HEARING AIDS

About 10% of the population suffers from some degree of hearing loss (1). The largest group (about 45%) concerns people older than 65 years. The group of 25 to 45-year-old persons includes up to 42%. One can classify different degrees of deafness by considering that the lowest stage is that of people having difficulty in understanding speech in a group conversation or among an audience listening to a speaker. The next stage concerns individuals that have difficulty hearing direct conversation, while the third stage involves individuals having difficulty hearing over the telephone but can hear amplified speech. The most severe stage is associated with individuals that cannot hear speech under any circumstances. Assuming that this classification concerns people that have acquired the hearing defect after learning to speak language by ordinary means, today all of them can benefit of a hearing aid to help them overcome their problem. Other individuals that are born deaf or who have acquired severe deafness sufficiently early in life to prevent them from learning speech through the usual means require special education techniques with associated devices and means.

Throughout history, different devices have been invented to help hearing-impaired people. First versions of hearing-aid devices used in the 18th century were often tapers or horns that guide the sound toward the vertex or the point supplied with a small opening placed on the ear. In the beginning of the 20th century, the first electronic hearing-aids were designed, using the diode tube and the triode, and were based on the telephone principle. Since the 1950s the continuous progress in electronic and mechanical engineering has given rise to different wearable and miniature devices. In the 1980s, great advances in integrated circuit (IC) technology allowed the design of much smaller devices and under-the-skin implantable devices. Nowadays, the rapid progress in surgery, in IC technology, in material study, and in different other related fields have given access to very sophisticated devices for different degrees of hearing loss.

AUDITORY PATHWAYS AND THE HEARING PROCESS

The auditory system is one of the most wonderful and complex systems of the body. Witness its sensibility to the random motion of air molecules in contact with the eardrum and its capability to support sounds as loud as the noise of a jet engine, or its capability to distinguish sounds in a noisy environment as in a discotheque. The system can be divided into four basic parts: (1) external ear, (2) middle ear, (3) inner ear and (4) nervous pathways that leads information to the brain for interpretation.

External Ear

As shown in Fig. 1, the external ear includes the pinna, the ear channel, and the eardrum (tympanic membrane). The basic task of this part is to collect and guide sound waves to the middle ear. The external ear has a frequency response with a 10 dB to 15 dB gain over frequencies ranging from 1.5 kHz to 7 kHz and presents two resonant frequencies at 2.5 kHz and 5 kHz.

Middle Ear

The middle ear is a small cavity separated from the external ear by the eardrum and from the internal ear by a bulkhead comprising two apertures covered by a flexible membrane: oval window and round window. The middle ear cavity is connected to the throat by the Eustachian tube—a passage that allows maintaining the pressure equilibrium on both sides of the eardrum.

The basic function of the middle ear is to adapt the air impedance to that of the liquids of the inner ear. This adaptation is performed by a system of three small bones: malleus, incus, and stapes, transferring mechanical vibrations caused by the sound wave on the eardrum to the oval window. This ossicular chain gives a mechanical amplification ratio of about 27 dB associated to the lever system that it forms and the surface ratio between the eardrum and the oval window (2). Two small muscles are also connected to this mechanical system to prevent damages when in the presence of loud sounds. However, these muscles attenuate low-frequency moderated sounds, and then act as a high-pass filter.

Inner Ear

The inner ear consists of a complex channel system grooved in the temporal bone and filled by a liquid called perilymph. This bony labyrinth contains a membranous one, filled by another liquid called endolymph. This system includes three basic parts: (1) semicircular canals, (2) the vestibule and (3) the cochlea. The two first parts are responsible of the head position and the body balance, while the last one deals with hearing. The cochlea is shaped as a snail's shell, coiled upon itself for two and a half turns. Two membranes, called Reissner's membrane and basilar membrane, divide the cochlea into three parallel canals: (1) the cochlear duct (limited by both membranes), (2) the vestibular canal (separated from the latter by the Reissner's membrane), and (3) the tympanic canal (separated from the cochlear duct by the basilar membrane). The duct is filled by the endolymph and the canals contain the perilymph and they communicate at the top of the cochlea. The tympanic canal ends at the round window while the vestibular one begins at the oval window. Hence, the mechanical movements transmitted by the ossicles of the middle ear to the oval window membrane are converted into hydraulic pressure waves traveling through the vestibular and the tympanic canals and around the cochlear duct. This will consequently reproduce the mechanical movements on the round window membrane. The combined fluid displacements induce undulations of the basilar membrane, which supports a structure known as the organ of Corti, the most important element in the entire hearing mechanism. This element, lying along one side of the basilar membrane, contains the cells that convert hydraulic pressure into electrical impulses to be sent to the brain (3). These cells are supplied with exciting hairs

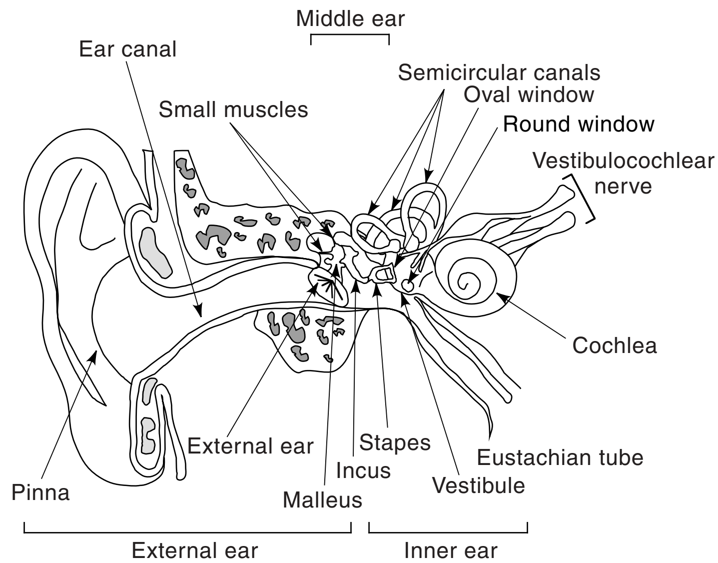

Figure 1. Structure of the human ear. There are three parts: (1) external ear, composed of the pinna, the ear canal, and the eardrum; (2) middle ear, containing the ossicles and the Eustachian tube; and (3) the inner ear, grooved in the temporal bone and mainly composed of the cochlea, which is connected to the auditory nerve fibers.

overhung by a flexible membrane called the tectorial membrane. Shearing action between the latter and the basilar membrane causes hairs to bend, and their cells then generate electrochemical signals for transmission by the auditory nerve to the central nervous system. Figure 2 represents a cross-section of the cochlea showing inner-ear structures.

Nervous Pathways

When the sound waves have been gathered by the external ear, and converted to mechanical energy, then to hydraulic pressure waves, and finally into electrical impulses by the middle and the inner ear, they are transmitted to the brain following the final link in the chain of hearing: the nervous pathways. As illustrated in Fig. 3, these pathways are enormously complex and contain different relay stations—cochlear nucleus, superior olive, inferior colliculus, and medial geniculate. There are also some descending nervous fibers, passing these relay stations and running from the brain back to the various parts of the ear. It is apparently the way that the brain directs partial or complete elimination of sound signals having no immediate importance.

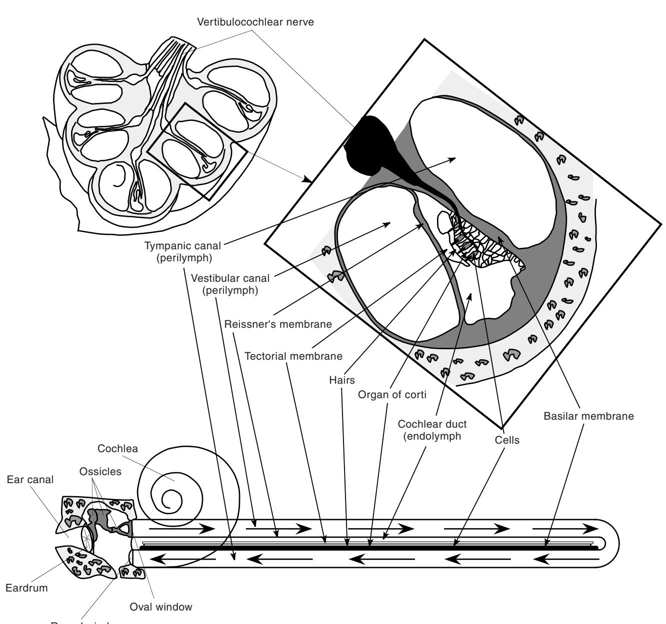

Figure 2. A cross-section of the cochlea. Shearing action between the tectorial membrane and the basilar membrane causes hairs to bend. As a result, an electrochemical signal is generated to travel along the auditory nerve toward the central nervous system.

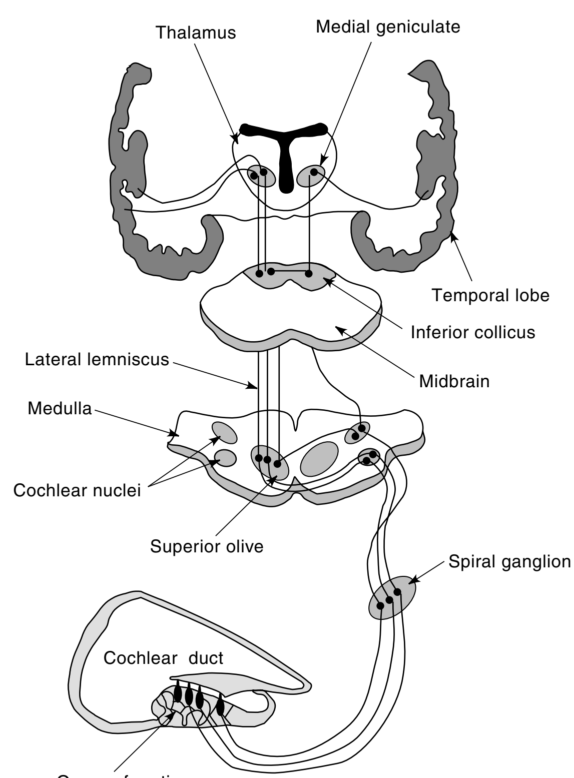

Figure 3. Nervous pathways of the auditory system contain different relay stations and involve ascending and descending nervous fibers.

HEARING DISORDERS

People with good hearing can detect a sound with intensity as low as 15 dB and tones at frequencies in the range of 16 Hz to 20 kHz. Figure 4 shows the threshold of hearing. The level of ordinary speech ranges from about 60 dB to about 80 dB over a frequency range running from about 100 Hz to nearly 8 kHz, with most used frequencies falling between 400 Hz and 3.4 kHz. The total deafness exists when sound cannot be heard at less than an average of 85 dB in speech frequencies (4,5).

At first thought, deafness may seem to be the simple inability to hear sounds of normal loudness. As a matter of fact, defective hearing also alters the quality of sound. In fact, speech sounds may be distorted, becoming muddy and unclear in a way that different words may be confused and indistinguishable or completely unintelligible unless their context makes them clear.

All hearing disorders can be classified into two categories: (1) conduction hearing losses and (2) sensorineural hearing losses. The first one is associated to the conductive structures of the ear, and so has its origins in the external and middle ears. Since these parts are dealing specially with amplification of the sound, the defects consist in a reduction of the sensitivity to all sounds independently of their frequencies. On the other hand, the sensorineural hearing losses arise in the inner ear or in the brain, as a result of a malfunction of some cells of the organ of Corti or some fibers of the auditory nerve or in the auditory cortex of the brain. Defects in these parts may affect hearing over all or a portion of audible frequencies.

Conduction hearing losses can have different causes as:

1. Blockage of the ear canal
2. Infection of the middle ear
3. Limitation in the middle ear bones movement due to some liquid accumulated or a muscle rigidity
4. Rupture of the eardrum that cannot heal
5. Bone growth freezing the stapes movements (otosclerosis)
6. Stiffening of incudo-stapedial joint (ankylosis)
7. Loss of incudo-stapedial joint (arrosion)
8. Loss of incudo-malleal joint
9. Loss, rupture, or fixation of ligaments

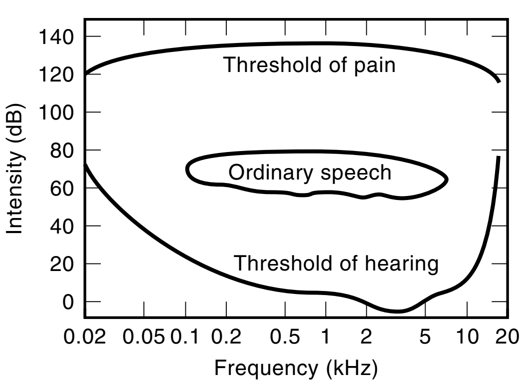

Figure 4. The threshold of hearing and pain. These thresholds vary with the frequency of the sound. The black region is the ordinary speech domain.

The second category of hearing impairments is related to the sensorineural defects. These can be caused by:

1. Illness, such as scarlet fever or meningitis
2. Sudden exposure to very loud noise, blast, or explosion
3. Long-term exposure to a noisy environment
4. Heavy drug use
5. Presbycusis
6. Congenital transmission

HEARING AIDS

Depending on the hearing loss class, different means can be used to remedy the situation. The remedy can consist of a simple cleaning, a surgically replacement of some of the middle ear bones by synthetic pieces, or having recourse to a hearing aid.

The technological advances that have taken place since the 1980s have provided numerous kinds of hearing aids. These device principles used depend on the degree of hearing loss. All of these corrective options can be classified as follows:

1. Sound amplifiers
2. Middle ear implants and bone conduction devices
3. Vibrotactile and electrotactile devices
4. Cochlear prostheses

Sound Amplifiers

The well-known sound amplifier placed in the external ear can be used only if the structures of the middle ear and the inner ear are not damaged. This kind of hearing aid is mainly composed of a microphone, an amplifier, and an output transducer (speaker). Its basic function consists of frequency shaping and amplification, in order to compensate for the deficiency in the individual hearing. Most of these devices are designed using traditional analog amplifiers. Today's advanced technologies have allowed reduction in size and power-consumption needs, to the extent that they are available in different molds, which fit behind the ear, in the ear at the pinna, or inserted in the auditory canal. This cosmetic consideration, making them more and more invisible, is the feature that increases their popularity in the deaf community. As a result, manufacturer's efforts have been invested in reducing device size rather than developing advanced signal processing, which would require a larger package. Thus these devices show many performance limitations that cannot be overcome without having recourse to modern signal processing techniques and sophisticated algorithms involving DSP.

The limitations in typical sound amplifier hearing aids can be summarized in restricted dynamic range, distortion, restricted bandwidth, and especially difficulty in understanding sounds in noisy environments. In general, the signal-to-noise ratio (SNR) needed by a hearing-impaired person to give speech intelligibility in noise comparable to that for speech in quiet is greater than that required by a normal hearing person (1).

The dynamic range in such hearing aid is bounded by noise at low sound levels and by amplifier saturation at high sound levels. A typical dynamic range is about 55 dB, which is about half the dynamic range of a normal ear. This can result in distortion for many sounds and can even affect monitoring of the user's voice. On the other hand, the high-frequency response of a typical sound amplifier hearing aid tends to fall off rapidly above about 5 kHz. This high-frequency value is insufficient for optimal speech intelligibility or music appreciation. However, increasing this frequency may give rise to distortion and can lead to feedback problems in the hearing aid. The latter is a major factor that limits the maximum gain and degrades the device frequency response. The problems related to the acoustic feedback depend on how the hearing aid is fit in the external ear and are most severe at high frequencies since this is where we find the highest gain (6). Some recent digital devices are using a feedback cancellation technique to allow increasing the gain of the device (7,8). This technique consists of estimating the feedback signal and to subtract it from the microphone input.

There are other techniques used to improve this kind of hearing aid. The so-called compression amplification is used to prevent amplifier saturation or to match the dynamic range of the amplified sound to that of the impaired ear. Currently used devices have up to three frequency bands on which one can apply this technique. At the same time, special attention is paid to minimize the attack time constants and the release times of electronic amplifiers to prevent their saturation (9). Many other techniques are also used to improve speech intelligibility by improving the SNR of different devices. However, results are still far from the ultimate goal, and improvements of the SNR didn't show significant improvements in speech intelligibility (10).

Middle-Ear Implants and Bone Conduction Devices

This kind of hearing aid can be used if the inner-ear structures are still viable. In this situation an implantable device is appropriate in the case where the hearing disorder cannot be remedied by a sound amplifier or when the latter shows very weak performances or medical contraindication. Among the problems that can be encountered with a sound amplifier and overcome by these devices are low fidelity, feedback, poor frequency response, and allergic reactions to the sound amplifier hearing aid packaging to be worn, such as ear molds (11,12).

The basic principle of the middle-ear implants consists of a controlled amplification of the ossicular chain movements. There exist two methods to achieve this. The first method uses a vibrator connected directly to the malleus or the stapes and the second one consists of controlling the ossicle movements by means of magnetic alloy pieces grafted to the ossicles. In either case electrical energy is transformed into mechanical energy.

The basic system of the first type of implant consists of a microphone placed in the external ear canal, an amplifier, a battery that can be transdermally charged, and a piezoelectric transducer (vibrator). The latter converts the voltage excitation into mechanical displacements (13,14).

The second way to control ossicle movements consists of using a coil implanted in the mastoid region near the middle-ear cavity to create an electromagnetic field that vibrates the ossicle supplied by the magnetic alloy piece. In this device an external unit including a microphone, an amplifier, and a battery, generates the signal that will control the mechanical movements of the magnetic alloy piece. This signal is conveyed to the implanted part by using a transformer with its secondary coil installed under the skin behind the ear in the mastoid region and connected to the coil responsible for magnetic alloy piece oscillations. These implantable pieces are totally biocompatible and do not necessitate any maintenance. They may be a wedge that will be fixed to the malleus, an incudo-stapedial ring, or a full incus replacement (15,16).

The other category of this type of devices is based on the skull bone conduction. The basic principle of these hearing aids can be summarized in the conversion of sounds into vibrations communicated directly to the skull of the device user. This is achieved by using a bone screw that is osseointegrated into the mastoid region or in some cases placed in the jaw. In general, a delay of about four months before using the device allows the osseointegration process to have sufficient stress transfer from the implant to the bone without progressive movements or abrasion (17,18).

As in the other systems, the device includes an external unit composed of a microphone, amplifiers, filters, power supply, and transducer monitoring the bone screw vibrations. In general, these vibrations are generated by a magnetic field varying according to the input signal. The transducer can be coupled to the bone screw directly or transdermally. In the case of a transcutaneous link a transformer with implanted secondary coil should be used to convey the magnetic field that will generate mechanical vibrations. However, for direct connection only one coil is sufficient to vibrate a piston attached to the bone screw.

This kind of hearing aid has shown performances up to 20 dB better than the best sound amplifier devices (19).

Vibrotactile and Electrotactile Devices

The basic principle of this kind of device is based on the fact that the skin is a spatially extended sense organ as is the eye and have some temporal processing capabilities similar to the ear. In fact, most of the qualities of the sensation produced when the ear is stimulated can be associated to similar counterparts in the skin sensation. Although these hearing aids are classified as assistive listening instruments, their performances demonstrated surprising results for sound and speech recognition when used alone (20,21).

Different schemes have been designed to be used as tactile hearing aids. They range from single-channel aids, providing minimal information on the sound and using simple processing of the speech signal, to multichannel ones, which incorporate frequency vocoders and use complex processing and sophisticated speech processing algorithms to bring out and provide as many sound features as possible (22). Regardless of the complexity of the device and the strategy of the signal processing, the aim of these hearing aids is to translate sounds into sensations to be perceived on any part of the deaf individual's body.

The tactile hearing aids can be divided in two classes. The first one includes vibrotactile devices, in which the acoustic signal is presented as a vibration to the skin by using a mechanical transducer. The second class involves electrotactile devices, in which the acoustic signal is converted to electrical current pulses stimulating the skin. One of the most important factors that must be taken into account is that the skin does not respond well to stimulation frequencies much above

600 Hz. Thus a special signal processing must be accomplished, to include high-frequency information of the sound inside this frequency band. Since the perception of sounds is no longer associated to the hearing system, this frequency transposition together with noise suppression techniques that can eliminate most of ambient background signal, give to these devices a great advantage over the bone conduction devices. According to these considerations, a simple system of a tactile hearing aid involves a microphone, power supply, amplifiers, filters, frequency transposition circuits, noise suppression circuits, and a transducer, which can be a mechanical vibrator or an electrode array delivering a current source electrical stimulus. Of course, a more complex system can include a DSP to perform more sophisticated signal processing strategies to extract different sound features. In either case the transducer is worn on the wrist and in some cases it can be located in other sites on the body as on the sternum or on the fingers.

To sum up, a tactile hearing aid may be able to provide deaf individuals with a lot of information on the sound as:

1. Onset and duration of the fundamental frequency of voicing and formant information
2. Rhythm of speech
3. Duration of a signal
4. Limited changes in loudness
5. Discrimination of common environmental sounds
6. Feedback regarding production parameters of the user's own speech
7. Complement to the visual sense (as in lip-reading), by coupling this skill with information regarding frication, nasality, and voicing

Despite all the benefits that can be accomplished by this kind of hearing aid, there has been no widespread use of them except for individuals who are blind besides being deaf. This is probably due to the fact that they are not using any part of the auditory system. This may be unacceptable for deaf individuals, especially when they are convinced that they can correct their hearing loss instead of looking for another means, different from that of a normal person, to hear. On the other hand, wearable tactile transducers may be reason enough to discourage deaf individuals from using these devices.

Cochlear Prostheses

As already mentioned, the final link in the chain of hearing is the nervous pathways. This means that, as in any other sense organ, the role of the ear consists in transforming the acoustic signal to electrical pulses traveling along nerve fibers toward the brain to be interpreted. The idea behind the use of cochlear prostheses is to deliver electrical pulses directly to the nerve fibers, according to the acoustic signal to be perceived. Thus these devices are intended for people suffering from sensorineural hearing loss and recognized as being totally or profoundly deaf.

Since this kind of device is designed to replace a major part of the hearing system, it's the most complex and the most expensive of all hearing aids. As Fig. 5 depicts, the basic constituents of the system can be summarized in a sound analyzer, a stimulus generator (generally surgically implanted under the skin), a communication link between these two parts, and an electrode array that delivers electrical pulses to auditory nerve fibers.

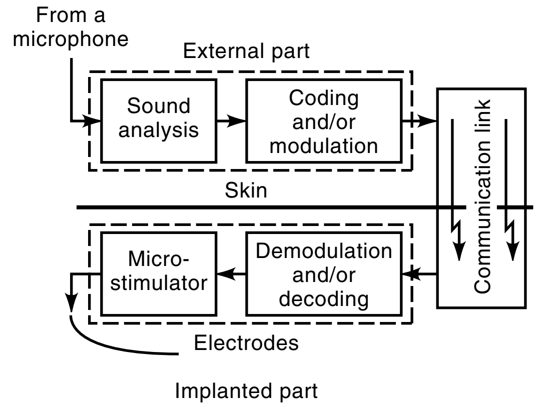

Figure 5. The basic constituents of a cochlear prosthesis. The sound is collected by a microphone, then processed to extract its characteristics. The appropriate actions to be taken by the implanted part will then be dispatched via the communication link.

Today, there exist different systems that are used worldwide and that are subject to continuous improvements. Besides using different processing algorithms, these systems differ, especially by their implanted part concept. The latter can be monoelectrode or multielectrode supplied by extracochlear, intracochlear, or modiolar electrodes, using a percutaneous plug or a transcutaneous link and delivering monopolar or bipolar stimulation. All this diversity arises from the fact that the stimulation techniques that should give maximum speech recognition are still ignored. Hence, each device is based on its designers' understanding of the human brain functions, and the way they try to emulate the external, middle, and inner ear functions. Regardless of the type of the device, results obtained by the same one with different individuals show considerable dissimilarities and complicate the situation by making each individual a unique case.

Cochlear prosthesis can be divided in two big categories: (1) monoelectrode/single-channel and (2) multielectrode/multichannel. The electrodes in both categories can be placed in different stimulation sites. When they are placed on the round window on the promontory or distributed over the cochlea, they are called extracochlear. In cases where they are inserted along the tympanic canal via the round window, they are called intracochlear. The last case corresponds to those impaled into the auditory nerve and called modiolar. Even though they can be justified by their contact with nerve fibers, modiolar electrodes are not suggested because of safety considerations since they damage the nerve and they risk affecting the facial nerve, which is very close to the auditory one. On the other hand, cochlear implants of both categories can either generate monopolar or bipolar stimulus. The first one is characterized by a common ground for all electrodes, that is, placed relatively far from the stimulation site. This mode can be used in case of small numbers of residual nerve fibers or hair cells since it spreads the stimulus over a large region. The bipolar mode consists of stimulating between two sites (positive/current source and negative/current sink) close to each other, which permits localization of the charge injection in a restricted region. With regard to the communication link between the sound analyzer and the implant, there exist basically two types. The transcutaneous one consists of an inductive link whose secondary coil is placed under the skin with the implant. The second one consists of a percutaneous plug and is less popular, since it exposes the individual to infection risks. However its transfer efficiency is far better than that of the first one. In both cases the communication link is intended to dispatch data and power to the internal part.

The category involving single-channel cochlear prostheses owes its success to the simplicity of its concept. Figure 6 depicts the basic block diagram of a design example of such devices, which achieved great success (23). This hearing aid uses a single electrode inserted in the tympanic canal via the round window and delivering an analog output signal that consists of a sine wave whose amplitude is overmodulated by the band-limited acoustic signal. It is a well-established fact that in this way, there is no means to achieve maximum speech discrimination nor of hoping to get further performance improvements. However, the quasi-absence of any signal processing and the very simple concept of the design allow a considerable reduction of the size of the device to the extent that it can be worn behind the ear as a sound amplifier hearing aid. On the other hand, their simplicity limits their costs especially with regard to the function they are intended to achieve. These two considerations make them very popular and increase their success.

The second category that includes multichannel cochlear implants offers much better performances and a lot of hope to reach optimum speech discrimination (24). This is due to their selectivity of the stimulated nerve fibers and the numerous possibilities of signal processing strategies that can be used for them. The most popular devices of this type are those using a powerful speech analyzer including a DSP, a transcutaneous communication link, and an intracochlear electrode array. However there is no consensus on the ideal number of electrodes needed. Because of their complexity due to the sophisticated operations that they are asked to perform and the different specialists (engineers, surgeons, psychologists, audiologists, speech-language pathologists, rehabilitation specialists, educators) involved to bring them into operation, these devices are the most expensive and necessitate a long rehabilitation time. Hence, their recipients should be carefully selected among the totally and profoundly deaf community. The selection criteria used by the majority of teams offering them are:

1. Being 18 years or older
2. Having acquired the hearing defect after learning to speak language by ordinary means
3. Being profoundly or totally deaf at both sides
4. Conventional hearing aids are not of any benefit
5. Being motivated to undertake a long rehabilitation program

In general, in multichannel cochlear prosthesis the microphone is located at the ear level to detect acoustic signals. Then the latter is communicated to the sound analyzer (external part), which processes it and dispatches appropriate control data to the implant (internal part) via the communication link. The implant consists of an electrical stimulus generator and is usually placed behind the ear, under the skin in the mastoid region. The generated stimulus is then presented to the electrodes, which are inserted into the cochlea via the round window. Since the nerve fibers ending in the organ of Corti, which is lying on the basilar membrane, are selectively excited depending on the acoustic signal perceived, the electrodes are distributed along the tympanic canal, close to this membrane. They are addressed according to the sound characteristics tending to emulate the effect of the combined liquid movements and the response of the hair cells.

It has been established that the stimulus should be a current waveform rather than a voltage one (25). This can be explained by safety considerations since the charge injected depends directly of the current level and thus can be better controlled. On the other hand, the current waveform used should be biphasic (completely balanced) to prevent any damage due to direct current accumulation and irreversible chemical reactions that occurs at the electrode–tissue interface. The most used current waveform consists of a rectangular pulse, allowing one to control the charge quantity by setting its amplitude and/or its width. To guarantee charge balancing, there's often a series capacitor connected to the current source output.

The ultimate goal of all cochlear implants is to find the ideal stimulation strategy that includes all of the speech signal features to provide all the information needed to the brain. The acoustic features of the sound waves have time and frequency specifications. The problem is that no one knows how these features should be presented to the inner ear, or what proportion of importance should be given to each one. Most of available speech processing algorithms are based on the three most important features: (1) peaks in vocal-tract transfer function (called formant positions), (2) vocal tract excitation rate (called pitch or fundamental frequency) and (3) the energy of the signal.

Two basic classes of sound analyzer are used for these devices. The first class adopts an analog approach to process and to extract the signal characteristics. An example of such processors is shown in Fig. 7. This one, called a compressed analog processor, uses automatic gain control circuitry to achieve the dynamic range compression of the sound. The bandpass filters extract the fundamental and the higher for-

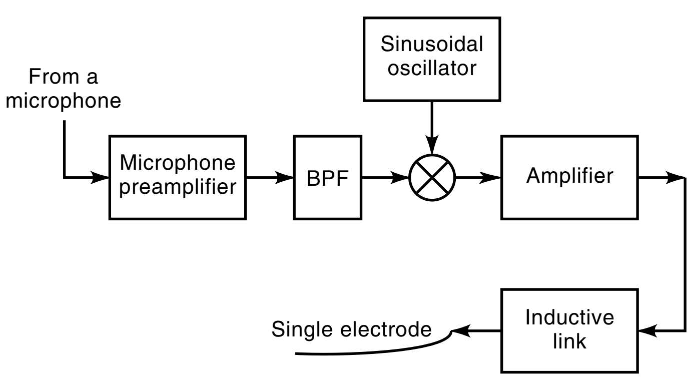

Figure 6. The block diagram of a single-channel cochlear prosthesis. The stimulus is an analog signal that consists of an amplitude overmodulated sine wave.

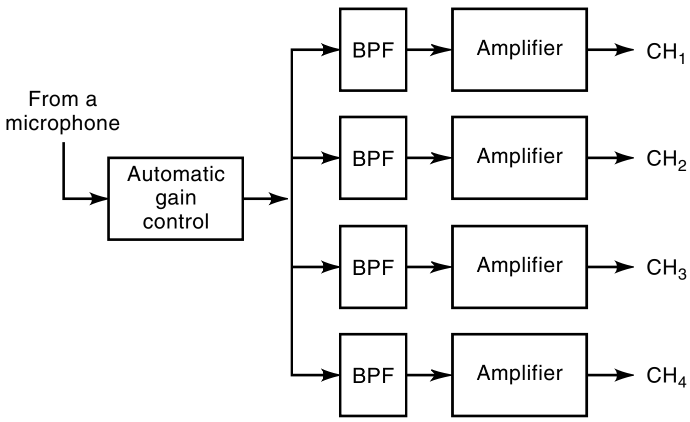

Figure 7. The block diagram of a compressed analog sound processor. The stimulus is delivered simultaneously over all channels.

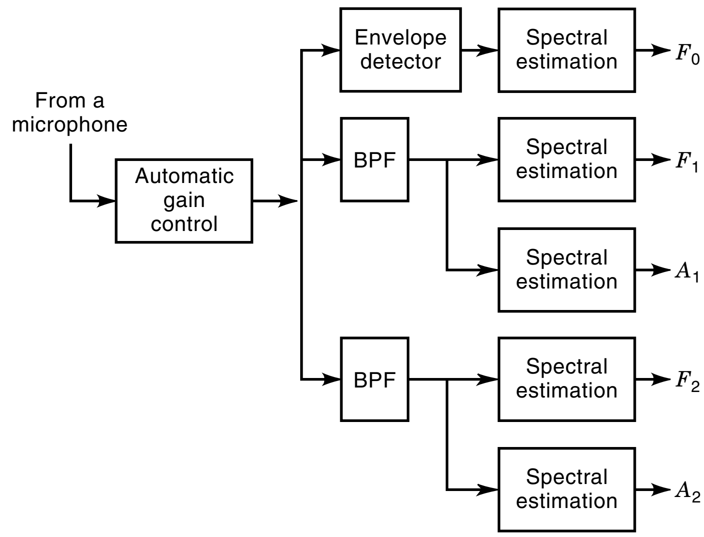

Figure 9. The block diagram of a feature-based sound processor. Only the fundamental frequency, the first formant and the second formant characteristics, are estimated.

mant information of the speech. The resulting signals are amplified and dispatched simultaneously to the cochlear electrodes (24,26). The problem that may be encountered when the stimuli are injected simultaneously is the interaction between the different channels inside the cochlea. Other processors propose an enhancement of this system to overcome this problem by using interleaved nonsimultaneous stimuli (26,27). These systems, called continuous interleaved sampling processors, have been designed with more than four channels to allow using more electrodes. Figure 8 shows such system. The half-wave rectifiers and the low-pass filters are used to extract the envelop of different signals collected at the output of different bandpass filters associated with each output channel. These envelop signals are then applied to a nonlinear mapping function to compress the dynamic range of the speech signal. Finally, each channel information modulates the amplitude of a biphasic current stream, including temporal offset to modulate the current pulse position of channel relative to each other.

The second class of sound processors is based on a digital approach. The system estimate some spectral features of the speech signal according to which the stimulus will be generated by the implanted part. Figure 9 shows an example in which the processor estimates the fundamental voice frequency (F_0), the first formant frequency and amplitude (F_1, A_1) and the second formant frequency and amplitude (F_2, A_2) (28,29). F_0 will be used to set stimulation frequency, F_1 and F_2 to address the affected electrodes and A_1 and A_2 to set the energy level. This concept has demonstrated some limitation,

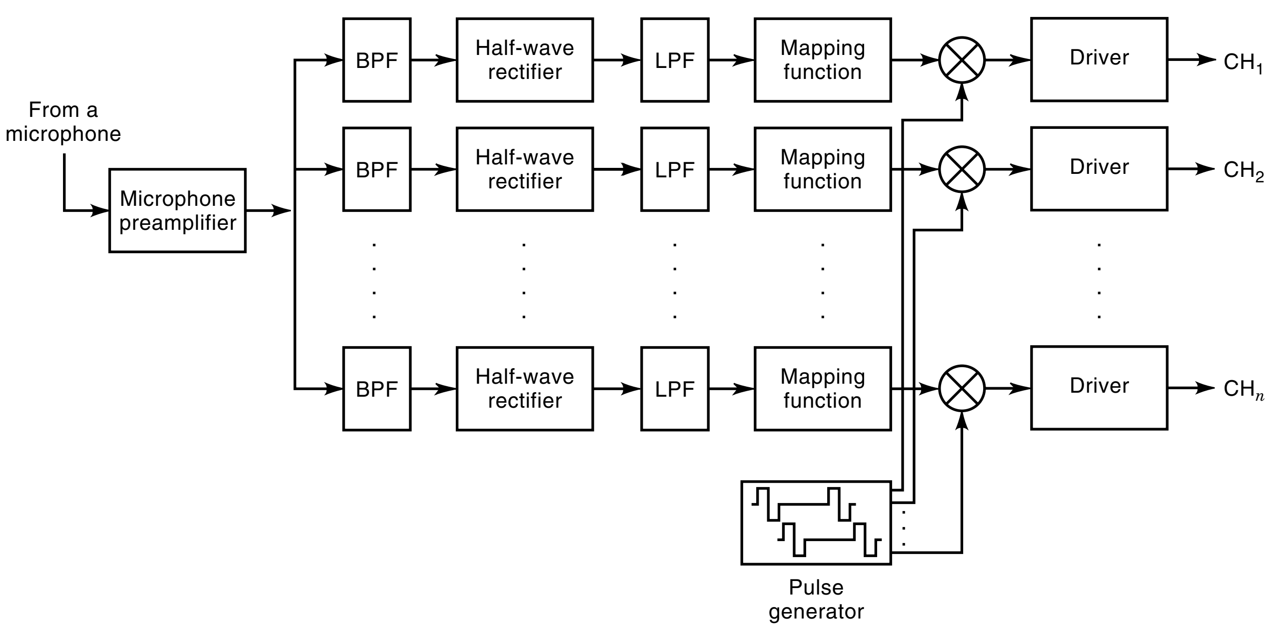

Figure 8. The block diagram of a continuous interleaved sampling sound processor. The current pulses are amplitude-modulated, including a temporal offset to prevent the interaction between channels.

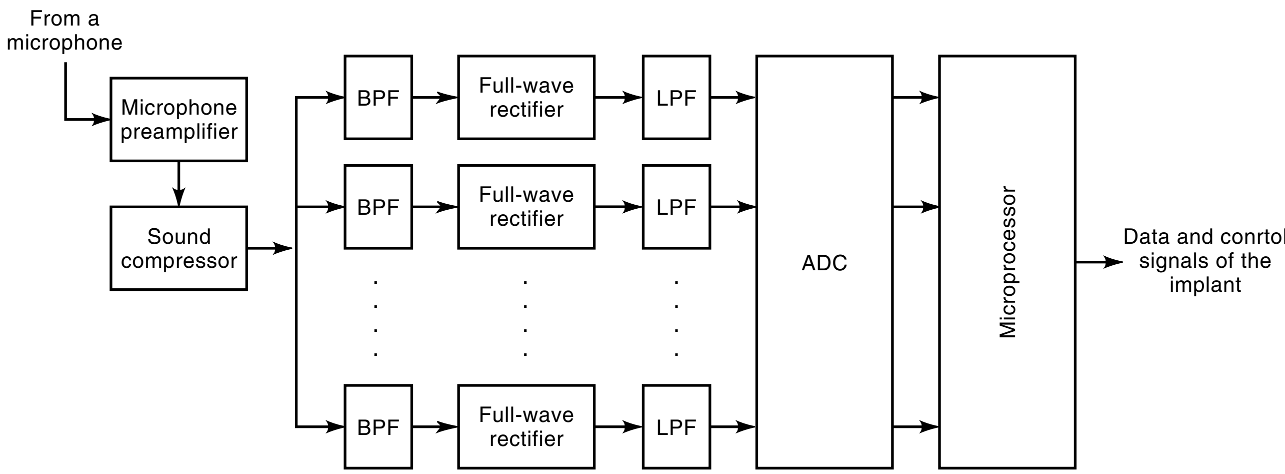

Figure 10. The block diagram of a spectral maxima sound processor. An advanced device involving a modern DSP.

since it is based on only a few number of speech features. Other alternatives have been proposed to improve this kind of processor by subdividing the sound frequency band into more subbands over which the sound spectrum is evaluated and then the most significant results over all the sound frequency band are used to generate stimuli. This kind of sound analyzer is called spectral maxima sound processor and is depicted in Fig. 10 (30). In this system, the dynamic range of the sound is compressed at the input, then the signal is passed through an analog signal processing part, consisting of filters and rectifiers, to be converted to digital data and to be processed by a powerful microprocessor.

It should be mentioned that, despite the complexity of these devices and the recent technological advances, there remains a lot of work to do to find the optimum stimulation strategies and to minimize their sizes. This will enable professionals to offer them to more deaf people including those born deaf and especially deaf children.

RECENT DEVELOPMENTS OF HEARING AIDS

The technological advances achieved since the invention of the transistor are progressing rapidly. The high-tech explosion of this era concerning all of the information technology systems is more and more present in daily life. As a result, a digital system revolution is extended by miniaturization, increase of circuit density and speed, and then by more sophisticated systems performing very complex computing and operations. This is directly reflected in the new and future hearing aids. The most affected systems are the sound amplifier devices and the cochlear prostheses. The former tend to extend their use by increasing their gains with better performance and the others should be accessible to more and other deaf people such as children, prelingually deafened, and even those who can obtain only marginal benefits from other amplification systems.

Advanced Sound Amplifiers Hearing Aids

Conventional sound amplifier devices that are using analog technologies cannot cope with intricacies of hearing and respond only partially to the needs of deaf people. The new digital audio techniques made available through modern integrated circuits can match more closely the individual needs and situation. One of the major capabilities of these new designs is programmability, giving access to up to 100 software-controlled parameters with regard to the maximum of 10 parameters available on their analog predecessors.

Most designers working on this kind of hearing aid concentrate on improvement of the feedback cancellation, compression amplification, and noise suppression, in order to increase their gains and to maximize speech quality. Besides working on the same old concept to improve it with digital approaches, there are other designers exploring other concepts such as by using microphone arrays.

The adaptive-feedback-cancellation technique, which consists of estimating the feedback signal and subtracting it from the microphone input, has shown 8 dB to 10 dB increase of the gain with digital designs (31,32). Furthermore, the latter allowed the implementation of multiband compression algorithms using different parameters to be chosen for optimum compression results. One objective of such systems is to place as much of the speech signal as possible within the residual hearing regions (33,34). A second objective consists in matching the loudness in the impaired ear to that in the normal one. All of these techniques have shown some improvements in speech quality (35,36). However, the major problem of intelligibility of speech in noise is still unsolved. Other means are used to treat this issue. In fact, instead of trying to reduce or to remove the noise, new techniques suggest enhancing the speech signal. This can be achieved by increasing the amplitude of spectral peaks while reducing the amplitude of the remaining frequency regions. This technique has not yet proven a significant enhancement of speech intelligibility, but many other techniques can be adopted, thanks to the flexibility and programmability of digital designs.

To overcome the problem of improving the quality of speech in noisy environments, other techniques based on focusing on the desired signal at the microphone input lead to use directional ones. This idea uses the fact that in many situations the desired acoustic signal comes from a single well-defined source while the noise sources are located throughout the area. Thus a directional microphone or a microphone array, built into an eyeglass frame, for example, is used to maintain high gain in the direction of the desired signal and to reduce gain for other direction sources (37,38). By using different possible signal processing strategies, these techniques showed up to 15 dB improvement in SNR. On the other hand, when these techniques are used with adaptive arrays, they can give greater improvements in SNR than when using fixed weight arrays (39).

Despite all of the research work undertaken to improve the quality of these devices, it still seems that because of the complexities of the auditory system and the nature of auditory impairments, advanced signal processing for hearing aids is a very difficult engineering problem. However, if deaf people can benefit even slightly from a new signal processing strategy, then this work is worthwhile.

Present and Future Directions for Cochlear Prostheses

One can say that cochlear prostheses have reached a robust childhood stage, since they have been shown to result in successful speech perception in profoundly deaf people. All of the specialists in auditory anatomy and physiology, otolaryngology, audiology, rehabilitation, education, speech-language pathology, bioengineering, psychology, and other related disciplines, as well as deaf people, are working together to extend their use and to make them accessible to a wider population of deaf people by establishing new selection criteria based on their success. The new digital processing techniques, together with the advanced integrated circuit technology, have allowed the implant and the sound processor sizes to be reduced, while at the same time increasing the power of the computing and processing strategies.

At the beginning, cochlear prostheses were intended for postlingually deafened adults. However, experiments of recent years have demonstrated that even prelingually and perilingually deafened people can benefit from these devices. Furthermore, experiments have involved a relatively large number of children with prelingual, perilingual, and postlingual deafness and have demonstrated that young people should be considered among cochlear prostheses candidates. The minimum age of the cochlear prostheses recipients was 24 months, but results showed that the most important factor that can affect the performance of such devices is the early detection of the hearing defect. This is justified by better results obtained with people having shorter duration of deafness. On the other hand, it seems that younger age implantation may limit the negative consequences of auditory deprivation. These results in children have been mainly reported for single-channel devices or feature-based devices only. Knowing the limitations of this type of devices, a lot of hope remains in using other more advanced devices, of course, by making them suitable for children.

Future developments of these devices concern all disciplines that are involved in their design stage, preimplantation stage, and postimplantation stage (40). Engineers should continue designing more flexible and powerful devices with respect to their safety considerations and by developing better signal processing and stimulation strategies to overcome the problem of noisy environments that still significantly detract from speech-perception abilities. Furthermore, since magnetic resonance imaging (MRI) is increasingly the diagnostic tool of choice for a variety of medical conditions, providing better resolution of soft-tissue structures, the material of cochlear implants should be chosen to be compatible and then not to contain any magnetic or ferrous metals. Some other design consideration, such as providing self-test capability in the device, should overcome the problems related to failure detection, especially in young children. This complementary circuitry does not need to be very complex; a simple detection of electrode failure such as open and short circuits may be of considerable help.

Specialists working with deaf people and involved in the preimplantation stage have much work to do to improve patient selection and future results. In fact, until now there is no residual hearing that is typically defined as profound hearing loss; moreover, the degree of preimplantation residual hearing does not predict postimplantation performances. Thus these specialists have to establish the critical distinction between the importance of residual pure tone sensitivity compared with that of overall residual auditory capacities. This should lead to well-defined audiometric criteria for candidacy, to better reflect functional auditory capacity.

The postimplantation working specialists are concerned with reeducation programs and performance evaluation. The biggest challenge for those scientists is to find a tailored rehabilitation program for every individual and to develop protocols to reflect therapies effective for various types of individuals receiving implants. This is due to the fact that therapeutic intervention may differ significantly in time and content for prelingually, perilingually, and postlingually deafened recipients. Studies of the relationship between the development of speech perception and speech production in cochlear prosthesis users must be continued. On the other hand, special programs should be established for children, depending on their ages and including educational programs, taking into account the suitable auditory and speech instruction using the auditory information offered by the implant. Language acquisition should be an outcome measure in young children. The combined speech perception, language production, and language comprehension rehabilitation is particularly challenging and should give better results after the childhood stage. More comparative studies on language development in children with normal hearing, children with hearing impairment using conventional hearing aids, and deaf children with cochlear prostheses should be conducted.

Other electrophysiological studies are also trying to solve the enigmatic side of the auditory system. Some of recent animal studies mention that electrical stimulation increases ganglion cell survival and also modifies the functional organization of the central auditory system. This issue should be developed to determine its implications in cochlear prostheses use.

All of these efforts should bring further and more precise explanations on the wide variation in performance across individual cochlear prosthesis users. At the same time, every specialist in a well-defined feature of the cochlear prosthesis system should continue to improve it. Taking advantage of new low-power and reduced-size components, some of these efforts may offer more compact sound processors to improve at least the size of this part of the system by making it better hidden and (Why not?) behind the ear or in-the-ear wearable.

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HEART, ARTIFICIAL. See ARTIFICIAL HEARTS AND ORGANS.
HEATING, ELECTRICAL. See INDUSTRIAL HEATING.
HEATING RESISTANCE. See RESISTANCE HEATING.
HEATING, SOLAR. See SOLAR HEATING.