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Encoding and Decoding of Auditory Signals in Relation to Human Speech and its Application to Human Cochlear Implants[1,2]

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Abstract. A description of the optimal electrode and stimulus configuration, based on mathematical model simulation of (1) the mechanical conduction and (2) the mechano-electrical transduction of the middle and inner ear, will be presented. It is assumed that a sensorineurally deaf patient, whose auditory nerve is at least partially intact, will be able to hear and understand, when an activation pattern in specific groups of remaining fibers of the hearing nerve can be reproduced in a manner equivalent to normal hearing. The placement of 2 bipolar concentric electrodes in the modiolus of the basal and middle cochlear turn in a deaf patient has provided the opportunity for testing of electrical hearing thresholds, differential thresholds, and electrode impedance changes over a 5-month period, with direct access to the electrodes through a skin window. The stimulator was housed in a pocket case to provide easy use in a daily environment. Bipolar pulses of 0.02–1.0 ms were triggered either by zero crossings or by a speech envelope controlled oscillator. It was found that loudness should be encoded by pulse amplitude or duration rather than by pulse repetition rate, since the charge per pulse was the primary variable of loudness sensation. The periodicity hearing allowed for electrical differential thresholds equivalent to or lower than the acoustical ones in normal persons in the 80- to 150-Hz region. For the optimal coding of speech, a multichannel system of 4–10 electrodes each for different nerve fiber groups will be necessary.

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Introduction

It is evident that the majority of studies aimed at the restoring of useful hearing by electrical stimulation of the auditory system in the deaf have remained mainly experimental [for reviews see *Ballantyne et al.*, 1978; *Bilger et al.*, 1977; *Dillier and Spillmann*, 1978]. The first human studies have been carried out primarily in California, initially at Stanford [*Simmons*, 1966], later at San Francisco [*Michelson*, 1971], at Los Angeles [*House and Urban*, 1973] and in Utah [*Eddington et al.*, 1978]. More recently, a number of groups became active in developing and implanting hearing prostheses in Australia [*Clark et al.*, 1977] and in Europe [*Chouard and MacLeod*, 1976].

One of the major long-range obstacles to the successful development of an electrical stimulation hearing aid is the *signal transformation* and the *encoding of the electrical signals* which are supposed to stimulate the nerve fibers. Until now, no such device meets the basic physiological requirements for speech sound encoding while fulfilling the needs of the deaf patient (external hardware case of pocket size) at the same time.

Method

Theoretical Aspects

Since a clear understanding of the physiological facts involved in hearing is a prerequisite for the successful electrical stimulation of hearing, the normal transfer mode of the auditory periphery has to be taken into account. A simple model of the signal transformation in the normal ear, based on the work of *Flanagan* [1962] with respect to the basilar membrane transfer function and on the work of *Weiss* [1966] with respect to the hair cell and neuron, has been implemented on a general purpose minicomputer (fig. 1). It consists of a number of parallel sections from three main blocks: (1) a place-dependent transfer function which accounts for the mechanical filtering of the middle ear and the tuning properties of the basilar membrane traveling wave; (2) a nonlinear place-independent transfer function to simulate the transduction of mechanical into electrical signals; (3) a stochastic threshold mechanism with refractory properties as a description of a single nerve fiber.

Each horizontal section (1–n) represents the activity at one specific location along the cochlear duct. The mechanical filter part implies the frequency selectivity of the middle ear and the basilar membrane, as revealed by mechanical tuning measures [*Helfenstein*,

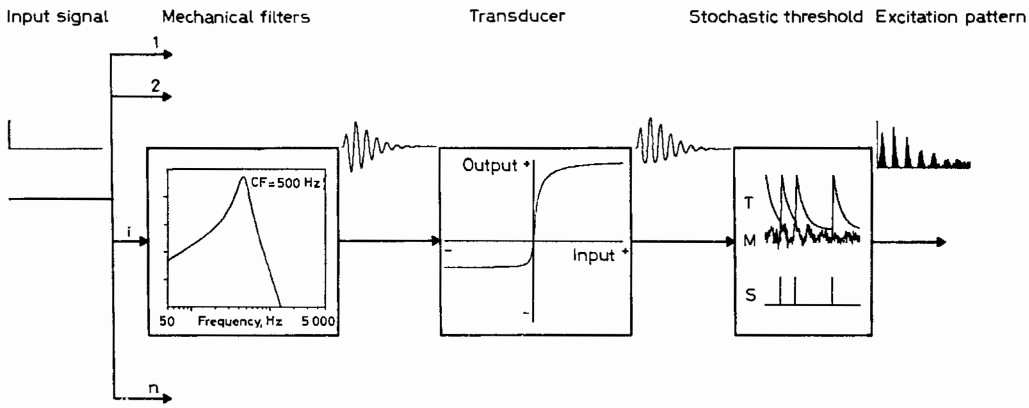

Fig. 1. Simplified model of the peripheral auditory system (refer to text for explanations).

1974; *Johnstone et al.*, 1970; *Rhode*, 1971]. The simulated input signal (ideal click pulse) is first transformed into a damped oscillation (impulse response). The second step, the transduction of mechanical into electrical signals, takes into account the nonlinear sensitivity of the hair cell [*Weiss et al.*, 1974]. Its output is an asymmetrical oscillation which represents the hypothetical presynaptic receptor potential. Gaussian noise is then added to simulate a stochastic synaptic transmitter release (M). The singal (T) is the hypothetical postsynaptic excitation threshold. After each spike (S) the nerve cell membrane becomes refractory. An exponential threshold decay is thought to occur [*Weiss*, 1966; *Biondi and Grandori*, 1975]. The excitation pattern of the hearing nerve can be displayed as a post-stimulus-time (PST) histogram in which the peak intervals still preserve the frequency of the chosen basilar membrane location (500 Hz).

Figure 2 shows the neural responses of different basilar membrane locations by ideal click stimulation. Clearly, the dependence of the peak intervals on the characteristic frequency is seen [*Kiang et al.*, 1965, p. 22]. Thus, the model seems to provide at least some approximation to the experimental data.

Although, on the other side, this model disregards many aspects of the complex actual situation (e.g., efferent innervation, differential role of inner and outer hair cells, bidirectional coupling of mechanical to electrical transduction, etc.), it may show the main features of the first processing steps of the auditory pathway. It can also provide some help in determining the transfer function necessary to produce an adequate excitation pattern with electrical stimulation.

The model responses for different intensities are shown in figure 3. An acoustical sine wave stimulation was simulated in the left part of the picture. The responses of a fiber

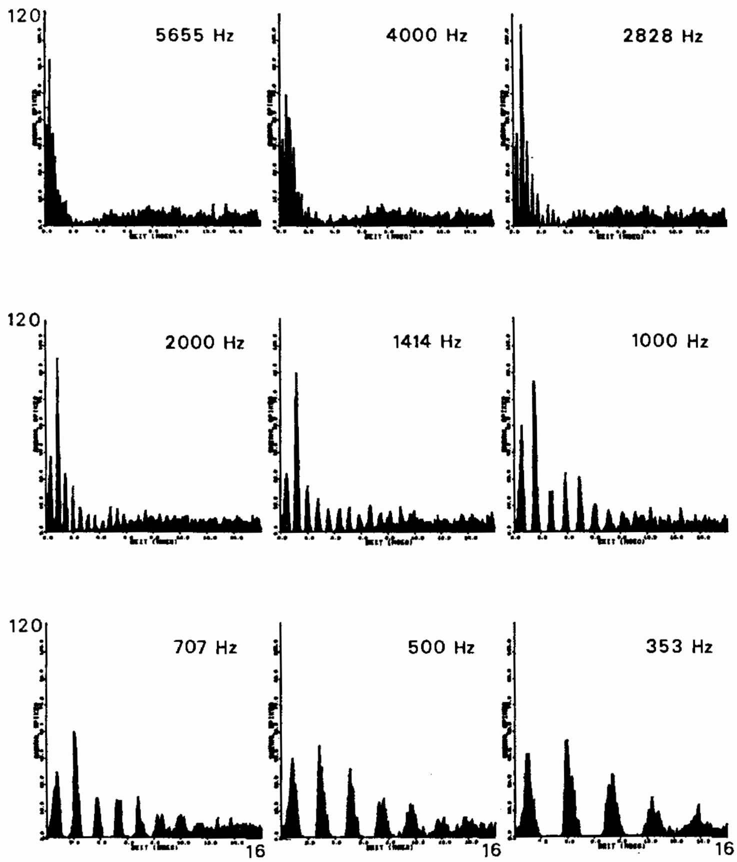

Fig. 2. Simulated click responses for different locations along the basilar membrane (indicated by their characteristic frequency in the upper right of each plot). Time scale 16 ms, Bin width 0.078 ms, 600 repetitions.

located at the 500-Hz region show wave-phase locked activity down to at least –30 dB. With electrical stimulation (right side of fig. 3), however, the dynamic range is restricted to less than 10 dB. Apart from the steeper slope of the input-output function, there is no mechanical filtering involved. Therefore, the frequency selectivity of the responses to electrical stimulation is practically absent. Since no synaptic transmission occurs, the spontaneous activity is missing and the PST peaks are much sharper than with acoustical stimulation, reflecting less jitter in phase lock histograms.

From these theoretical considerations, the signal encoding in the auditory periphery can be described in a first approximation as a short-time

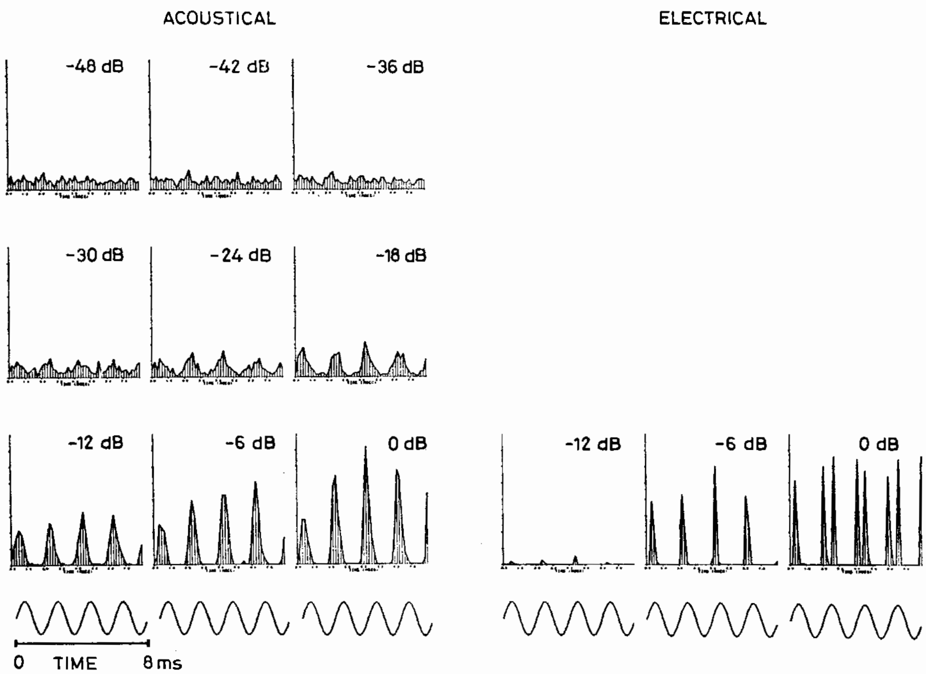

Fig. 3. Model responses for sine wave stimulation of different intensities (0 dB level was arbitrarily chosen at saturation intensity). Left side: acoustical stimulation. Ratio of noise standard deviation to resting potential for model neuron (s/R_r) = 0.45. Right side: electrical stimulation. Ratio s/R_r = 0.1. Time scale 8 ms, Bin width 0.156 ms, 600 repetitions.

spectral analysis by a set of n channels of narrow band filters followed by nonlinear compression. The output signals are each proportional to the firing probabilities of small nerve fiber groups at appropriate locations. The time course of these probability functions depends on the input signal and on the filter and nonlinear characteristics. Periodicities and amplitude changes of the input signal are represented in the firing probability (PST) histogram.

Problem Definition

To reproduce an optimal nerve excitation by electrical stimulation, the input signal would have to be split in the same n narrow band filtered signal channels which after nonlinear compression should be transformed into a coding pattern appropriate for auditory nerve stimulation.

Three problems need to be solved if an electrical hearing prosthesis is expected to work in a physiological way: (1) *signal transformation* (acoustic →electric); (2) *information extraction,* redundancy reduction; (3) *transcutaneous transmission,* nerve stimulation.

In the neurosensorily deaf patient, the neural prosthesis designer is confronted with the dual task of introducing a signal into the auditory system and of generating a pattern of neural activity which is meaningful to auditory perception.

This project is based upon the hypothesis that a sensorineurally deaf patient with at least partially intact nerve fibers and functional central auditory system will be able to hear when an activation pattern in the remaining fibers of the hearing nerve can be reproduced similar to the activation pattern in normal hearing.

In an attempt to apply this hypothesis to man, a series of questions must be answered:

(a) How many nerve fibers will be functioning in congenital or acquired deafness and how will the inner ear structures and the auditory nerve tolerate the implantation of metal electrodes and the subsequent electrical stimulation?

(b) Which activity pattern in the hearing nerve and at higher levels in the auditory pathway reproduces most closely the activity pattern of the acoustical stimulation?

(c) How can the tonotopical organization of the cochlea, i.e., the place principle of frequency detection, be artificially introduced, since periodicity hearing alone will not be sufficient to produce even a very restricted speech discrimination [*Bilger et al.,* 1977]?

(d) What is the optimal stimulus and coding scheme for a given electrode configuration?

(e) How can these systems be surgically placed at optimal locations with minimal damage to important structures?

(f) What constitutes a rehabilitation program for an implant patient which provides him with as much profit from the prosthesis as possible while helping him to learn to use the new sensory input?

Keeping in mind the complexity of the problem, we have defined as our immediate goal the development and evaluation of a practical hearing prosthesis which allows the profoundly deaf patient to reestablish contact with the acoustic environment. Using a limited number of channels, such a

Table I. Results of electrical stimulation in the outer ear canal (Brenner electrode)

Aetiology	Positive	%	Negative	Total
Toxic	5	100	0	5
Menière	3	100	0	3
Otosclerosis	3	100	0	3
Traumatic	3	100	0	3
Vascular	4	80	1	5
Congenital	7	54	6	13
Infectious	9	50	9	18
Tumor	0	0	1	1
Total	34	67	17	51

prosthesis will provide a coarse representation of normal sounds. Through extensive testing and study of the auditory performance in a small group of totally deaf patients, criteria for signal encoding should be identified which might enable second-generation multichannel prostheses to provide at least some speech intelligibility.

Practical Aspects

To obtain some information about the state of the hearing nerve in profound deafness, experiments with electrical stimulation through electrodes placed on the promontory and in a saline solution in the ear canal [Brenner electrode, *Stevens and Jones,* 1939] have been performed with 51 deaf and 10 normal-hearing subjects. The results are summarized, according to the underlying pathology, in table I. Although the number of cases in each etiological condition is small, a dependence of the test result on the type of the disease is likely to exist. Toxic deafness (n = 5), Menière's disease (n = 3), otosclerosis with loss of response to bone conduction (n = 3), and traumatic hearing loss by transverse fracture of the petrous bone (n = 3) range over vascular (sudden deafness, n = 4), congenital (n = 7), and infectious (meningitic deafness, n = 9) hearing loss. Overall, 67% of the deaf patients experienced hearing sensations by electrical stimulation in at least one ear. Figure 4 shows the areas of electrical sensation thresholds with sine wave stimulation of 50–12 000 Hz. The *electrophonic hearing* is characterized by a nearly normal electrical frequency discrimination. Electrode-tissue interface effects are thought to be responsible for this phenomenon [*Flottorp,* 1976]. Contrary to this type of response, which depends on the integrity of

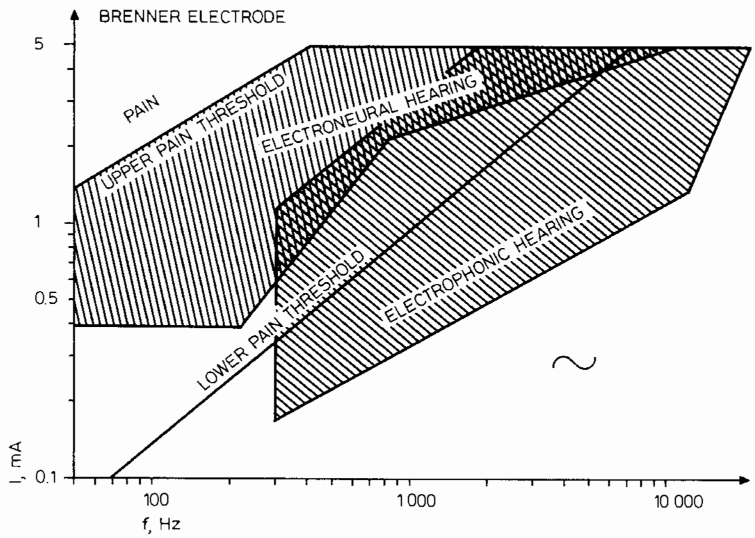

Fig. 4. Areas of electrical sensation thresholds by electrical sine wave stimulation with ear canal saline electrodes. The 5-mA limit indicates the maximal output of the current stimulator.

the hair cells, *electroneural hearing* is most frequent in profound to total deafness. In general, electroneural hearing requires higher current amplitudes. Usually, pain and other unpleasant sensations (vertigo, muscle twitching) were elicited at levels only slightly above or even below hearing thresholds.

The threshold increase with rising frequency is characteristic for sine wave stimulation and does not occur with pulse stimulation since the *charge per period* seems to be the main parameter for auditory stimulation. In cases without hearing by ear canal or promontory stimulation, potential hearing through implanted electrodes could not be excluded. Also, more specific psycho-electrical tests (e.g., differential thresholds, speech discrimination) could not be based on ear canal or promontory electrode testing. These preliminary noninvasive tests, therefore, could only give a crude estimation of the potential effects of electrical stimulation in the hearing nerve. To achieve a specific excitation of only small groups of nerve fibers, an electrode must be placed as close to them as possible, preferably in the cochlear modiolus, in order to reduce current density in neighboring nerves or other excitable tissues to a negligible minimum.

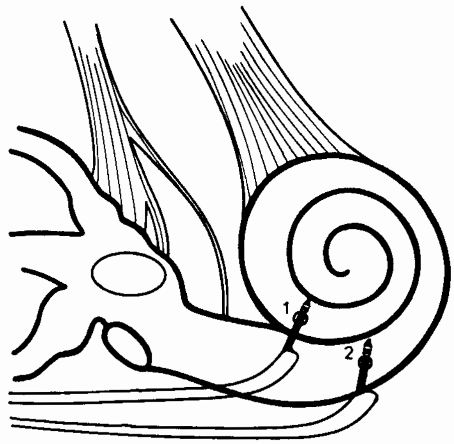

Fig. 5. Placement of two bipolar electrodes through separate fenestrations in the cochlear wall of the basal and middle turn.

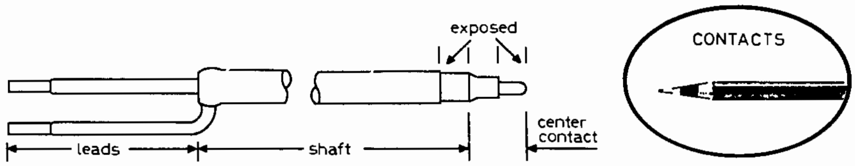

Fig. 6. Dimensions of the electrode (Rhodes Neurological Electrode NE-100): shaft diameter 0.5 mm; shaft length 5.0 mm; lead diameter 0.2 mm; contact diameter: 0.2 mm center, 0.5 mm outer; contact length: 0.5 mm center, 0.5 mm outer.

A 50-year-old patient had lost his hearing bilaterally after meningitic infection 40 years ago. Electrodes were placed through two separate fenestrations in the cochlear wall of the basal and middle turn (fig. 5). They penetrated the scala vestibuli to reach the nerve fibers in the modiolus, thus providing some tonotopy as well as a stimulation close to the excitable cochlear nerve tissue, which allowed for the use of lower current levels than with electrodes in the scala tympani, inserted through the round window. Two commercial nerve electrodes for chronic stimulation were used (fig. 6). They are bipolar, concentric, of pure platinum. Transcutaneous communication is accomplished through a hardwire multichannel coupling encapsulated in a pyrolitic biocarbon skin window. To carry out the psycho-electric experiments, a hardwire coupling and a direct access to the electrodes was considered to be necessary. However, this skin button technique will only be an intermediate stage, if the experiments can show practical ways of signal encoding in terms of speech discrimination.

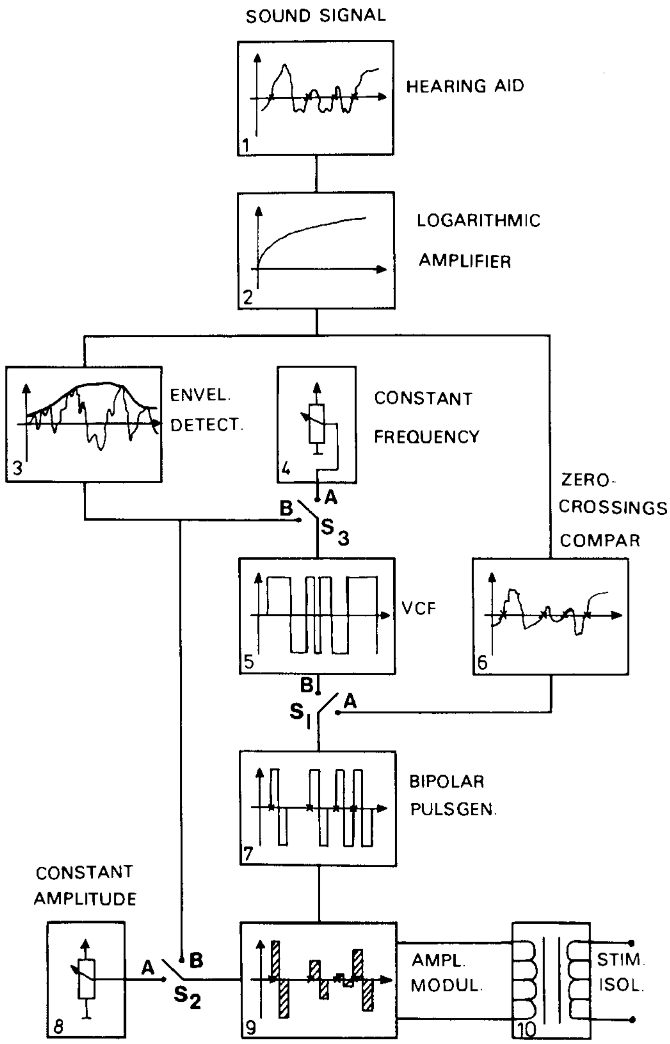

Fig. 7. Block diagram of the portable signal-processing device.

Signal-Processing Aspects

Signals from a microphone in a standard behind-the-ear-aid case are amplified, compressed, and transmitted to the signal transformation subsystem (fig. 7, top). The electrical output is then passed to a logarithmic

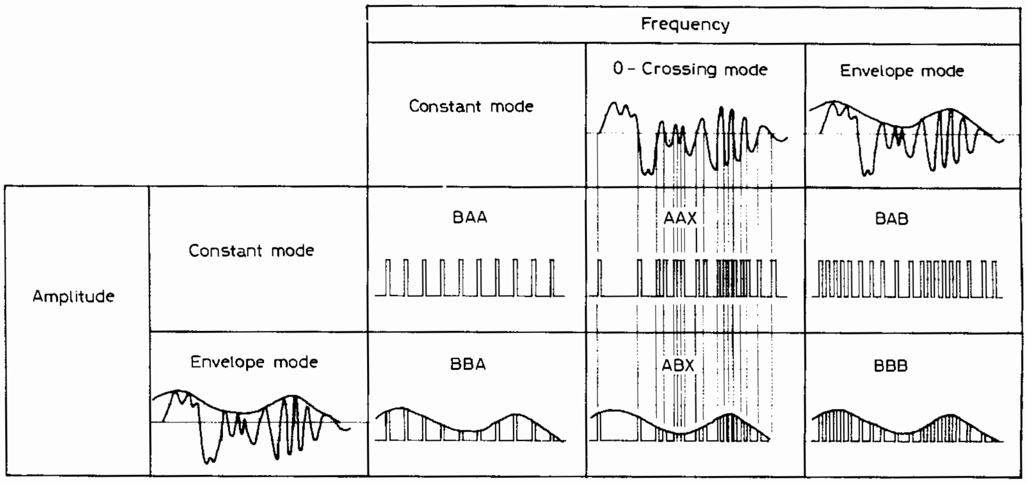

Fig. 8. Coding options which can be selected by a switch. Symbols A and B correspond to switch positions S_1–S_3 in figure 7.

amplifier for additional amplitude compression with an adjustable slope and offset according to specific threshold criteria.

The next step is the detection of positive going zero crossings in a comparator circuit to eliminate low-amplitude noise. Each zero crossing initiates one bipolar pulse of constant duration (S1: A). Alternatively, the output of the log amplifier can be passed through a lowpass filter (30 Hz) to extract the speech amplitude envelope. This low-frequency signal can then be used to modulate either the pulse repetition rate (S1: B, S3: B), the pulse amplitude (S2: B), or both at the same time (S1: B, S2: B, S3: B) (fig. 8).

Combined amplitude and frequency modulation would provide a more gradual adjustment of the intensity range than amplitude modulation alone if an increased stimulation frequency is perceived as an increase in loudness. The discrete bipolar pulse is triggered either by the zero crossing comparator circuit (S1: A) or by the output of a voltage-controlled oscillator (S1: B). This oscillator is in turn controlled either by the speech envelope detecting circuit output (S3: B) or by a manually set constant-voltage source (S3: A). Hence, the *timing of the pulses* may be determined by any of three ways (fig. 8): (1) a pulse is generated each time the acoustic input signal realizes one full period of oscillation (AAX, ABX); (2) pulses are generated at a frequency proportional to the instantaneous acoustic input level (BAB, BBB); (3) pulses are generated at a fixed rate independent of the acoustic signal (BAA, BBA).

The *pulse amplitude* may in turn be modulated by either of two modes: either it is controlled by a manually set constant-voltage source (AAX, BAA, (BAB) or it is controlled by the output of the speech envelope detector (ABX, BBA, BBB).

Results

2 weeks after the implantation the first experiments were performed, connecting the electrode contacts in the skin window to an optoelectronically isolated constant-current stimulator. Both electrodes had been checked regarding subjective evoked hearing perceptions, electrical threshold values and impedances. With both electrodes, clear and distinct reproducible hearing perceptions could be produced which have been characterized as engine noise, sibiliant, knocking or singing sounds. Bursts of biphasic pulses were presented with an on-off ratio of 1:1 and a burst frequency of 3 Hz. The burst rhythm could be well described. Stimulation of the basal turn electrode (electrode 2) gave higher-pitched whistling sound sensations (described as 'zing-zing'), whereas stimulation of electrode 1 (located in the middle turn) produced lower-pitched sounds or noise (characterized as 'flug-flug'). Sometimes, with electrode 2 two tones could be recognized simultaneously at a medium intensity.

Parallel *threshold curves* between 10 and 1 000 Hz repetition rate were found for different *pulse durations*. Therefore, thresholds were nearly independent of repetition rate. With continuous stimulation, however, the thresholds rose for repetition rates above 120 Hz and showed pronounced directional hysteresis. Even with very high burst rates (100 Hz) the thresholds ranged still below the continuous stimulation thresholds (fig. 9). Continuous stimulation is not as easily detected as intermittent stimulation at higher frequencies. At low burst rates (2–5 Hz) the thresholds were practically identical.

Amplitude and frequency *difference limens* were measured by a computer-controlled stimulation-response procedure (fig. 10). For *frequency difference limen* (DL) measurements, tone pairs of randomly set second frequency were generated. The influence of amplitude and frequency can be

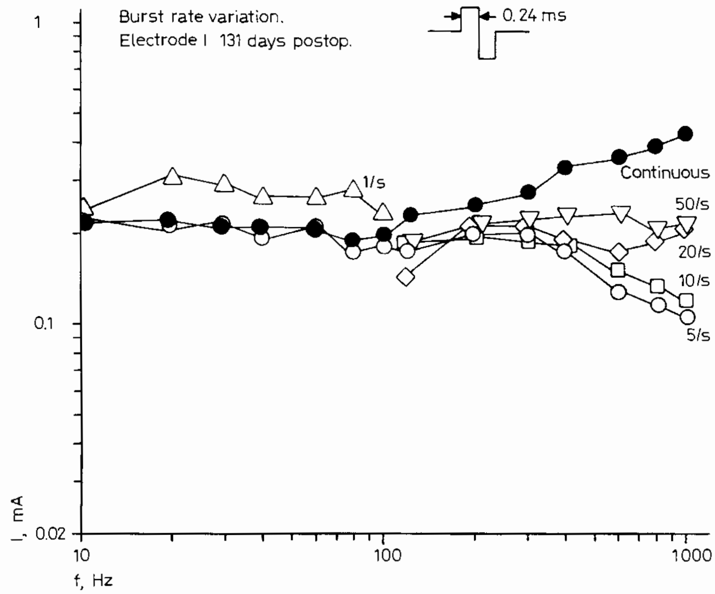

Fig. 9. Pulse amplitude at hearing threshold as a function of repetition rate (10–1 000 Hz) and rate of interruption (on-off ratio 1 : 1). ● = Continuous stimulation.

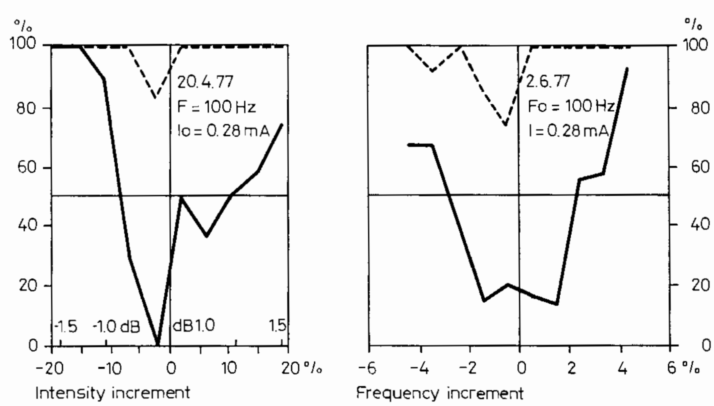

Fig. 10. Intensity and frequency difference limen (DL) measurements. Correct and false responses to the presentation of 100 tone pairs with randomly set second tone frequency or intensity as a function of increment size (10 different increment steps equally distributed among 100 trials). 100 bursts; 8 ms duration; 400-ms interval. — = Correct responses (%); – – – = 100%-wrong responses (%).

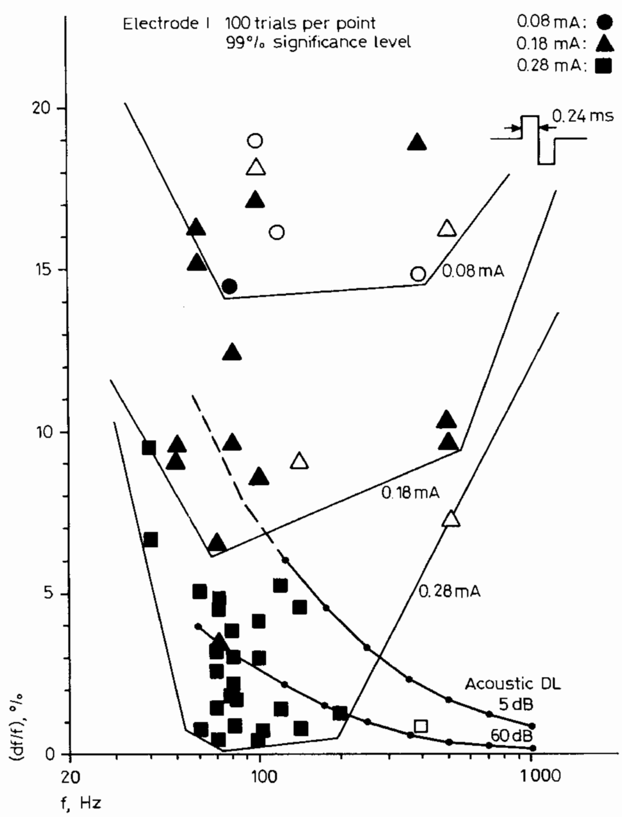

Fig. 11. Comparison of electrical frequency DLs with acoustical DLs. Three different pulse amplitude levels giving different DLs (tentatively indicated by thin lines). Filled symbols represent detected steps of both increments and decrements, open symbols indicate unsymmetrical responses (only one stimulus polarity – either positive or negative – is detected). [Acoustic DL: *Shower and Biddulph,* 1931.]

seen in figure 11. With the highest amplitudes and at frequencies below 120 Hz, electrical DLs were equal or even better than acoustical DLs in normal subjects, taking advantage of the periodicity principle of frequency encoding. *Intensity DLs* measured by the same procedeure were found to be 0.8–1.2 dB which compare to those found in normals by acoustical stimulation.

Discussion

The evaluation experiments should give answers to the following questions:

(a) What are the changes of electrical electrode properties after implantation in the human body and after long-term electrical stimulation?

(b) What are the optimal stimulus parameters; how do they vary over time?

(c) How much information can be transmitted over a single electrode channel, i.e., how well can a patient detect amplitude and frequency differences?

(d) What are the consequences for further proceeding in regard to speech encoding?

The repeated measurements of *electrode impedances* at different intervals revealed no general changes of electrical properties over the 5-month time period. However, the hearing thresholds varied considerably. In this case, it is suspected that only relatively few nerve fibers were viable and probably had no direct contact with the stimulating electrodes. Small changes of the tissue structure around the electrode tips could therefore have caused big changes of local current densities. The most effective stimulus parameters were found to be the *pulse amplitude* and the *pulse repetition rate*. The pulse duration is complementary to the amplitude in an intermediate range and does not seem to noticeably influence the subjective quality of the sounds.

The *channel capacity,* i.e., the maximal error-free information transmission rate, is determined by the bandwidth and the signal to noise ratio [*Hill et al.,* 1968]. It is estimated in our measurements to about 600–800 bits/s. Although it is possible to get a 70% speech syllable discrimination score with a channel capacity of only 1 000 bits/s [*Flanagan,* 1972], it must be clearly understood that in these cases the signals have to be optimally coded. With single channel electrical stimulation of the hearing nerve, this is not attainable. An increase in channel capacity is only possible with multichannel systems. The ideal channel number would probably be 4–10 independent electrodes each stimulating a different nerve fiber group in the main speech frequency region. Of primary importance will be the best possible coding of speech signals and their distribution to different channels.

Conclusions

The experiments have shown the usefulness of the periodicity principle of hearing (below 200 Hz) as well as of the place principle of frequency encoding for the electrical stimulation of the auditory nerve. Experiences with implants in the USA and in France have shown that electrically produced sound sensations were consistently rated most suitable by these patients. Rhythm, modulation and stress of the voice can be discriminated with a single stimulation channel. Whereas this information has shown many benefits for the deaf patient, multiple-channel implants did not seem to be more effective than single ones until now. Several questions still are left open, mainly interelectrode electrical insulation within the transcutaneous or subcutaneous connection and within the destination place of the electrode surfaces, namely the cochlear turn or the modiolus. Since speech discrimination certainly requires a set of independent channels, further research should be carried out on the development of adequate electrode materials, insulation, and on transmission link systems. Therefore, patient selection has to take into account that cochlear implantation has remained an experimental procedure, for which the ethical guidlines of the 'First International Conference on Electrical Stimulation of the Acoustic Nerve in Man', held in San Francisco in 1974 [*Merzenich et al.,* 1974], are as valid now as before.

Résumé

Nous décrivons une configuration optimale à la fois d'une électrode et d'un stimulus, basée sur une simulation par un modèle mathématique de la conduction mécanique de la transduction mécano-électrique de l'oreille moyenne et interne. Nous supposons évidemment qu'un patient atteint d'une surdité de perception grave et dont le nerf auditif est au moins partiellement intact, sera capable d'entendre et de comprendre, lorsqu'un «pattern» d'activation dans des groupes spécifiques de fibres nerveuses subsistant peut être créé ayant un sens équivalent à celui de l'ouïe normale. Ayant implanté chez un patient deux électrodes (bipolaires concentriques) sur la columelle au niveau de la base et du tour moyen de la cochlée, nous avons pu mesurer des seuils électriques absolus et différentiels ainsi que des changements d'impédance de l'électrode pendant une période de plus de 5 mois, grâce à un accès direct aux électrodes à travers la peau. Le stimulateur était situé dans une petite boîte de poche de manière à en faciliter l'usage quotidien; des impulsions de 0.02–1 ms étaient déclenchées soit par le passage de la ligne zéro soit par un oscillateur contrôlé par l'enveloppe du signal acoustique de la parole. L'expérience a montré que la sonie devait être codée en fonction de l'amplitude ou de la durée de l'impulsion plutôt que par son taux de répétition, la charge par impulsion étant le variable primaire de la sonie. Le principe de

périodicité nous a semblé par ailleurs pouvoir être retenu en dessous de 150 Hz car nous avons observé, pour cette zone fréquentielle, des seuils différentiels électriques de même niveau ou même inférieurs aux seuils différentiels acoustiques d'une personne à audition normale. Nous pensons d'autre part qu'un codage optimal de la parole nécessite un système à canaux multiples, 4-10 électrodes stimulant un groupe différent de fibres du nerf auditif.

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