of the second implant [209]. The current bilateral implants are essentially two independent devices, showing mostly the better ear or the head shadow effect [210]. However, encouraging results have been produced by coordinating the use of two implants in the laboratory, showing significantly improvement in detecting interaural timing differences [211] and binaural masking level differences [212]. These recent advances have increased the likelihood of success in using two implants to produce truly functional binaural hearing. A second area of active research in extending the cochlear implant's utility is combined acoustic and electric hearing [213,[214]]. Many hearing-impaired persons have significant residual hearing at low frequencies, for example, below 300 Hz. These low-frequency sounds are not even transmitted by the telephone because they provided little or no speech intelligibility [215]. Surprisingly, these low-frequency acoustic sounds, when combined with a cochlear implant, can improve the implant performance in noise by 10 dB in SNR [216-[218]]. All manufacturers are conducting clinical trials in combined acoustic and electric hearing with the expectation that this new technology will penetrate new markets.

It is clear that one has to go beyond cochlear implants to address the full spectrum of hearing disorders. Figure 22 shows presently available options in treating hearing impairment. The

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

traditional hearing aid will likely occupy the largest market, helping millions with moderate to severe hearing impairment. The middle ear implant uses similar cochlear implant technology but uses mechanical output to directly drive the cochlea, which will likely help those with conductive or mixed hearing loss, with unilateral loss, and with chronic dysfunction of the middle ear [219]. The auditory nerve implant is also revived to provide low stimulation current and sharp spatial selectivity, potentially solving the poor pitch perception problem in electric hearing [220, [221]]. Hardware from the microphone to the implant packaging will become better and smaller, making totally implantable devices feasible and effective [222].

For those who have an ossified cochlea, no auditory nerve, or acoustic tumors, the cochlear implant is not going to provide any help. The stimulation site has to be moved to the central auditory nervous system. The auditory brainstem implant (ABI) stimulates the first stage in the auditory brainstem (i.e., the cochlear nuclei) has been ongoing for more than 20 years with several hundred patients, but has produced mixed results [223,[224]]. Recent data suggest that the likelihood of success of the ABI depends on the presence of acoustic tumors, which has somehow affected the integration of the central nerve pathways critical to speech perception [225]. Because of its relatively easy surgical access and clearly defined anatomical structure, the inferior colliculus (IC) has also been proposed as an effective stimulation site [226]. Several patients have received the IC implant recently, producing useful hearing [227].

As the most successful neural prosthesis, the cochlear implant has also served as a model for the design and evaluation of other neural prostheses. One notable example is development of a retinal implant by Second Sight, which was based on a 16-channel cochlear implant (Advanced Bionics Corporation Clarion C1 device) and had been implanted in at least six subjects blind as a result of retinitis pigmentosain [179]. In the retinal implant, a camera replaced the microphone while a surface electrode array arranged in a 4 × 4 pattern replaced the 16-electrode cochlear electrodes. The RF link protocol and the internal receiver and stimulator remained basically unchanged. The other example is the vestibular implant, which is used to restore balance function [228,[229]]. No human vestibular implantation has been reported but the first vestibular implant used in a human clinical trial will likely be adapted from the cochlear implant. After all, the fundamental principles are the same for all neural prostheses, involving electric stimulation of the nerve.

XI. Summary

Electric stimulation had a humble beginning as dangerous direct current stimulation almost fried Volta’s brain. Thanks to persistent and outstanding collaborative work by engineers, scientists, physicians, and entrepreneurs, safe and charge-balanced stimulation has now provided or restored hearing to more than 120,000 people worldwide. The present review has systematically and comprehensively presented the cochlear implant system design and specifications, identified key subsystem components and functions, provided valuable system integration and evaluation information, and discussed broad perspective and impact beyond the cochlear implant. It is fair to conclude that the cochlear implant not only has a long and distinguished history but also a bright future as it continues to broaden its utility for treatment of a wide range of hearing impairment and to serve as a model to guide development of other neural prostheses.

Acknowledgments

The authors thank J. Roland, W. Luxford and T. Karkaraventos for temporal bone studies, and T. Lu, T. Chua, Q. Tang, P. Lin, P. Leake, A. Copeland and K. Olsen for support and comments. The work was supported in part by the National Institutes of Health, U.S. Department of Health & Human Services (RO1-DC002267, RO1-DC008858, P30-DC008369, and HHS-N-263-2007-00054-C) and The Chinese Academy of Sciences Pilot Project of the Knowledge Innovation Program (KGCX2-YX-607).

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

References

1. Volta A. On the electricity excited by mere contact of conducting substances of different kinds. Royal Soc Philos Trans 1800;90:403–431.
2. Fletcher, H. Speech and Hearing in Communication. Vol. Second. New York: D Van Nostrand Company, Inc.; 1953.
3. Flanagan JL, Golden RM. Phase Vocoder. Bell System Technical Journal 1966;45:1493–1509.
4. Dudley H. The vocoder. Bell Labs Record 1939;17:122–126.
5. Shannon RV, Zeng FG, Kamath V, Wygonski J, Ekelid M. Speech recognition with primarily temporal cues. Science 1995;270:303–4. [PubMed: 7569981]
6. Wilson BS, Finley CC, Lawson DT, Wolford RD, Eddington DK, Rabinowitz WM. Better speech recognition with cochlear implants. Nature 1991;352:236–8. [PubMed: 1857418]
7. Zeng FG, Nie K, Stickney GS, Kong YY, Vongphoe M, Bhargave A, Wei C, Cao K. Speech recognition with amplitude and frequency modulations. Proc Natl Acad Sci U S A 2005;102:2293–8. [PubMed: 15677723]
8. Flottorp G. Effect of Different Types of Electrodes in Electrophonic Hearing. J Acoust Soc Am 1953;25:236–245.
9. Simmons FB. Electrical stimulation of the auditory nerve in man. Arch Otolaryngol 1966;84:2–54. [PubMed: 5936537]
10. Stevens S, Jones R. The mechanism of hearing by electrical stimulation. J Acoust Soc Am 1939;10:261–269.
11. Stevens S. On hearing by electrical stimulation. J Acoust Soc Am 1937;8:191–195.
12. Jones R, Stevens S, Lurie M. Three mechanisms of hearing by electrical stimulation. J Acoust Soc Am 1940;12:281–290.
13. Djourno A, Eyries C. Auditory prosthesis by means of a distant electrical stimulation of the sensory nerve with the use of an indwelt coiling. Presse Med 1957;65:1417. [PubMed: 13484817]
14. Djourno A, Eyries C, Vallancien P. Preliminary attempts of electrical excitation of the auditory nerve in man, by permanently inserted micro-apparatus. Bull Acad Natl Med 1957;141:481–3. [PubMed: 13500039]
15. Djourno A, Eyries C, Vallancien B. Electric excitation of the cochlear nerve in man by induction at a distance with the aid of micro-coil included in the fixture. C R Seances Soc Biol Fil 1957;151:423–5. [PubMed: 13479991]
16. Simmons FB, Epley JM, Lummis RC, Guttman N, Frishkopf LS, Harmon LD, Zwicker E. Auditory nerve: Electrical stimulation in man. Science 1965;148:104–6. [PubMed: 14258725]
17. Michelson RP. The results of electrical stimulation of the cochlea in human sensory deafness. Ann Otol Rhinol Laryngol 1971;80:914–9. [PubMed: 5127759]
18. House WF, Urban J. Long term results of electrode implantation and electronic stimulation of the cochlea in man. Ann Otol Rhinol Laryngol 1973;82:504–17. [PubMed: 4721186]
19. Doyle JH, Doyal JB, Turnbull FM. Electrical stimulation of the eighth cranial nerve. Arch Otolaryngol 1964;80:388–391. [PubMed: 14198701]
20. Luxford, W.; Brackmann, DE. The history of cochlear implants. In: Gray, RF., editor. Cochlear implants. San Diego, CA 92105: College-Hill Press; 1985. p. 1-26.
21. Chouard CH. The surgical rehabilitation of total deafness with the multichannel cochlear implant. Indications and results. Audiology 1980;19:137–45. [PubMed: 6892765]
22. Eddington DK, Dobelle WH, Brackmann DE, Mladevosky MG, Parkin JL. Auditory prosthesis research with multiple channel intracochlear stimulation in man. Ann Otol Rhinol Laryngol 1978;87:1–39. [PubMed: 736424]
23. Hochmair-Desoyer IJ, Hochmair ES, Burian K, Fischer RE. Four years of experience with cochlear prostheses. Med Prog Technol 1981;8:107–19. [PubMed: 6895542]
24. Kiang NY, Moxon EC, Levine RA. Auditory-nerve activity in cats with normal and abnormal cochlea. In: Sensorineural hearing loss. Ciba Found Symp 1970:241–73. [PubMed: 5210916]
25. Garud R, Rappa MA. A Socio-Cognitive Model of Technology Evolution: The Case of Cochlear Implants. Organization Science 1994;5:344–362.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

26. Bilger RC. Psychoacoustic evaluation of present prostheses. Ann Otol Rhinol Laryngol Suppl 1977;86:92–104. [PubMed: 405915]
27. Bilger RC, Hopkinson NT. Hearing performance with the auditory prosthesis. Ann Otol Rhinol Laryngol Suppl 1977;86:76–91. [PubMed: 405914]
28. Bilger RC, Black FO. Auditory prostheses in perspective. Ann Otol Rhinol Laryngol Suppl 1977;86:3–10. [PubMed: 405908]
29. Bilger RC, Black FO, Hopkinson NT. Research plan for evaluating subjects presently fitted with implanted auditory prostheses. Ann Otol Rhinol Laryngol Suppl 1977;86:21–4. [PubMed: 405906]
30. Bilger RC, Stenson NR, Payne JL. Subject acceptance of implanted auditory prosthesis. Ann Otol Rhinol Laryngol Suppl 1977;86:165–76. [PubMed: 405905]
31. Bilger RC. Electrical stimulation of the auditory nerve and auditory prostheses: a review of the literature. Ann Otol Rhinol Laryngol Suppl 1977;86:11–20. [PubMed: 405903]
32. Van de Ven, A.; Polley, D.; Garud, R.; Venkataraman, S. The Innovation Journey. Oxford, UK: Oxford University Press; 1999.
33. Cochlear Implants. NIH Consensus Statement May 2-4;1988 7(2)
34. Eddington DK. Speech discrimination in deaf subjects with cochlear implants. J Acoust Soc Am 1980;68:885–91. [PubMed: 6893461]
35. Wilson BS, Lawson DT, Finley CC, Wolford RD. Coding strategies for multichannel cochlear prostheses. Am J Otol 1991;12(Suppl):56–61. [PubMed: 2069190]
36. Zeng FG, Shannon RV. Loudness-coding mechanisms inferred from electric stimulation of the human auditory system. Science 1994;264:564–6. [PubMed: 8160013]
37. Dorman MF, Soli S, Dankowski K, Smith LM, McCandless G, Parkin J. Acoustic cues for consonant identification by patients who use the Ineraid cochlear implant. J Acoust Soc Am 1990;88:2074–9. [PubMed: 2269723]
38. Shannon RV. Temporal modulation transfer functions in patients with cochlear implants. J Acoust Soc Am 1992;91:2156–64. [PubMed: 1597606]
39. An SK, Park SI, Jun SB, Lee CJ, Byun KM, Sung JH, Wilson BS, Rebscher SJ, Oh SH, Kim SJ. Design for a simplified cochlear implant system. IEEE Trans Biomed Eng 2007;54:973–82. [PubMed: 17554817]
40. Zeng FG. Trends in cochlear implants. Trends Amplif 2004;8:1–34. [PubMed: 15247993]
41. Balkany T, Hodges A, Menapace C, Hazard L, Driscoll C, Gantz B, Kelsall D, Luxford W, McMenomy S, Neely JG, Peters B, Pillsbury H, Roberson J, Schramm D, Telian S, Waltzman S, Westerberg B, Payne S. Nucleus Freedom North American clinical trial. Otolaryngol Head Neck Surg 2007;136:757–62. [PubMed: 17478211]
42. Koch DB, Osberger MJ, Segel P, Kessler D. HiResolution and conventional sound processing in the HiResolution bionic ear: using appropriate outcome measures to assess speech recognition ability. Audiol Neurootol 2004;9:214–23. [PubMed: 15205549]
43. Arnoldner C, Riss D, Brunner M, Durisin M, Baumgartner WD, Hamzavi JS. Speech and music perception with the new fine structure speech coding strategy: preliminary results. Acta Otolaryngol 2007;127:1298–303. [PubMed: 17851892]
44. Edgerton BJ, Doyle KJ, Brimacombe J, Danley MJ, Fretz RJ. THE EFFECTS OF SIGNAL PROCESSING BY THE HOUSE–URBAN SINGLE-CHANNEL STIMULATOR ON AUDITORY PERCEPTION ABILITIES OF PATIENTS WITH COCHLEAR IMPLANTS. Annals of the New York Academy of Sciences 1983;405:311–322. [PubMed: 6575654]
45. Clark GM, Blamey PJ, Brown AM, Gusby PA, Dowell RC, Franz BK, Pyman BC, Shepherd RK, Tong YC, Webb RL, et al. The University of Melbourne--nucleus multi-electrode cochlear implant. Adv Otorhinolaryngol 1987;38:V–IX. 1–181. [PubMed: 3318305]
46. Patrick JF, Busby PA, Gibson PJ. The development of the Nucleus Freedom Cochlear implant system. Trends Amplif 2006;10:175–200. [PubMed: 17172547]
47. Hochmair I, Nopp P, Jolly C, Schmidt M, Schosser H, Garnham C, Anderson I. MED-EL Cochlear implants: state of the art and a glimpse into the future. Trends Amplif 2006;10:201–19. [PubMed: 17172548]

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

48. James CJ, Skinner MW, Martin LF, Holden LK, Galvin KL, Holden TA, Whitford L. An investigation of input level range for the nucleus 24 cochlear implant system: speech perception performance, program preference, and loudness comfort ratings. Ear Hear 2003;24:157–74. [PubMed: 12677112]
49. Spahr AJ, Dorman MF, Loiselle LH. Performance of patients using different cochlear implant systems: effects of input dynamic range. Ear Hear 2007;28:260–75. [PubMed: 17496675]
50. Zeng FG, Grant G, Niparko J, Galvin J, Shannon R, Opie J, Segel P. Speech dynamic range and its effect on cochlear implant performance. J Acoust Soc Am 2002;111:377–86. [PubMed: 11831811]
51. Zeng FG. Temporal pitch in electric hearing. Hear Res 2002;174:101–6. [PubMed: 12433401]
52. Green T, Faulkner A, Rosen S. Enhancing temporal cues to voice pitch in continuous interleaved sampling cochlear implants. J Acoust Soc Am 2004;116:2298–310. [PubMed: 15532661]
53. Laneau J, Wouters J, Moonen M. Relative contributions of temporal and place pitch cues to fundamental frequency discrimination in cochlear implantees. J Acoust Soc Am 2004;116:3606–19. [PubMed: 15658711]
54. Loizou PC. Signal-processing techniques for cochlear implants. IEEE Eng Med Biol Mag 1999;18:34–46. [PubMed: 10337562]
55. Rubinstein JT. How cochlear implants encode speech. Curr Opin Otolaryngol Head Neck Surg 2004;12:444–8. [PubMed: 15377959]
56. Ay SU, Zeng F-G, Sheu BJ. Hearing with bionic ears. IEEE Circuits and Devices 1997;5:18–23.
57. Fant, G. Acoustic theory of speech production. The Hague: Mouton; 1970.
58. Zeng FG. Role of temporal fine structure in speech perception. J Acoust Soc Am 2008;123:3710.
59. Seligman PM, Patrick JF, Tong YC, Clark GM, Dowell RC, Crosby PA. A signal processor for a multiple-electrode hearing prosthesis. Acta Otolaryngol Suppl 1984;411:135–9. [PubMed: 6596837]
60. Blamey PJ, Dowell RC, Clark GM, Seligman PM. Acoustic parameters measured by a formant-estimating speech processor for a multiple-channel cochlear implant. J Acoust Soc Am 1987;82:38–47. [PubMed: 3624639]
61. Patrick JF, Clark GM. The Nucleus 22-channel cochlear implant system. Ear Hear 1991;12:3S–9S. [PubMed: 1955088]
62. Skinner MW, Holden LK, Holden TA, Dowell RC, Seligman PM, Brimacombe JA, Beiter AL. Performance of postlinguistically deaf adults with the Wearable Speech Processor (WSP III) and Mini Speech Processor (MSP) of the Nucleus Multi-Electrode Cochlear Implant. Ear Hear 1991;12:3–22. [PubMed: 2026285]
63. Van Tasell DJ, Soli SD, Kirby VM, Windin GP. Speech waveform envelope cues for consonant recognition. The Journal of the Acoustical Society of America 1987;82:1152–1161. [PubMed: 3680774]
64. Rosen S. Temporal information in speech: Acoustic, auditory and linguistic aspects. Philos Trans R Soc Lond B Biol Sci 1992;336:367–73. [PubMed: 1354376]
65. McDermott HJ, McKay CM, Vandali AE. A new portable sound processor for the University of Melbourne/Nucleus Limited multielectrode cochlear implant. J Acoust Soc Am 1992;91:3367–71. [PubMed: 1619114]
66. Nie K, Barco A, Zeng FG. Spectral and temporal cues in cochlear implant speech perception. Ear Hear 2006;27:208–17. [PubMed: 16518146]
67. Zeng FG, Shannon RV. Loudness balance between electric and acoustic stimulation. Hear Res 1992;60:231–5. [PubMed: 1639732]
68. Wilson BS, Finley CC, Farmer JC Jr, Lawson DT, Weber BA, Wolford RD, Kenan PD, White MW, Merzenich MM, Schindler RA. Comparative studies of speech processing strategies for cochlear implants. Laryngoscope 1988;98:1069–77. [PubMed: 3172953]
69. Lai WK, Bogli H, Dillier N. A software tool for analyzing multichannel cochlear implant signals. Ear Hear 2003;24:380–91. [PubMed: 14534409]
70. Zeng FG, Galvin JJ 3rd. Amplitude mapping and phoneme recognition in cochlear implant listeners. Ear Hear 1999;20:60–74. [PubMed: 10037066]
71. Fishman KE, Shannon RV, Slattery WH. Speech recognition as a function of the number of electrodes used in the SPEAK cochlear implant speech processor. J Speech Lang Hear Res 1997;40:1201–15. [PubMed: 9328890]

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

72. Litvak L, Saoji A, Spahr A, Kulkarni A, Marzalek M. Use of simultaneous stimulation to represent fine structure in cochlear implant processors. J Acoust Soc Am 2008;123:3055.
73. Wilson BS, Schatzer R, Lopez-Poveda EA, Sun X, Lawson DT, Wolford RD. Two new directions in speech processor design for cochlear implants. Ear Hear 2005;26:73S–81S. [PubMed: 16082269]
74. Green T, Faulkner A, Rosen S, Macherey O. Enhancement of temporal periodicity cues in cochlear implants: effects on prosodic perception and vowel identification. J Acoust Soc Am 2005;118:375–85. [PubMed: 16119358]
75. Laneau J, Wouters J, Moonen M. Improved music perception with explicit pitch coding in cochlear implants. Audiol Neurootol 2006;11:38–52. [PubMed: 16219993]
76. Lan N, Nie KB, Gao SK, Zeng FG. A novel speech-processing strategy incorporating tonal information for cochlear implants. IEEE Trans Biomed Eng 2004;51:752–60. [PubMed: 15132501]
77. Rubinstein JT, Wilson BS, Finley CC, Abbas PJ. Pseudospontaneous activity: stochastic independence of auditory nerve fibers with electrical stimulation. Hear Res 1999;127:108–18. [PubMed: 9925022]
78. Loeb GE. Are cochlear implant patients suffering from perceptual dissonance? Ear Hear 2005;26:435–50. [PubMed: 16230894]
79. Nie K, Stickney G, Zeng FG. Encoding frequency modulation to improve cochlear implant performance in noise. IEEE Trans Biomed Eng 2005;52:64–73. [PubMed: 15651565]
80. Throckmorton CS, Selin Kucukoglu M, Remus JJ, Collins LM. Acoustic model investigation of a multiple carrier frequency algorithm for encoding fine frequency structure: implications for cochlear implants. Hear Res 2006;218:30–42. [PubMed: 16797896]
81. Crosby, PA.; Daly, CN.; Money, DK.; Patrick, JF.; Seligman, PM.; Kuzma, JA. Cochlear implant system for an auditory prosthesis. United States Patent. 4532930. 1985.
82. Hochmair, IJ.; Hochmair, ES. Transcutaneous power and signal transmission system and methods for increased signal transmission efficiency. United States Patent. 5070535. 1991.
83. Zierhofer CM, Hochmair ES. Geometric approach for coupling enhancement of magnetically coupled coils. Biomedical Engineering, IEEE Transactions on 1996;43:708–714.
84. Shannon RV, Adams DD, Ferrel RL, Palumbo RL, Grandgenett M. A computer interface for psychophysical and speech research with the Nucleus cochlear implant. J Acoust Soc Am 1990;87:905–7. [PubMed: 2307784]
85. Robert, ME.; Wygonski, J. House Ear Institute Nucleus Research Interface (HEINRI) Users Guide. Los Angeles, CA: House Ear Institute; 2002.
86. Daly, CN.; McDermott, H. Embedded data link and protocol. United States Patent. 5741314. 1998.
87. Vandali AE, Whitford LA, Plant KL, Clark GM. Speech perception as a function of electrical stimulation rate: using the Nucleus 24 cochlear implant system. Ear Hear 2000;21:608–24. [PubMed: 11132787]
88. Friesen LM, Shannon RV, Cruz RJ. Effects of stimulation rate on speech recognition with cochlear implants. Audiol Neurootol 2005;10:169–84. [PubMed: 15724088]
89. Plant K, Holden L, Skinner M, Arcaroli J, Whitford L, Law MA, Nel E. Clinical evaluation of higher stimulation rates in the nucleus research platform 8 system. Ear Hear 2007;28:381–93. [PubMed: 17485987]
90. Sokal NO, Sokal AD. Class E-A new class of high-efficiency tuned single-ended switching power amplifiers. Solid-State Circuits, IEEE Journal of 1975;10:168–176.
91. Kazimierczuk M, Puczko K. Exact analysis of class E tuned power amplifier at any Q and switch duty cycle. Circuits and Systems, IEEE Transactions on 1987;34:149–159.
92. Hahn, TW.; Griffith, GA. Power transfer circuit for implanted devices. United States Patent. 6212431. 2001.
93. Shaquer, CM. Transcutaneous power optimization circuit for cochlear implant. United States Patent. 6810289. 2004.
94. Zierhofer, CM. Multichannel cochlear implant with neural response telemetry. United States Patent. 6600955. 2003.
95. Schulman, JH.; Whitmoyer, DI.; Gord, JC.; Strojnik, P. Multichannel cochlear implant system including wearable speech processor. United States Patent. 5603726. 1997.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

96. McDermott H. An advanced multiple channel cochlear implant. IEEE Trans Biomed Eng 1989;36:789–97. [PubMed: 2744794]
97. Shannon RV. A model of safe levels for electrical stimulation. IEEE Trans Biomed Eng 1992;39:424–6. [PubMed: 1592409]
98. McCreery DB, Agnew WF, Yuen TG, Bullara L. Charge density and charge per phase as cofactors in neural injury induced by electrical stimulation. IEEE Trans Biomed Eng 1990;37:996–1001. [PubMed: 2249872]
99. Huang CQ, Shepherd RK, Carter PM, Seligman PM, Tabor B. Electrical stimulation of the auditory nerve: direct current measurement in vivo. IEEE Trans Biomed Eng 1999;46:461–70. [PubMed: 10217884]
100. Huang CQ, Shepherd RK, Seligman PM, Clark GM. Reduction in excitability of the auditory nerve following acute electrical stimulation at high stimulus rates: III. Capacitive versus non-capacitive coupling of the stimulating electrodes. Hear Res 1998;116:55–64. [PubMed: 9508028]
101. Ghovanloo M, Najafi K. A compact large voltage-compliance high output-impedance programmable current source for implantable microstimulators. IEEE Trans Biomed Eng 2005;52:97–105. [PubMed: 15651568]
102. Gord, JC. Programmable current output stimulus stage for implantable device. United States Patent. 6181969. 2001.
103. Baker MW, Sarpeshkar R. Feedback Analysis and Design of RF Power Links for Low-Power Bionic Systems. Biomedical Circuits and Systems, IEEE Transactions on 2007;1:28–38.
104. Neihart NM, Harrison RR. Micropower circuits for bidirectional wireless telemetry in neural recording applications. Biomedical Engineering, IEEE Transactions on 2005;52:1950–1959.
105. Money, D. High compliance output stage for a tissue stimulator. United States Patent. 6289246. 2001.
106. Karunasiri, RT. Digitally controlled RF amplifier with wide dynamic range output. United States Patent. 7016738. 2006.
107. Abbas PJ, Brown CJ, Shallop JK, Firszt JB, Hughes ML, Hong SH, Staller SJ. Summary of results using the nucleus CI24M implant to record the electrically evoked compound action potential. Ear Hear 1999;20:45–59. [PubMed: 10037065]
108. van der Beek FB, Boermans PP, Verbist BM, Briaire JJ, Frijns JH. Clinical evaluation of the Clarion CII HiFocus 1 with and without positioner. Ear Hear 2005;26:577–92. [PubMed: 16377994]
109. Miller CA, Abbas PJ, Brown CJ. An improved method of reducing stimulus artifact in the electrically evoked whole-nerve potential. Ear Hear 2000;21:280–90. [PubMed: 10981604]
110. van den Honert C, Stypulkowski PH. Characterization of the electrically evoked auditory brainstem response (ABR) in cats and humans. Hear Res 1986;21:109–26. [PubMed: 3754550]
111. Miller CA, Abbas PJ, Rubinstein JT. An empirically based model of the electrically evoked compound action potential. Hear Res 1999;135:1–18. [PubMed: 10491949]
112. Hashimoto T, Elder CM, Vitek JL. A template subtraction method for stimulus artifact removal in high-frequency deep brain stimulation. J Neurosci Methods 2002;113:181–6. [PubMed: 11772439]
113. Heffer LF, Fallon JB. A novel stimulus artifact removal technique for high-rate electrical stimulation. J Neurosci Methods 2008;170:277–84. [PubMed: 18339428]
114. Griffith, GA.; Faltys, MA. Remote control unit for use with an implantable neural stimulator. US Patent. 7092763. Aug 15. 2006
115. Stakhovskaya O, Sridhar D, Bonham B, Leake PA. Frequency Map for the Human Cochlear Spiral Ganglion: Implications for Cochlear Implants. JARO. 2006
116. Dorman M, Spahr A, Gifford R, Loiselle L, McKarns S, Holden T, Skinner M, Finley C. An electric-to-place map for a cochlear implant patient with hearing in the nonimplanted ear. JARO. 2007
117. Boex C, Baud L, Cosendai G, Sigrist A, Kos M, Pelizzone M. Acoustic to Electric Pitch Comparisons in Cochlear Implant Subjects with Residual Hearing. J Assoc Res Otolaryngol 2006:1–15. Epub.
118. Escude B, James C, Deguine O, Cochard N, Eter E, Fraysse B. The Size of the Cochlea and Predictions of Insertion Depth Angles for Cochlear Implant Electrodes. Aud Neurotol 2006;11 (suppl 1):27–33.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

119. Adunka O, Kiefer J. Impact of electrode insertion depth on intracochlear trauma. Otolaryngol Head Neck Surg 2006;135:374–82. [PubMed: 16949967]
120. Wright CG, Roland PS, Kuzma J. Advanced bionics thin lateral and helix II electrodes: a temporal bone study. Laryngoscope 2005;115:2041–2045. [PubMed: 16319621]
121. Rebscher SJ, Heilmann M, Bruszewski W, Talbot NH, Snyder RL, Merzenich MM. Strategies to Improve Electrode Positioning and Safety in Cochlear Implants. IEEE Trans Biomed Eng 1999;46:340–352. [PubMed: 10097469]
122. Rebscher SJ, Talbot N, Bruszewski W, Heilmann M, Brasell J, Merzenich MM. A Transparent Model of the Human Scala Tympani Cavity. J Neurosci Meth 1996;64:105–114.
123. Chen BK, Clark GM, Jones R. Evaluation of trajectories and contact pressures for the straight nucleus cochlear implant electrode array - a two-dimensional application of finite element analysis. Medical Engineering and Physics 2003;25:141–147. [PubMed: 12538068]
124. Rebscher SJ, Hetherington A, Bonham B, Wardrop P, Whinney D, LEAKE PA. Considerations for design of future cochlear implant electrode arrays: electrode array stiffness, size and depth of insertion. J Rehab Res Dev 2008;45 in press.
125. Kha HN, Chen BK, Clark G, Jones R. Stiffness properties for Nucleus standard straight and contour electrode arrays. Medical Engineering and Physics 2004;26:677–685. [PubMed: 15471696]
126. Roland JT. A model for cochlear implant electrode insertion and force evaluation: results with a new electrode design and insertion technique. Laryngoscope 2005;115:1325–1339. [PubMed: 16094101]
127. Wardrop P, Whinney D, Rebscher SJ, Luxford W, Leake P. A temporal bone study of insertion trauma and intracochlear position of cochlear implant electrodes. II: Comparison of Spiral Clarion and HiFocus II electrodes. Hear Res 2005;203:68–79. [PubMed: 15855031]
128. Balkany TJ, Eshraghi AA, Yang N. Modiolar proximity of three perimodiolar cochlear implant electrodes. Acta Otolaryngol 2002;122:363–9. [PubMed: 12125990]
129. Fraysse B, Macias AR, Sterkers O, Burdo S, Ramsden R, Deguine O, Klenzner T, Lenarz T, Rodriguez M, von Wallenberg E, James C. Residual hearing conservation and electroacoustic stimulation with the nucleus 24 contour advance cochlear implant. Otol Neurotol 2006;27:624–33. [PubMed: 16868510]
130. James C, Fraysse B, Deguine O, Lenarz T, Mawman D, Ramos A, Ramsden R, Sterkers O. Combined Electroacoustic Stimulation in Conventional Candidates for Cochlear Implantation. Aud Neurotol 2006;11(suppl 1):57–62.
131. Fraysse B, Dillier N, Klenzner T, Laszig R, Manrique M, Perez CM, Morgon AH, Muller-Deile J, Macias AR. Cochlear Implants for Adults Obtaining Marginal Benefit from Acoustic Amplification. Am J Otol 1998;19:591–597. [PubMed: 9752966]
132. Gantz B, Turner C, Gfeller K. Acoustic Plus Electric Speech Processing: Preliminary Results of a Multicenter Clinical Trial of the Iowa/Nucleus Hybrid Implant. Aud Neurotol 2006;11(suppl 1): 63–68.
133. Gantz B, Turner C, Gfeller K, Lowder M. Preservation of hearing in cochlear implant surgery: advantages of combined electrical and acoustical speech processing. Laryngoscope 2005;115:796–806. [PubMed: 15867642]
134. Adunka O, Unkelbach M, Mack M, Hambek M, Gstoettner W, Kiefer J. Cochlear implantation via the round window minimizes trauma to cochlear structures: a histologically controlled study. Acta Otolaryngol 2004;124:807–812. [PubMed: 15370564]
135. Lenarz T, Stover T, Buechner A, Paasche G, Briggs R, Risi F, Pesch J, Battmer R. Temporal Bone Results and Hearing Preservation with a New Straight Electrode. Aud Neurotol 2006;11(suppl 1): 34–41.
136. Briggs RJS, Tykocinski M, Xu J, Risi F, Svehla M, Cowan R, Stover T, Erfurt P, Lenarz T. Comparison of Round Window and Cochleostomy Approaches with a Prototype Hearing Preservation Electrode. Aud Neurotol 2006;11(suppl 1):42–48.
137. Roland JT, Zeitler D, Jethanamest D, Huang TC. Evaluation of the short hybrid electrode in human temporal bones. Otol Neurotol 2008;29:482–488. [PubMed: 18317395]

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

138. Wardrop P, Whinney D, Rebscher SJ, Roland JTJ, Luxford W, Leake PA. A temporal bone study of insertion trauma and intracochlear position of cochlear implant electrodes. I: Comparison of Nucleus banded and Nucleus Contour electrodes. Hear Res 2005;203:54–67. [PubMed: 15855030]
139. Skinner M, Ketten D, Holden L, Harding G, Smith P, Gates G, Neely J, Kletzker G, Brunsden B, Blocker B. CT-derived estimation of cochlear morphology and electrode array position in relation to word recognition in Nucleus-22 recipients. J Assoc Res Otolaryngol 2002;3:332–350. [PubMed: 12382107]
140. Nadol JB, Shiao JY, Burgess BJ, Ketten DR, Eddington DK, Gantz BJ, Kos I, Montandon P, Coker NJ, Roland JT, Shallop JK. Histopathology of cochlear implants in humans. Annals of Otol Rhinol Laryngol 2001;110:883–891.
141. Richter B, Aschendorff A, Lohnstein P, Husstedt H, Nagursky H, Laszig R. Clarion 1.2 standard electrode array with partial space-filling positioner: radiological and histological evaluation in human temporal bones. J Laryngol Otol 2002;116:507–13. [PubMed: 12238669]
142. Richter B, Aschendorff A, Lohnstein P, Husstedt H, Nagursky H, Laszig R. The nucleus contour electrode array: a radiological and histological study. Laryngoscope 2001;111:508–513. [PubMed: 11224784]
143. Briggs R, Tykocinski M, Saunders E, Hellier W, Dahm M, Pyman B, GM C. Surgical implications of perimodiolar cochlear implant electrode design: avoiding intracochlear damage and scala vestibuli insertion. Cochlear Implants International 2001;2:135–149. [PubMed: 18792095]
144. Tykocinski M, Saunders E, Cohen LT, Treaba C, Briggs RJS, Gibson P, Clark GM, Cowan RSC. The contour electrode array: safety study and initial patient trials of a new perimodiolar design. Otol Neurotol 2001;22:33–41. [PubMed: 11314713]
145. Tykocinski M, Cohen LT, Pyman BC, Roland T, Treaba C, Palamara J, Dahm MC, Shepherd RK, Xu J, Cowan RS, Cohen NL, Clark GM. Comparison of Electrode Position in the Human Cohclea Using Various Perimodiolar Electrode Arrays. Am J Otol 2000;21:205–211. [PubMed: 10733185]
146. Gstoettner W, Plenk H, Franz P, Hamzavi J, Baumgartner W, Czerny C, Ehrenberger K. Cochlear Implant Deep Electrode Insertion: Extent of Insertional Truama. Acta Otolaryngol 1997;117:274–277. [PubMed: 9105465]
147. Roland JT. A model for cochlear implant electrode insertion and force evaluation: results with a new electrode design and insertion technique. Laryngoscope 2005;115:1325–1339. [PubMed: 16094101]
148. Stover T, Issing P, Graurock G, Erfurt P, ElBeltagy Y, Paasche G, Lenarz T. Evaluation of the advance off-stylet insertion technique and the cochlear insertion tool in temporal bones. Otol Neurotol 2005;26:1161–70. [PubMed: 16272935]
149. Rebscher S, Hetherington A, Bonham B, Wardrop P, Whinney D, LEAKE PA. Considerations for design of future cochlear implant electrode arrays: electrode array stiffness, size and depth of insertion. J Rehab Res Dev 2008;45 in press.
150. Reefhuis J, Honien M, W CG, Mann EA, Biernath KR, C P, E J, B C. Bacterial meningitis among children with cochlear implants beyond 24 months after implantation. Pediatrics 2006;117:284–289. [PubMed: 16390918]
151. Leake, P.; Rebscher, SJ. Anatomical considerations and long-term effects of electrical stimulation. In: Fay, RR., editor. Cochlear Implants: Auditory prostheses and electric hearing, Springer handbool of auditory research. New York: 2004. p. 101-148.
152. Rebscher SJ, Snyder RL, Leake PA. The effect of electrode configuration and duration of deafness on threshold and selectivity of responses to intracochlear electrical stimulation. J Acoust Soc Am 2001;109:2035–2048. [PubMed: 11386556]
153. Frijns JHM, B JJ, Grote JJ. The importance of human cochlear anatomy for the results of modiolus-hugging multichannel cochlear implants. Otology and Neurotology 2001;22:340–349. [PubMed: 11347637]
154. Frijns JHM, d Snoo SL, t Kate JH. Spatial Selectivity in a Rotationally Symmetric Model of the Electrically Stimulated Cochlea. Hear Res 1996;95:33–48. [PubMed: 8793506]
155. Nadol J, Young Y. Survival of spiral ganglion cells in profound sensorineural hearing loss: implications for cochlear implantation. Ann Otol Rhinol Laryngol 1989;98:411–416. [PubMed: 2729822]

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

156. Skinner MW, Holden TA, Whiting BR, Voie AH, Brunsden B, Neely JG, Saxon EA, Hullar TE, Finley CC. In vivo estimates of the position of advanced bionics electrode arrays in the human cochlea. Ann Otol Rhinol Laryngol Suppl 2007;197:2–24. [PubMed: 17542465]
157. Finley CC, Holden TA, Holden LK, Whiting BR, Chole RA, Neely GJ, Hullar TE, Skinner MW. Role of Electrode Placement as a Contributor to Variability in Cochlear Implant Outcomes. Otol Neurotol. 2008 in press.
158. Aschendorff A, Kubalek R, Turowsi B. Quality control after cochlear implant surgery by means of rotational tomography. Otol Neurotol 2005;26:34–7. [PubMed: 15699717]
159. Aschendorff A, Kromeier J, Klenzner T. Quality control after insertion of the nucleus contour and contour advance electrode in adults. Ear Hear 2007;28:75S–79S. [PubMed: 17496653]
160. Skinner M, Holden L, Whiting B. In vivo estimates of the position of advanced bionics electrode arrays in the human cochlea. Ann Otol Rhinol Laryngol 2007;197:2–24.
161. Balkany TJ, Eshraghi AA, Yang N. Modiolar proximity of three perimodiolar cochlear implant electrodes. Acta Otolaryngol 2002;122:363–9. [PubMed: 12125990]
162. Lauridsen O, Gunthersen C, Bonding P, Tos M. Experiments with a Thinfilm Multichannel Electrode for Cochlear Implantation. Acta Otolaryngol 1983;95:219–226. [PubMed: 6687652]
163. White RL, Roberts LA, Cotter NE, Oh-Hyun K. Thin-Film Electrode Fabrication Techniques. Ann New York Acad Sci 1983;405:183–190. [PubMed: 6575643]
164. Bonham B, Litvak L. Current Focusing and Steering: Modeling, physiology and psychophysics. Hear Res. 2008 in press.
165. Firszt J, Koch D, Downing M, Litvak L. Current steering creates additional percepts in adult cochlear implant recipients. Otol Neurotol. 2007
166. Busby P, Plant K. Dual electrode stimulation using the nucleus CI24RE cochlear implant: Electrode impedance and pitch ranking studies. Ear Hear 2005;26:504–511. [PubMed: 16230899]
167. Koch D, Downing M, Osberger M, Litvak L. Using current steering to increase spectral resolution in CII and HiRes 90K Users. Ear Hear 2007;28:38S–41S. [PubMed: 17496643]
168. Plant K, Law MA, Whitford L, Knight M, Tari S, Leigh J, Pedley K, Nel E. Evaluation of streamlined programming procedures for the Nucleus cochlear implant with the Contour electrode array. Ear Hear 2005;26:651–68. [PubMed: 16378000]
169. Wakefield GH, van den Honert C, Parkinson W, Lineaweaver S. Genetic algorithms for adaptive psychophysical procedures: recipient-directed design of speech-processor MAPs. Ear Hear 2005;26:57S–72S. [PubMed: 16082268]
170. Fu QJ, Galvin JJ 3rd. Maximizing cochlear implant patients' performance with advanced speech training procedures. Hear Res. 2007
171. Potts LG, Skinner MW, Gotter BD, Strube MJ, Brenner CA. Relation between neural response telemetry thresholds, T- and C-levels, and loudness judgments in 12 adult nucleus 24 cochlear implant recipients. Ear Hear 2007;28:495–511. [PubMed: 17609612]
172. Smoorenburg GF, Willeboer C, van Dijk JE. Speech perception in nucleus CI24M cochlear implant users with processor settings based on electrically evoked compound action potential thresholds. Audiol Neurootol 2002;7:335–47. [PubMed: 12401965]
173. Seyle K, Brown CJ. Speech perception using maps based on neural response telemetry measures. Ear Hear 2002;23:72S–79S. [PubMed: 11883770]
174. Spivak LG, Chute PM, Popp AL, Parisier SC. Programming the cochlear implant based on electrical acoustic reflex thresholds: patient performance. Laryngoscope 1994;104:1225–30. [PubMed: 7934592]
175. Clement RS, Carter PM, Kipke DR. Measuring the electrical stapedius reflex with stapedius muscle electromyogram recordings. Ann Biomed Eng 2002;30:169–79. [PubMed: 11962769]
176. Finley CC, Holden TA, Holden LK, Whiting BR, Chole RA, Neely GJ, Hullar TE, Skinner MW. Role of Electrode Placement as a Contributor to Variability in Cochlear Implant Outcomes. Otol Neurotol. 2008
177. Skinner MW, Ketten DR, Holden LK, Harding GW, Smith PG, Gates GA, Neely JG, Kletzker GR, Brunsden B, Blocker B. CT-derived estimation of cochlear morphology and electrode array position in relation to word recognition in Nucleus-22 recipients. J Assoc Res Otolaryngol 2002;3:332–50. [PubMed: 12382107]

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

178. French NR, Steinberg JC. Factors governing the intelligibility of speech sounds. J Acoust Soc Am 1947;19:90–119.
179. de Balthasar C, Patel S, Roy A, Freda R, Greenwald S, Horsager A, Mahadevappa M, Yanai D, McMahon MJ, Humayun MS, Greenberg RJ, Weiland JD, Fine I. Factors affecting perceptual thresholds in epiretinal prostheses. Invest Ophthalmol Vis Sci 2008;49:2303–14. [PubMed: 18515576]
180. Cohen NL, Roland JT Jr, Marrinan M. Meningitis in cochlear implant recipients: the North American experience. Otol Neurotol 2004;25:275–81. [PubMed: 15129105]
181. Whitney CG. Cochlear implants and meningitis in children. Pediatr Infect Dis J 2004;23:767–8. [PubMed: 15295228]
182. Reefhuis J, Honein MA, Whitney CG, Chamany S, Mann EA, Biernath KR, Broder K, Manning S, Avashia S, Victor M, Costa P, Devine O, Graham A, Boyle C. Risk of bacterial meningitis in children with cochlear implants. N Engl J Med 2003;349:435–45. [PubMed: 12890842]
183. Wei BP, Shepherd RK, Robins-Browne RM, Clark GM, O'Leary SJ. Effects of inner ear trauma on the risk of pneumococcal meningitis. Arch Otolaryngol Head Neck Surg 2007;133:250–9. [PubMed: 17372082]
184. Guidance for Industry and FDA Staff - Recognition and Use of Consensus Standards. U.S. Food and Drug Administration; Sep 17. 2007
185. Harnack D, Winter C, Meissner W, Reum T, Kupsch A, Morgenstern R. The effects of electrode material, charge density and stimulation duration on the safety of high-frequency stimulation of the subthalamic nucleus in rats. J Neurosci Methods 2004;138:207–16. [PubMed: 15325129]
186. Mendes GC, Brandao TR, Silva CL. Ethylene oxide sterilization of medical devices: a review. Am J Infect Control 2007;35:574–81. [PubMed: 17980234]
187. Lazzi G. Thermal effects of bioimplants. IEEE Eng Med Biol Mag 2005;24:75–81. [PubMed: 16248120]
188. Guidance for Industry and FDA Premarketand Design Control Reviewers: Medical Device Use-Safety: Incorporating Human Factors Engineering into Risk Management. U.S. Food and Drug Administration; Jul 18. 2000
189. Cheng AK, Rubin HR, Powe NR, Mellon NK, Francis HW, Niparko JK. Cost-utility analysis of the cochlear implant in children. JAMA 2000;284:850–6. [PubMed: 10938174]
190. Zeng, FG. Compression and cochlear Implants. In: Bacon, SP.; Popper, AN.; Fay, RR., editors. Compression: From cochlea to cochlear Implants, vol. 17, Springer Handbook of Auditory Research. New York: Springer-Verlag; 2003. p. 184-220.
191. Snyder RL, Middlebrooks JC, Bonham BH. Cochlear implant electrode configuration effects on activation threshold and tonotopic selectivity. Hear Res 2008;235:23–38. [PubMed: 18037252]
192. Bierer JA, Middlebrooks JC. Auditory cortical images of cochlear-implant stimuli: dependence on electrode configuration. J Neurophysiol 2002;87:478–92. [PubMed: 11784764]
193. Oxenham AJ, Bacon SP. Cochlear compression: perceptual measures and implications for normal and impaired hearing. Ear Hear 2003;24:352–66. [PubMed: 14534407]
194. Fu QJ. Temporal processing and speech recognition in cochlear implant users. Neuroreport 2002;13:1635–9. [PubMed: 12352617]
195. Nelson DA, Donaldson GS, Kreft H. Forward-masked spatial tuning curves in cochlear implant users. J Acoust Soc Am 2008;123:1522–43. [PubMed: 18345841]
196. Smith ZM, Delgutte B, Oxenham AJ. Chimaeric sounds reveal dichotomies in auditory perception. Nature 2002;416:87–90. [PubMed: 11882898]
197. Shannon RV. Recent advances in cochlear implants. J Acoust Soc Am 1990;87:421–2. [PubMed: 2299048]
198. Wilson BS, Dorman MF. The surprising performance of present-day cochlear implants. IEEE Trans Biomed Eng 2007;54:969–72. [PubMed: 17554816]
199. Fu QJ, Shannon RV. Phoneme recognition by cochlear implant users as a function of signal-to-noise ratio and nonlinear amplitude mapping. J Acoust Soc Am 1999;106:L18–23. [PubMed: 10462822]

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

200. Dorman MF, Loizou PC, Fitzke J, Tu Z. The recognition of sentences in noise by normal-hearing listeners using simulations of cochlear-implant signal processors with 6-20 channels. J Acoust Soc Am 1998;104:3583–5. [PubMed: 9857516]
201. Stickney G, Zeng F, Litovsky R, Assmann P. Cochlear implant speech recognition with speech masker. J Acoustic Society of America. 2003 in revision.
202. Kong YY, Cruz R, Jones JA, Zeng FG. Music perception with temporal cues in acoustic and electric hearing. Ear Hear 2004;25:173–85. [PubMed: 15064662]
203. Zatorre RJ, Belin P, Penhune VB. Structure and function of auditory cortex: music and speech. Trends Cogn Sci 2002;6:37–46. [PubMed: 11849614]
204. Han D, Zhou N, Li Y, Chen X, Zhao X, Xu L. Tone production of Mandarin Chinese speaking children with cochlear implants. Int J Pediatr Otorhinolaryngol 2007;71:875–80. [PubMed: 17376544]
205. Wei CG, Cao K, Zeng FG. Mandarin tone recognition in cochlear-implant subjects. Hear Res 2004;197:87–95. [PubMed: 15504607]
206. Vongphoe M, Zeng FG. Speaker recognition with temporal cues in acoustic and electric hearing. J Acoust Soc Am 2005;118:1055–61. [PubMed: 16158660]
207. Xin L, Fu QJ, Galvin JJ 3rd. Vocal emotion recognition by normal-hearing listeners and cochlear implant users. Trends Amplif 2007;11:301–15. [PubMed: 18003871]
208. Gfeller K, Witt S, Woodworth G, Mehr MA, Knutson J. Effects of frequency, instrumental family, and cochlear implant type on timbre recognition and appraisal. Ann Otol Rhinol Laryngol 2002;111:349–56. [PubMed: 11991588]
209. Summerfield AQ, Marshall DH, Barton GR, Bloor KE. A cost-utility scenario analysis of bilateral cochlear implantation. Arch Otolaryngol Head Neck Surg 2002;128:1255–62. [PubMed: 12431166]
210. Schleich P, Nopp P, D'Haese P. Head shadow, squelch, and summation effects in bilateral users of the MED-EL COMBI 40/40+ cochlear implant. Ear Hear 2004;25:197–204. [PubMed: 15179111]
211. Laback B, Majdak P. Binaural jitter improves interaural time-difference sensitivity of cochlear implantees at high pulse rates. Proc Natl Acad Sci U S A 2008;105:814–7. [PubMed: 18182489]
212. Long CJ, Carlyon RP, Litovsky RY, Downs DH. Binaural unmasking with bilateral cochlear implants. J Assoc Res Otolaryngol 2006;7:352–60. [PubMed: 16941078]
213. von Ilberg C, Kiefer J, Tillein J, Pfenningdorff T, Hartmann R, Sturzebecher E, Klinke R. Electric-acoustic stimulation of the auditory system. New technology for severe hearing loss. ORL J Otorhinolaryngol Relat Spec 1999;61:334–40. [PubMed: 10545807]
214. Gantz BJ, Turner C. Combining acoustic and electrical speech processing: Iowa/Nucleus hybrid implant. Acta Otolaryngol 2004;124:344–7. [PubMed: 15224850]
215. Chang JE, Bai JY, Zeng FG. Unintelligible low-frequency sound enhances simulated cochlear-implant speech recognition in noise. IEEE Trans Biomed Eng 2006;53:2598–601. [PubMed: 17152439]
216. Kong YY, Stickney GS, Zeng FG. Speech and melody recognition in binaurally combined acoustic and electric hearing. J Acoust Soc Am 2005;117:1351–61. [PubMed: 15807023]
217. Dorman MF, Gifford RH, Spahr AJ, McKarns SA. The benefits of combining acoustic and electric stimulation for the recognition of speech, voice and melodies. Audiol Neurootol 2008;13:105–12. [PubMed: 18057874]
218. Turner C, Gantz B, Vidal C, Behrens A, Henry B. Speech recognition in noise for cochlear implant listeners: benefits of residual acoustic hearing. J Acoust Soc Am 2004;115:1729–35. [PubMed: 15101651]
219. Ricketts, TA. Middle ear implants. Vol. 6. New York: Westminister Publications, Inc.; 2002.
220. Middlebrooks JC, Snyder RL. Auditory prosthesis with a penetrating nerve array. J Assoc Res Otolaryngol 2007;8:258–79. [PubMed: 17265124]
221. Badi AN, Kertesz TR, Gurgel RK, Shelton C, Normann RA. Development of a novel eighth-nerve intraneural auditory neuroprosthesis. Laryngoscope 2003;113:833–42. [PubMed: 12792319]

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

222. Briggs RJ, Eder HC, Seligman PM, Cowan RS, Plant KL, Dalton J, Money DK, Patrick JF. Initial clinical experience with a totally implantable cochlear implant research device. Otol Neurotol 2008;29:114–9. [PubMed: 17898671]

223. McCreery DB. Cochlear nucleus auditory prostheses. Hear Res 2008;242:64–73. [PubMed: 18207678]

224. Schwartz MS, Otto SR, Shannon RV, Hitselberger WE, Brackmann DE. Auditory brainstem implants. Neurotherapeutics 2008;5:128–36. [PubMed: 18164492]

225. Colletti V. Auditory outcomes in tumor vs. nontumor patients fitted with auditory brainstem implants. Adv Otorhinolaryngol 2006;64:167–85. [PubMed: 16891842]

226. Lenarz T, Lim HH, Reuter G, Patrick JF, Lenarz M. The auditory midbrain implant: a new auditory prosthesis for neural deafness-concept and device description. Otol Neurotol 2006;27:838–43. [PubMed: 16936570]

227. Lim HH, Lenarz T, Joseph G, Battmer RD, Samii A, Samii M, Patrick JF, Lenarz M. Electrical stimulation of the midbrain for hearing restoration: insight into the functional organization of the human central auditory system. J Neurosci 2007;27:13541–51. [PubMed: 18057212]

228. Della Santina CC, Migliaccio AA, Patel AH. A multichannel semicircular canal neural prosthesis using electrical stimulation to restore 3-d vestibular sensation. IEEE Trans Biomed Eng 2007;54:1016–30. [PubMed: 17554821]

229. Merfeld DM, Haburcakova C, Gong W, Lewis RF. Chronic vestibulo-ocular reflexes evoked by a vestibular prosthesis. IEEE Trans Biomed Eng 2007;54:1005–15. [PubMed: 17554820]

NIH-PA Author Manuscript
NIH-PA Author Manuscript
NIH-PA Author Manuscript

Biographies

Fan-Gang Zeng (S'88-M'91-SM'07) is a Professor in Anatomy and Neurobiology, Biomedical Engineering, Cognitive Sciences and Otolaryngology - Head and Neck Surgery at the University of California, Irvine. He has published more than 70 peer-reviewed journal articles, 20 book chapters, and 2 books including a volume on cochlear implants in Springer Handbook of Auditory Research (Springer-Verlag, New York). He holds 10 U.S. Patents and has given more than 100 invited presentations worldwide. He is on the editorial board for IEEE Transactions on Biomedical Engineering, Journal of Association for Research in Otolaryngology, Journal of Otology, Audiology and Neurotology, and Hearing Research. He

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

has reviewed grants for National Institutes of Health, National Science Foundation, National Natural Science Foundation of China, Natural Sciences and Engineering Research Council of Canada, and British Wellcome Trust. He is on the Advisory Board of Apherma Corporation, Sunnyvale, CA, Nurotron Biotech Inc., Irvine, CA, and ImThera Inc., San Diego, CA.

Stephen Rebscher was born in Redwood City, CA. He received the B.S., degree with highest honors in biological sciences at San Jose State University in 1976 and received the M.A. degree in biological and natural sciences in 1979. He has been a research specialist in the Department of Otolaryngology at the University of California in San Francisco since 1979. His primary area of interest is the development of safe and effective cochlear implant electrode arrays. To

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

this end he has worked to better understand the mechanics and effects of stimulation of prototype electrode arrays and to evaluate the safety of currently available cochlear implants. He is a co-inventor of the UCSF/Advanced bionics clinical cochlear implant system and co-inventor on several patents describing the design and development of cochlear implant electrode arrays.

William "Van" Harrison is currently Executive Chairman of the Board, Managing Director, Chief Technology Officer of SILERE Medical Technology, Inc., a Washington State neuro-stimulation company. He started as a design engineer at Hewlett Packard and his career has transcended over 30 years of pioneering product development and technology advances in

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

medical devices. As a vice president for 10 years at the Advanced Bionics Corporation, he led the engineering team that developed the HR-90k cochlear implant, which was regarded as the most advanced and sophisticated neuro-stimulation device ever developed and commercially distributed at that time. Prior to his tenure at Advanced Bionics, he co-founded and managed Acoustic Imaging Corporation, a highly regarded diagnostic ultrasound imaging company.

Xiaoan Sun received his Ph.D. in Electrical Engineering from Southern Methodist University, Dallas, Texas. He joined the Research Triangle Institute, Research Triangle Park, North Carolina as an electrical engineer in auditory prosthesis research. His work in RTI included signal processing strategy, digital signal processing hardware and software development for cochlear implants, cochlear implant research interface development, safety protection hardware development for direct current stimulation of the inner ear. Since 2006 he has been the technical director of Nurotron Biotechnology Inc., Irvine, California. He has worked on circuit development for cochlear implants, transcutaneous power and data transmission, high efficiency RF power amplifier, RF receiver, system design and verification.

Haihong Feng received the B.S., M.S. and Ph.D. degrees in Underwater Acoustics Engineering from Harbin Engineering University, Harbin, China, in 1988, 1991, and 1996, respectively. From 1996 to 1999, he was an Associate Professor at the Department of Underwater Acoustics Engineering, Harbin Engineering University, China. From 1999 to 2003, he was the Vice Director of National Key Laboratory of Underwater Acoustics Technology, Harbin Engineering University, China. From 2004 to 2008, he was the Director of Shanghai Acoustics Laboratory of Institute of Acoustics, Chinese Academy of Sciences (IACAS). His research interests include underwater acoustics engineering, cochlear implants, and speech signal processing.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

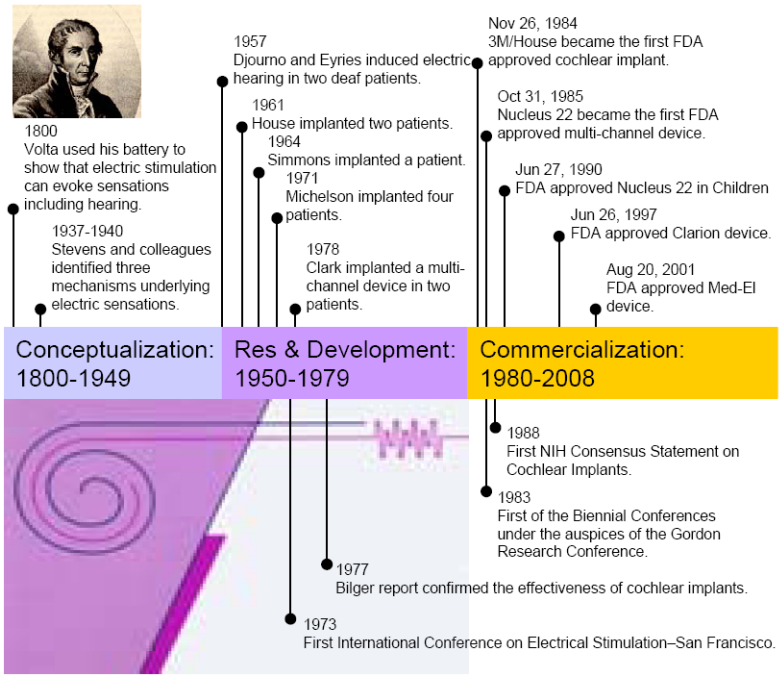

Figure 1.

Three phases defining the major events in the development of cochlear implants. The conceptualization phases demonstrated the feasibility of electric stimulation. The research and development phase legitimized the utility and safety of electric stimulation. The commercialization phase saw a wide spread use of electric stimulation in treating sensorineural hearing loss.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

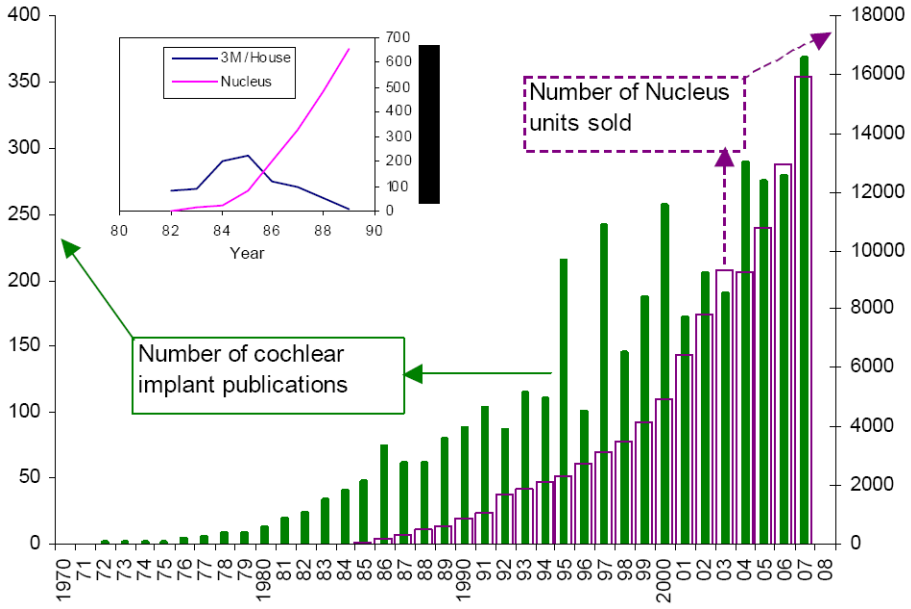

Fig. 2.

Exponential growth of cochlear implant research and sales. Note the 10 year delay for sales growth. The data of annual publications on cochlear implants (filled green bars with the unit on the left y-axis) were collected using keywords (cochlear AND implant) in PubMed (http://www.pubmed.gov) on June 19, 2008. The sales data (open purple bars with the unit on the right y-axis) were disclosed in Cochlear Annual Report (http://www.cochlear.com.au). Insert: Annual sale number of 3M/House single-electrode (blue line) and Nucleus multi-electrode (purple line) cochlear implants between 1982 and 1989 [25].

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

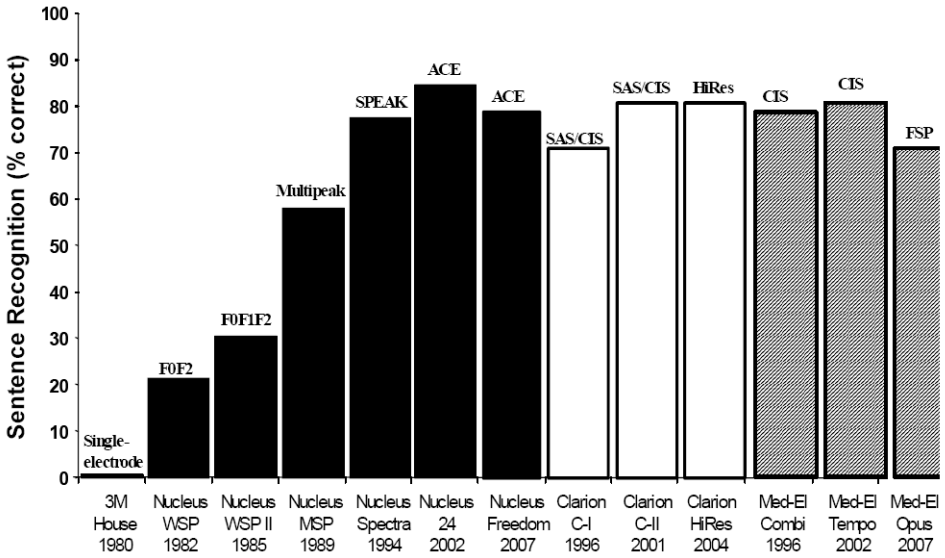

Fig. 3.

Sentence recognition scores with a quiet background as a function of time for the 3M/House single-electrode device (first column), the Cochlear Nucleus device (filled bars), the Advanced Bionics Clarion device (open bars), and the Med-El device (shaded bars). Previous results before 2004 were summarized in Zeng [40]. The latest results were obtained in the following references: Nucleus Freedom [41], Clarion HiRes system [42], and Med-El Opus device [43]. For a detailed description of the acronyms, see Section III.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

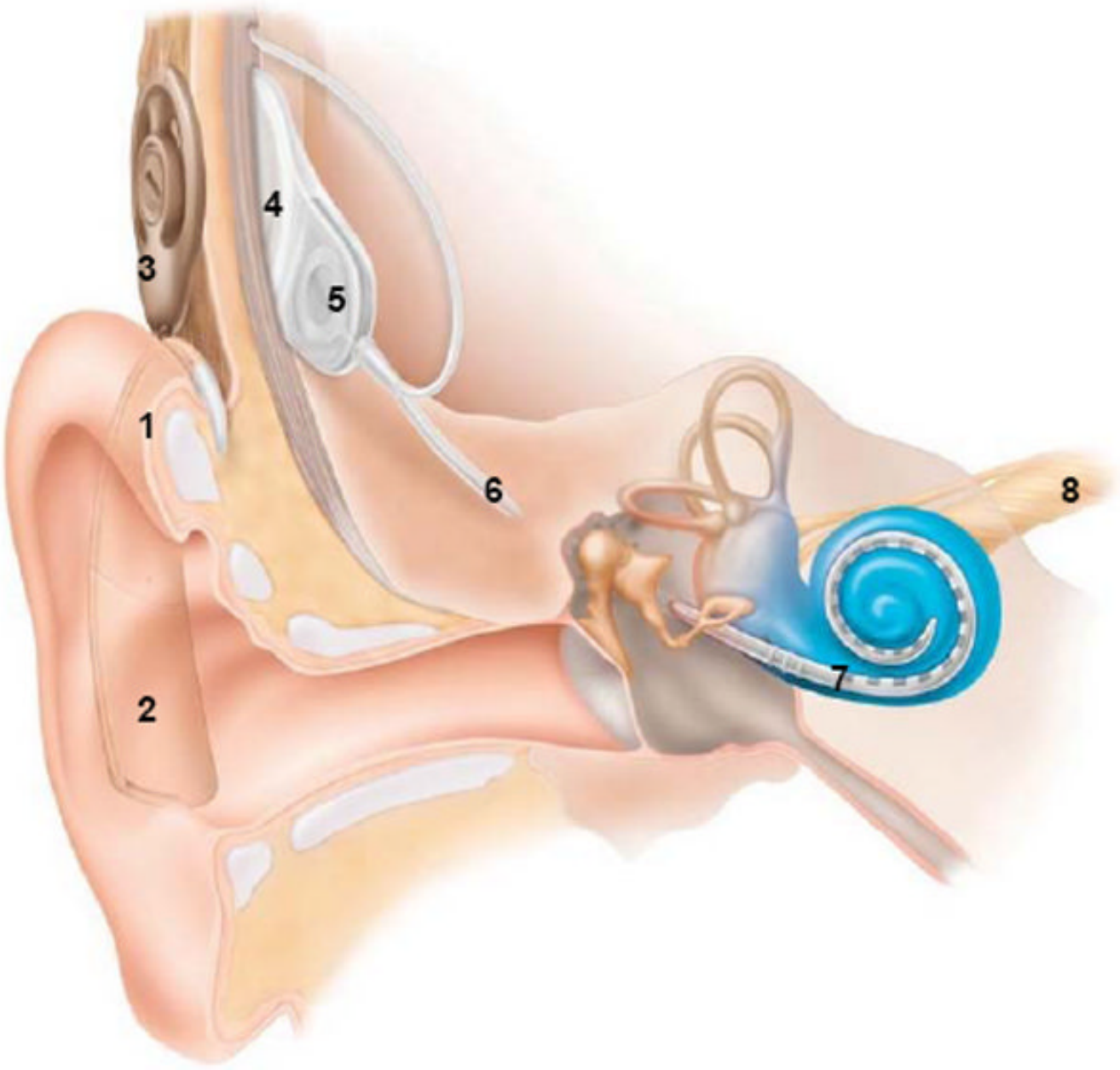

Fig. 4.
A typical modern cochlear implant system that converts sound to electric impulses delivered to the auditory nerve (www.cochlear.com).

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

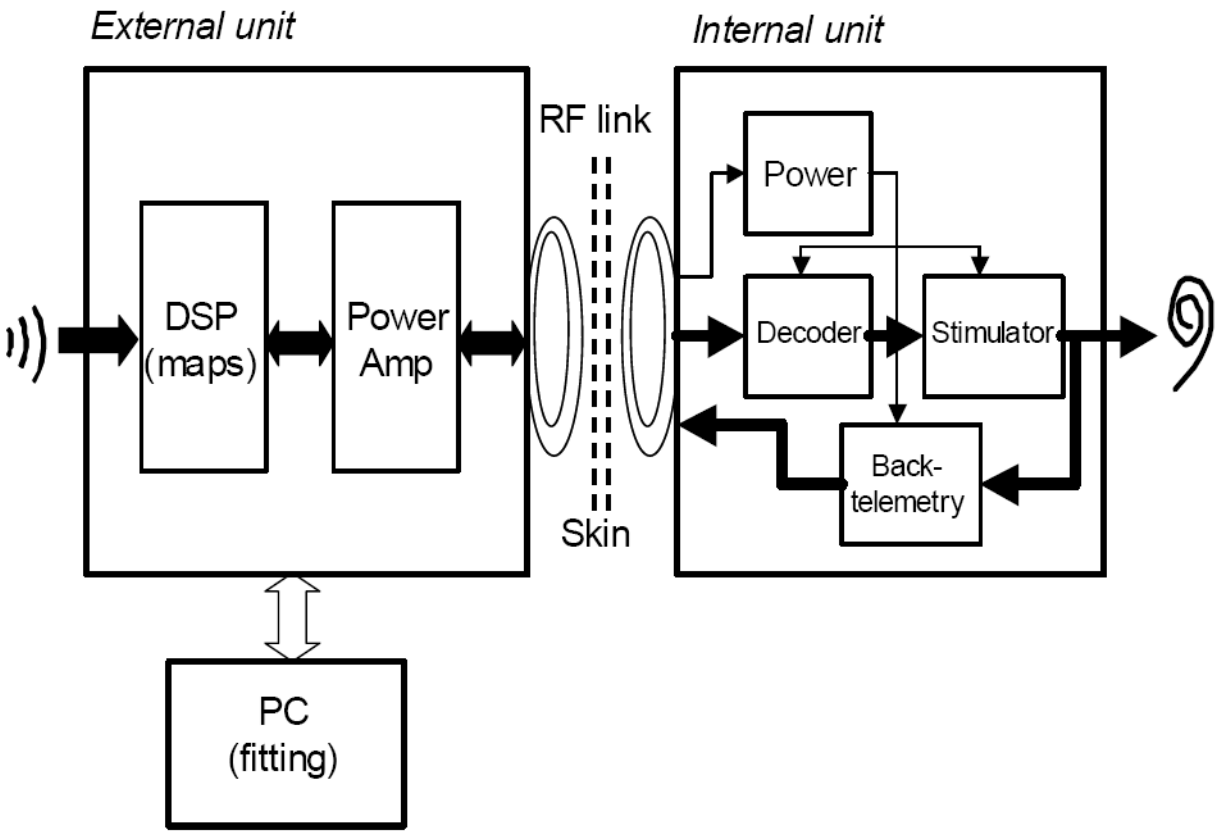

Fig. 5.

Architecture and functional block diagram of a modern cochlear implant. The single-electrode systems had an analog external unit and contained no internal active circuits [44]. The Utah six-electrode, four-channel system had a percutaneous plug and contained no internal circuits [22]. The Nucleus 22 device had essentially all components of a modern system, except for the back telemetry circuits [45].

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

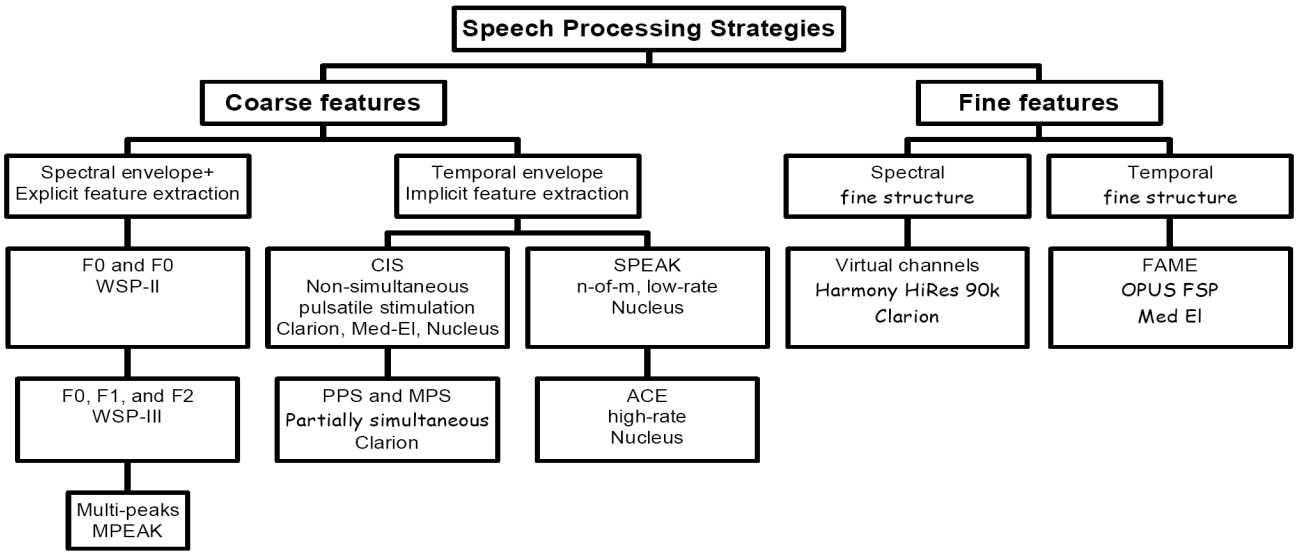

Fig. 6.
Classifications of signal processing strategies in cochlear implants.

NIH-PA Author Manuscript
NIH-PA Author Manuscript
NIH-PA Author Manuscript

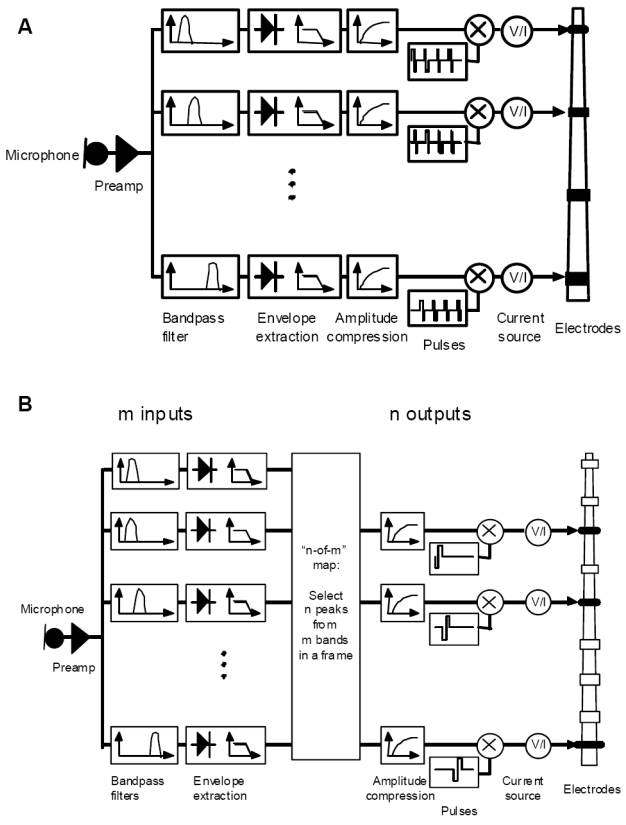

Fig. 7.

A. Block diagram and signal processing in the Continuous-Interleaved-Sampling (CIS) strategy. B. Block diagram of the “n-of-m” strategy.

NIH-PA Author Manuscript
NIH-PA Author Manuscript
NIH-PA Author Manuscript

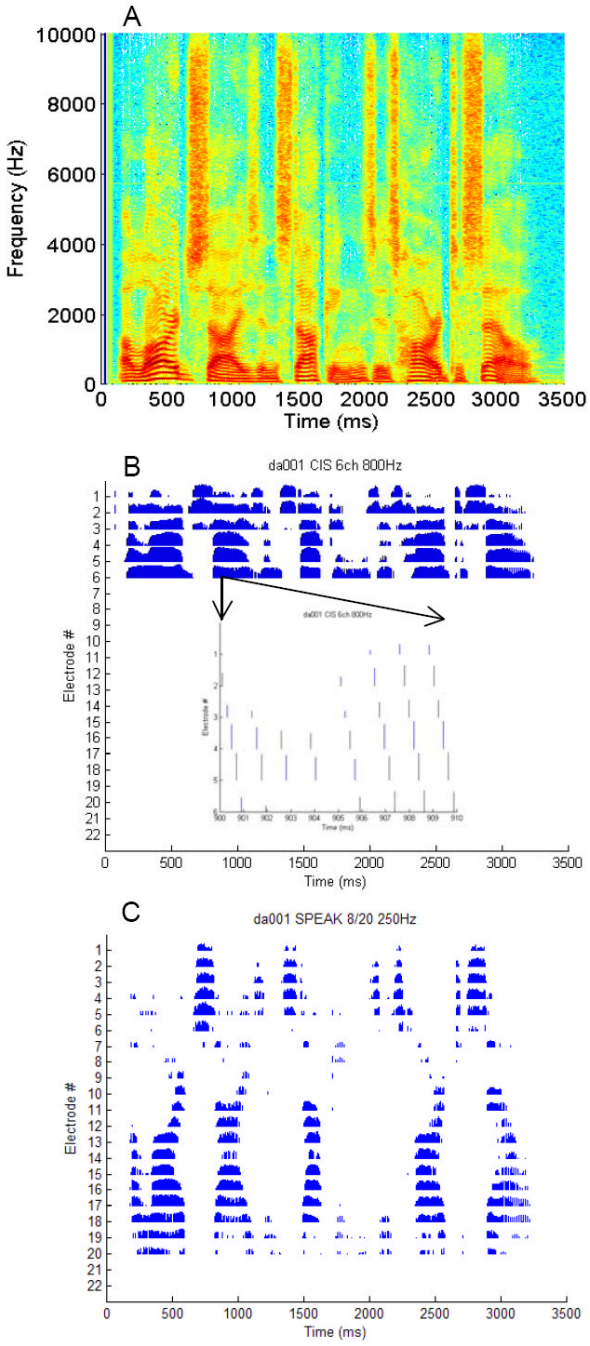

Fig. 8.

Spectrogram (panel A) and electrodograms of CIS (panel B) and SPEAK (panel C) for the sentence: “A large size in stockings is hard to sell”.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

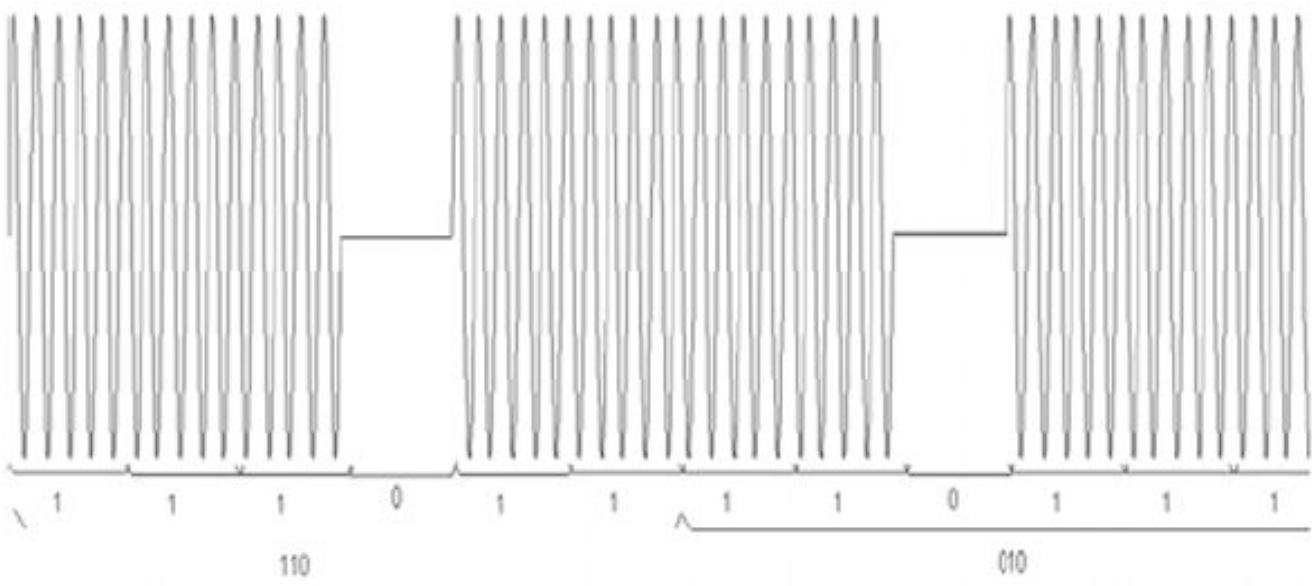

Fig. 9.

RF transmission coding in the Nucleus Freedom System. Waveform shows the original RF signal, with its amplitude being modulated on and off. The numbers below the waveform show the raw bits recovered from the RF signal. The raw bits are grouped into 6, coding the discrete data token.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

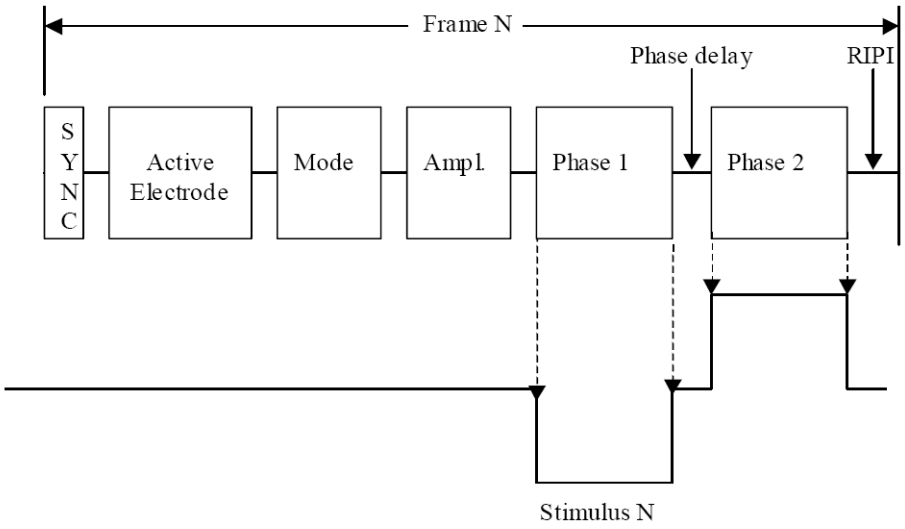

Fig. 10.
The expanded mode frame coding scheme in the Nucleus system.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

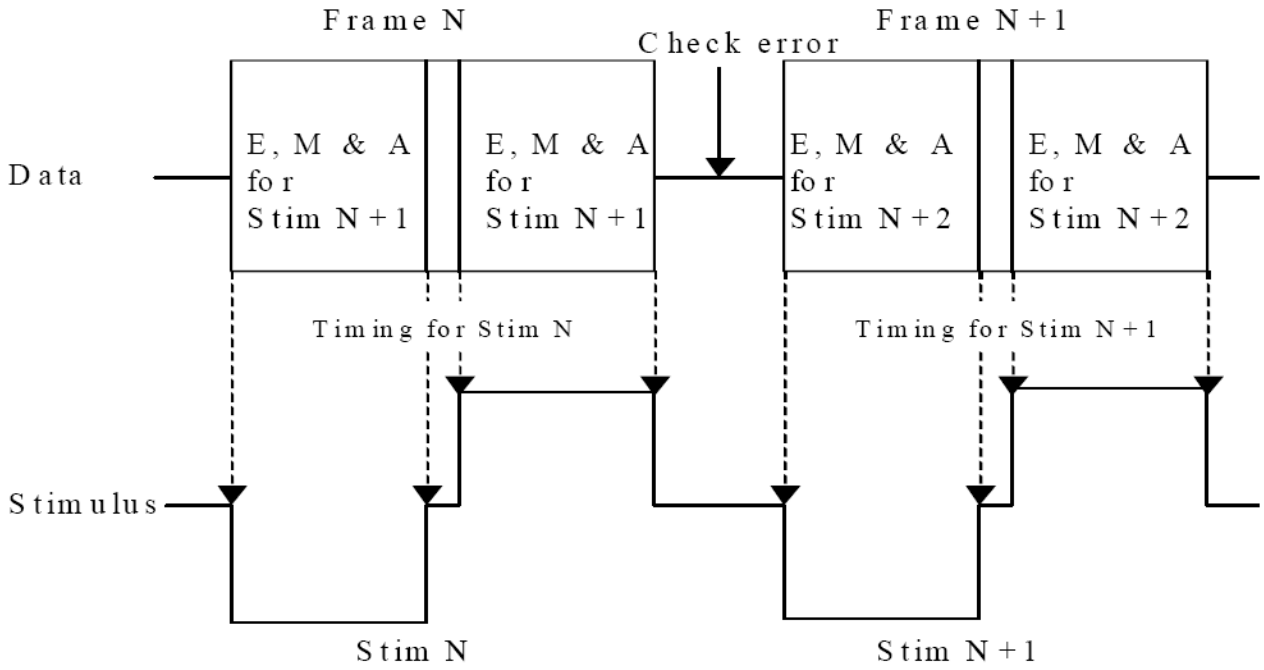

Fig. 11.
The embedded mode frame coding scheme.

NIH-PA Author Manuscript
NIH-PA Author Manuscript
NIH-PA Author Manuscript

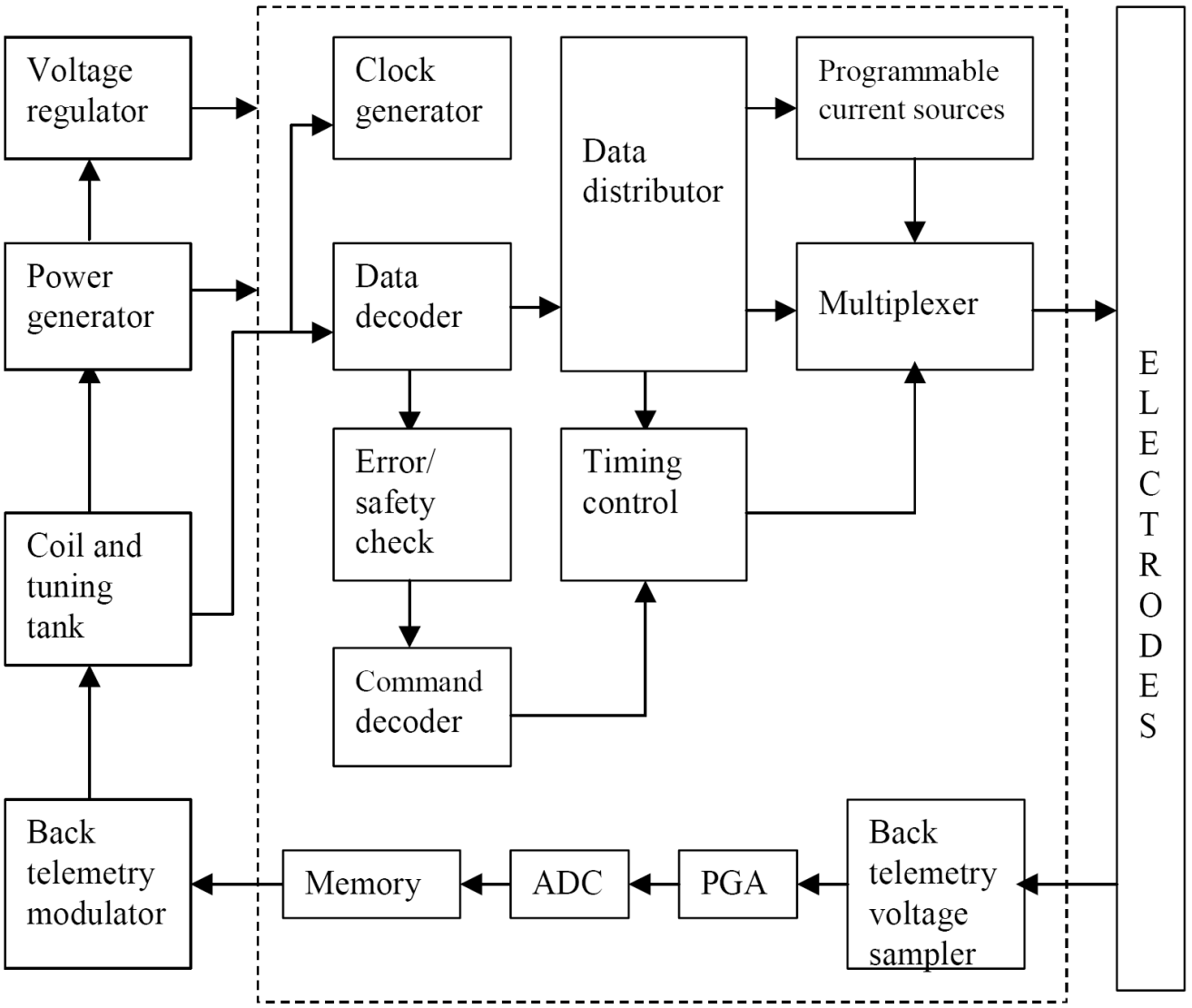

Fig. 12.
Block diagram of the cochlear implant internal unit.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

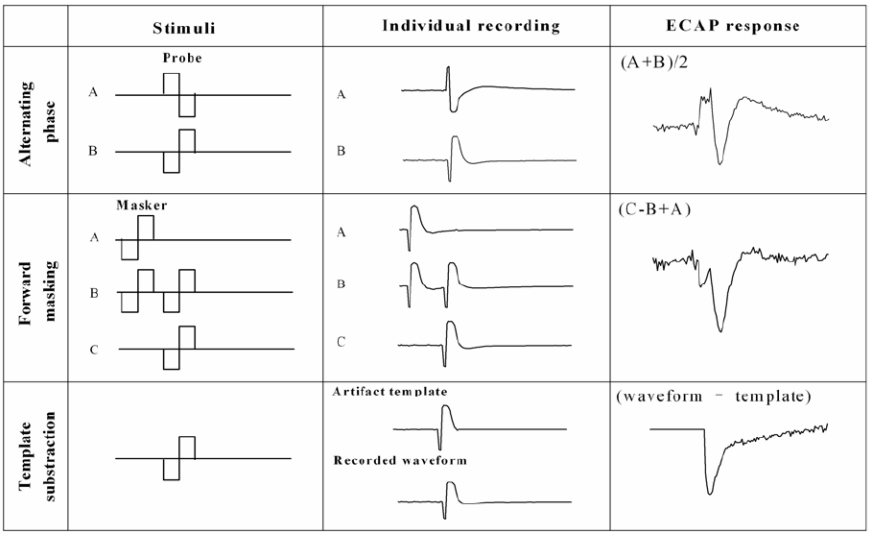

Fig. 13.

Three electric artifact removal techniques (three rows). The individual recordings and the derived ECAP responses were obtained in an actual Clarion HiRes 90k user (Qing Tang, personal communication).

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

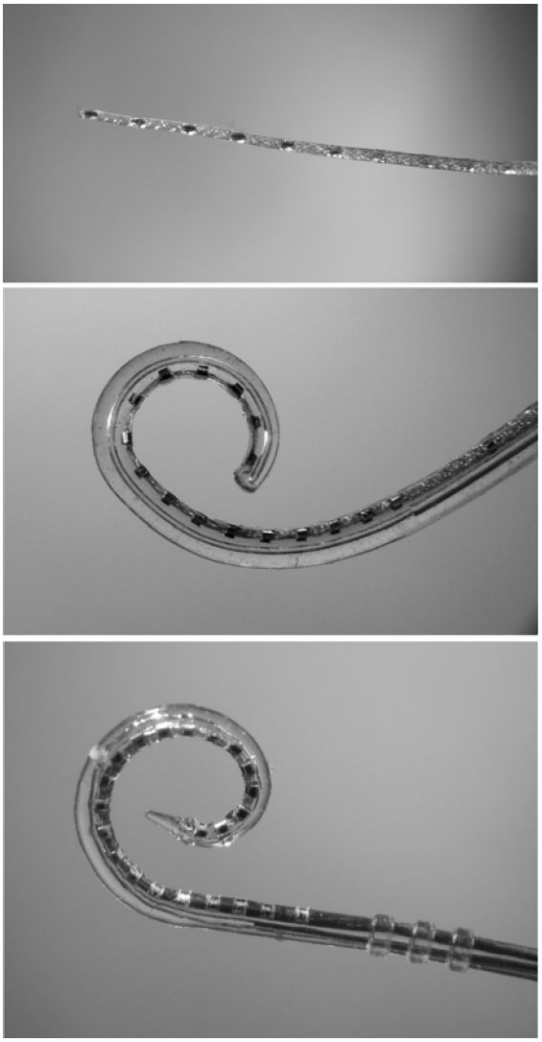

Fig. 14.
Top: Med-El Combi 40+™' Middle: Advanced Bionics Helix™ ; Bottom: Cochlear Contour™ electrode arrays.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

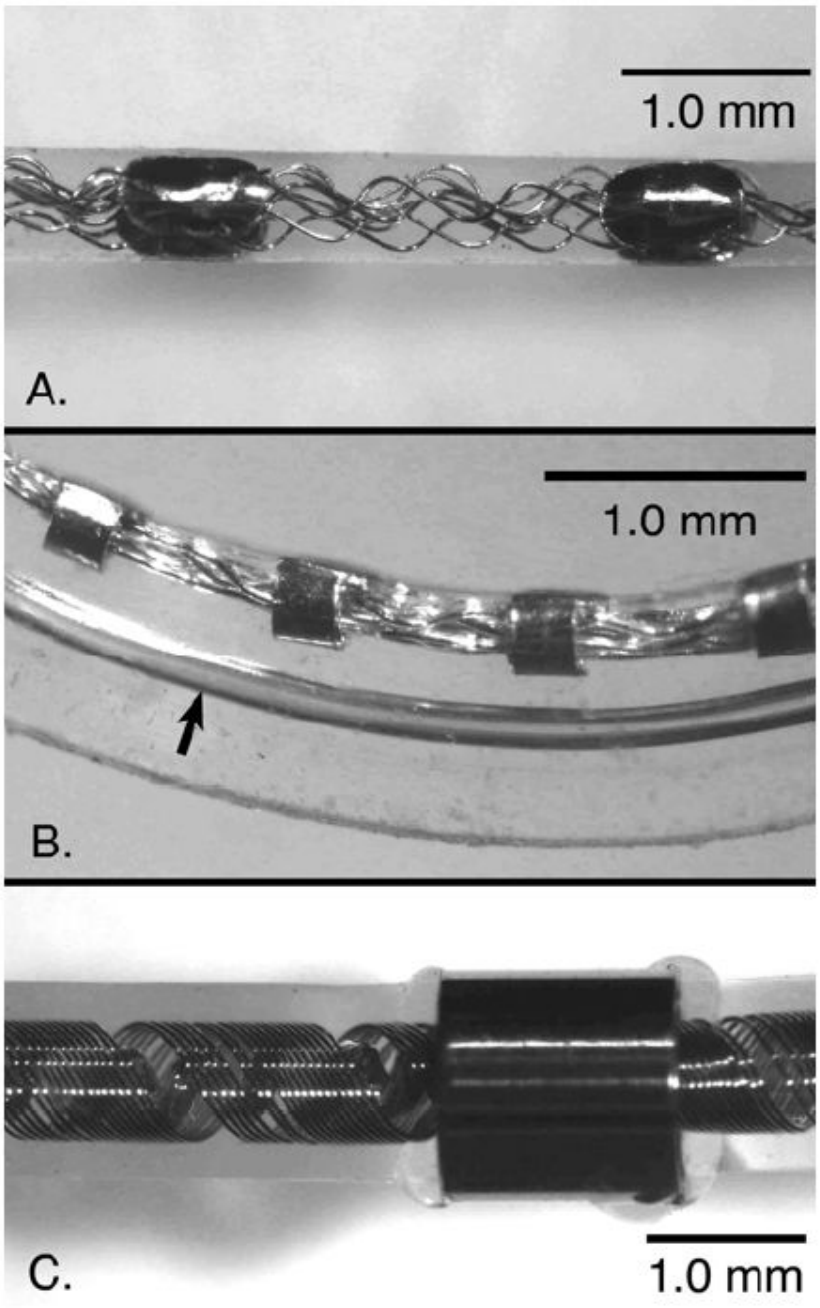

Fig. 15.

Panel A: Med-El Combi 40+™; Panel B: Advanced Bionics Helix™ ; and Panel C: helical winding connecting the implanted stimulator and the electrode array in the Advanced Bionics Helix™ and 1J devices.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

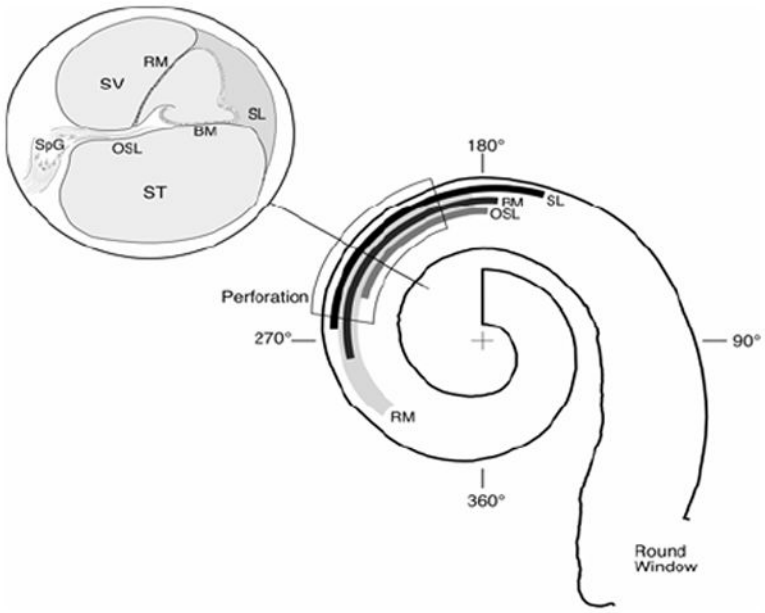

Fig. 16.

To define the initial site and extent of damage to each structure within the cochlea, we dissected and analyzed 13 epoxy embedded temporal bones with traumatic insertions of the Cochlear Contour array. The bars labeled SL, BM, OSL and RM represent the region of injury to the spiral ligament, basilar membrane, osseous spiral lamina and Reissner's membrane respectively. The location of each of these structures is shown in the inset cross section of the scala tympani. The outlined box between 180° and 270° indicates the region where the electrode perforated the basilar partition. The distance over which the electrode perforated the partition ranged from 7° to 185° with a mean distance of 64.7°. Beyond this perforation the electrode remained in either the scala tympani or scala vestibuli with minimal damage. SpG = Spiral Ganglion, ST = Scala Tympani, SV = Scala Vestibuli.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

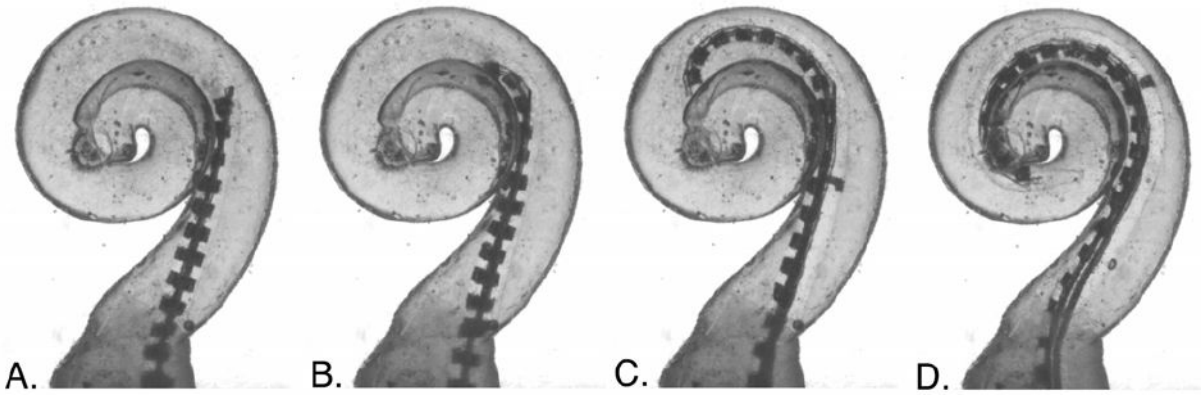

Fig. 17.

Images illustrating the intended insertion path of a pre-molded spiral electrode using the AOS technique. A Cochlear Contour electrode is shown as an example. The electrode array loaded on a straight stylet is shown in image (a) prior to advancing the array. As the stylet is held in position, the electrode is gently pushed off of the stylet and resumes its pre-molded shape (b-d). In this ideal case there is no contact with the lateral wall of the scala tympani. One possible mode of damage with this technique occurs if the electrode and stylet are advanced too far into the ST prior to advancing the array. In this case the straight stylet and electrode will contact the outer wall resulting in trauma similar to that of earlier straight electrodes. Studies at UCSF have shown that the distance from the RW to the first turn of the cochlea varies by more than 50% and that the marker molded in the Contour array may be well outside the cochleostomy when this contact and damage occurs in many subjects.

NIH-PA Author Manuscript
NIH-PA Author Manuscript
NIH-PA Author Manuscript

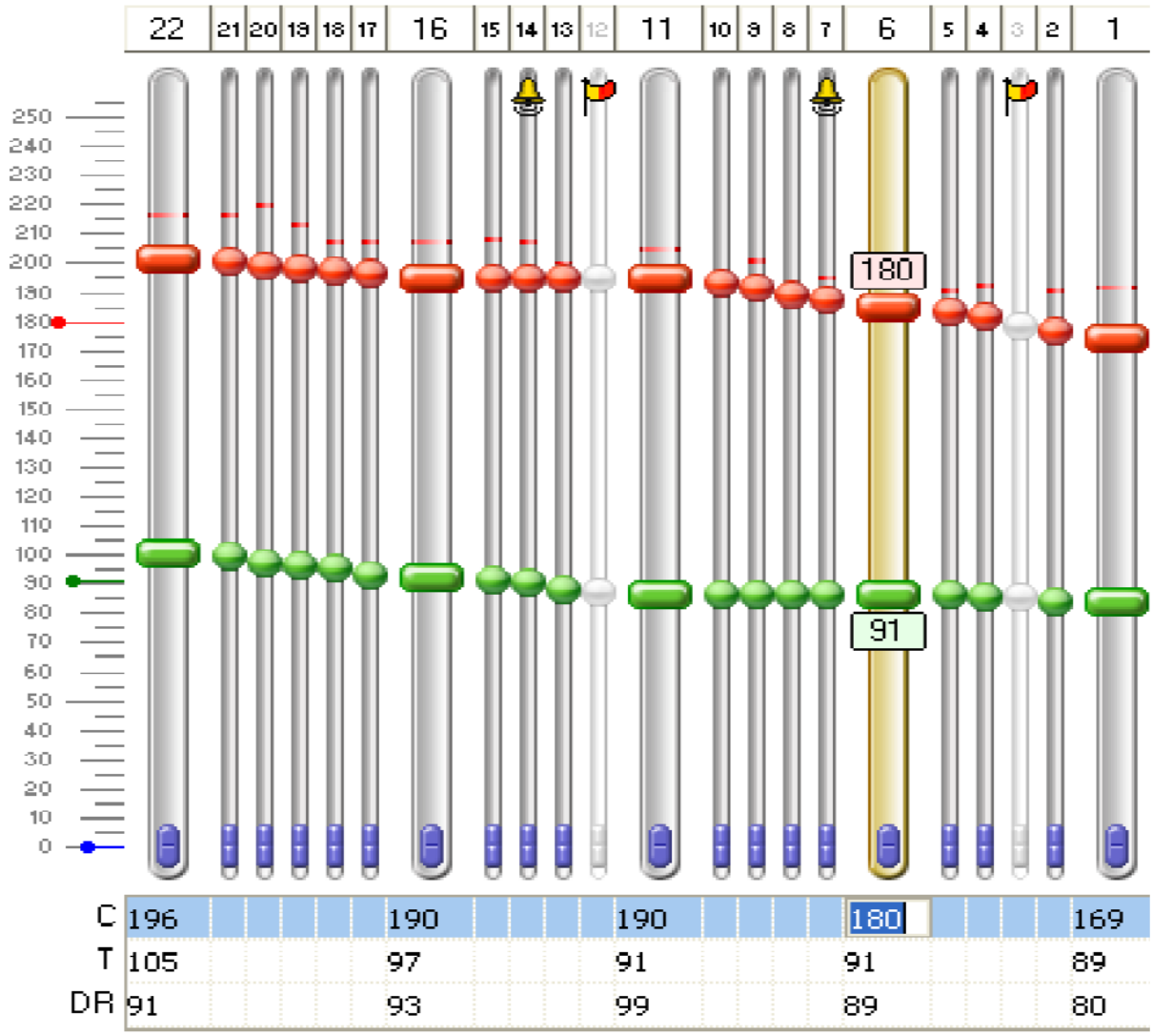

Fig. 18.
A section of Nucleus Freedom Custom Sound fitting program.

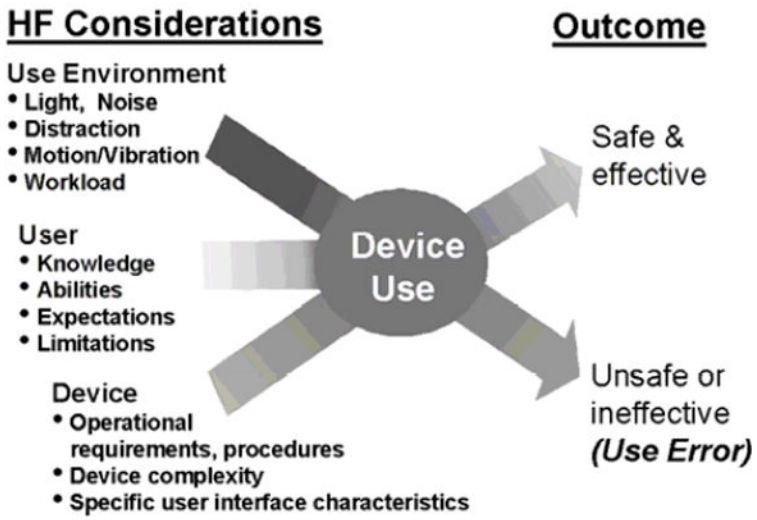

Fig. 19.
Interaction of human factors that results in either safe and effective use, or unsafe or ineffective use [from [188]].

NIH-PA Author Manuscript NIH-PA Author Manuscript NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

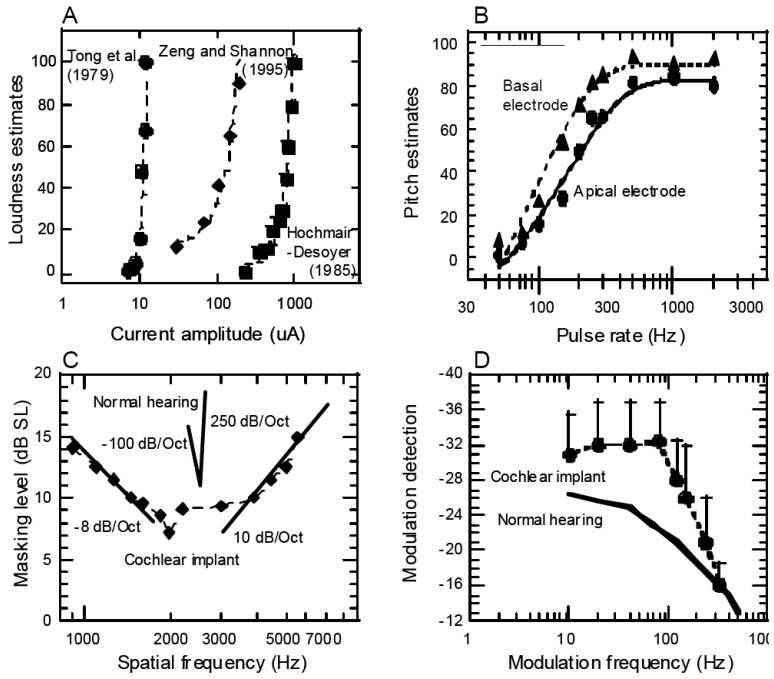

Fig. 20.

Basic psychophysical performance in cochlear implants: A. Loudness growth as a function of current level from three studies [67]; B. Pitch estimate as a function of electric stimulation rate (x-axis) and place (symbols) [51]; C. Spatial tuning curves in a cochlear implant user (squares superimposed with the two shallow lines) and a normal-hearing listener (steep lines without symbols) [195]; and D. Temporal modulation detection as a function of modulation frequency in an average cochlear implant user (circles) and an average normal-hearing listener (thick line) [38].

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

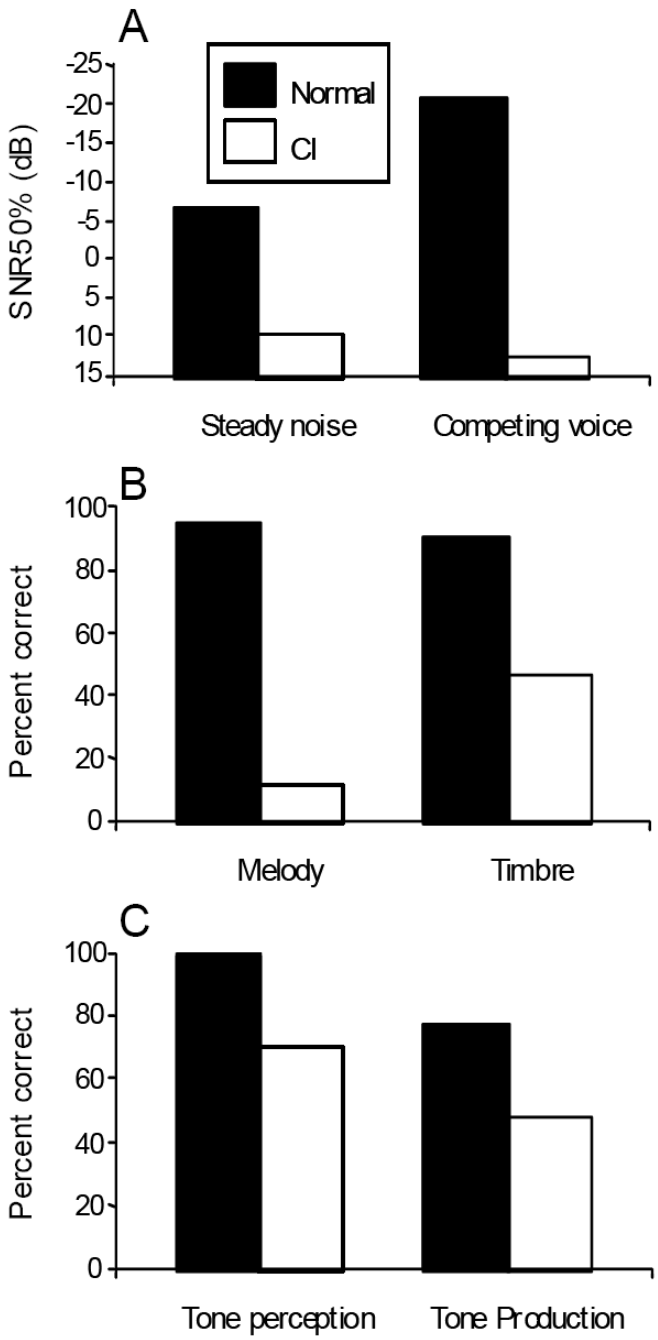

Fig. 21.

Challenges facing the present cochlear implant users (open bars); as a control, the comparative normal data are presented as the filled bars. A. Speech recognition in noise, showing signal-to-noise ratio (SNR) needed to achieve 50% correct performance in the presence of either steady-state noise or a competing voice [7]. Music perception, showing melody recognition [202] and timbre recognition [208]. C. Tonal language processing, showing Mandarin tone perception[205] and production[204].

NIH-PA Author Manuscript
NIH-PA Author Manuscript
NIH-PA Author Manuscript

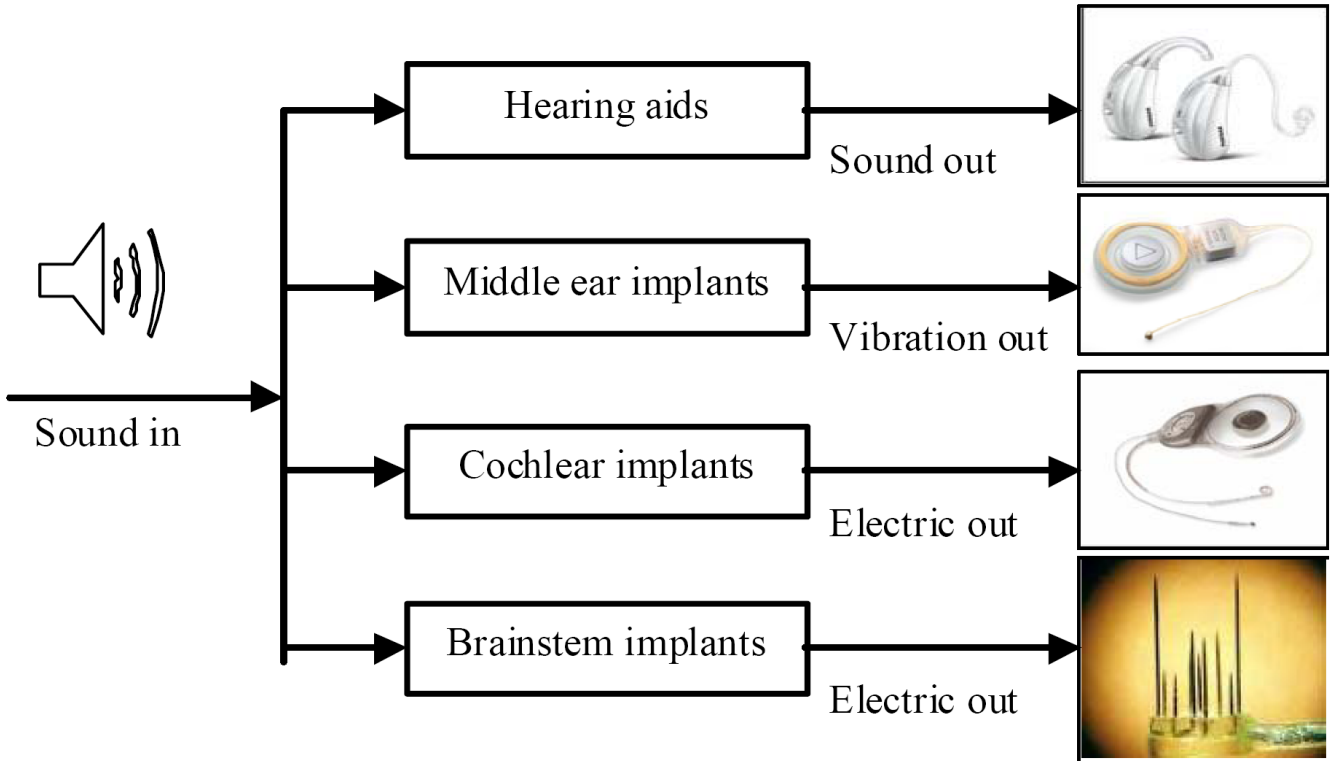

Fig. 22.

Treatment of hearing impairment using hearing aids, middle ear implants, and cochlear implants. The hearing aids shown are Exélia micro from Phonak (www.phonak.com). The middle ear implant shown is Soundbridge from Med-El (www.medel.com). The cochlear implant shown is Nucleus-24 from Cochlear (www.cochlear.com). The stimulation site of a brainstem implant may be cochlear nucleus or inferior colliculus. The brainstem implant shown is a penetrating electrode array that stimulates the cochlear nucleus (developed by Huntington Medical Research Institutes: www.hmri.org).

NIH-PA Author Manuscript
NIH-PA Author Manuscript
NIH-PA Author Manuscript

Table I

System and functional specifications of three major cochlear implant systems. Data Sources: Cochlear Freedom [46]; Clarion HiRes 90K (www.bionicear.com); Med El Opus processor and Sonata implant [47].

Components	Parameters	Nucleus Freedom	Clarion HiRes 90K	MED-EL MAESTRO
External unit	Name and key features	Freedom: Omni or direct mics 4 sound fields IDR (-75 dB) Freq range: 100-8000 Hz 3 Zinc Air batteries (3-5 days)	Harmony: Omni mic Dual-loop AGC IDR (20-80 dB) Freq range: 150-8000 Hz Lion batteries (14-24 hours)	OPUS2: Omni mic Dual-loop AGC IDR (-75 dB) Freq range: 70-8500 Hz 3 zinc-air batteries (3-5 days)
	Processing strategies	CIS SPEAK ACE	CIS MPS HiRes Fidelity 120	CIS+ HD CIS FSP
	Number of maps	4	6	4
RF link	RF carrier	5 MHz	49 MHz	12 MHz
	Data rate	0.5 MB/Sec	1MB /Sec	0.6 MB/Sec
Internal unit	Number of electrodes	22	16	12
	Number of current sources	1	16	24
	Current range	0 - 1.75 mA	0 -1.9 mA	0 -1.2 mA
	Total stimulation rate	32 KHz	83 KHz	51 KHz
	Simultaneous stimulation	No	Yes	Yes
Back telemetry	Impedance measure	Yes	Yes	Yes
	Electric field imaging	No	Yes	Yes
	Neural telemetry	Yes	Yes	Yes

NIH-PA Author Manuscript
NIH-PA Author Manuscript
NIH-PA Author Manuscript

Table II

RF data transmission specifications in three major cochlear implant manufacturers.

Manufacturers	Bit Coding	Modulation	Carrier Frequency	Data Rate	Additional timing for bit decoding
Nucleus Freedom	ON-OFF coding	ASK	5MHz	500 KBits/s	Needed
Clarion HiRes90k	Pulse width coding	ASK	49MHz	1.09 MBits/s	Not needed
Med El Sonata	Manchester coding	ASK	12MHz	600 KBits/s	Not needed

NIH-PA Author Manuscript NIH-PA Author Manuscript NIH-PA Author Manuscript

Table III

Design parameters for currently available clinical and experimental cochlear implant devices from the 3 major cochlear implant manufacturers.

	Current electrode arrays					Experimental electrode arrays		
Parameters	**Advanced Bionics HiFocus 1J**	**Advanced Bionics HiFocus Helix**	**Cochlear Contour Advance**	**Med-El Combi 40+**	**Med-El FlexSoft**	**Cochlear Hybrid**	**Cochlear Hybrid-L**	**Med-El FlexEAS**
Active Length	17mm	13.25mm	15.5mm	26.4mm	26.4mm	6mm	15mm	20.9mm
Total Length	20mm	20mm	25mm	31.5mm	31.5mm	6mm/10mm	16mm	25mm
Carrier Material	Silicon rubber (LSR-70)		Silicon rubber (LSR-30)	Silicon rubber (LSR-40)		Silicon rubber (LSR 30)		Silicon rubber (LSR-40)
Carrier Diameter (base to tip)	0.8-0.4mm	1.16-0.66mm 1.2-0.7mm	0.8-0.5mm	0.8×0.78mm at base 0.58×0.48mm at apex			0.35×0.25 mm at tip	0.8×0.78mm at base 0.58×0.35mm at apex
Number of Electrodes	16	16	22	12 pairs	7 basal pairs + 5 apical singles	6	22	7 basal pairs + 5 apical singles
Spacing	1.1mm	0.85mm	0.75mm	2.4mm	2.4mm	0.75mm	0.75mm	1.9mm
Shape	Straight	Pre-curved	Pre-curved	Straight	Straight	Straight	Straight	Straight
Stylet	No	Yes	Yes	No	No	No	No	No

IEEE Rev Biomed Eng. Author manuscript; available in PMC 2009 November 25.

NIH-PA Author Manuscript

NIH-PA Author Manuscript

NIH-PA Author Manuscript

Table IV

List of biocompatible materials and their applications.

Materials		**Applications**
Metal	Titanium	Case; Encapsulation
	Platinum	Electrode
	Iridium	Electrode
	Zirconium	Case
	Gold	Coil; Encapsulation
Non metal	Ceramic	Case; Feedthroughs
	Glass	Feedthroughs
	Silicone rubber	Carrier; Encapsulation
	Parylene	Insulation coating
	Teflon	Insulation coating