# A Versatile System for the Generation and the Development of Speech Coding Strategies in Cochlear Implants

Norbert Tönder, Rainer Hartmann,* and Rainer Klinke

***Abstract*—A signal processor-based (DSP32C) stimulation system for cochlear implants has been developed. This system allows up to 14 stimulation channels (place principle) and conserves the time structure of the acoustic signal (periodicity principle). Any combination of these features is possible; in compressed analogue mode, pulsatile mode, or combinations of both. The system also takes electrical crosstalk into account caused by the electrical properties of the inner ear and compensates for it. Lateral inhibition strategies improve spatial resolution. Finally, the system offers an interface for the generation of quick continuous interleaved sampling strategies (CIS); up to 11 000 pulses/s per channel.**

***Index Terms*— Channel interaction, cochlear implant, coding strategies, signal processor, speech processing.**

## I. Introduction

PATIENTS suffering from sensorineural hearing loss can regain some auditory capabilities by electrical stimulation of the remaining auditory nerve fibers. For this purpose, the impulse patterns in the auditory nerve, naturally occurring as a consequence of the transduction process, have to be evoked by electrical stimulation. In order to simulate the neural code present under natural circumstances, a set of electrodes implanted into the inner ear for the selective stimulation of different groups of nerve fibers is used. For optimal performance, electrical stimulation should stimulate certain groups of nerve fibers selectively, mimicking the place-principle (tonotopy) present in the cochlea [1]. This requires a multichannel device. The time-structure of the acoustic signal (phase-lock) up to at least 3.5 kHz has also to be preserved in the neuronal pattern [2]–[4]. Thus, the stimulus code has to make use of both the place principle and the periodicity detection [5], [6]. The ultimate goal is coding of speech signals and the transmission of important daily life acoustic events in order to enable the patients to perform almost normal auditory communication and contacts with their environment. Because of technical difficulties the implants and speech processors presently available provide only limited aspects of the natural neuronal code.

Manuscript received March 6, 1997; revised December 1, 1997. This work was supported by the Deutsche Forschungsgemeinschaft (DFG) under Grant SFB269. *Asterisk indicates corresponding author.*

N. Tönder and R. Klinke are with the Physiologisches Institut III der J.W. Goethe-Universität Frankfurt am Main, D-60590 Germany.

*R. Hartmann is with the Physiologisches Institut III, J.W. Goethe-Universität, Theodor-Stern-Kai 7, D-60590 Frankfurt/M, Germany (e-mail: R.Hartmann@em.uni-frankfurt.de).

Publisher Item Identifier S 0018-9294(98)03718-5.

Currently available cochlear implants can be categorized into two types [7]–[9].

1) Single-channel systems which only transmit the temporal code [10]–[13].
2) Multichannel devices which realize tonotopy by stimulation frequency specific cochlear positions and which also transmit the temporal code, at least in the lower-frequency range [14]–[17].

These present multichannel systems can either use purely pulsatile or analogue coding, or a combination of the two for a various number of channels [18], [19].

In general, the present systems are not very flexible. There is normally only one coding principle available to the patient. Because of this, it is very difficult to compare and analyze different coding strategies and their effects, except for cases the stimulation electrodes are directly accessible in percutaneous systems [20]. It is even more complicated to optimize the code used for a patient.

The aim of the present paper is to describe a versatile system capable of real-time processing of acoustic signals. Many requirements for a cochlear implant have become apparent from animal experimentation or from the experience of patients. The implementation of a dynamic range that is comfortable to a patient and the avoidance of channel interactions within the cochlear tissues which may lead to unwanted loudness summation and thus discomfort are examples of such requirements [21]. With the system proposed here, different coding strategies for cochlear implants can be tested both in animal experiments and in patients. The system is based on one signal processor so that it can be further developed into a portable device.

For good speech communication, the frequency range from 300 Hz to 3.5 kHz is required. This range comprises 14-critical bands [22] on which the present system is conceptually based.

The concept of the software aims at an optimal utilization of the resources of the signal processor and also an easy adaptation for different target hardware.

## II. General Concept

The system has to accomplish three tasks and is, therefore, divided into three different functional units (Fig. 1).

1) Extraction of information from the signal. This unit is termed "*front-end*."
2) Transformation of this information into electrical stimulation patterns. This unit is termed "*back-end*."

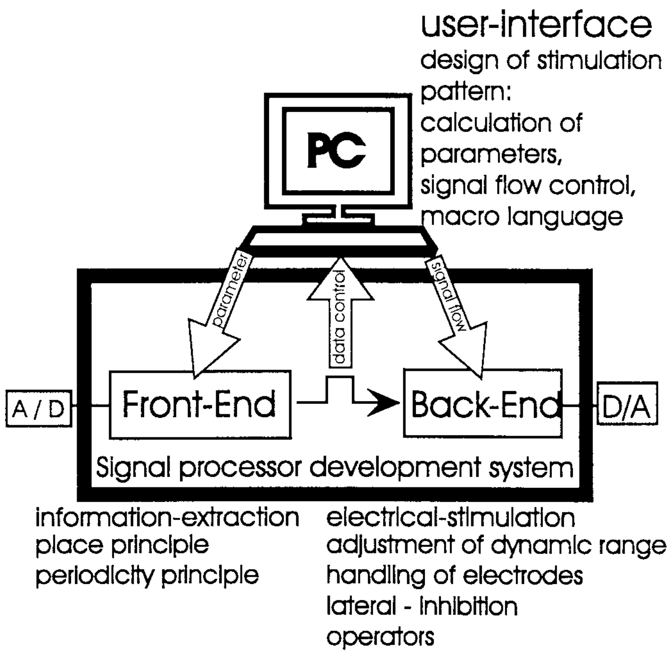



Fig. 1. General set-up of the system. The front-end serves for the signal analysis, the back-end provides the intended stimulation pattern wanted. Front-end and back-end are implemented on the signal processor development system and controlled by the user interface.

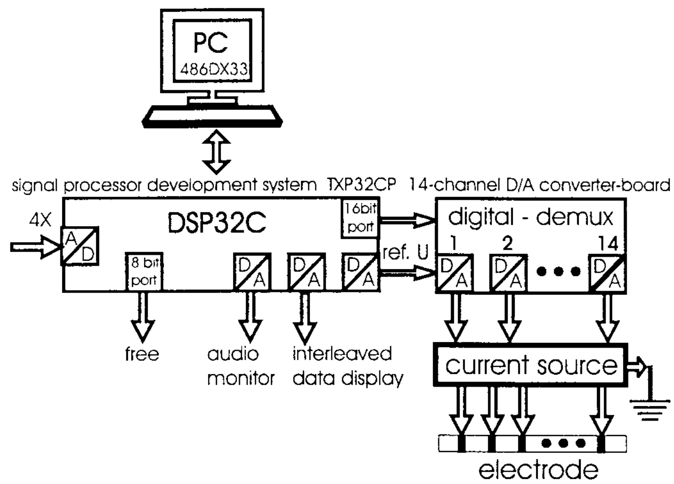



Fig. 2. Schematic diagram of the hardware. The system consists of a PC equipped with a signal processor development system from AT&T (TXP32CP) with the signal processor DSP32C and a 14-channel D/A converter-card for the control of the current sources.

3) The user is provided with versatile options to enter data to produce virtually any type of stimulation pattern. This unit is referred to as the "*user interface*."

Front-end and back-end units are realized on the signal processor development system, the user interface is implemented on a PC (IBM compatible, 486).

The acoustic signal to be processed is picked up by a microphone and digitized. Subsequently, information is extracted in a manner similar to the processes performed by the inner ear. The different frequency bands, set by the program, sound pressure and phase information are extracted. In addition there is a function capable of avoiding crosstalk between neighboring stimulation electrodes. This problem is served in the inner ear because of the good electrical conductivity of the cochlear tissues. These functions are carried out in the front-end.

From this information, a complex stimulation pattern for multichannel electrical stimulation of the cochlea is generated in a second processing step in the back-end. The back-end contains three further subunits to handle the problems of electrical stimulation.

1) A signal router for introduction of *"lateral inhibition"* [23]–[25]. This is a strategy to improve channel separation which is taken from neurophysiology.
2) A lookup-table for adaptation of the dynamics of the different channels to the dynamic range acceptable to the patient.
3) A D/A-server for the free assignment of individual electrode channels.

The combination of these three functional subunits allows the formation of operators which enable the user to design virtually any stimulation pattern. Front-end and back-end functions are controlled by the user interface.

The high versatility of digital systems, such as the one proposed here, is due to the ability to change the signal processing parameters over a wide range. Additionally, these systems provide an option for the control of signal flow through the functional units in the back-end. The signal flow is exclusively controlled by the user interface and there are no predetermined signal processing paths.

Data vectors permit communication between the functional units and subunits. The following definitions are used: Data elements represent the outputs of the individual channels of units and subunits. They are renewed every cycle. Data vectors are composed of data elements. The units and subunits can access the individual data elements for each channel separately.

## III. Implementation

### *A. Hardware*

The system is based on the TXP32CP development system (AT&T) with the DSP32C signal processor (Fig. 2). The signal processor is equipped with a 32-b floating-point multiplier and a 40-b floating-point adder with pipelining. Cycle time is 80 ns which means that the DSP32C has an arithmetic dynamic range of 1500 dB with 25 Mflops. This is sufficient for signal processing in real-time. The development system is equipped with four 12-b A/D converters for signal acquisition. In addition there are three D/A converters for signal output. Two of those are used for on-line control functions. The third is used for the generation of the reference voltage for a 14-channel 12-b D/A converter-board, which is controlled by a fast (10 Mword/s) 16-b parallel port of the DSP32C. Each channel of the converter board is connected to a current source (custom-made, $\pm 15$ V, 1 mA) and can be connected to the individual channels of a multichannel electrode for electrical stimulation. In addition, there is an 8-b parallel port for data output. The development system and the 14-channel D/A converter-board are realized as a PC ISA-bus interface.

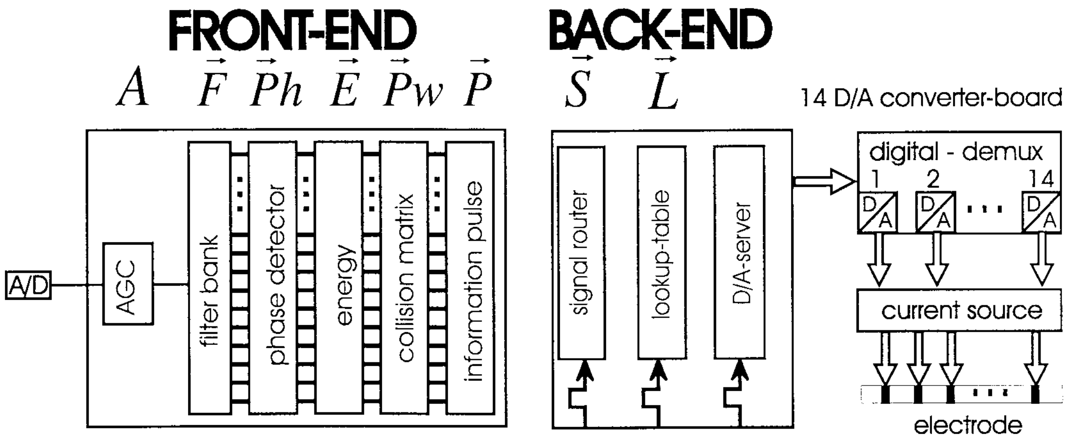



Fig. 3. Signal processing path. The front-end extracts information about frequency ($\vec{F}$), phase ($\vec{Ph}$), and energy ($\vec{E}$) and calculates an information pulse ($\vec{P}$). The collision matrix, decides in which channels an eventually simultaneous pulse generation has to be transformed into a serial one. The back-end has free access to all data vectors and generates the electrode signal.

## *B. Front-End*

The signal analysis is schematically shown in Fig. 3. The signal is first digitized and then processed by an automatic gain control (AGC) (data element A in Fig. 3) in order to be able to cope with different sound intensities. This is particularly important for speech recognition. Different parameters are available: *level* for the loudness desired, *attack* and *release* for the reaction time to sound pressure increase and decrease, respectively. A multichannel filter bank then subdivides the signal into several frequency bands. Each filter output is stored as a corresponding data element of the data vector $\vec{F}$. The order of each filter and its transfer function are set by coefficients in the user interface. There are 630 filter coefficients available, which can be allotted to the individual filters. Up to 23 filter-channels can be used. For each channel, the phase information is extracted and stored in the data vector $\vec{Ph}$. The energy contained in the different frequency bands is determined from the envelope of the filter output and available as data vector $\vec{E}$. $\vec{Ph}$ and $\vec{E}$ are summarized as data vector $\vec{P}$. The data thus generated will be termed the *information pulse*. The time structure of these information pulses is as follows:

At phase-time zero the data elements of $\vec{P}$ are set to zero.

The time structure of the individual data elements of $\vec{P}$ are controlled by four parameters: threshold, dead-time, and positive- and negative-stimulation time. In addition, a collision matrix to avoid simultaneous stimulation of neighboring channels, thus minimizing crosstalk, is present. The minimum distance between electrodes which can be activated simultaneously is termed the *crosstalk-radius*. According to Merzenich [26], this *crosstalk-radius* for bipolar stimulation approximates 2 mm in a human cochlea.

1) A threshold can be set for every channel. Below this threshold no entry is made into the data elements.
2) Within a *"dead-time"* no further entries can be made into the data elements. Thus, a maximal entry rate can be set for each filter channel.
3) The time the data element preserves a positive value can be set.
4) The time it preserves a negative value can be set.

The collision-matrix determines which data elements of $\vec{P}$ can be simultaneously entered and in which elements the simultaneous entry has to be transformed into a serial entry (Fig. 4). If this is necessary, the filter-channel with the lower channel-number has priority. As the frequency bands in the individual filter channels are determined by filter coefficients, the weighting of the individual frequency bands can be selected freely.

## *C. Back-End*

This functional unit consists of three subunits to control the signal flow (Fig. 3):

- a signal router with data vector $\vec{S}$;
- a lookup-table with data vector $\vec{L}$;
- a D/A-server.

The system can be adapted to any type of electrode, any electrode position and any stimulation site, (e.g., intracochlear or brainstem nuclei stimulation [27]–[30]). Each of these subunits has a variable number of channels that have access to any data element of the front-end and the back-end.

*1) Signal Router:* The major task of the signal router is the generation of place-related and phase-oriented (PrPo) stimulation patterns by access to the front-end data elements. Here, the decisions on, whether the stimulation is to be purely pulsatile using data elements $\vec{P}$, analogue using data elements $\vec{F}$, or using envelope data $\vec{E}$ are made.

Alternatively, data element A generated by the AGC can be used for a single-channel analogue stimulation. As every channel of the signal router has separate access to the data elements, any mixed or superposed types of stimuli can be generated. For example, three channels could provide analogue stimuli, four other channels pulsatile and two further channels superposed analogue/pulsatile stimuli. For the realization of the multichannel pulsatile PrPo stimulation pattern, the signal router uses the data elements of vector $\vec{P}$. The signal router determines the weightings for the individual filter channels and defines—through weighted superposition—the stimulation mode (monopolar-multipolar [31]). Fig. 5 shows an example

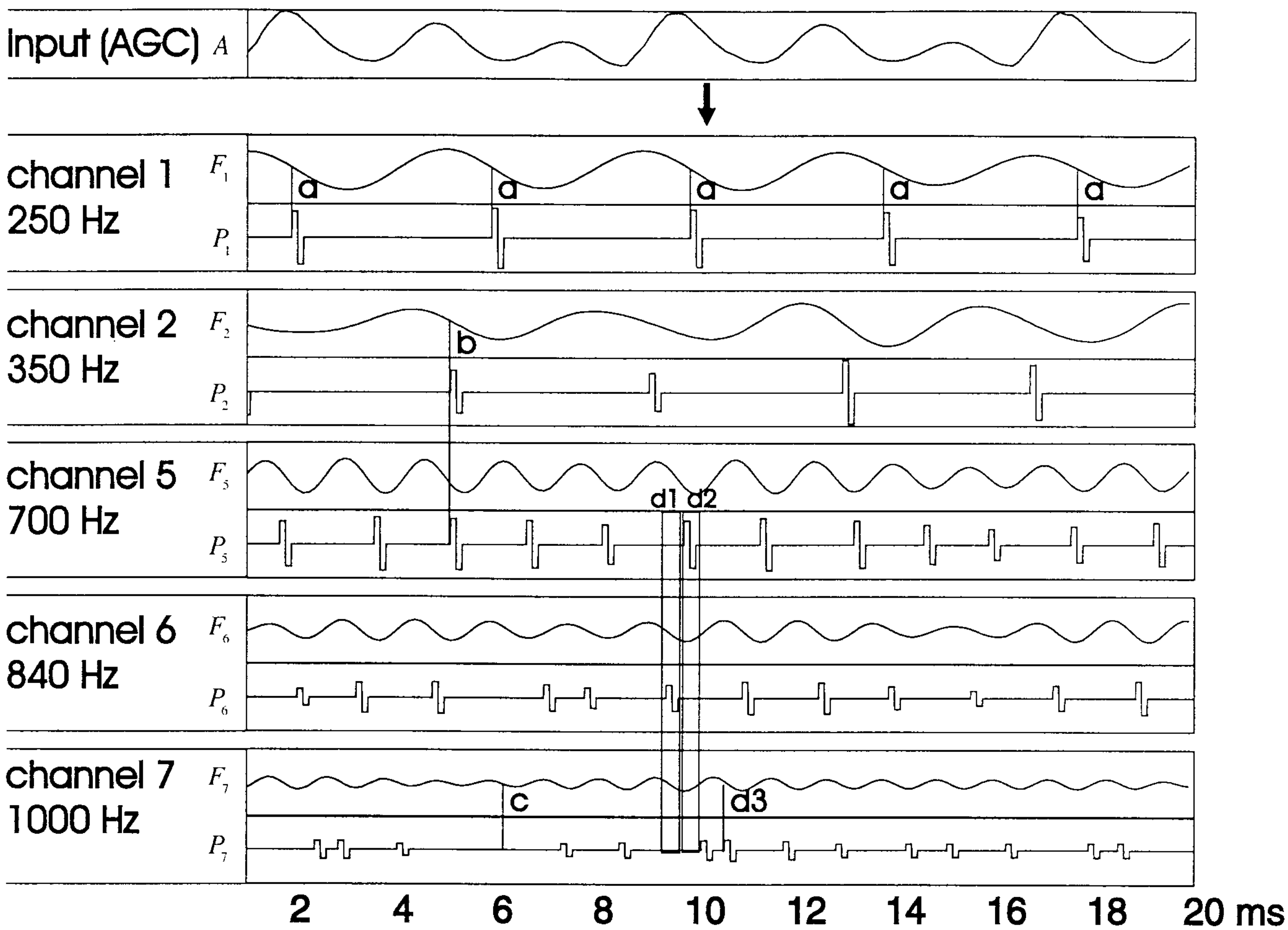



Fig. 4. Display of acoustic input signal and filter outputs and information pulses. The figure shows five out of 14 channels, crosstalk radius of three. (a) Channel 1 illustrates the phase-lock, (b) simultaneous formation of information pulses in channels with distance $> 3$ in channels 2 and 5. Conversion of simultaneous into serial stimulation in channels 5, 6, and 7 (d1–d2). In d1 a phase-locked information pulse is generated, which delays channels 5 and 7. Following channel 6, the channel with the lowest channel number (5) is stimulated. Only after the release of the information pulse in channel 5 channel 7 is unblocked. The subsequent information pulse (d3) phase-locked again. (c) The freely definable threshold can be seen in channel 7.

of bi- and tripolar stimulation modes. The signal router can for example build up a Genin-matrix [32] or virtual pulses [33].

The stimulation mode determines the number of electrodes used to deliver a stimulation event. With monopolar stimulation, a common ground electrode is used. In the case of bipolar stimulation, two electrode channels are used to apply a spatially localized stimulus. For lateral inhibition, at least three electrode channels (tripolar) are necessary. The stimulation mode can be enlarged freely up to an $n$-polar stimulation by the signal router.

In addition to the above, a logical operator can be implemented. As the three functional units and subunits are not strictly coupled, the logical operators *or* and *and* can be set up through the use of multiplication and addition. For this purpose, any data elements are logically interpreted by the lookup-tables. This means that below a certain threshold the values are set to zero, above this threshold to one (corresponding to a Schmitt trigger). Subsequently, the *and* can be realized by the use of multiplication and the *or* by use of addition. Thus, any stimulation pattern can be generated.

The signal router is equipped with 20 channels and has simultaneous access to 920 data elements. Any number of products of weighted or unweighted data elements can be superimposed. The data output is through data vector $\vec{S}$.

*2) Lookup-Table:* The lookup-tables are necessary for the adjustment of the dynamic range possible with electrical stimulation. There are 23 lookup-tables (12-b) available. With the aid of the lookup-tables, data elements are replaced according to instructions which can be set up freely by the user interface. The lookup-tables are primarily used for dynamic range compression. Second, with the aid of the lookup-tables the logic operations *and* and *or* of the signal router can be *normalized* or *Boolean complemented*.

*3) D/A-Server:* The D/A-server also consist of 20 channels and assigns the individual electrode channels to the D/A converters. This subunit also controls the 8-b and 16-b parallel ports as well as the on board 12-b D/A converters. Further more this subunit can interfere with the signal processing in the front-end and initiate the generation of information pulses at any time.

### D. User Interface

The user interface is implemented on a PC and allows the design of virtually any coding strategy for stimulation of the

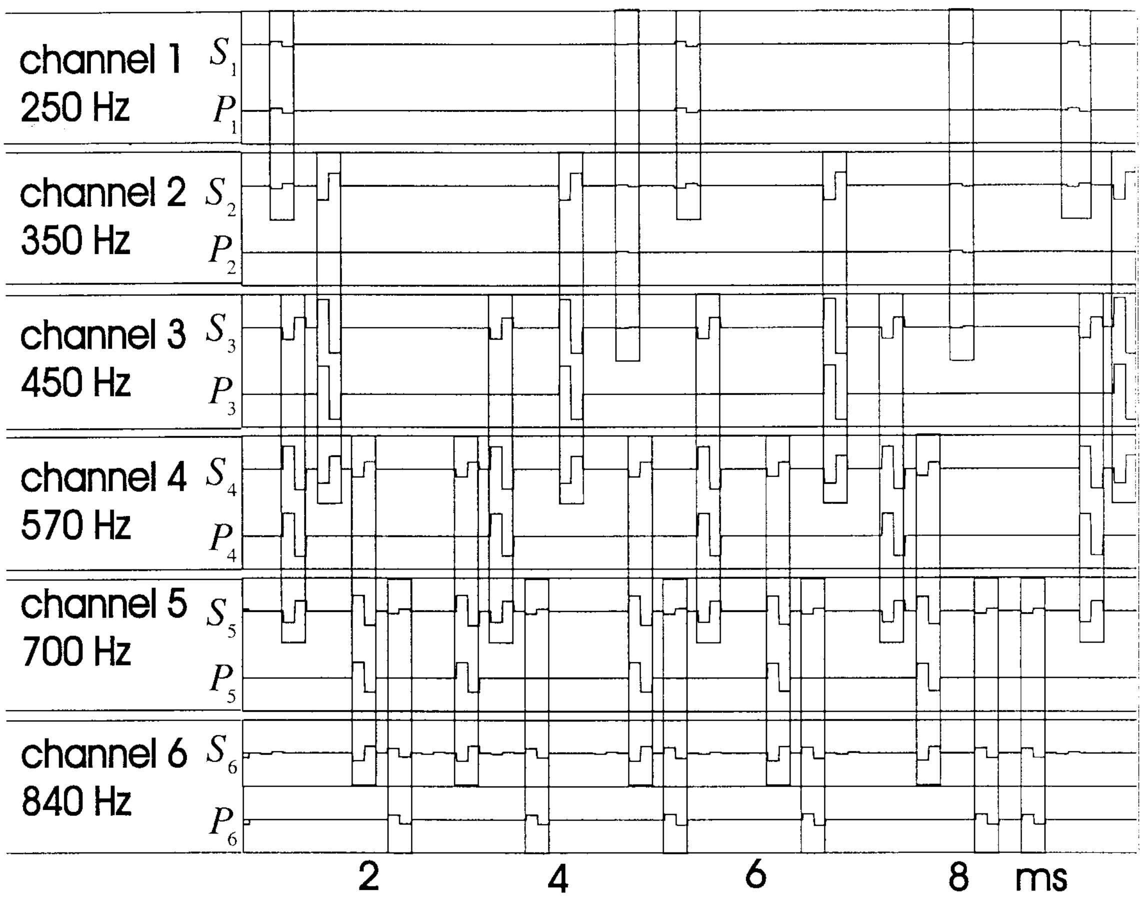



Fig. 5. Illustration of six channels out of 14 of a tripolar PrPo pulsatile stimulation, crosstalk radius three. Channel 1 is driven in a bipolar, the other channels in a tripolar mode. Information pulse displayed with the output of the signal router. The rectangles mark the information pulse $P_i$ with the corresponding signal router outputs.

auditory nerve. For the use of different stimulation strategies, the parameters of the front-end and the lookup-tables of the back-end have to be calculated and the signal flow has to be specified. The different coding strategies can be stored as a file and be used during an experiment or for a patient.

The system can be used off-line and on-line. When used off-line the stimulation patterns are downloaded to the signal processor development system and the system runs independently of the user on the signal processor. In the on-line mode, an interactive design and adaptation of coding strategies is possible. There are four possibilities for monitoring the functional units.

1) An audio monitor connected to data elements via a D/A converter.
2) A single-channel oscilloscope displaying whole data vectors realized by an interleave strategy provided by the signal processor.
3) The display of data elements or data vectors on the user interface oscilloscope. The time and frequency domains can be displayed on-line.
4) The storage of data which provide the base for later mathematical evaluation.

The signal flow between data elements and the individual channels of units and subunits can be determined in two different ways: either by a macro language or by a graphic signal path. The signal path displayed on the monitor represents the units and subunits, including their data vectors, which can be linked freely by lines. This allows the signal flow to be monitored graphically at any point.

## IV. Performance

The frequency range within which the system is supposed to work is crucial for the realization and applicability of the different coding strategies. The sampling rate has to be at least twice the highest possible frequency (corner frequency) to be processed. The complexity of the signal processing algorithms depends on the number of channels to be processed simultaneously, the number of filters set up in the front-end, and the number of operators in the back-end. The corner frequency is also limited by the complexity of the signal processing.

As the filters are defined by coefficients, the total number of down-loaded coefficients determines the computational effort. Fig. 6(a) illustrates the corner frequency of the front-end as a function of the number of filters. For this figure, a fourth-order filter bank was chosen. The figure shows that the corner frequency decreases with increasing number of channels. To show the influence of filter-order for an eight-channel filter bank, the order was varied [Fig. 6(b)]. Again, with increasing number of coefficients the corner frequency decreases. Thus,

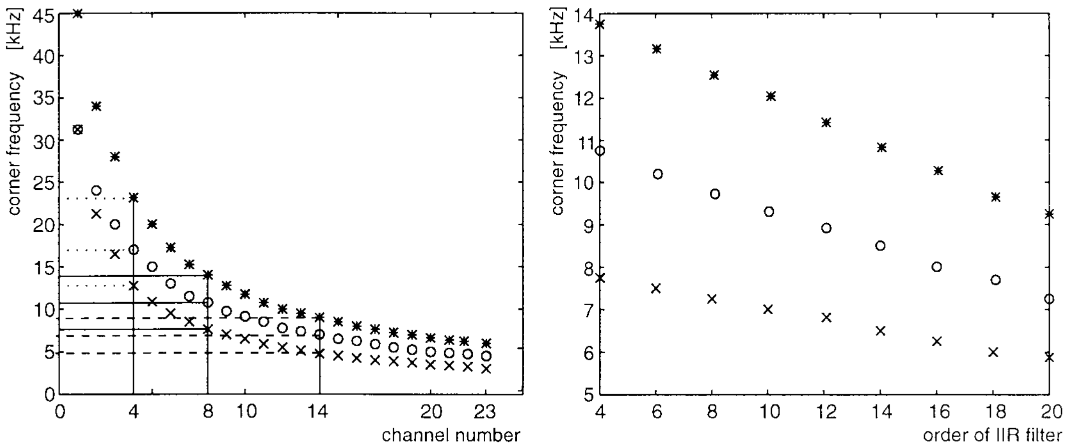



Fig. 6. (a) Corner frequency as a function of channel number of the front-end, fourth-order filter bank, ($*$) without generation of an information pulse, ($\circ$) with generation of an information pulse, collision matrix set, ($\times$) with generation of an information pulse without collision matrix. (b) Corner frequency as a function of filter order; eight channels used.

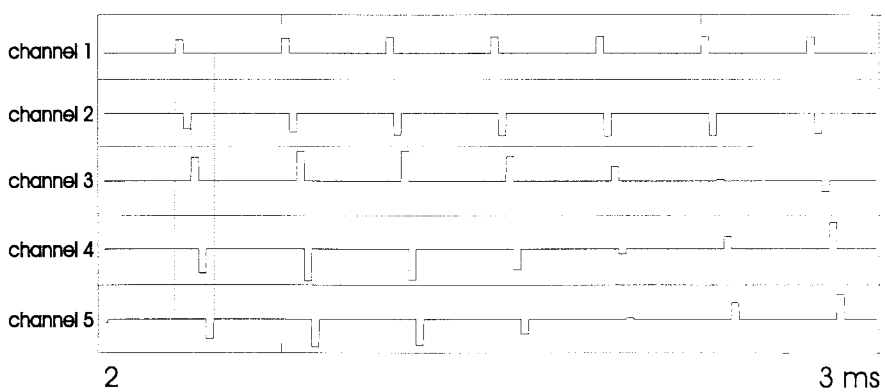



Fig. 7. Time structure of data outputs in the continuous interleaved sampling mode in five out of 14 channels. The analogue value is coded in the lower 12-b of the parallel port, the upper 4-b contain the channel information.

the user has the choice between a filter bank of higher-order filters or a filter bank with a greater number of channels.

For the activation of the stimulation electrodes, functional units in the back-end are necessary. In addition, a lookup-table for dynamic compression and a D/A-server channel are required for each electrode channel. The signal router determines the stimulation mode for each channel separately. The stimulation mode (mono-, bi-, tripolar, up to $n$-polar) determines the calculation time necessary for the signal router. The expression $n$-polar means that each individual information pulse is fed into each electrode channel.

When discussing the corner frequency, three different modes of action have to be considered.

### *A. Continuos Interleaved Sampling (CIS)*

If the CIS strategy is wanted [34] no information pulse is generated, the filter outputs are sampled directly and used for interleaved stimulation (Fig. 7). All necessary information is extracted by the front-end. The output consists of time and the amplitude of the stimuli. However, the stimulation pulses proper are generated by an external system (e.g., a cochlear implant). Its hardware is provided with the stimulation time, the electrode channel, and the compressed amplitude. The sum of single stimulation pulses/s generated is a consequence of the corner frequency multiplied by the number of stimulation channels used. Therefore, the frequency range transmittable by each channel equals the frequency range of the system. This implies that the output of the individual filters can be used directly as a stimulus after appropriate dynamic compression. As no information pulses are generated in this case, the calculation time is minimal. In Fig. 8 the corner frequency for a 14-channel filter bank is shown as a function of the stimulation mode used (from mono- up to 14-polar). The identical mode was used for all stimulation channels. It can be seen that an upper corner frequency of 3.5 kHz can be guaranteed up to 11-polar stimulation.

The following two modes B and C represent approximately the upper and lower limits of the corner frequency.

### *B. Place-Related and Phase-Oriented Single-Pulse Stimulation*

The front end extracts all information and generates the data vector $\vec{P}$ (information pulse). The collision matrix allows only one stimulation at one instant. Stimulation pulses due

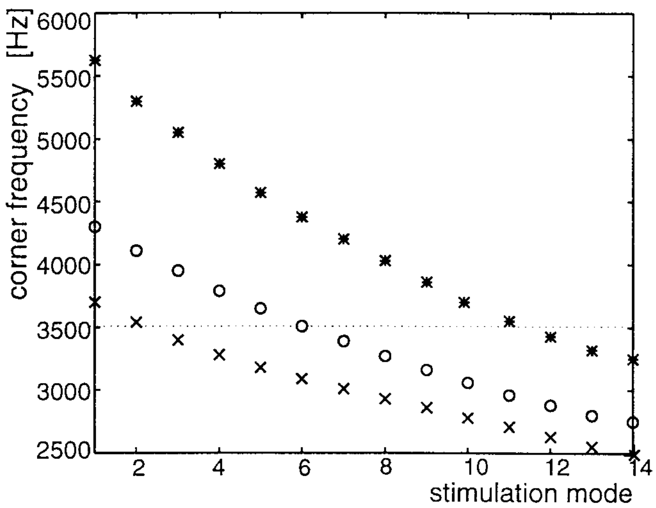



Fig. 8. Corner frequency as a function of stimulation mode (= number of elektrodes) For $*$, $\circ$, and $\times$ see Fig. 6. A 14-channel filter bank fourth-order filter. In the interleaved mode ($*$) the system transmits a frequency range of 3.5 kHz up to an 11-polar stimulation. Pulsatile stimulation in the single pulse mode ($\circ$) reaches a bandwidth of 3.5 kHz with six-polar stimulation. With phase-locked stimuli, only bipolar stimulation is possible.

simultaneously are edited to serial pulses (see Fig. 4). With this mode, 14 stimulation channels with up to six-polar stimulation can be realized with a corner frequency of 3.5 kHz.

### C. Place-Related and Phase-Locked Pulsatile Stimulation

The following coding strategy comes closer to the neuronal exciting patterns evoked in the auditory nerve by speech stimuli [35]–[38]. The front-end extracts all information and the data vector $\vec{P}$ defines an information pulse. The collision matrix allows the generation of stimuli in all channels simultaneously. In this case 14 channels can be realized with bipolar stimulation (Fig. 8) at 3.5 kHz.

The corner frequency which can be achieved lies between operation modes B and C (Sections IV-B and IV-C), depending on the number of channels that are simultaneously activated (PrPo multipulse stimulation). As already mentioned, adjacent stimulation channels must not normally be activated simultaneously for biological reasons. The minimal distance between the electrodes which can be activated simultaneously is termed the *crosstalk-radius*. If, for example the crosstalk-radius is set to three electrodes, a maximum of five simultaneous stimulation pulses are possible with a 14-channel filter bank. Thus for practical use, the system can provide quadripolar stimulation at 3.5 kHz if the crosstalk-radius is three electrodes and 14 stimulation channels are available.

In Fig. 9(a), the corner frequency is illustrated as a function of the filter bank used for monopolar, bipolar, and tripolar stimulation modes. It can be seen that the corner frequency changes as a function of number of stimulation channels and stimulation mode, respectively. A corner frequency of 3.5 kHz is maintained in every case. Fig. 9(b) illustrates the $n$-polar mode with 8–13 channels. This implies that each individual information pulse is fed into each electrode channel. A significant reduction of the corner frequency occurs as the number of data accesses of the signal router increases proportional to the second power of the channel number. Thus, with $n$-polar stimulation a maximum of 11 channels can be used in mode C (place-related and phase-locked pulsatile stimulation mode).

### D. Influence of Logic Operators

The system allows linear and logic operators to be defined and applied to the data elements. The linear operations are addition, multiplication and the logic *and*/*or*. Fig. 10 shows the corner frequency as a function of the operator used. The filter bank of the front-end in this particular example consists of four or eight channels with fourth-order filters and stimulation mode C. In this figure, $F$ signifies the corner frequency of the front-end mode C [compare Fig. 6(a)]. In $D$ the 20 D/A-servers and in $L$ the 23 lookup-tables are added. The corner frequency becomes smaller with higher complexity.

The corner frequency is represented in the right part of Fig. 10 as a function of the operators, including the D/A-server and the lookup-tables. It can be seen, that with 3.5 kHz and a four-channel filter bank, 450 logic operators can be used, whereas with an eight-channel filter bank 270 operators are possible.

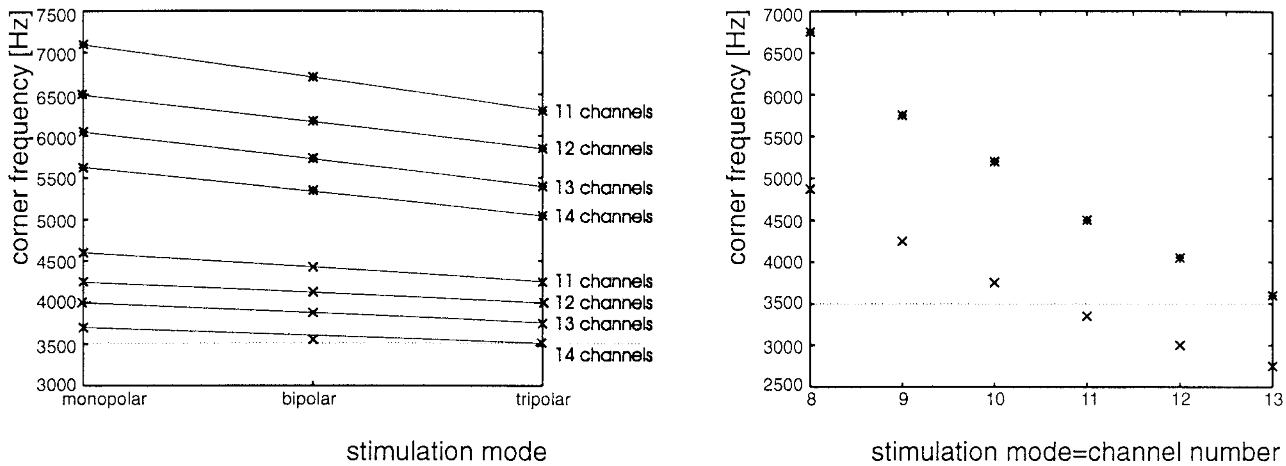
## V. Discussion

### A. Adaptability to External Hardware

The high versatility of the system allows its use for animal experimentation as well as for patients. In animal experiments, the stimulation electrodes are directly coupled to the current sources which are linked to the D/A converter-board. This allows simultaneous control of the current in the different electrode channels. In this configuration, full use is made of the versatility of the system. Likewise, the present system could be used in conjunction with the human Symbion system [39].

In patients with transcutaneous signal transfer, direct access to the individual electrodes is impossible. Here, the system offers three possibilities for connection to an implanted device. First, the current readings for the individual electrode channels are provided simultaneously in analogue form via the 14-channel D/A converter-board. Second, the stimulation pattern generated can be edited via a 16-b parallel port in digitized form. The electrode channel is coded with the upper 4 b and the lower 12 b code the current values. The patient's device has to be able to accept a free definition of the electrode channel including stimulation amplitude for the coding of pulsatile stimulation. The stimulus generation proper has then to be performed by the patient's implant. Electrode channel, instant of stimulation and stimulation current for the pulsatile stimuli are set by the present system. An example of a compatible interface is the MED-EL COMBI-40+ device [40].

Full use of the versatility of the present system can be made if the stimulation current of the individual electrode channels can be controlled freely and independently. If a patient's implant is equipped with digital control of all electrode channels and current values, the data can be transmitted serially by the 16-b parallel port. To guarantee simultaneous setup of the




(a) (b)

Fig. 9. (a) Corner frequencies of the filter banks with 11–14 channels for mono-, bi-, and tripolar stimulation mode and interleaved ($*$) or pulsatile ($\times$) stimulation. (b) Corner frequency as a function of number of channels (8–13) with $n$-polar stimulation mode.

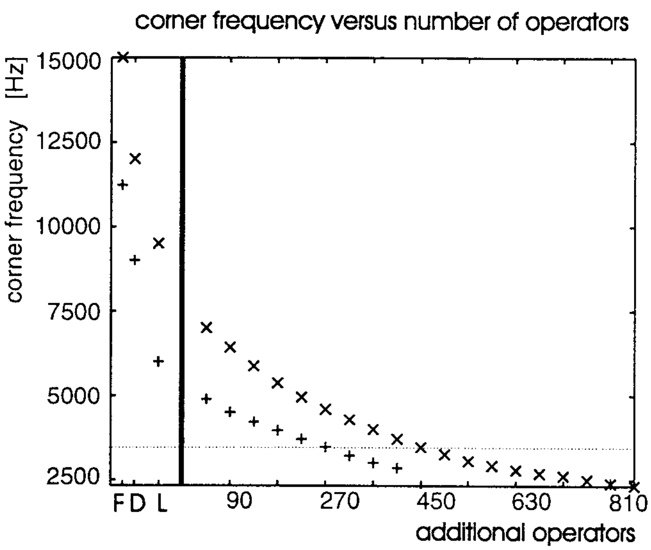



Fig. 10. Corner frequency as a function of the operators used for a four-channel ($\times$) and eight-channel ($+$) filter bank in phase-lock mode ($*$ in Figs. 6 and 7), fourth-order filter. Left $F$: filter bank only. $D$: filter bank and additionally 20 D/A servers. $L$: filter bank, D/A-servers and 23 lookup tables. Right: Frequency range as a function of additional number of logic operators of the signal router.

current values in spite of serial data transfer the timing for the validity of the data transferred is set via the 8-b parallel port.

### B. Choice of the DSP-Processor

The system enables the user to integrate new types of functions into the stimulation program without having to change the hardware. The chosen solution requires only that new processor software is implemented if new technologies are to be adapted.

The choice of the DSP32C was based on the following considerations: The DSP32C allows 25 Mflops, which is sufficient for real-time operations. It also uses a floating-point processor the quantization noise of which can be neglected. As future signal processors are expected to be largely equipped with floating point arithmetic, the transfer of software can easily be undertaken. The development system used, the TXP32CP is comparable to the Ariel- or Loughboury development systems. Implementation with development hardware in other laboratories should therefore be relatively easy.

### C. Choice of Software

The signal processor development system is programmed in in-line assembler, i.e., there are no explicit function calls. This has the advantage of an optimal utilization of the performance of the processor. However, it requires programming effort, particularly when implementation on other signal processor development systems is intended. To minimize this effort, only in-line functions are stored in the memory of the processor. There is no link between these functions and the main program. The functions are linked to the signal processor by the signal flow control. The clarity of these functions allows an easy transfer to the assembler of a different signal processor as they are independent of the main program. New features are simply established by the addition of appropriate standardized functions, operated by means of the signal flow control. The control of the entire signal processing is based on the user interface programmed in C++. By appropriate compilation, the user interface can be implemented in any target hardware.

The idea to gain independence of the signal processor development system by programming in C was rejected. In this case, namely, it would be useless to write a program that maps well to the hardware of the DSP32C. Hardware-independent programming, on the other hand, would reduce the performance by about 90%, which would severely limit the possibilities of real-time processing.

One of the main advantages of the present system is the possibility of setting up the signal flow by a mouse-controlled input. The data flow is represented by lines on the screen linking the respective functional symbols with their data vectors which gives high versatility. The graphical display reduces the number of connections possible, since too many lines cannot be resolved by the eye. However, the channels presently available for the functional units have been found to be sufficient for the design of stimulation patterns. Only 14 D/A converters for controlling the current sources have been realized as no more were needed. There are no such limitations for the software.

### *D. Place-Related and Phase-Oriented Multipulse Stimulation (PrPo)*

The PrPo stimulation shown in Fig. 5; stimulating with one pulse/channel and one pulse/phase has not yet been realized in clinical devices. In fact, there may be a disadvantage of this coding strategy because of slow pulse repetition rate with low-frequency stimuli: Single pulses with low repetition rate could eventually be resolved by the patient and be perceived as rattling sounds [41]. A further disadvantage is the phase shift of the individual pulses. These shifts are produced by the collision matrix in order to avoid crosstalk between neighboring stimulation channels. The dissolution of the strict phase-lock, however, can also have beneficial effects. With natural sound stimulation the phase-lock of action potentials is much weaker than with electrical pulsatile stimulation. Thus, the collision matrix counteracts the "hypersynchronization" in the trains of action potentials evoked by electrical stimulation [42], [43].

### *E. Continuous Interleaved Sampling (CIS)*

Speech recognition scores increase with higher sampling rates using a CIS strategy [44]–[46]. The MED-EL COMBI 40+ system can use up to 12 channels with a maximum pulse repetition rate of 18 kpulses/s at present. As a consequence, the phase information in the 12-channel mode can only be maintained up to a frequency of 750 Hz (half sampling rate) and in the six-channel mode up to 1500 Hz. In channels which represent higher frequencies, no phase information can be transformed.

The system here described is able to provide a maximum pulse repetition rate up to 154 kpulses/s with 14 channels, as can be seen in Fig. 8. Each channel can sample with 11 kpulses/s and, thus, transmit phase information of up to 5.5 kHz.

In conclusion, the present system provides the possibility of maintaining phase information with high stimulation rates and in all the frequency bands (CIS) for the first time. It can also generate phase-locked pulses in the different frequency bands.

The system allows the free design of new stimulation patterns and can be used for experimental studies in both animals and patients.

Any details will be provided by the authors.

## ACKNOWLEDGMENT

The authors would like to to thank Dr. M. Müller for critically reading the manuscript and F. Hartwich for helpful suggestions in programming.

## REFERENCES

[1] G. v. Bekesy, *Experiments in Hearing*. New York: McGraw-Hill, 1960.
[2] M. C. Vivion, M. M. Merzenich, P. A. Leake-Jones, M. White, and M. Silverman, "Electrode position and excitation patterns for a model cochlear prosthesis," *Ann. Otol. Rhinol. Laryngol. Suppl.*, vol. 82, pp. 19–20, 1981.
[3] G. Langner, "Periodicity coding in the auditory system," *Hear. Res.*, vol. 60, pp. 115–142, 1992.
[4] C. H. Chouard and R. Sposetti, "Environmental and electrophysiological study of absolute pitch," *Acta Otolaryngol. (Stockh.)*, vol. 111, pp. 225–230, 1991.
[5] D. K. Eddington, "Speech discrimination in deaf subjects with cochlear implants," *J. Acoust. Soc. Amer.*, vol. 68, no. 3, pp. 885–891, 1980.
[6] N. Dillier, H. Bogli, and T. Spillmann, "Speech encoding strategies for multielectrode cochlear implants: A digital signal processor approach," *Progress in Brain Res.*, vol. 97, pp. 301–311, 1993.
[7] R. A. Schindler and M. M. Merzenich, *Cochlear Implants*. New York: Raven, 1985.
[8] J. H. M. Frijns, "Cochlear implants a modeling approach," Masters thesis, Rijksuniversiteit Leiden, Leiden, the Netherlands, 1995.
[9] R. P. Michelson, M. M. Merzenich, R. A. Schindler, and D. N. Schindler, "Present status and future development of the cochlear prosthesis," *Ann. Otol. Rhinol. Laryngol.*, vol. 84, pp. 494–498, 1975.
[10] A. J. Fourcin, S. M. Rosen, B. C. J. Moore, E. E. Douek, G. P. Clarke, H. Dodson, and L. H. Bannister, "External electric stimulation of the cochlea: Clinical, psychophysical, speech-perceptual and histological findings," *Audiol.*, vol. 13, pp. 85–107, 1979.
[11] J. R. Walliker, R. Rosen, E. E. Dousek, A. J. Fourcin, and B. C. J. Moore, "Speech signal presentation to the totally deaf," *J. Biomed. Eng.*, vol. 5, pp. 316–320, May 1983.
[12] W. F. House, K. Berliner, W. Crary, M. Graham, R. Luckey, N. Norton, W. Selters, H. Tobin, and J. Urban, "Cochlear implants," *Ann. Otol. Rhinol. Laryngol.*, vol. 85, pp. 1–93, 1976.
[13] E. S. Hochmair, "An implantable current source for electrical nerve stimulation," *Int. J. Artif. Organs*, vol. 2, pp. 255–261, 1979.
[14] W. F. House, "Cochlear implants—Past, present and future," *Cochlear. Implants: New Perspectives*, vol. 48, p. 3, 1993.
[15] G. M. Clark, R. Black, D. J. Dewhurst, I. C. Forster, J. F. Patrick, and Y. C. Tong, "A multiple-electrode hearing prosthesis for cochlear implantation in deaf patients," *Med. Progr. Technol.*, vol. 5, pp. 127–140, 1977.
[16] G. M. Clark, "Electrical stimulation of the auditory nerve: The coding of frequency, the perception of pitch and the development of cochlear implant speech processing strategies for profoundly deaf people," *Clin. Exp., Pharmacol. Physiol.*, vol. 23, pp. 766–776, 1996.
[17] R. A. Schindler and D. K. Kessler, "Clarion cochlear implant: Phase I investigational results," *Amer. J. Otol.*, vol. 14, no. 3, pp. 263–272, 1993.
[18] N. Tönder, R. Hartmann, and R. Klinke, "Sprachkodierung für Mehrkanal-Hörprothesen auf dem DSP32C-Signalprozessor," in *Proc. Fortschritte in der Akustik-DAGA 1993*, pp. 776–779.
[19] ——, "Sprachkodierung für Mehrkanal-Hörprothesen auf dem DSP32C-Signalprozessor," *Biomedizinische Technik*, band 41, pp. 600–601, 1996.
[20] B. S. Wilson, C. C. Finley, J. C. J. Farmer, D. T. Lawson, B. A. Weber, R. D. Wolford, P. D. Kenan, M. W White, and M. M. Merzenich, "Comparative studies of speech processing strategies for cochlear implants," *Laryngoscope*, vol. 98, pp. 1069–1077, 1988.
[21] M. W. White, M. M. Merzenich, and J. Gardi, "Multichannel cochlear implants: Channel interactions and processor design," *Archs Otolar.*, vol. 110, pp. 493–501, 1984.
[22] E. Zwicker, *Psychoakustik*. Berlin, Germany: Springer-Verlag, 1982.
[23] R. Hartmann and R. Klinke, "Response characteristics of nerve fibers to patterned electrical stimulation," in *Cochlear Implants. Models of the Electrically Stimulated Ear*. New York: Springer-Verlag, 1990, pp. 135–160.
[24] D. Mortazavi, "Untersuchungen zur Einschränkung des elektrischen Feldes und Verbesserung der Kanaltrennung bei intracochleärer elektrischen Reizung," Masters thesis, Johann Wolfgang Goethe–Universität, Frankfurt, Germany, 1995.
[25] T. Ifukube and R. L. White, "A speech processor with lateral inhibition for an 8 channel cochlear implant and its evaluation," *IEEE Trans. Biomed. Eng.*, vol. BME-34, pp. 876–882, Nov. 1987.
[26] M. M. Merzenich, "Coding of sound in a cochlear prosthesis—Some theoretical and practical considerations," in *Cochlear Prosthesis*, Ch. W. Parkins and S. W. Andersen, Eds. New York: New York Acad. Sci., 1983, pp. 502–508.
[27] J. Tillein, N. Tönder, W. Stöcker, R. Hartmann, and R. Klinke, "Electrical stimulation of nerve cells in brain slices: A model for multi-channel-auditory prosthesis," in *Proc. 5th Vienna Int. Workshop on Functional Electrostimulation (FES)*, 1995, pp. 189–192.

[28] ——, "Studying the efficiency of multi-channel-electrodes for cochlear prosthesis using the brain slice technique," presented at *4th Seminar on Audiology and Phoniatric*, Bozen/South Tyrol, Dec. 13–16, 1995.

[29] ——, "Spatial resolution of multi-channel auditory prosthesis tested with rat brain slices," in *Proc. 24th Göttingen Neurobiology Conf. 1996*, N. Elsner and H. U. Schnitzler, Eds., vol. II, p. 459.

[30] ——, *Untersuchungen Zur Verbesserung der Lokalen Reizauflösung bei Elektrischer Reizung in Cochlea-und Hirnstammimplantaten (in vitro Experimente)*. Zürich, Switzerland: BMT-Kongress, 1996.

[31] R. Hartmann and R. Klinke, "Impulse patterns of auditory nerve fibers to extra- and intracochlear electrical stimulation," *Acta Otolaryngol. (Stockh.)*, vol. 469, pp. 128–134, 1990.

[32] J. Genin, "Electrical interactions: Measurement and correction," in *Cochlear Implant: Current Situation*, P. Banfai, Ed. Erkelenz, Germany: Rudolph Bermann Gmbh, 1988, pp. 359–364.

[33] B. S. Wilson, D. T. Lawson, M. Zerbi, and C. C. Finley, "Recent developments with the CIS strategies," in *Advances in Cochlear Implants: Proc. 3rd Int. Cochlear Implant Conf.*, I. J. Hochmair-Desoyer, Ed., 1994, pp. 103–112.

[34] B. S. Wilson, C. C. Finley, D. T. Lawson, R. D. Wolford, D. K. Eddington, and W. M. Rabinowitz, "Better speech recognition with cochlear implants," *Nature*, vol. 352, pp. 236–238, 1991.

[35] P. Lutter, E. H. Mark, and F. Rattay, "Speech coding in the auditory nerve," submitted for publication.

[36] F. Rattay and P. Lutter, "Speech sound representation in the auditory nerve: Computer simulation studies on inner ear mechanisms," submitted for publication.

[37] M. Knauth, R. Hartmann, and R. Klinke, "Discharge pattern in the auditory nerve evoked by vowel stimuli—A comparison between acoustical and electrical stimulation," *Hear. Res.*, vol. 74, pp. 247–258, 1994.

[38] B. Delgutte, "Representation of speech-like sounds in the discharge patterns of auditory-nerve fibers," *J. Acoust. Soc. Amer.*, vol. 68, pp. 843–857, 1980.

[39] M. F. Dorman, M. T. Hannley, G. A. McCandless, and L. M. Smith, "Auditory phonetic categorization with the symbion multichannel cochlear implant," *J. Acoust. Soc. Amer.*, vol. 84, no. 2, pp. 501–510, 1988.

[40] C. M. Zierhofer, I. J. Hochmair-Desoyer, and S. Hochmair, "Electronic design of a cochlear implant for multichannel high-rate pulsatile stimulation strategies," *IEEE Trans. Rehab. Eng.*, pp. 112–116, 1995.

[41] P. Pialoux and C. H. Chouard, "Physiological and clinical aspects off the rehabilitation of total deafness by implantation of multiple intracochlear electrodes," *Acta Otolaryngol. (Stockh.)*, vol. 81, p. 436, 1976.

[42] R. Hartmann, G. Topp, and R. Klinke, "Discharge patterns of cat primary auditory fibers with electrical stimulation of the cochlea," *Hear. Res.*, vol. 13, pp. 47–62, 1984.

[43] R. K. Shepherd, C. L. Maffi, S. Hatsushika, E. Javel, Y. C. Tong, and G. M. Clark, "Temporal and spatial coding in auditory prostheses," in *Information Processing in the Mammalian Auditory and Tactile Systems*, M. J. Rowe and L. M. Aitkin, Eds. New York: Liss, 1990, pp. 281–293.

[44] E. Hochmair, "Kodierungsstrategien für cochlea-implantate," *Biomedizinische Technik*, pp. 22–27, 1996.

[45] B. S. Wilson, D. T. Lawson, C. C. Finley, and R. D. Wolford, "Coding strategies for multichannel cochlear protheses," *Amer. J. Otol.*, vol. 12 (Suppl. 1), pp. 55–60, 1991.

[46] N. Dillier, W. K. Lai, and H. Bögli, "A high spectral transmission coding strategy for a multi-electrode cochlear implant," in *Advances in Cochlear Implants*, I. J. Hochmair-Desoyer and E. S. Hochmair, Eds. Wien, Austria: Manz, 1993, pp. 152–157.

[47] P. Pohl, O. Peter, C. Zierhofer, R. Kaufmann, T. Czylok, I. J. Hochmair-Desoyer, and E. Hochmair, "Comparisom of combined analog and pulsatile coding strategies" in *Advances in Cochlear Implants*, I. J. Hochmair-Desoyer and E. S. Hochmair, Eds. Wien, Austria: Manz, 1993, pp. 158–165.


**Norbert Tönder** was born in Flensburg, Germany, in 1963. He received the diploma in physics in 1994 from the Goethe-Universität, Frankfurt am Main, Germany. He is currently working toward the Ph.D. degree in physics at the institute "Physiology III" of the Goethe-Universität.

His research interests are electrophysiology, mathematical modeling of electrical stimulation, signal processing, and imaging.

**Rainer Hartmann** was born in 1944 in Bludenz, Austria. He received the Dipl. Ing. degree in communication engineering at the Technical University, Aachen, Germany, with special interest in psychoacoustics. In 1971 he started neurophysiological experiments at the Free University of Berlin with system-analysis of the afferent/efferent pathways in the vestibular system; graduation to Dr. Ing. at the Technical University of Berlin, Berlin, Germany.

Since 1977, he has pursued basic neurophysiological research on electrical stimulation of deaf ears for cochlear implants at the Goethe-Universitaet, Frankfurt am Main, Germany.

Dr. Hartmann is a Member and Project Leader in research projects sponsored by the Deutsche Forschungsgemeinschaft.

**Rainer Klinke** was born in Landsberg, Germany, in 1936. He studied medicine and mathematics at the Universities of Erlangen, Vienna, and Heidelberg, where he received the M.D. degree.

Following medical training he worked in a number of physiology departments and was appointed Associate Professor at the Free University of Berlin, Germany. In 1977 he accepted a chair in Physiology at the Goethe-Universität Frankfurt am Main, Germany. His main interest are neurophysiology and sensory physiology particularly of the auditory system. Presently his research is devoted to the basic physiology of cochlear implants and to neuronal plasticity in the central auditory system. He is coauthor and editor of a number of textbooks in physiology and member of the editorial boards of, among others, *Hearing Research*.