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Pitch perception and signal processing in electric hearing

[Tonhöhenwahrnehmung und Signalverarbeitung bei elektrischem Hören]

Dissertation
zum Erwerb des Doktorgrades der Humanbiologie
an der Medizinischen Fakultät der
Ludwig-Maximilians-Universität zu München

vorgelegt von
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aus
München

Jahr 2004

Mit Genehmigung der Medizinischen Fakultät der Universität München

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Tag der mündlichen Prüfung:	15.12.2004

INTRODUCTION

The loss of no other sense organ reduces the quality of life more than the loss of the sense of hearing. Deaf patients not only loose the pleasure of hearing their own child, a little bird and music etc. but they completely loose the possibility to communicate acoustically with their social environment. Those patients grown up in a deaf environment are able to manage the communication by use of sign language and nowadays more and more via fax, short messages and internet. However, there is another group of patients deafened by a progressive or sudden hearing loss who were able to hear normally for a long period of time and who were grown up in an environment based on oral communication. This patient group suffers immensely because they are completely cut from their social environment. The problem arises with the onset of progressive hearing loss and the use of hearing aids. Patients are withdrawing more and more from oral communication, first from the contact with unfamiliar persons, then from the contact with groups, then also from the contact with familiar persons.

There are several categories of hearing loss. Mild hearing loss is defined as an average pure tone threshold at 500, 1000 and 2000 Hz by 26 to 40 dB, moderate hearing loss by 41 to 55 dB, a moderate to severe hearing loss by 56 to 70 dB, severe hearing loss by 71 to 90 dB and profound hearing loss by more than 91 dB (Goodman, 1965). Hearing loss can be influenced by several factors. There is conductive hearing loss which is associated with damages in the outer or middle ear as an ossification of the middle ear bones or the accumulation of fluid behind the eardrum. Conductive damages are reducing the hearing by maximally 60 dB and can mostly be treated surgically. Permanent conductive hearing losses are reducing the transmission of energy to the cochlea and can generally be corrected by the amplification of the sound by a hearing aid. Other damages occur in the inner ear and are described as sensorineural hearing losses. Mainly, there is a damage of the inner and/or outer hair cells. The loss of hair cells reduces the ability of the inner ear to transduce the mechanical movement within the cochlea to neural activity in the auditory nerve. The major cause of damage to hair cells is exposure to noise. Medical conditions that can cause damage of hair cells include Menier's disease, ototoxic drugs, viral and bacterial infections or lack in the autoimmune system. Other damages of the inner ear are caused by a loss of the intracochlear fluid, an ossification of the cochlea, otitis media, craniocerebral injury, barotraumas or acoustic neuromas.

If the amount of hearing loss is that severe that amplification of the sound with a hearing aid in best conditions results in an insufficient level of speech perception (Lenarz et al., 2002), a cochlear implant is indicated for postlingually deafened adults with severe to profound hearing loss. Cochlear implants are directly stimulating the auditory nerve and, that way, bypass the mechanical-neural mechanism of the organ of Corti including the inner and outer hair cells. Modern cochlear implants provide electrical stimulation via an electrode array with a number of electrodes. The most current implant types are the CI24RCA by Cochlear (Melbourne, Australia), the HiRes90K by Advanced Bionics (Sylmar, United States of America) and the COMBI 40+ by MED-EL (Innsbruck, Austria). The implants differ mainly in the number of stimulating electrodes and their intracochlear position (Fig. 1). The CI24RCA consists of 22 intracochlear electrodes which are spaced 0.75 mm and are positioned between an 8-mm and a 23.75-mm distance from the round window when the electrode array is fully inserted (up to last stiffening ring). The Hires90K consists of 16 electrodes which are spaced 1.1 mm and are positioned between a 7-mm and a 23.5-mm distance from the round window when the array is fully inserted (up to shoulder of array). The COMBI 40+ consists of 12 electrodes which are spaced 2.4 mm and are positioned between a 3.9 and 30.3 mm distance from the round window when the array is fully inserted (up to silicone ring).

The electrode array is inserted into the scala tympani of the cochlea by a small hole, called cochleostomy, near the round window. The array is connected with a receiver stimulator unit which is embedded into the temporal bone behind the ear. The receiver includes a magnet to fix the external equipment at the head. The external equipment consists of a speech processor which is worn behind the ear and a communication coil. The acoustic signal is detected by a microphone which is part of the speech processor. The speech processor converts the acoustic signal into electrical stimulation pulses which are delivered to the receiver under the skin by the communication coil with an opposing magnet.

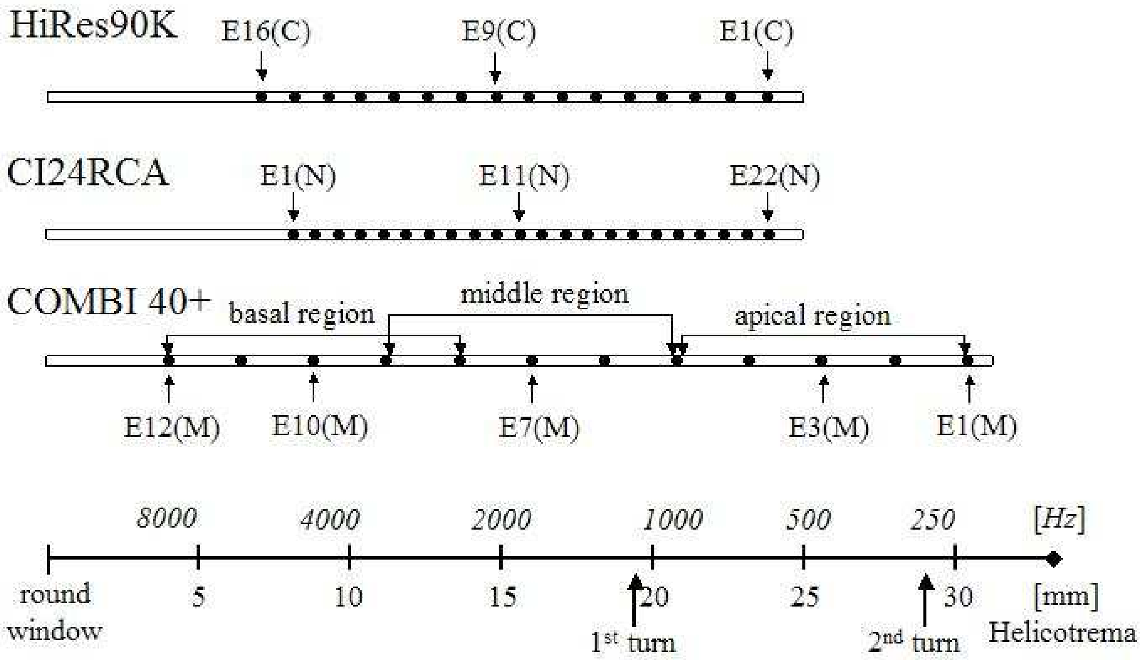

FIGURE 1. *Schematic drawing of three different electrode arrays of the cochlear implants HiRes90K by Advanced Bionics (Sylmar, United States of America), CI24RCA by Cochlear (Melbourne, Australia) and COMBI 40+ by MED-EL (Innsbruck, Austria). The electrode arrays with the numbering of the electrodes are shown according to their position along the cochlea. The distance in mm from the round window as well as the best frequencies along the cochlea according to the frequency-place allocation in normal hearing (Zwicker & Fastl, 1999) is indicated. The different defined cochlear regions for experiment 1 (page 16) are noted.*

About four weeks after the implantation of the internal components, the speech processor is individually adjusted. For each electrode, the current is slowly increased until the threshold of hearing is just reached. This is called the threshold level (THR). The current amplitude is then increased until the maximum comfortable level (MCL) is reached. The stimulation of a single electrode is referred as a perception of a tone. The THR and MCL are individual for each recipient and electrode. They define the dynamic range of each electrode. The electrodes are stimulating different regions of the cochlea. Similar to the frequency-place allocation in normal hearing (Zwicker & Fastl, 1999), different regions in the cochlea evoke different pitch perceptions. The pitch is increasing from the apex to the base of the cochlea. This tonotopy is implemented in the speech processing strategy. The incoming acoustic signal is band pass filtered and the filtered signals are then coded to stimulate according electrodes. Low frequency filters are allocated to electrodes in the apical region, high frequency filters are allocated to electrodes in the basal region. The energy of the incoming acoustic signal in each band is mapped for each electrode between THR and MCL level. That means the frequency characteristic of the acoustic signal is presented by the place and the amplitude of stimulation.

Most cochlear implant recipients reach a high level of speech perception, namely about 45% correct for monosyllables, about 80% correct for word recognition and sentence recognition with a great interindividual variance (Fettermann & Domico, 2002; Gstöttner et al., 2000; Hamzavi et al, 2001; Helms et al, 1997; Pasanisi et al., 2003; Valimaa & Sori, 2001). The success of the cochlear implant for adults correlates with the duration of deafness (Friedland et al., 2003; Gomaa et al., 2003, Hamzavi et al., 2003), sentence recognition before implantation (Gomaa et al., 2003) and factors like residual hearing, age at implantation and nerve survival. For the majority of recipients it enhances the quality of life because it allows the way back to oral communication with the environment. However, it can not replace a normal hearing ear. Most recipients complain about poor speech recognition in noise. The average result for a sentence test in noise (Oldenburger Satztest, Wagener et al., 1999a-c) for 12 subjects with excellent speech perception in quiet and regular telephone use is 0.16 dB signal to noise ratio measured for a speech recognition level for 50% correct words (Nobbe & Baumann, 2004). Normal hearing listeners reach a signal to noise ratio of about -8 dB tested with both ears and measured with speech and noise signals from a front loudspeaker (Beutelmann et al., 2003). This effect is partly due to the fact that most recipients are implanted only monolateral due to the high costs of a cochlear implant system. Several studies show that the speech perception performance is enhanced for subjects with bilateral implantation (Litovsky et al., 2004; Nopp et al., 2004; Schön et al., 2002; Tyler et al., 2003; van Hoesel et al., 1993; van Hoesel & Tyler, 2003). The detection of speech in noise is also affected by the limitations in the speech processing strategy and the electrode array design. Depending on the implant design, the acoustic signal is analyzed by 12 to 22 band pass filters and this way, the spectral information is reduced. Fine spectral changes in the signal can not be transmitted by the speech processor. Moreover, it is not guaranteed that all electrodes on the array evoke different pitch perceptions. Several studies with the Nucleus CI22M (Clark, 1987) have shown that for this implant type with an electrode distance of 0.75 mm, there are electrodes indiscriminable in pitch (Busby & Clark, 1996; Collins et al., 1997; Donaldson & Nelson, 2000; Henry et al. 2000; McKay et al., 1999; Nelson et al., 1995; Pfingst et al., 1999; Zwolan et al., 1997). The electrode discrimination ability increased with electrode distance. The effect of indiscriminable electrodes causes a further loss of the fine spectral information in the incoming signal.

Another limiting factor for the appraisal of cochlear implants is the perception of music. Cochlear implant recipients often complain about the poor sound quality of music with the implant. They mostly report being unable to recognize familiar songs due to the very different sound quality and transmission of melody contours. Leal et al. (2003) reported that 38% of their examined subjects (a total of 29) do not enjoy listening to music with their device. 86% of them report that they have reduced their listening habits after the implantation. Tyler et al. (2000) have analyzed the experience of cochlear implant recipients with music.

They found that 83% of the recipients report a decline in musical enjoyment and half of them report that the sound of music is unpleasant or difficult to follow. Several studies have tested melody recognition and pitch discrimination with cochlear implant users. Nobbe et al. (2004) tested forty cochlear implanted subjects with familiar German melodies and report recognition scores between 12 and 90%. Gfeller et al. (2000) have compared the recognition of familiar melodies by cochlear implant recipients with normal-hearing subjects. They found that the 49 cochlear implant users recognized in average only 13% in contrast to the normal-hearing subjects who were able to recognize 55% of the melodies. Pijl & Schwarz (1995) report for seventeen subjects an averaged recognition of familiar melodies of 44% when presented directly to the electrode array, but recognition was compromised when the stimuli were presented after being filtered by the speech processor. Fujita & Ito (1999) also found great interindividual differences for the melody recognition of their eight subjects. They as well as the authors cited above indicated a greater recognition for those melodies with recognizable rhythmic patterns.

The insufficient presentation of musical sounds is also caused by the limitations of the implant device and the speech processing strategy. To differentiate between different musical tones and instruments (Kong et al., 2004) and for the speech perception of tonal languages (Fu et al., 2002) much more spectral fine structure is needed than for the representation of speech signals. Again the representation of spectral fine structure is limited by the number of independent channels on the electrode array. The number of independent channels is generally not restricted by hardware components as the lack of technical possibilities but by the fact that the stimulation of each electrode causes an electric field which is not stimulating the auditory nerve with the same accuracy as the fine tuned inner hair cells (Kral et al., 1998). There are different approaches to enhance the precision of the electrical stimulation. One strategy to decrease the neural spread of masking is the use of electrode arrays which curve around the modiolus after the insertion into the cochlea (Cohen et al., 2001; Fayad et al., 2000; Kuzma &

Balkany, 1999). This way, the electrodes are closer to the auditory nerve and are expected to stimulate a smaller and more precise number of auditory nerve fibers. However, the effect of this new electrode array design is discussed (Boëx et al., 2003) and the method seems insufficient to facilitate an increase in the number of independently discriminable electrodes on the array. That means that if current electrode arrays and ways of stimulation do not provide a larger number of independent channels, other methods have to be examined to provide a better presentation of the spectral fine structure in a signal with the current system. A better presentation of the spectral fine structure is expected to result in a better speech perception in noise and a better representation of musical sounds and melody contours.

In normal hearing the sensation of pitch is coded with place and time information in the auditory nerve. The traveling wave of the acoustic signal is first transformed into mechanical oscillation in the middle ear and is then transmitted to the inner ear at the oval window which is a thin membrane at the basal end of the cochlea. The cochlea consists of three channels, the scala vestibuli, scala media and scala tympani (Fig. 2).

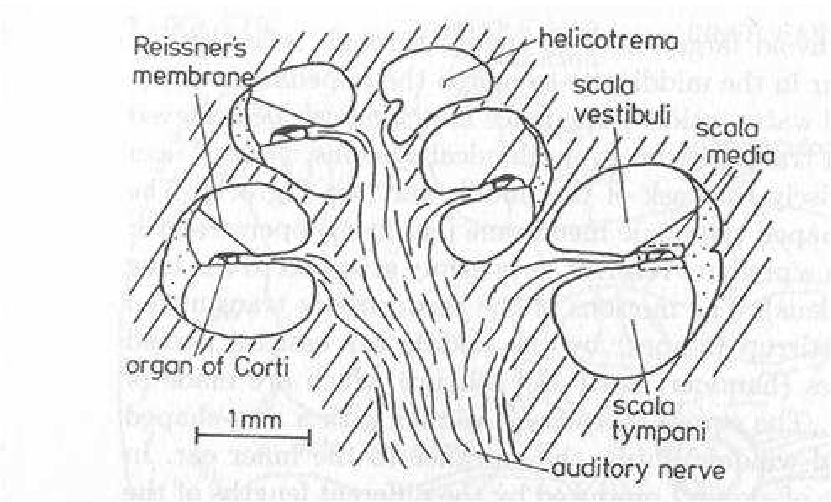

FIGURE 2. *Schematic drawing of the cross section of the cochlea with the scala vestibule, scala media and scala tympani and the enervation of the auditory nerve (taken from Zwicker & Fastl, 1999).*

The oval window is connected with the scala vestibuli. At the apical end of the cochlea, the scala vestibuli is connected with the scala tympani. Both scalas are filled with perilymph. In normal hearing the oscillation of the oval window causes an oscillation of the perilymph and the basilar membrane. The basilar membrane oscillates with maxima at different places for different oscillation frequencies due to its decreasing stiffness from base to apex. In this way, the incoming signal is coded into different places of stimulation along the cochlea. High frequencies evoke a maximal oscillation frequency at the base, low frequencies at the apex and both are stimulating different auditory nerve fibers. The frequency of the incoming signal is also coded by time information. The oscillation frequency of the basilar membrane is transmitted by the firing rate of the neurons connected with the inner hair cells. This effect is limited to stimulation rates up to 2000 Hz due to the refractory time (about 0.5 ms) of the auditory nerve fibers (van den Honert et al., 1997; Abbas et al., 1999).

Current signal processing strategies in cochlear implants are based on pitch changes due to different places of stimulation. The stimulation rate at each electrode is constant. That means that the place of stimulation does not correspond to the rate of stimulation. Research is done to enlarge the pitch spectrum of cochlear implants with including not only changes in the place of stimulation but also in the rate of stimulation (for example Fearn, 1999).

This thesis elaborates different factors to influence the pitch perception with electrical stimulation in order to create a transformed speech processing strategy based on the MED-EL COMBI 40+ implant. This implant provides an especially deep insertion of the electrode array into the cochlea (up to 30.3 mm from the round window, Fig. 1) and a wide electrode spacing (2.4 mm). When fully inserted, the distribution of electrodes along the cochlea in the COMBI 40+ allows a more detailed analysis of the effect of electrode position along the cochlea on pitch than the other implant types. Furthermore, with the deep insertion of the electrode array and the stimulation of apical regions in the cochlea, it is possible to reduce the mismatch between place and rate of stimulation. In the apical region of the cochlea where low frequency sounds cause a maximal oscillation of the basilar membrane in normal hearing, the stimulation with a low pulse rate might lead to a more distinct pitch and a better representation of the signal.

This thesis is structured as following:

Chapter I (General Method) presents the subjects participating in the hearing experiments and describes the customized interface and software which was applied to stimulate a certain electrode of the MED-EL COMBI 40+ implant.

Chapter II (Hearing Experiments) describes experiments conducted to examine the parameters influencing pitch perception with the cochlear implant. **Experiment 1** (page 16) starts with a pitch ranking task for different electrodes of the MED-EL COMBI 40+ implant in order to find out if the electrode spacing is wide enough to provide discriminable electrodes based on pitch perception. In **experiment 2 to 5** (page 24 to 47) the influence of pulse rate is tested for four test electrodes along the array. The pitch height (**experiment 2**, page 24) and the sound quality (**experiment 3**, page 30) are rated for different pulse rates in order to investigate the upper limit of pitch changes depending on pulse rate at different electrode locations within the cochlea. In **experiment 4** (page 36) the work is then extended to an experiment investigating the just noticeable difference in pulse rate (pulse rate discrimination limen) in order to evaluate the accuracy of rate changes for a possible implementation in a ‘rate strategy’. In **experiment 5** (page 47) the test is repeated with amplitude modulated stimuli. **Experiment 6** (page 51) investigates the exact evoked pitch height depending on electrode position in the cochlea in a binaural electric-acoustic experiment with subjects with residual hearing whereat the acoustic frequency at the non-implanted ear was adjusted.

Chapter III (page 64) outlines the development of a new speech processing strategy whereby the results of the hearing experiments are incorporated.

The transformed speech processing strategy is tested for speech and music perception in **Chapter IV** (page 73) in comparison with the classical speech strategy for this implant.

In Chapter V (page 93) a comprehensive discussion of the results of the hearing experiments about pitch perception as well as of the results of the new speech processing strategy is carried out. The conclusions of this thesis are then related to the outcome of studies on these topics published by other research groups.

In **Chapter VI** (page 108) the main outcomes of this thesis are briefly summarized in English and in German.

I. GENERAL METHOD

1. Participants

Sixteen subjects participated in the hearing experiments. All of the subjects used the MED-EL COMBI 40+ device daily. The average age of subjects at the time of implantation was 54 years, with a range from 30 to 78 years (Table I).

TABLE I. *Demographical data of the subjects.*

Index	Sex	Age	Cause of deafness	Duration of deafness [m]	CI use Exp. 1 [m]	CI use Exp. 2 [m]	CI use Exp. 3 [m]	CI use Exp. 4 [m]	CI use Exp. 5 [m]	CI use Exp. 6 [m]	CI use Rate-CIS [m]
S1	M	64	Progr. degen.	3	39	43	45	48	53	60	64
S2	M	33	Trauma, progr. degen.	60	-	33	35	38	-	50	-
S3	M	65	Progr. degen., SHL	24	-	17	19	-	-	-	-
S4	F	78	Progr. degen.	8	30	-	-	-	-	41	-
S5	M	64	Cochl. Otosclerosis	360	27	31	33	36	41	-	52
S6	F	49	Progr. degen.	13	14	18	20	23	-	-	39
S7	F	38	Progr. degen.	33	17	21	23	26	31	38	42
S8	M	68	Toxic	5	-	55	57	58	-	-	-
S10	F	34	Sudden hearing loss	2	3	7	9	-	-	-	28
S11	F	30	Congenital, syndromal	72	47	-	-	-	-	-	-
S12	M	64	Progr. degen.	19	4	-	-	-	-	-	29
S13	M	55	Progr. degen.	552	-	26	28	31	-	43	-
S14	M	56	Sudden hearing loss	8	-	33	35	-	-	-	-
S15	M	51	Cochl. Otosclerosis	10	-	-	-	-	-	6	10
S16	M	65	Sudden hearing loss	3	-	-	-	-	-	-	12
S17	F	38	Congenital	12	-	-	-	-	-	-	51

All subjects had a profound to total sensorineural hearing loss in the implanted ear prior to implantation. Four of the subjects (S2, S11, S13 and S15) had sufficient residual hearing in the contralateral ear to make use of a hearing aid. In each subject, the position of the electrode array was radiologically examined using Stenvers' view x-ray scans of the subject's cochlea. The distance between neighboring electrode contacts was constant; no overlapping or kinking of the array could be identified. For most of the subjects, the electrode array was fully inserted into the cochlea with the exception of S3 where there were E11 and E12 outside the cochlea. Prior to data collection, the subjects had been using their implants for 4 to 51 months at the time of the beginning of the listening experiments and were average to ‘star’ users (from occasional to regular telephone use). The subjects received an allowance for participation in the study and gave informed consent. The design of the study was approved by the local ethical committee.

2. Stimulation hardware

The MED-EL COMBI 40+ implant allows stimulation at an overall pulse rate of 18,180 biphasic pulses per second (pps) in monopolar mode with an extra-cochlear reference electrode which is normally located under the temporalis muscle (Zierhofer et al., 1995). Whenever possible, biphasic current pulses with a pulse duration of 26.7 µs per phase were used. The pulse duration of three subjects had to be increased in order to achieve comfortable loudness at all electrodes (S5: pulse duration 26.7-35 µs per phase; S12: 40-70 µs per phase; S13: 26.7-40 µs per phase). The current amplitude was equal for each phase with the negative phase leading.

The stimuli were generated on an IBM-compatible PC using ‘Matlab’® software and stored as matrices with channel of stimulation, current amplitude, current range, pulse duration, and minimal pulse distance as parameters. For the channel of stimulation, one of the 12 channels could be chosen. The current delivered to the electrode was defined with the current amplitude and the current range. The current amplitude is binary coded with 7 bits between 0 and 127. There are four logarithmically equally spaced, partially overlapping current ranges which are coded between 0 and 3. Thus, a current range can be thought of as a base amplification of the biphasic pulse amplitude. The pulse duration is indicated as a value in the range between 16 and 255. A pulse duration of 26.7 µA corresponds to 16. The minimal pulse is determined from the onset of the current biphasic pulse to the onset of the following stimulation pulse. It is indicated as bit-value in the range between 33 and 1023. Thus, it defines the stimulation rate. The conversion of bit-value to minimal pulse distance limits the stimulation rate for each channel between 586 and 1818 pps. The stimulation matrix was then transmitted via a RS232 serial line with a transmission rate of 115,200 Baud to a proprietary Interface (Research Interface Box, manufactured at the University of Technology Innsbruck, Austria).

```
download and stimulation tool V.2.05 for the RIB
>>>>>>>> SERIAL PORT: 2
>>>>>>>> BAUD RATE: 115200
>>>>>>>> COMMAND: fitting D:/Andrea~1/map_fi~1/Huber15.t12
>>>>>>>> COMMAND: load D:/Andrea~1/sti_fi~1/test


24 pulses:
1    0.000 ms:   ch  1   amp   93   rng 2   wid  16   md   33
2    0.055 ms:   ch  1   amp   93   rng 2   wid  16   md   33
3    0.110 ms:   ch  1   amp   93   rng 2   wid  16   md   33
4    0.165 ms:   ch  1   amp   93   rng 2   wid  16   md   33
5    0.220 ms:   ch  1   amp   93   rng 2   wid  16   md   33
6    0.275 ms:   ch  1   amp   93   rng 2   wid  16   md   33
7    0.330 ms:   ch  1   amp   93   rng 2   wid  16   md   33
8    0.385 ms:   ch  1   amp   93   rng 2   wid  16   md   33
9    0.440 ms:   ch  1   amp   93   rng 2   wid  16   md   33
10    0.495 ms:   ch  1   amp   93   rng 2   wid  16   md   33
11    0.550 ms:   ch  1   amp   93   rng 2   wid  16   md   33
12    0.605 ms:   ch  1   amp   93   rng 2   wid  16   md   33
13    0.660 ms:   ch  1   amp   93   rng 2   wid  16   md   33
14    0.715 ms:   ch  1   amp   93   rng 2   wid  16   md   33
15    0.770 ms:   ch  1   amp   93   rng 2   wid  16   md   33
16    0.825 ms:   ch  1   amp   93   rng 2   wid  16   md   33
17    0.880 ms:   ch  1   amp   93   rng 2   wid  16   md   33
18    0.935 ms:   ch  1   amp   93   rng 2   wid  16   md   33
19    0.990 ms:   ch  1   amp   93   rng 2   wid  16   md   33
20    1.045 ms:   ch  1   amp   93   rng 2   wid  16   md   33
21    1.100 ms:   ch  1   amp   93   rng 2   wid  16   md   33
22    1.155 ms:   ch  1   amp   93   rng 2   wid  16   md   33
23    1.210 ms:   ch  1   amp   93   rng 2   wid  16   md   33
24    1.265 ms:   ch  1   amp   93   rng 2   wid  16   md   33
>>>>>>>> COMMAND: Execute 0 0 1
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FIGURE 3. *Information delivered to the Research Interface box for the stimulation of a single channel with a fixed amplitude, pulse width and stimulation rate.*

Figure 3 shows the resulting stimulation data, as it is encoded in the stimulation data file, along with hardware diagnostic information (serial port number, baud rate) and all commands contained in the command file. This listing is generated by a download tool (rib.exe) switched to diagnostic mode. All parameters for each stimulation pulse (ascending number) are listed, including the time offset from the beginning of stimulation, to facilitate a verification of the stimulation pattern. In this example electrode E1 (‘ch 1’) is stimulated at a constant amplitude (‘amp 90’) in a constant current range (‘rng 2’), a minimal pulse duration (‘wid 16’) and a constant minimal distance between pulses (‘md 33’) which results in a stimulation rate of 1515 pps. For some of the hearing experiments it was necessary to deliver a pulse rate lower than determined by the maximally available pulse distance (md 1023 corresponding to 586 pps). In order to achieve a lower pulse rate, a technique comparable to down sampling was applied as following: the next multiple *n* of the desired rate which could be realized by setting the minimal distance accordingly was chosen. Then, only the *n*-th pulse was presented with normal amplitude, the amplitude value of all following pulses of that cycle was set to the minimum (‘amp 0’). All applied pulse rates were controlled by means of a frequency counter prior to the experiments. Attachment 1 shows an example for the presentation of three stimuli separated by gaps. Stimulus 1 (start at pulse 1) and 2 (start at pulse 18) have the same stimulation rate which is in the range between 586 and 1818 pps, stimulus 3 (start at pulse 35) is presented with a lower stimulation rate, therefore the minimal pulse distance has changed and every second amplitude is set to 0.

3. Comfortable listening levels

McKay et al. (1999) and Pfingst et al. (1999) reported an influence of overall stimulation level on electrode discrimination. Thus, in the first test session, the stimulation level on each electrode was adjusted to a comfortable level. To assess comfortable listening level, the patients scaled the loudness elicited by each electrode from 0 (no hearing sensation) to 50 (uncomfortably loud). The levels were measured using an ascending-descending technique, where current was increased until stimulation became too loud and then decreased to a level corresponding to comfortable loudness (25 on the loudness scale). The comfortable listening levels were further adjusted to ensure that all electrodes were approximately of equal loudness. This was done by a paired comparison with an overlap of one electrode for all electrodes. Finally, all electrodes were presented sequentially to control the listening levels again. The order of electrodes was in an apical-basal direction. Comfortable loudness levels were obtained for each stimulation rate used in the experiments (see below). During loudness balancing subjects were asked to indicate electrodes eliciting unpleasant or unclear pitch perceptions. No electrodes with differences in sound quality were found.

II. HEARING EXPERIMENTS

1. Experiment 1: Electrode discrimination

The technique of the cochlear implant on what concerns the transmission of spectral information to the auditory nerve is based on the tonotopic organization in the cochlea. Different places of stimulation are evoking different pitch perceptions. In this experiment the discrimination of single electrodes based on pitch perception is examined. If all electrodes of the array are discriminable, different electrodes will activate neurons in tonotopically disparate regions of the cochlea. Nadol et al. (1989) have shown that in individual deaf ears the pattern of neural survival may be different and related to age, duration of deafness and etiology. The average spiral ganglion cell distribution shows a peak along the segment in the section between 6 to 15 mm from the basal end of the cochlea. The electrodes of the COMBI 40+ are stimulating the cochlea between a 3.9-mm and 30.3-mm distance to the round window. Therefore electrode discrimination (*ED*) is tested for three reference electrodes and the four adjacent electrodes positioned in different cochlear regions, apical, medial and basal. Another tested parameter in experiment 1 is the effect of stimulation rate on *ED*. In the normally applied CIS strategy (Wilson et al. 1991, see also chapter III.1., page 64) all electrodes are stimulated at the high stimulation rate of 1515 pps. In the apical region of the cochlea where low frequency sounds have their best oscillatory frequencies in normal hearing (Zwicker & Fastl, 1999), the stimulation with a low pulse rate might lead to a more distinct pitch. Conversely, in the basal region of the cochlea where high frequencies show a peak in the envelope of the traveling wave in normal hearing, the stimulation with a low pulse rate might lead to a less distinct pitch. Consequently, the changes in pitch strength which might occur due to this rate-place mismatch along the cochlear region might influence the capability of *ED*. Experiment 1 was therefore conducted with three different stimulation rates, 1515, 500 and 250 pps. A more detailed overview of experiment 1 can be found in a recent publication of Baumann & Nobbe (2004a).

a) Participants

Eight subjects participated at the electrode discrimination experiment (S1, S4, S5, S6, S7, S10, S11 and S12). The electrode array of those eight subjects was fully inserted into the cochlea and the electrodes stimulated regions from 1¾ to 2¼ complete turns of the cochlea in all subjects (Fig. 4).

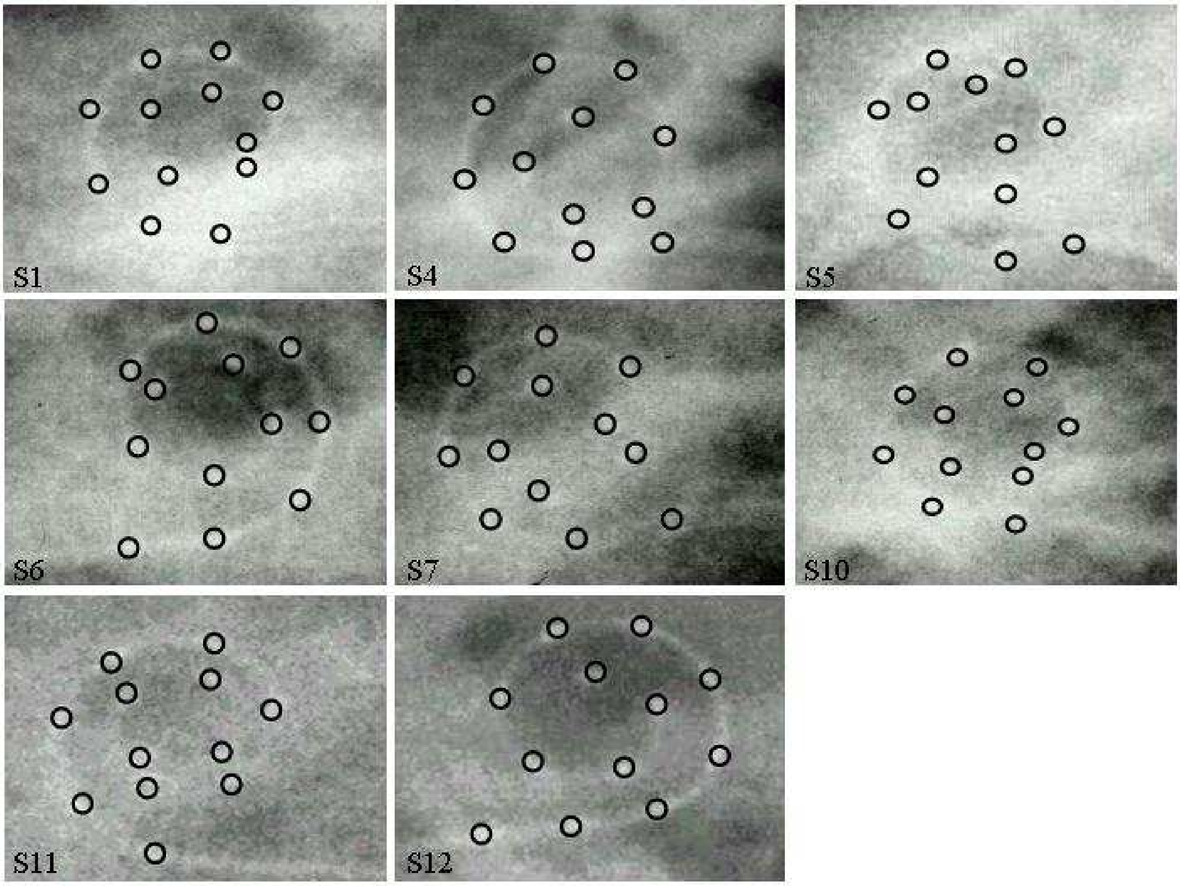

FIGURE 4. *Stenvers' view x-ray scans of the eight subjects participating in experiment 1. The positions of the electrodes are highlighted for a better overview.*

b) Procedure

A two alternative forced-choice (2AFC) procedure was used to measure electrode discrimination. In one interval a reference electrode was stimulated and in the other interval one of the associated probe electrodes was stimulated. The subject was asked to indicate the interval containing the higher pitch. Electrode discrimination was assessed by calculating the percentage *ED* of electrode pitches judged in the expected order, i.e. higher pitch for the more basal electrode. No feedback as to correct or incorrect responses was given.

Three reference electrodes where chosen, located apically, medially and basally on the electrode array (E3, E7, E10, Fig. 1, page 3). These reference electrodes were compared with the two adjacent apical and basal probe electrodes. Tests for the different reference electrodes were always in the same order from the apical to the middle to the basal reference electrode. The order in which probe electrodes were tested for a certain reference electrode was randomized. Each combination of reference and probe electrode was presented five times within one test block and each block was tested three times so that 15 estimates for each combination of reference and probe electrode were obtained.

Each pair of intervals consisted of two 500 ms long stimuli separated by a 500 ms quiet gap. To investigate the influence of pulse rate with which the electrodes were stimulated, tests were performed at pulse rates of 1515 pps (the pulse rate normally used in the MED-EL CIS strategy), 500 pps, and 250 pps. The pulse rate was held constant within one block and varied randomly throughout the three repetitions of each block. To prevent any residual loudness differences between electrodes from affecting test results, one electrode in each interval was stimulated at a comfortable level, and the other was stimulated at 90% of the current amplitude required for comfortable loudness. Prior to testing, a training run containing each combination of reference and probe electrode was performed.

c) Statistics

Significant discrimination between the probe and reference electrode was achieved, if *ED* was greater or equal to 86.67% correct (probe and reference electrodes for 13 out of 15 trials judged in the expected order). This threshold was calculated based on the confidence interval for the binomial distribution for performance better than chance (50% correct).

To assess whether or not electrode discrimination differs between the apical, middle and basal region of the cochlea, a 2-way repeated-measures ANOVA was performed on a data set where *ED* was averaged across rates for each probe electrode (so that *ED* results as a function of reference electrode and probe electrode). To investigate the effect of rate on *ED* in the different regions of the cochlea, a 2-way repeated-measures ANOVA was applied to a data set where *ED* was averaged across reference electrodes (so that *ED* results as a function of reference electrode and rate). If significant differences were indicated by ANOVA ($p \leq 0.05$), the Tukey test was used for *post-hoc* comparisons between factor levels.

d) Results

Scores for *ED* are plotted for the eight subjects in Fig. 5. The results are grouped for the different reference electrodes (E3, E7, E10) and the respective probe electrodes. The parameter is pulse rate. The dotted line indicates the threshold for significant discrimination (*ED* = 87.67%). Although *ED* reaches high values for the majority of the subjects, there is high within-subject variability in some subjects. Listeners S1, S4 and S5 are top performers with nearly perfect *ED* at all electrodes. Listeners S6 to S12 reach significant electrode discrimination on a range of electrodes. These subjects show differences in *ED* between neighboring electrodes at some reference electrodes. Concerning the comparison of different cochlear regions, one listener (S11) shows a tendency for poorer electrode discrimination in the apical region compared to other regions. Three listeners (S10, S6 and S12) have reduced electrode discrimination in the middle region and one listener shows reduced electrode discrimination in the basal region (S7).

A summary for *ED* is plotted in Fig. 6. Bars show the percentage of conditions across all tested conditions in which *ED* was at or above the significance level. In the group of top performers, one listener (S1) could discriminate all probe electrodes from the respective reference electrodes, resulting in score of 100% significant electrode discrimination across comparisons.

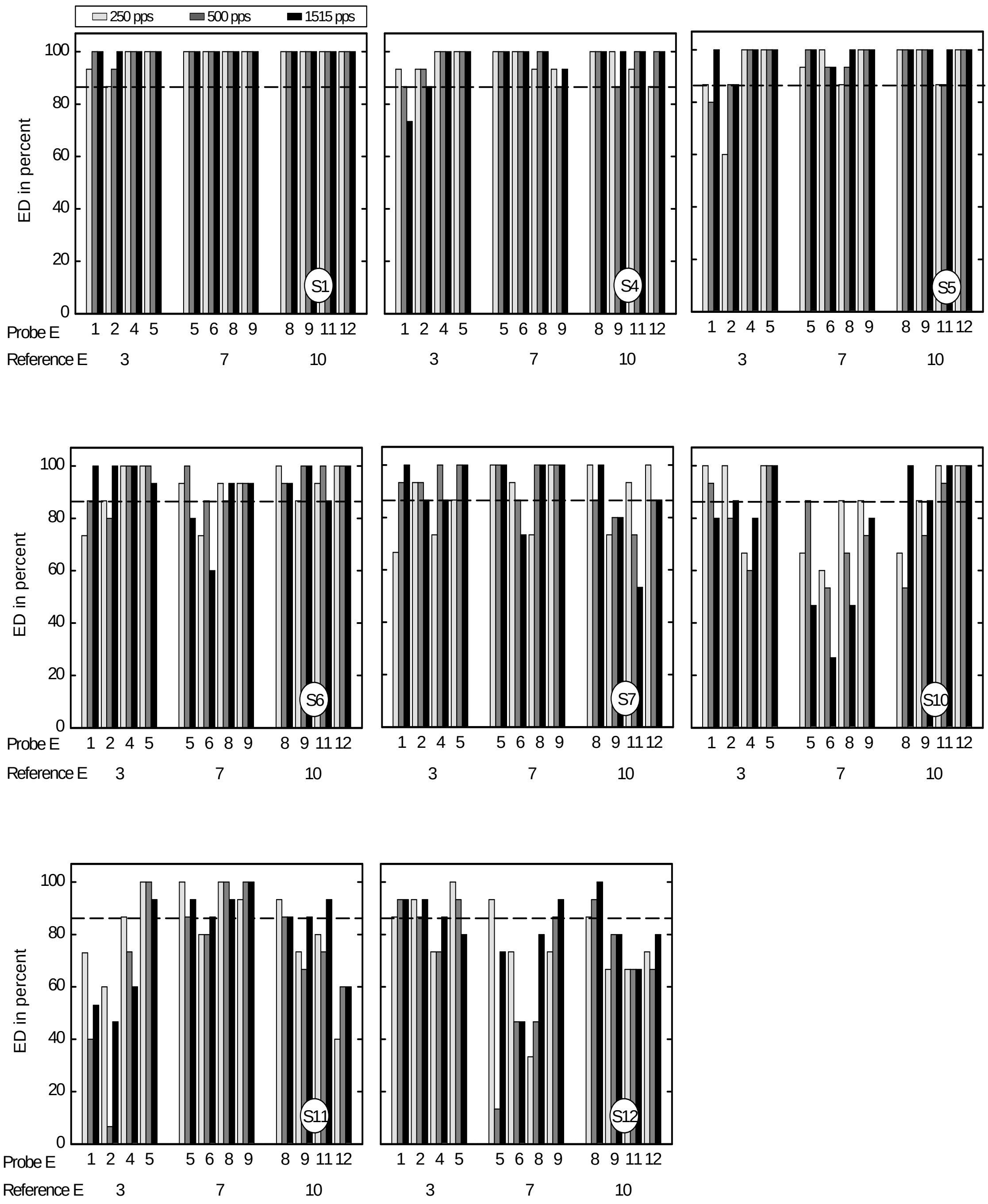

FIGURE 5. *Individual results of the electrode discrimination experiment 1. The percent correct score for electrodes judged in the right order is plotted as a function of test electrode; the parameter is pulse rate; the results are grouped for the three reference electrodes E3, E7 and E10.*

Two other top performers (S4 and S5) had excellent results for almost all conditions tested, resulting in a score for significant electrode discrimination across comparisons of over 94%. The other subjects showed significant electrode discrimination in 41.67% to 86.11% of the tested conditions.

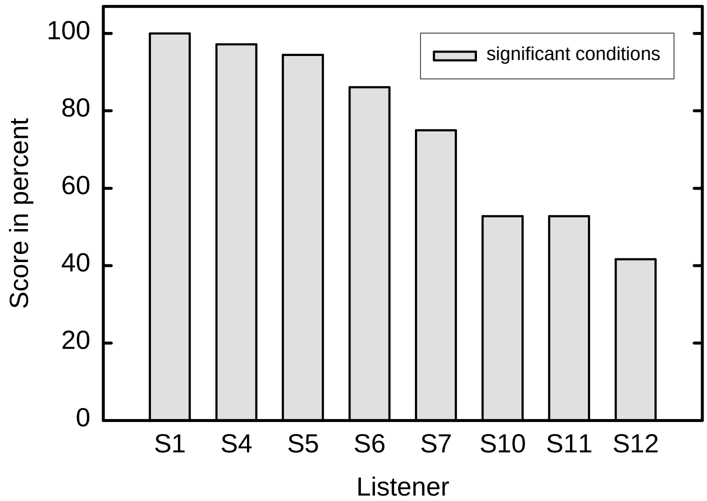

FIGURE 6. *Overview over the performance of all subjects. The bars indicate the percentage of conditions over all tested combinations of pulse rate, test and reference electrode which were discriminated significantly.*

Despite the considerable variability between individual results, the mean *ED* was calculated to provide an overview over the average ability to discriminate adjacent electrodes. Figure 7(a) shows the mean *ED* averaged across rates for each probe electrode and reference electrode to assess the effect of probe electrode position in relation to the reference electrode. The data show that in general the pattern of results does not vary largely between probe electrode positions. Using the statistics described above, no significant effect of reference electrode could be found ($F = 0.165$, $DF = 2$, $p = 0.849$), but the distance between the reference and the probe was significant ($F = 6.118$, $DF = 3$, $p = 0.004$). *Post-hoc* tests showed that here the only significant difference exists when comparing RefE+2 to RefE-1.

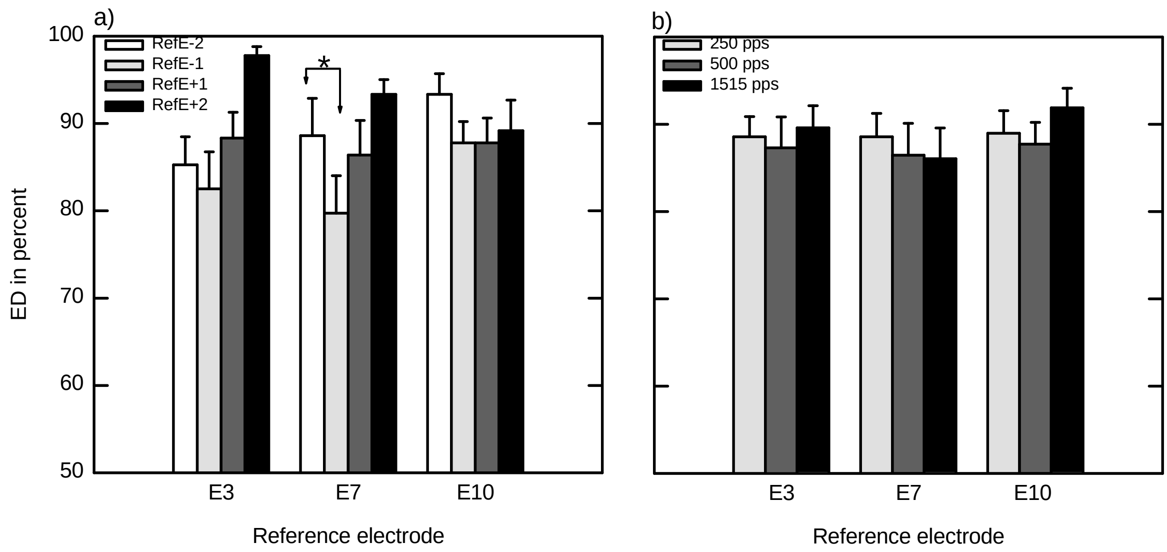

FIGURE 7. (a) *ED as a function of electrode separation and reference electrode (averaged over pulse rates),* (b) *ED as a function of pulse rate and reference electrode (averaged over electrode separations). Error bars indicate the standard error of the mean.*

This means that in particular there was no significant difference between *ED* resulting from adjacent electrodes positioned apically or basally to the reference electrode (i.e. RefE-2 *vs.* RefE-1, and RefE+2 *vs.* RefE+1). In other words, there was no significant effect of distance between the reference electrode and the probe electrode. To assess the group effect of rate in relation to the reference electrode, Figure 7(b) shows mean *ED* averaged across probe electrodes for each reference electrode and rate. Again, the pattern of results does not vary much between reference electrodes. Statistical evaluation did not reveal any significant effect of reference electrode ($F = 0.165$, $DF = 2$, $p = 0.849$) or rate ($F = 0.826$, $DF = 2$, $p = 0.458$). In neither analysis was a significant interaction between the factors involved ($F = 1.438$, $DF = 6$, $p = 0.223$ for reference x distance, Fig. 7(a); $F = 0.399$, $DF = 4$, $p = 0.807$ for reference x rate, Fig. 7(b)).

e) Discussion

Experiment 1 has shown that the tested electrodes could be discriminated significantly by the average user. Furthermore, there was no difference in *ED* between the different tested regions of the cochlea. The results for the different tested pulse rates for stimulation revealed that *ED* is independent of stimulating pulse rate. That means that the electrode spacing of 2.4 mm of the COMBI 40+ electrode array is wide enough to evoke different pitch sensations when stimulating different electrodes.

This result is conform with former research (Busby & Clark, 1996; Collins et al., 1997; Nelson et al., 1995; Pfingst et al., 1999; Tong & Clark, 1985) with a different electrode array, the CI22M of Cochlear (Melbourne, Australia). This electrode array consists of 22 electrodes with a spacing of 0.75 mm. Nelson et al. (1995) have tested the *ED* for electrodes with different spatial separations (0.75 to 3 mm) in 12 subjects. The changes in pitch sensitivity with spatial separation show that the performance grows with increased spatial separation. Only two subjects reach near perfect performance at a spatial separation of 0.75 mm. Two criterion performance levels were chosen at a sensitivity $d' = 2$ and $d' = 3$. The spatial separation required to reach those performance levels ranged between 0.47 mm and 8.71 mm for $d' = 2$ (average 4.01 mm) and between 3.41 mm and 13.48 mm for $d' = 3$ (average 6.57 mm). The sensitivity d' was not calculated for the results of experiment 1. However, the decision criterion in experiment 1 of 87.67% correct for a significant performance is based on the confidence interval for the binomial distribution for performance better than chance (50% correct) with $p < 0.01$, which is a very strict criterion. That means that the average result for *ED* of Nelson et al. (1995) is probably in the same range for the CI22M as the results of experiment 1 with the COMBI 40+ electrode array. Busby & Clark (1996) measured electrode discrimination for electrode separations up to 4.5 mm using a 4AFC procedure in early deafened subjects. They randomized the level by 0 to 20 current units, which approximately corresponds to the 10%-level roving used in experiment 1.

Applying the significance criterion formulated above (87.67% correct) to their data shows that a spatial separation of about 2.5 mm would result in significant electrode discrimination for this patient group. The results cited above and the results of experiment 1 show that an electrode separation of 2.4 mm provides perceptually distinguishable information for each electrode for average cochlear implant user.

The fact that *ED* is independent of cochlear region was also observed by Nelson et al. (1995) and Donaldson & Nelson (2000). Other investigators have observed poorer electrode discrimination in the more basal region of the electrode array of the CI22M (Henry et al., 2000; Pfingst et al., 1999; Zwolan et al., 1997). In summary, it does not appear as if *ED* is strongly dependent on cochlear region. In particular, it is no more difficult to discriminate electrodes in the apical cochlear region than it is for electrodes in the middle and basal regions. The amount of residual neural structures in the apical cochlear region is often discussed. The data presented in experiment 1 clearly show that, at least in the average subject, residual neural structures in the apex of the cochlea do exist and are appropriate for electrical stimulation in that they are sensitive to pitch changes provided by different places of stimulation.

2. Experiment 2: Scaling of pitch height

Experiment 1 (page 16) has shown that a different place of stimulation evokes a different pitch perception. This happens in a way that the perceived pitch is increasing for places of stimulation changing from the apical to the basal region which is consistent with the tonotopic organization of the cochlea in normal hearing (Zwicker & Fastl, 1999). Besides the place of stimulation, another parameter is supposed to change the pitch perception for electrical stimulation, namely the rate of stimulation. Experiment 2 is examining the effect of rate changes on the perceived pitch for different cochlear regions. Due to the frequency-place transformation in the inner ear of normal hearing subjects, it is expected that low pulse rates have a higher influence on pitch perception at apical electrode positions and that high pulse rates have a higher influence on pitch perception at basal electrode positions. It is estimated that above a certain pulse rate the pitch height would be constant (saturation rate). This saturation rate is expected to be lower at an apical electrode and higher at a basal electrode and it is expected that the slope of the psychometric function up to the saturation rate would be shallower at an apical electrode and steeper at a basal electrode.

a) Participants

Ten subjects took part in the pitch scaling experiment (S1, S2, S3, S5, S8, S6, S7, S8, S13 and S14). The electrode was not fully inserted into the cochlear for S3 where there were E11 and E12 external to the scala tympani (controlled by Stenvers' view x-ray scans).

b) Procedure

A two interval numerical estimation procedure was used in the pitch scaling task. The subjects were instructed to assign a numerical value in the range of 0 to 50 to the pitch of a stimulus delivered with varying pulse rate. A low pitch was assigned with a low number and a high pitch was assigned with a high number to assess the pitch estimation of the subject. Subjects were directed to avoid the extreme positions of the scale, namely 0 and 50, in order to have enough room left for their whole range of pitch heights. The target stimulus was presented after the presentation of a reference stimulus. Both stimuli had a duration of 500 ms and were separated by a gap of 500 ms. The reference stimulus was set on a pitch height of 'middle' (25). To determine the reference stimulus, prior to the experimental runs a sequence of three stimuli was presented to the subjects: the most apical electrode E1 at the lowest pulse rate applied in the experiment (100 pps, presumably lowest pitch), the reference stimulus at an electrode position varying between E2 and E11 at 800 pps and the most basal electrode E12

(E10 for S3) at 800 pps (presumably highest pitch). The electrode position of the reference stimulus was switched using an ascending-descending technique until the subject assigned a pitch height of 25 to the reference stimulus in relation to the first (lowest) and third (highest) stimulus. The target stimulus was presented at seven different pulse rates, 100, 141, 200, 238, 400, 566 and 800 pps. Presentation order was randomized within a block of trials. Each block consisted of nine estimates for each stimulus. Four different electrodes, E1, E3, E7 and E10 were tested. Within one run, the electrode position was fixed and the pulse rate was altered. The electrodes were tested in a fixed order (E1, E7, E3, E10) since no sequence effect was expected. A training block with all pulse rates at all test electrodes was presented prior to data collection. Final scores were determined by calculating the arithmetic mean of nine estimates recorded for stimulation of each condition. Data were collected within one session.

c) Results

The individual results of the perceived pitch height estimates are plotted in Fig. 8. The averaged pitch estimate of each subject is plotted as a function of pulse rate. The different symbols represent the estimates for electrodes E1, E3, E7 and E10. The error bar shows the range of individual standard error. The individual reference electrode is noted for each subject at the bottom of the figure.

Three of the listeners (S1, S2, S3) show almost no influence of pulse rate on their pitch estimations at all test electrodes. The estimates for each electrode lay within a small range, which results in flat curves with only small standard errors. The pitch scaling of listener S3 displays no significant dependency on pulse rate at electrode E10. These three listeners estimate the pitch evoked by E1 at nearly the same height as the pitch evoked by E3. There are two listeners with influence of pulse rate on pitch perception at E1 and E3, but hardly any influence of pulse rate at E7 and E10 (S6 and S8).

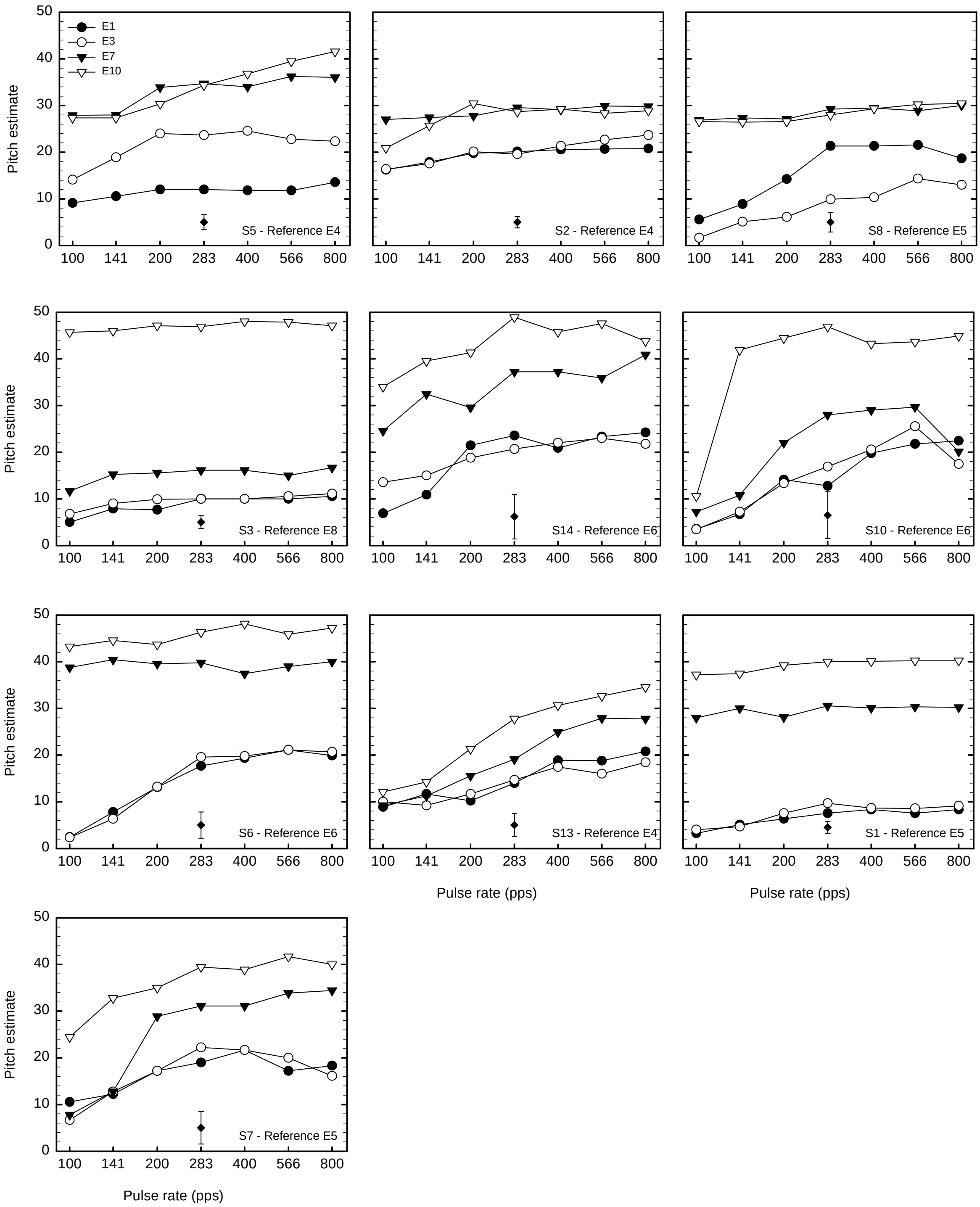

FIGURE 8. *Individual results for the pitch scaling experiment. The estimated pitch height is shown as a function of pulse rate; the parameter is electrode number. The mean standard deviation and the reference electrode are indicated for each subject.*

For listener S6 the pitch of E1 is estimated higher than the pitch of E3. This phenomenon was already observed in a previous experiment of electrode discrimination and could not be further explained via Stenvers' view x-ray scans of the electrode array. Listener S6 estimates the pitch of E1 and E3 at nearly the same height. The other five listeners (S5, S7, S10, S13 and S14) show increasing pitch estimation with increasing pulse rate at all electrodes. The estimates for the different electrodes differ inter-individually to a large extent. The pulse rate where no further increase of pitch height was perceived (saturation rate) is between 141 and 400 pps. At some electrodes the estimates between 100 and 141 pps are not significantly different.

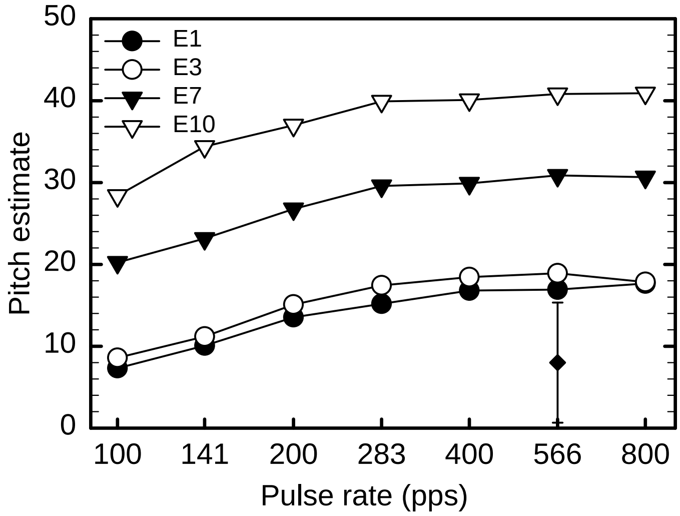

FIGURE 9. *Average pitch scaling results for nine subjects (S8 was excluded due to a pitch reversal between E1 and E3) in the same format as Fig. 8.*

Figure 9 shows the averaged estimates of nine listeners plotted in the same format as in Fig. 8. The data of S8 was excluded due to the described pitch reversal between E1 and E3. The results show that the averaged pitch estimates are increasing with increasing pulse rate at all electrodes. Pitch is increasing significantly up to a pulse rate of 283 pps. The averaged pitch estimates for E1 and E3 show only a small but statistically significant difference for pulse rates in between 100 and 566 pps (t-test $p<0.05$). The estimates for 800 pps show no significant difference, which might be due to the pitch scaling of subjects S5, S7, S10. For these three listeners, the perceived pitch height at E3 decreases between 566 and 800 pps.

d) Discussion

Experiment 2 shows that the influence pulse rate variation is limited to pulse rates up to about 283 pps. Furthermore, the effect of pulse rate on pitch perception could be observed at all stimulated electrodes. The results were independent of place of stimulation in the cochlea. An increasing pitch perception with increasing pulse rate has been previously observed (Hochmair-Desoyer et al., 1983; Shannon, 1983; Tong & Clark, 1985; Pijl & Schwarz, 1995; Fearn & Wolfe, 2000; Zeng, 2002). Asymptotic pitch sensation with increasing pulse rate was found between 200 and 600 pps. Hochmair-Desoyer et al. (1983) and Wilson et al. (1997) reported about three subjects with no sign of saturation for pitch sensation with increasing pulse rates up to 500 to 1000 Hz. Even in experiment 2, the pulse rate for asymptotic pitch varied between subjects and at single electrodes. The averaged data over all ten subjects showed a saturation rate of about 283 pps for temporal pitch perception at four examined electrodes. This is consistent with the results of four subjects at two electrodes in a recent study by Zeng (2002). He found an increased pitch perception with increasing frequency up to roughly 300 Hz. Considering the amount of collected data in these two studies and the carefully conducted loudness balancing prior to testing, on average a saturation pulse rate of about 300 pps for temporal pitch perception conveyed by electric pulses seems to be proved.

The results of experiment 2 also show that the pitch perception increases with increasing electrode number from an apical to a basal region in the cochlea. However the pitch difference between the more apical electrodes E1 and E3 is small in comparison to the pitch difference between electrodes E3, E7 and E10. The pitch difference between E1 and E3 which is an electrode distance of two electrodes only amounts about 1 to 2 pitch units, the pitch difference between E3 and E7 which corresponds to an electrode distance of four electrodes amounts about 12 pitch units, a comparably large difference. The pitch difference between E7 and E10 which corresponds to an electrode distance of three electrodes amounts about 10 pitch units. That means that the pitch difference of two adjacent electrodes would correspond to about 3 pitch units on the pitch scale which is a relatively small perceptual change but seems to be adequate for an electrode distance of 2.4 mm.

3. Experiment 3: Scaling of sound quality

During the pitch height scaling in experiment 2 (page 24) subjects often reported difficulties in judging the pitch height of stimuli with a low pulse rate. Therefore a scaling of sound quality depending on pulse rate was conducted. Due to the experiences in experiment 2, an effect on sound quality was expected for pulse rates below 300 pps. Furthermore, the experiment was conducted at different places of stimulation similar to experiment 2. The expectation was that low pulse rates would have less influence on sound quality at more apical electrodes where the neurons of the spiral ganglion cells are tuned to low frequencies in normal hearing according to the frequency-place transformation (Zwicker & Fastl, 1999).

a) Participants

Ten subjects took part in the pitch scaling experiment (S1, S2, S3, S5, S8, S6, S7, S8, S13 and S14). The electrode was not fully inserted into the cochlear for S3 where there were E11 and E12 external to the scala tympani (controlled by Stenvers’ view x-ray scans).

b) Procedure

A single interval line length scaling procedure was used to judge the sound quality of stimuli with varying pulse rate and electrode position. All ten subjects of experiment 2 participated also in experiment 3. The subjects were instructed to assign the sound quality of the stimulus by touching on a scale between the endpoints ‘extremely buzzy’ and ‘extremely clear’ on a touch screen (see Fig. 10). The line was internally scaled from 0 (extremely buzzy) to 27 cm (extremely clear). Poor sound quality was assigned by touching on a position located towards the left end, better sound quality by touching towards the right end of the scale. Eight different pulse rates were presented within one block. Stimuli were presented at 100, 119, 141, 168, 200, 238, 566 and 800 pps in random order. The sound quality was tested at four different electrodes according to experiment 1 (page 16), at E1, E3, E7 and E10. Nine estimates for each pulse rate were recorded within one block. The blocks were ordered according to stimulated electrodes, E1, E7, E3 and E10. Prior to the experimental runs, a training session was conducted whereby all electrodes were stimulated once at all pulse rates applied in the experiment. The final score was calculated as the arithmetic mean of nine estimates. All conditions were tested within one session.

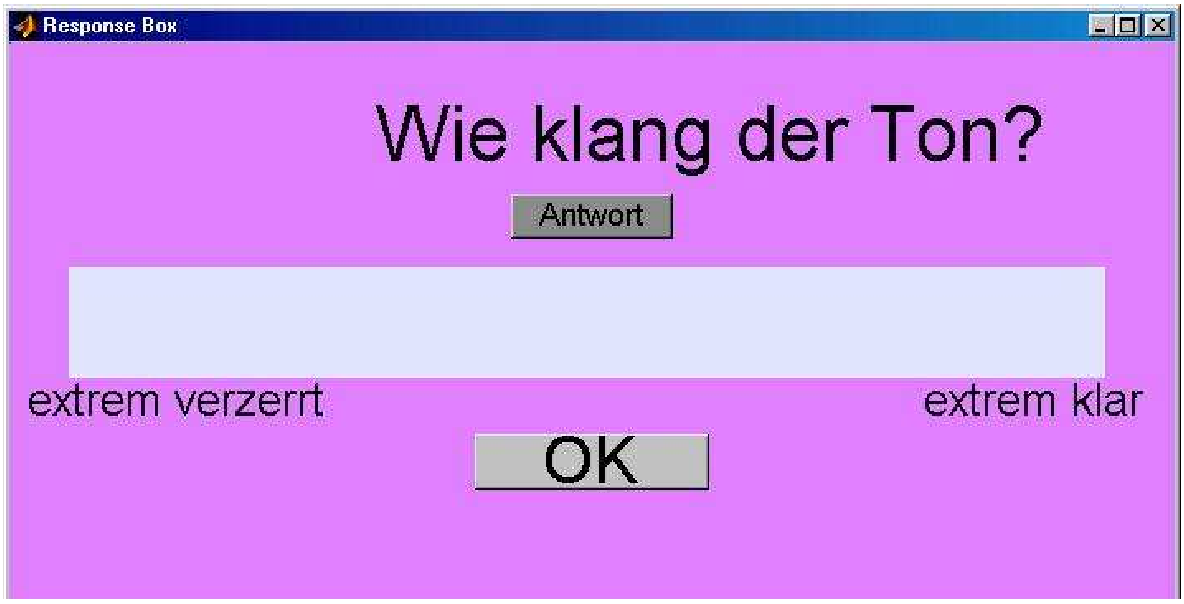

FIGURE 10. *Screen copy of the TFT touch screen used for the scaling of sound quality (experiment 3, line length method). The task of the subject was to indicate the sound quality of the stimulus between extremely buzzy (left side) and extremely clear (right sight) by pointing at a position on the grey bar. After the scaling the ‘OK’ button was pressed to confirm the input.*

c) Results

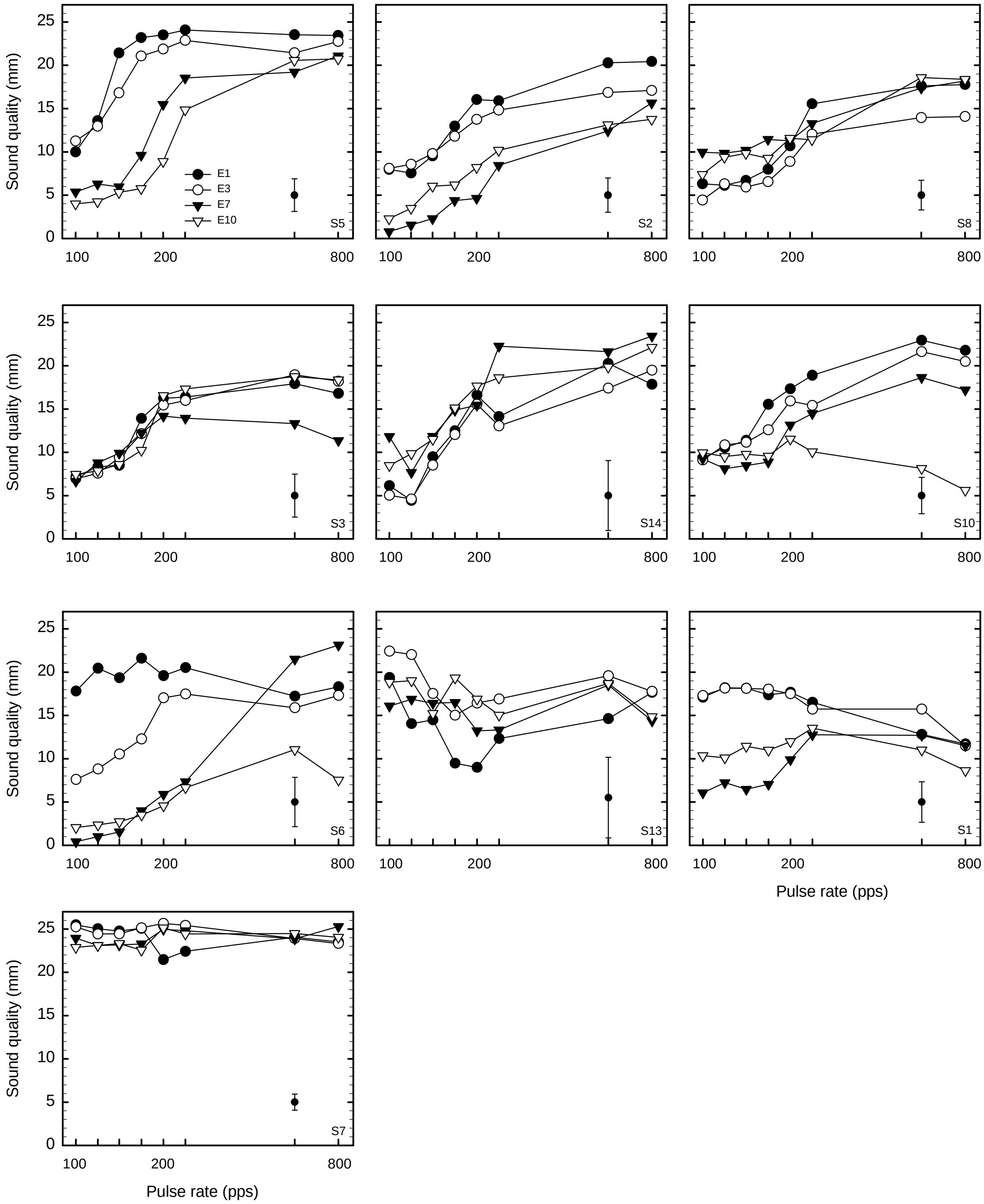

FIGURE 11. *Individual results for the sound scaling experiment. The sound quality in* mm *line length is plotted as a function of pulse rate; the parameter is electrode number.*

The averaged estimated sound quality in line length units (0 cm: extremely buzzy; 27 cm: extremely clear) is plotted as a function of pulse rate in Fig. 11 for the individual subjects; the parameter is electrode number. The inter-individual results vary considerably. There are three listeners with significant (t-test with 95% confidence interval) influence of pulse rate on sound quality at all electrodes (S2, S5, S8). Four listeners show significant influence of pulse rate on sound quality at least at one single electrode (S3, S6, S10, S14) and three listeners show no significant influence (S1, S7, S13). The majority of the subjects judge the sound of the lowest pulse rate as lowest perceived sound quality. At most of the electrodes an increasing sound quality with increasing pulse rate can be observed. The individual sound quality functions exhibit a split into two regions for seven out of ten subjects: one region below 200 pps with sound quality depending on pulse rate and another region above 200 pps with hardly changing sound quality estimates. Regarding the individual results, for example the estimates of listener S5, sound quality reaches a maximum at 168 pps at E1 and E3, and at 200 and 566 pps at E7 and E10, respectively. The estimates of listener S10 show a dependency of sound quality on pulse rate at E1, E3 and E7. Sound quality increases up to a pulse rate of 566 pps. At the more basal electrode E10 however, the estimates of S10 are independent of pulse rate and much lower as for the other electrodes. For listener S7 the estimates at all electrodes are independent of pulse rate and equally high for all pulse rates and electrodes. The averaged estimates of listener S13 as well as of listener S1 show no significant influence on pulse rate, partly due to the large intra-individual variation.

Figure 12 shows the averaged results over all listeners and pulse rates for the four test electrodes. The averaged sound quality estimates are increasing with increasing pulse rate at all electrodes. There are significant increases in sound quality judgments between 119 and 168 pps and between 238 and 566 pps at E1. Sound quality increases significantly between 100 and 119 pps, between 141 and 200 pps and between 238 and 566 pps at E3. At E7 and E10, sound quality increases significantly between 141 and 566 pps, respectively. There is no influence on the averaged sound quality judgments for the highest pulse rates (566 and 800 pps) applied in the experiment for all test electrodes. That means that sound quality estimates saturate at 566 pps independent of electrode location. The averaged sound quality estimates for the apical electrodes E1 and E3 are significantly higher than the estimates for the more basal electrodes E7 and E10 at pulse rates up to 238 pps. Significant sound quality differences between E1 - E3 and E7 - E10 can only be observed at single pulse rates.

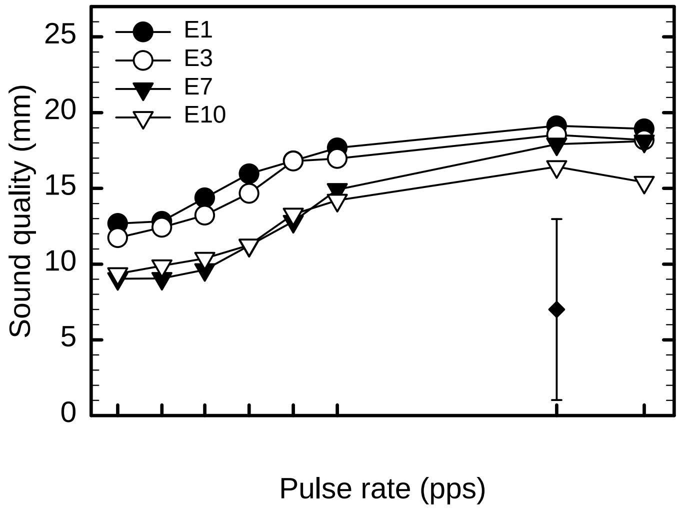

FIGURE 12. *Average sound scaling results for ten subjects in the same format as Fig. 11.*

d) Discussion

Sound quality is increasing with increasing pulse rate up to about 566 pps. This means that changes in pulse rate are always resulting in changes in sound quality. This effect is hardly described in the literature. In a recent study, Fearn & Wolfe (2000) did a quality rating for stimuli with changing pulse rate in six subjects implanted with the CI22M. Each stimulus was presented twice and should be rated on a line between two bipolar quality words like ‘like-dislike’, ‘mechanical-natural’, ‘clear-fuzzy’ etc. The mean of eight positions was taken to give a quality rating of the sound. The results show that sound quality is increasing with increasing pulse rate between 100 and 400 pps. Fearn & Wolfe (2000) also observed that more basal electrodes were judged lower in sound quality than more apical electrodes. This effect occurred up to 1000 pps. For an electrode distance between the most apical and most basal electrode of 11.25 mm the difference in sound quality was 40 cu on a scale between 0 and 100. In the present sound quality experiment, a difference in sound quality between more apical and more basal electrodes can also be observed. However, the difference is much smaller: For an electrode distance between the most apical (E1) and most basal electrode (E10) of 24 mm, the sound quality difference is 3 to 4 cm on a scale between 0 and 27 cm. This would correspond to only 14.8 cu on a scale between 0 and 100. In the present study the stimuli were only judged between extremely buzzy and extremely clear. The data of Fearn & Wolfe (2000) also include sensations like ‘pleasant’, ‘mechanical’, ‘natural’, and ‘musical’. Most cochlear implant patients have been deaf or have had a profound hearing loss before implantation. Therefore especially the stimulation of more basal electrodes often evokes unpleasant pitch sensations and most cochlear implant patients prefer the sound of more apical electrodes. This effect might influence the data of Fearn & Wolfe (2000) and cause the difference in the sound quality rating compared to the results of experiment 3.

The effect of poorer sound quality at low pulse rates at more basal electrodes might be due to the mechanism of tonotopic allocation at the spiral ganglion. In the more apical region, more neurons of the auditory nerve tuned to low frequencies might exist than in the more basal region. However, current studies do not report a decrease in sound quality with high pulse rates at more apical electrodes and the sound quality of very high pulse rates with varying electrode position was not examined yet.

In normal hearing a distinct change of sound quality depending on the modulation frequency of the stimulus is described as roughness (Zwicker & Fastl, 1999). For a 100%-amplitude modulated stimulus with a carrier frequency of 1 kHz maximal roughness is perceived for a modulation frequency of 70 Hz. For modulation frequencies higher than 70 Hz, roughness decreases up to about 400 Hz. In the psychoacoustic literature it is described that the sensation of pitch strength is related with stimulus frequency. The pitch strength of a pure tone is increasing with increasing frequency up to about 750 Hz (Zwicker & Fastl, 1999). This effect might also generally contribute to a change in sound quality with increasing pulse rate. The sensation in this experiment might be a mixture of pitch strength and roughness.

4. Experiment 4: Pulse rate discrimination

Experiment 2 (page 24) has investigated the subjective pitch height judgment depending on stimulation rate at single electrodes. There is an increase of pitch height with increasing stimulation rate. However, the average pitch height increases only about 10 pu between the low pulse rate (100 pps) and the pulse rate for saturating pitch height (283 pps). Furthermore, the interindividual results vary considerably. One group of subjects shows an increase of pitch height with increasing stimulation rate only at single electrodes, some other subjects show only a weak increase of pitch height with increasing stimulation rate (4 pu). It is interesting to find out whether the individual slopes of the curves only vary in a scaling experiment or if the subjects with a weak increase are not able to detect small rate differences even in an objective task. Experiment 4a investigates the detection of small rate differences at two electrodes. This way, the amount of rate changes necessary to evoke differences in pitch height in the interesting range between 100 and 283 pps can be figured out for a possible implementation in a speech strategy. Furthermore, it is tested whether rate changes above 283 pps do effectively not evoke any pitch differences as it was found in experiment 2 (page 24). A more detailed overview of experiment 4 can be found in the publication by Baumann & Nobbe (2004b).

a) Participants

Seven subjects participated in the pulse rate discrimination experiment (S1, S2, S5, S6, S7, S8 and S13).

b) Stimuli

Biphasic pulse trains with varying stimulation rate were used in experiment 4. Phase duration was 26.7 µs for most subjects, except for S5 and S13 where phase duration had to be increased to 36.7 µs to achieve a comfortable loudness level. Due to the influence of stimulation rate on the loudness of the stimuli, comfortable listening levels were measured for several pulse rates (100, 141, 200, 283, 400, 566, and 800 pps). As the pulse rate of the stimuli was in the continuous range of 100 to 800 pps, the level of the stimulus was set to the comfortable loudness measurement of the closest measured pulse rate. To avoid the influence of any residual loudness cues, a roving level paradigm was utilized (details outlined in experiment 4(a).

c) Procedure

A three interval, two alternative forced-choice procedure with feedback was used to measure the difference limen. Similar to Shackleton & Carlyon (1994) the base rate (for pulses and amplitude modulation), R_0, was the same in the first and one of the second or the third intervals, and either higher or lower by ΔR_0 in the other interval. Therefore, the two standard intervals consisted of the first and either the second or the third stimulus, whereas the target interval was located at the second or third interval. Details are outlined in Fig. 13(a) and 13(c). The base rate used in each trial was randomly chosen from a rectangular distribution of width ±10% in 1%-steps centered on the nominal base rate R_0. For the target and standard intervals $\pm\Delta R_0/2$ was added to R_0 for the standard intervals and $\pm\Delta R_0/2$ was subtracted from R_0 for the target interval. The sign of $\pm\Delta R_0/2$ was randomly selected on each trial. An adaptive two-down one-up procedure was used, dividing ΔR_0 by 1.41 after two consecutive correct responses and multiplying ΔR_0 by 1.41 after one incorrect response. After three reversals the factor was reduced to 1.19. One run ended after ten reversals.

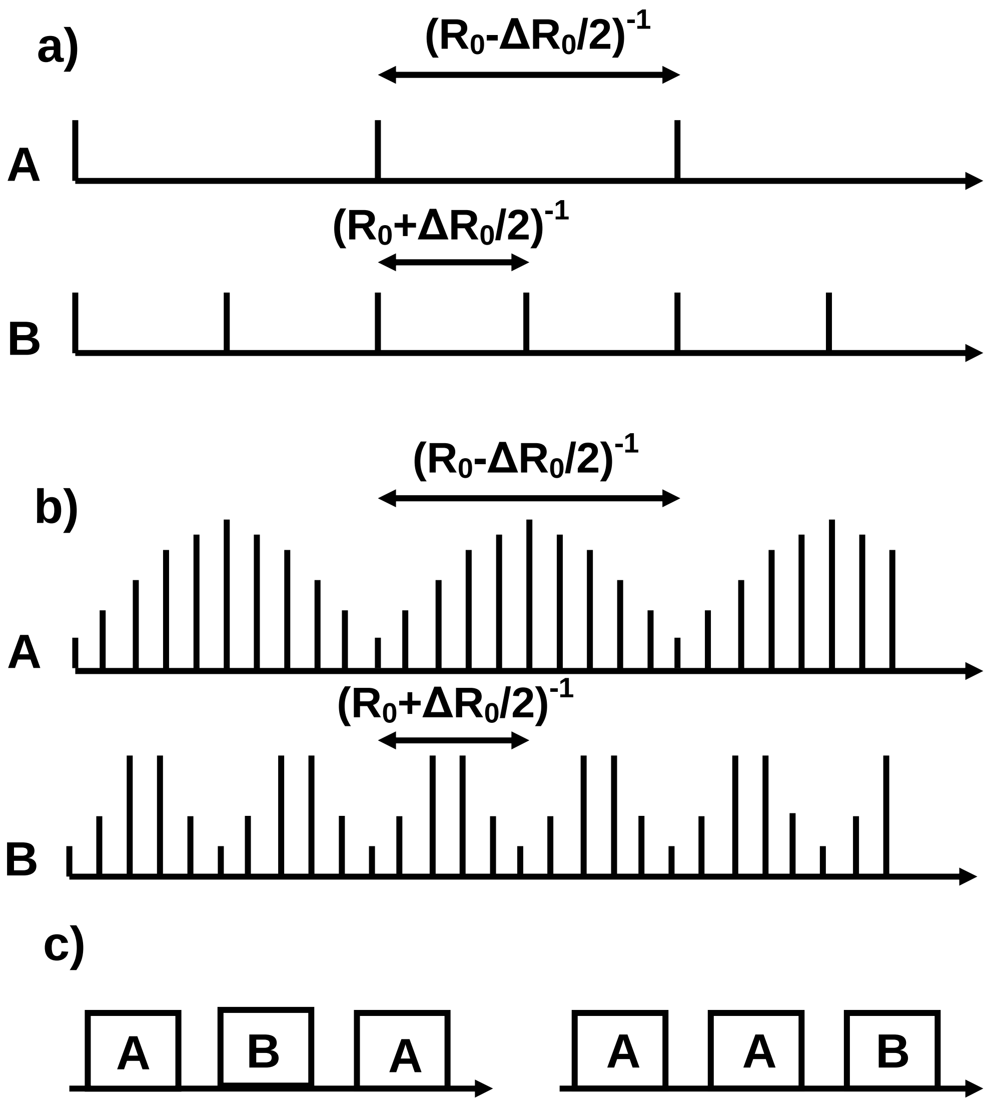

FIGURE 13. *Schematic drawing of the presented stimuli. In this case the rate of the standard stimulus (A) is calculated by subtracting $\Delta R_0/2$ from the base rate R_0, the target stimulus rate (B) is calculated by adding $\Delta R_0/2$ to the base rate R_0. Fig. 13(a) shows the stimuli used in experiment 4 to determine the PRDL, Fig. 13(b) shows the amplitude modulated stimuli utilized in experiment 5 (page 47). The sequence of presentations of standard and target stimuli is shown in Fig. 13(c).*

The last six reversals were used for data calculation. Threshold of each subject was obtained by calculating the arithmetic mean of the last six reversals of six different runs. The standard deviation of these 36 data points was calculated for each condition.

d) Experiment 4a: Pulse rate difference limen (PRDL) with roving level

i) Method

A preliminary study (experiment 4a) with four listeners (S1, S5, S6, S7) investigated the influence of an alternating stimulus level in every test interval (roving level paradigm) in pulse rate discrimination. Three conditions were tested. In the base condition the PRDL was measured at E3 for a base rate of 200 pps without roving level. A roving level of ±5% and ±10% was applied for the second and third condition, respectively. Thereby, listeners were encouraged to pay attention to the pitch differences and not to remaining loudness cues. The roving was distributed randomly between the three stimuli in one trial, as follows: one stimulus was presented at the comfortable loudness level, a second stimulus was played at a level softer than the comfortable loudness level and a third stimulus was played at a level louder than the comfortable loudness level. The roving level was calculated by adding ±5% of the dynamic range to the comfortable loudness level for the second condition or ±10% of the dynamic range for the third condition.

ii) Results

The results of experiment 4a show a strong dependency of PRDL on roving level. All listeners performed significantly better without roving level. In Fig. 14, the average PRDL in pps is plotted as a function of the amount of roving in percent of the dynamic range that was added to the comfortable listening level. The individual PRDLs with the standard deviation for the four listeners are presented by different open symbols. The filled circle with the connected line shows the arithmetic mean of the individual results with the standard deviation. PRDL increases with increasing roving parameter. Applying a roving level of ±5% increases PRDL by 20 pps compared to the presentation without roving level. Threshold further increases by 27 pps for the 10%-roving condition. The PRDL of listener S7 shows a very strong roving effect. For listener S1 the PRDLs for the 5%- and 10%-roving conditions hardly change.

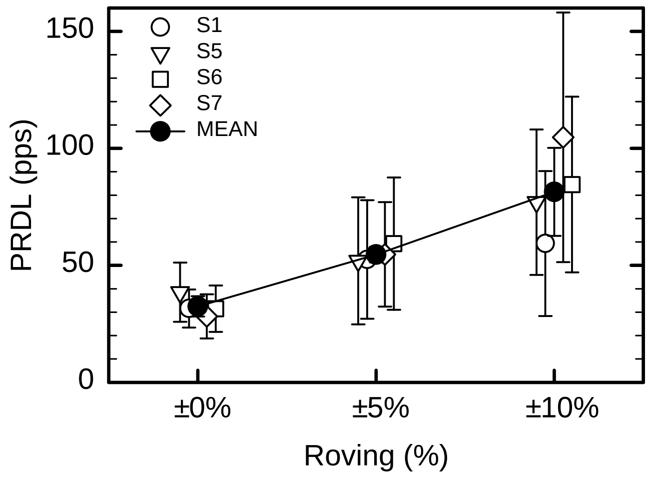

FIGURE 14. *Average PRDL in pulses per second (*pps*) as a function of roving level. The individual PRDLs for the four listeners with the standard deviation are presented by different open symbols. The filled circle with the connected line shows the arithmetic mean of the individual results with the standard deviation.*

In order to control the subjects' decision criteria and to be sure that the influence of possibly remaining loudness cues is eliminated, a roving level paradigm should be applied in the following experiments. The amount of roving, however, should be set cautiously because listeners reported to have difficulties in performing the experimental task within the 10%-roving condition, as the level of some stimuli was too soft. They also sometimes reported the perception of a slightly different pitch sensation in every interval, which might be traced back to the fact that there is possibly a small influence of stimulus level on perceived pitch (Townshend et al., 1987). In order to minimize the disturbing effect of stimulus level alteration on pitch discrimination, the 5%-roving condition was chosen for experiments 4b and 5 because within this condition loudness cues were presumably excluded. In this condition, the within subject as well as the between subject differences were within a reasonable range.

e) Experiment 4b: Pulse rate difference limen (PRDL)

i) Method

The PRDL of seven listeners (S1, S2, S5, S6, S7, S8, S13) was measured in the apical (E3) and basal region (E10) of the electrode array at four different base pulse rates, R_0 = 200, 283, 400, and 566 pps. The 5%-roving level procedure as described for experiment 4a was applied.

ii) Results

The individual PRDL results are plotted as a function of base pulse rate (R_0) in Fig. 15. The different symbols and line styles represent the different listeners. All listeners show an increasing PRDL with increasing base pulse rate. Fig. 15(a) shows the PRDL at E3 (apical electrode location). The results vary considerably between subjects. There are two top performers with regard to small PRDLs (S2 and S13), who demonstrate less influence of base pulse rate on PRDL up to 400 pps. Interestingly, for listener S13 the PRDL increases considerably at 566 pps, whereas the PRDL for S2 shows no influence of base pulse rate between 400 and 566 pps. For listener S5, PRDL increases considerably between 200 and 283 pps, but shows only a small increase with increasing pulse rate between 283 and 566 pps. Fig. 15(b) shows the PRDL at E10 (basal electrode location). The results are similar to the PRDLs obtained at the apical electrode location, except that the between subject variation at 566 pps is in a smaller range than at E3. This is due to the performance of listener S2 who shows an influence of electrode position on PRDL which is considerably higher at E10 than at

E3. Vice versa, the PRDL of S1 is considerably smaller at E10 than at E3. In contrast to E3, the results of listener S5 do not show a ceiling effect starting at 283 pps.

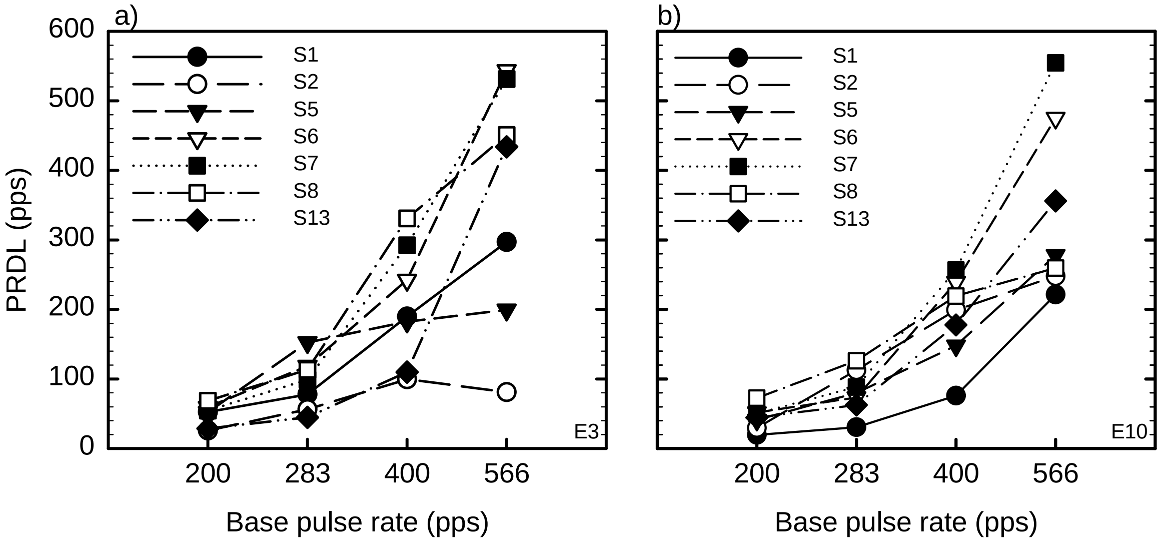

FIGURE 15. *Individual PRDL in pulses per second (*pps*) as a function of base rate at the apical electrode E3* (a) *and the basal electrode E10* (b)*.*

Figure 16 shows the PRDL results for each base rate averaged over all listeners. The increasing PRDL with increasing base pulse rate can be observed for either the basal or apical electrode. The averaged PRDL at E10 are slightly smaller than at E3. A statistical t-test comparison ($p<0.05$) does not show a significant difference. Regarding the individual results, four out of seven subjects show no difference between the apical and the basal electrode, one subject performs better at the apical electrode, one subject performs better at the basal electrode and one listener showed no consistent better performance at one or the other electrode. The standard deviation increases with increasing base pulse rate.

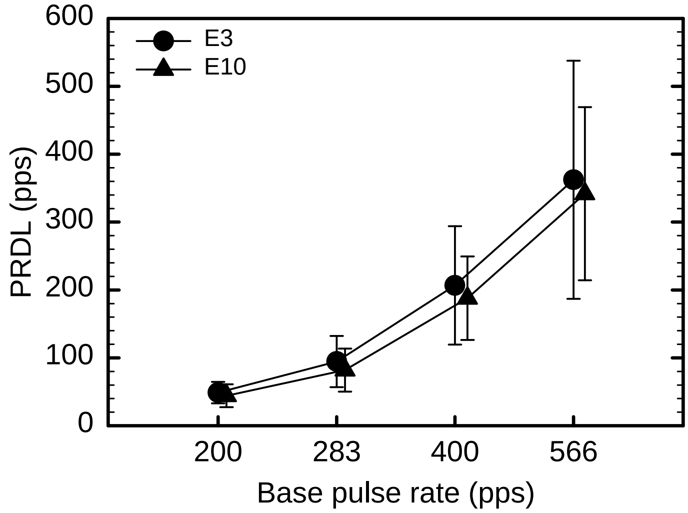

FIGURE 16. *Average PRDL (n = 7) in pulses per second (*pps*) as a function of base rate for E3 and E10, with the standard deviation.*

f) Discussion

Experiment 4 reveals that rate changes of about 23% are discriminable at a base rate of 200 pps. With increasing base rate, the PRDL increases (31% at 283 pps, 49% at 400 pps, 62% at 566 pps). The discrimination is independent of electrode position in the cochlea. These results are consistent with the most recent study by Zeng (2002). He investigated the PRDL in four listeners (3 Nucleus CI22M, 1 Ineraid implant) and reported an average PRDL of 40 Hz at 200 Hz (20%) and 135 Hz at 300 Hz (67.5%). No difference in PRDL was found between the tested most apical and basal electrodes for these implant types.

Previously published reports about pulse rate discrimination vary considerably. Townshend et al. (1987) studied PRDL on one electrode position in three subjects implanted with an 8-electrode device und reported PRDLs between 9% and 50% (base rate at 200 pps). McDermott & McKay (1997) measured the PRDL on three electrode positions in one highly skilled cochlear implant user (a former piano tuner) implanted with the Nucleus CI22M. They reported a PRDL of 11.3% at an apical and middle test electrode and of 5.4% at a basal test electrode. This variability might be attributed to the experimental paradigm utilized in these studies without thorough loudness balancing or level roving, and the fact that the subjects had highly varying etiologies of their hearing loss as well as different experience in psychophysical or speech tests. Van Hoesel & Clark (1997) measured the PRDL in two subjects, at two electrodes and at two implanted ears each. They found PRDLs between about 8% and 23% at a base rate of 200 pps and 12% to 55% at a base rate of 300 pps. This corresponds to the best performing subjects in experiment 4.

The results of experiment 4 can be compared with acoustic frequency discrimination in a limited range; only if the presented acoustic stimuli have been designed carefully to change only the temporal cues pertained to the stimulus and if changes in the excitation pattern are absent. Kaernbach & Bering (2001) used high pass filtered click sequences to explore the temporal mechanism involved in the pitch of unresolved harmonics in normal hearing subjects. The fundamental frequency (F0) of high-pass filtered, low-pass masked click trains was varied from 100 to 250 Hz. They reported frequency difference limens (FDLs) between 1.15% (3 kHz cut off frequency) and 1.5% (6 kHz cut off frequency) for a base frequency of 250 Hz. Carlyon & Deeks (2002) found a FDL of 5% at a base rate of 200 Hz for band pass filtered pulses (between 3.9 and 5.4 kHz) in alternating phase with three normal hearing subjects.

That means that compared to the frequency discrimination in normal hearing, the performance of cochlear implant subjects for rate discrimination is poor. One possible explanation is the fact that cochlear implant users have different kind of damages in the inner ear. Many studies with hearing impaired subjects have shown that the mechanisms which provide a high level of frequency discrimination capability are damaged (for example McDermott et al., 1998; Moore & Glasberg, 1986, Moore & Peters, 1992; Simon & Yund, 1993; Turner & Nelson, 1982; Tyler et al., 1983). The FDL for hearing impaired subjects is elevated. This can be attributed to the loss of sharp mechanical tuning of the basilar membrane which is often damaged by the loss of outer hair cells. McDermott et al. (1998) compared the FDL of normal hearing listeners and hearing impaired with a steeply sloping sensorineural hearing loss. The FDL amounts 2 to 4% at 250 Hz for hearing impaired in contrast to 1.2% for normal listeners. The frequency discrimination of pure tones of hearing impaired subjects is reduced; however, it is still considerably better than the average PRDL derived with electrical stimulation. That means that the loss of mechanical tuning can not completely explain the poor PRDL of cochlear implant subjects.

Furthermore, a broader excitation pattern resulting in an increased spread of neural excitation has been shown in tank and temporal bone studies (Kral et al., 1998) as well as in cochlear implant subjects (Shannon, 1990) compared to normal hearing. This can be traced back to the distribution of the electric potential and the corresponding current path. More centrally located pitch processing units are expecting a sharply tuned input from only a few neural fibers. A broader range of stimulated neurons might cause a loss of neural tuning and might deteriorate the detection of small pulse rate changes.

McKay & Carlyon (1999) explain the limited temporal discrimination with individual factors such as the reduced numbers of spiral ganglion cells or associated changes in the peripheral neural auditory system. In the subject group there are two subjects with a very short duration of deafness before implantation and no etiology of auditory neuropathy, S1 and S8. These subjects should show a nearly normal distribution of spiral ganglion cells and therefore the peripheral neural structures necessary for normal temporal pitch analysis should be provided. The results of those subjects are not supporting the neurologically motivated explanations, because they do not show improved PRDL. An additional argument against this neuropathological explanation of poor frequency discrimination is the observation that all of the subjects in the present study show a high level of speech recognition in noise with speech reception thresholds ranging from -0.35 to -2.4 dB (calculated as the signal to noise ratio at 50% speech perception over one test list with thirty sentences consisting of five words each) derived with a German sentence test (Oldenburger Satztest). Berlin et al. (2003) report a complete loss of speech reception in hearing impaired patients with auditory neuropathy with a mechanically intact cochlea. Therefore, it seems unlikely that the speech recognition capabilities of subjects of the present study have been reduced by a major degeneration of spiral ganglion cells or the whole auditory nerve.

An alternative explanation for the poor discrimination of pulse rate changes might arise from current cochlear implant stimulation strategies and electrode arrays which employ unnatural patterns of neural excitation. McKay & Carlyon (1999) have supposed that poor frequency discrimination is caused by the missing reproduction of phase relationships between different cochlear places (which occur in acoustic hearing due to traveling wave mechanics) and the additional mismatch of electrical rate with the corresponding characteristic frequency. Although a deteriorative effect of this rate-place mismatch can not be completely excluded, the absence of any significant influence of place of stimulation on PRDL suggests that the missing correspondence between rate and place does not degrade temporal pitch discrimination to a large extent.

Additionally, electrical stimulation with pulse trains causes a more deterministic response of the auditory nerve than acoustic stimulation. Hartmann et al. (1984) have examined period histograms and interspike interval histograms of the auditory nerve in cats. They found that the response of the auditory nerve for electrical stimulation was highly synchronized, whereas the response for acoustic stimulation was more stochastic (100-Hz stimulation frequency). Litvak et al. (2001) observed the responses of the auditory nerve fibers to unmodulated and modulated high-rate modulated electrical pulse trains in deafened cats. They found responses to modulated pulse trains that resembled responses to tones in intact ears. However, these responses were only observed in a limited range of modulation depths and presentation levels. Nevertheless, their results suggest that the coding of complex stimulus waveforms might improve with signal processing strategies for cochlear implants using a desynchronizing pulse train.

5. Experiment 5: Modulation rate difference limen ('MRDL')

Experiment 4 (page 36) has shown that the ability of cochlear implant users to detect small rate changes when stimulating a single electrode is poor. At a base rate of 200 pps the pulse rate difference limen amounts 23%. This is considerably higher than just noticeable frequency differences in impaired hearing and in normal hearing with band passed filtered signals. The reduced rate discrimination can be traced back to several factors: the loss of mechanical fine tuning, the broader excitation pattern, the degeneration of spiral ganglion cells, the loss of phase relationships and the high synchronization of the response of the auditory nerve. Litvak et al. (2001) have suggested a desynchronizing pulse train in order to get a more stochastic response of the auditory nerve for electrical stimulation. Theoretically, amplitude modulated stimuli with high carrier rate might also reduce the strong synchronization of responses due to a more probabilistic excitation. In experiment 5, the pulse rate discrimination experiment was repeated with amplitude modulated stimuli. A better detection of modulation frequency differences was expected. A more detailed overview of experiment 5 can be found in the publication by Baumann & Nobbe (2004b).

a) Participants

Three subjects (S1, S5, S7) took part at the modulation rate discrimination experiment.

b) Method

The pulse rate of the carrier was set to 5081 pps to avoid aliasing effects. A sinusoidal amplitude modulation of the carrier pulse (example is given in Fig. 13(b)) was applied with base modulation rates of 200, 283, 400, and 566 Hz. For each base modulation rate, a measurement of the comfortable loudness level was conducted. Similar to experiment 4 (page 36), the level of the stimulus between these base modulation rates was set to the comfortable loudness measurement of the closest modulation base pulse rate. The modulation depth of the stimuli was obtained in the following way. Prior to the experiment, the threshold for an unmodulated 5081-pps stimulus was measured for each subject. The current of the stimulus was then modulated between this threshold and the current required for a comfortable listening level for a modulated stimulus. The onset and offset was modulated by a Gaussian window with a rise time of 25 ms (between 10 and 90% of the stimulus' amplitude).

The results of the previous experiment indicated no dependency of the pulse rate discrimination limen on electrode position. Therefore, experiment 5 measured MRDL only in the apical region (E3) at four different modulation base rates (R_0 = 200, 283, 400, and 566 pps). The carrier pulse rate was 5081 pps (an overview over the stimuli is given in Fig. 13(b). The same procedure as in experiment 4 (page 36) with a roving level of 5% was applied.

c) Results

Figure 17 compares the results of the three participating listeners for the PRDL (filled symbols) and MRDL (open symbols) obtained from E3. In general, the difference limen for the amplitude modulated stimuli is higher than for the unmodulated pulse trains for all subjects and all conditions. In order to detect differences, a large amount of modulation rate change is required above a modulation base rate of 200 pps. A comparison of the averaged data for the three subjects for both experiments is shown in Fig. 18. The amplitude modulated stimuli show a steeper increase of DL with increasing base rate than the unmodulated pulse sequences. Due to smaller interindiviual differences, the DL of the amplitude modulated stimuli displays a smaller standard deviation.

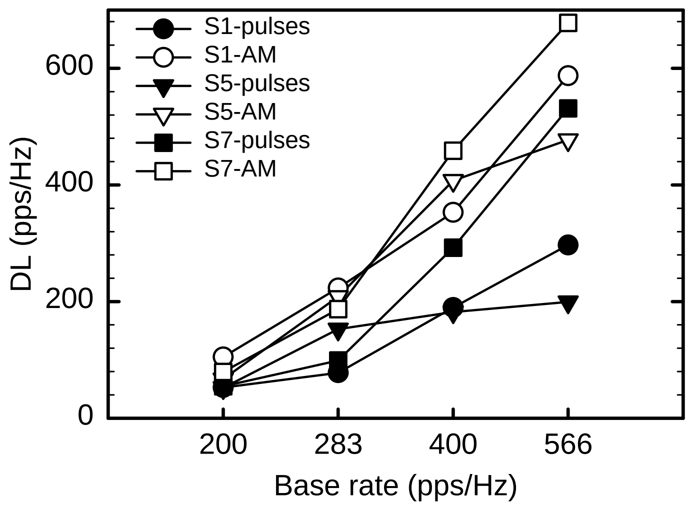

FIGURE 17. *Individual PRDL (filled symbols) and AMDL (open symbols) in* pps*, respectively* Hz*, as a function of base rate for three listeners at electrode E3.*

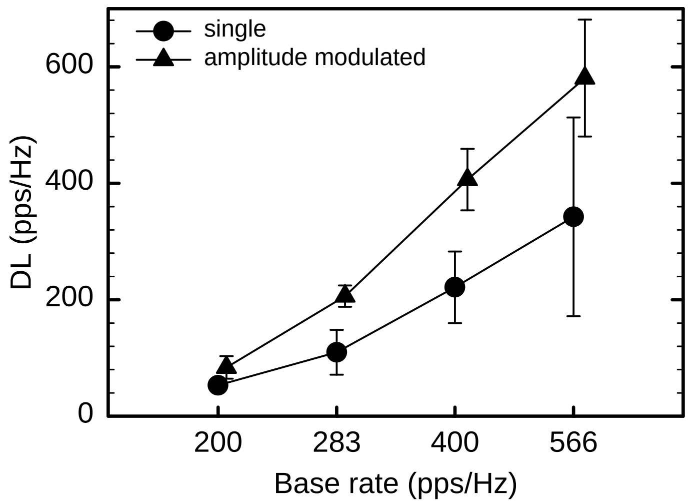

FIGURE 18. *Average PRDL and AMDL in* pps*, respectively* Hz*, as a function of base rate with standard deviation at electrode E3.*

d) Discussion

Against the expectations the MRDL was higher than the PRDL. It amounts to 41% at a modulation base rate of 200 pps. This is not consistent with the results of previous research (McDermott & McKay, 1997). McDermott & McKay (1997) examined the MRDL in one subject using amplitude modulated pulse trains with a carrier frequency of 1200 pps. At a modulation rate of 200 pps, the subject was able to discriminate a MRDL between 4.2% and 26.7% depending on electrode position. In contrast, the best performing subject of the present study had shown a MRDL of 33.4%. This discrepancy might be caused by differences in the experimental setup, namely the lower carrier pulse rate (1200 pps) and the absence of stimulus level roving, so that the better performance of the subject might arise from small changes of residual loudness, roughness, or timbre. McDermott & McKay (1997) also reported that the average difference limen was smaller for unmodulated than for amplitude-modulated pulse trains. This is consistent with the results of the present study and also in agreement with data obtained from normal hearing subjects (Formby, 1985).

A recent study by Oxenham et al. (2004) can also be compared with experiment 5. They used so called transposed tones to dissociate temporal from place information in normal hearing subjects. The transposed tones where generated by multiplying a half-wave rectified low-frequency sinusoid with a high-frequency sinusoidal carrier. This way the information of low-frequency sinusoids was presented at locations tuned to high frequencies. In a 3 AFC test, the detection of modulation frequency changes between 55 and 320 Hz was tested. They found increased thresholds for transposed tones in comparison with the results for pure tones with the same procedure. For a modulation rate of 200 pps, the FDL amounted between 5 and 12% depending on the carrier frequency. With increasing carrier frequency, the threshold increased by 7%. This is in contrast to the results of experiment 5 where there was no difference depending on stimulated cochlear region. Furthermore, in Oxenham et al. (2004), threshold was decreasing with increasing modulation frequency between 55 and 320 Hz for both, transposed and pure tones. Contrarily, in electrical stimulation threshold increases with increasing base rate (experiment 4, page 36) and modulation base rate (experiment 5) and it can be assumed that rate changes are noticeable on average up to 300 pps. The elevated threshold in experiment 5 in comparison with the results obtained by Oxenham et al. (2004) may again be caused by the broader range of neural excitation, the degeneration of spiral ganglion cells and the loss of the support of the cochlear mechanical fine tuning in the cochlear implant subjects. However, it remains unclear why threshold is decreasing with increasing modulation frequency.

6. Experiment 6: Binaural pitch adjustment

The influence of place and rate on the pitch perception with electrical stimulation was examined in experiments 1 to 5. However, the absolute range of the perceived pitch remains unclear. The pitch height evoked by each electrode on the array is mainly influenced by the place of stimulation and changes only by a small amount with the rate of stimulation. For pulse rates higher than approximately 300 pps, there is a saturation in the pitch height (experiment 3, page 30). A common assumption is that with electrical stimulation rates above pitch saturation the perceived pitch corresponds to the best frequency of the auditory nerve neurons according to the frequency-place allocation for normal hearing derived for example by Greenwood (1990). According to Greenwoods function, the most apical electrode on the array, E1, located at a 30.3-mm distance from the round window might evoke a pitch perception in electrical hearing which corresponds to about a 140 Hz pure tone in normal hearing. In the same way, electrode E2 might evoke a pitch perception corresponding to a 300 Hz pure tone and electrode E6 a pitch perception corresponding to a 1.3 kHz pure tone. The most basal electrode E12 might evoke a pitch perception corresponding to an 8 kHz pure tone. It seems obvious, that the perceived pitch evoked by each electrode should be utilized individually for the signal processing strategy. Since in the current implementation for the COMBI 40+ the signal is filtered by 12 band pass filters and the information of each band pass filter is then transmitted to an electrode location inside the cochlea (see Fig. 1, page 3), an exact allocation of the spectral information in the signal to an electrode with the corresponding pitch perception might contribute to a better acceptance of the sound of a cochlear implant and might enhance the representation of spectral information. In the current speech processing strategy for the COMBI 40+ the band pass filters are allocated to electrode positions on a logarithmic scale in order to approximate the frequency-place transformation. However, up to now it is unclear whether the electrical stimulation with a fixed and relatively high rate evokes a pitch perception of which the increment and range can be compared with the frequency-place transformation in normal hearing.

One way to examine the exact evoked pitch perception with electrical stimulation is to test subjects with residual acoustical hearing at the non-implanted ear. Most of those subjects have residual hearing in the lower frequency range up to 1 kHz. Regarding Greenwoods frequency-place map (Greenwood, 1990), the apical electrodes of the COMBI 40+ might evoke pitch perceptions which can be matched within the range of acoustical hearing at the non-implanted ear. Therefore, an experiment was designed in which subjects with residual hearing had the task to compare the pitch height of acoustic and electrical stimulation. Namely they had to adjust the frequency of a pure tone at the non-implanted ear in a way to match the pitch with the perception elicited by electrical stimulation of a certain apical electrode with a fixed stimulation rate.

a) Participants

The six participating subjects (S1, S2, S3, S7, S13 and S15) had residual hearing at the non-implanted ear with hearing losses between 45 and 100 dB HL at 125 Hz and between 70 and 105 dB HL at 1000 Hz. One subject (S4) had a limited residual hearing range up to only

400 Hz. The individual pure tone audiograms are shown in Fig 19. Three subjects were regularly using a hearing aid at the contralateral ear (S2, S13, S15).

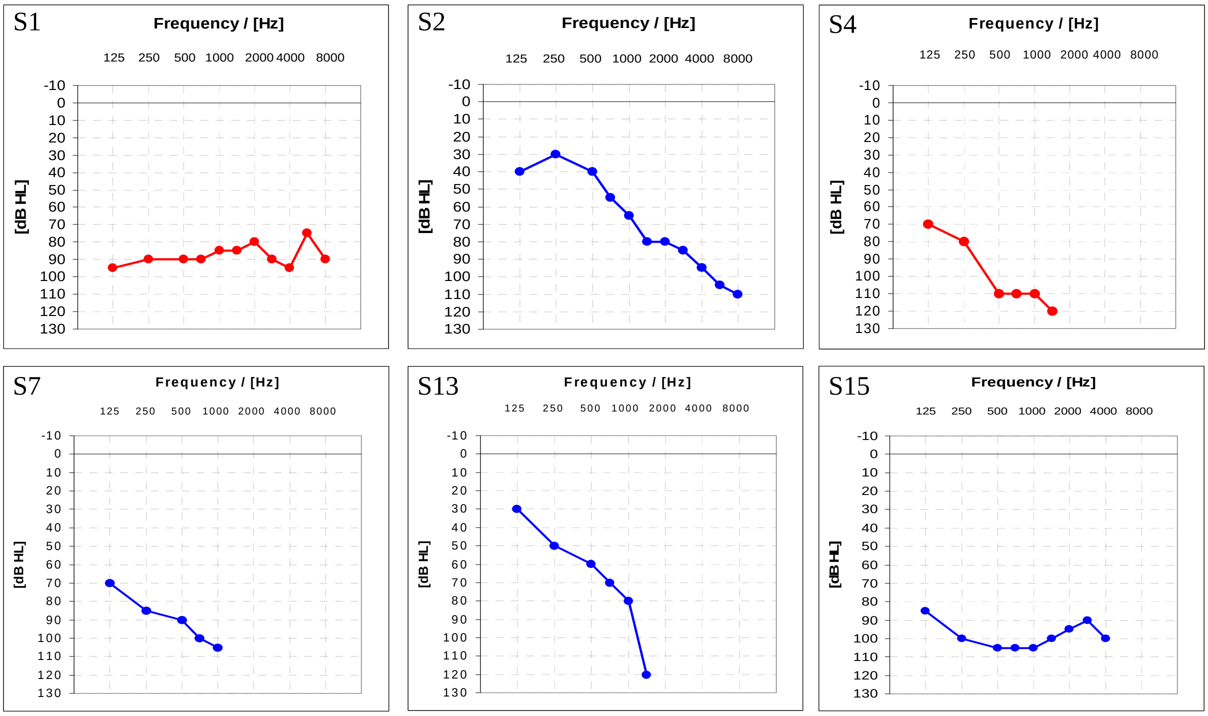

FIGURE 19. *Individual pure tone audiograms of the six subjects with residual hearing at the non-implanted ear participating in experiment 6.*

b) Procedure

The pitch-matching task was performed at different apical electrodes and with varying start frequencies of the acoustic stimulus. The stimulated electrode was chosen randomly. The stimuli for the implanted and the residual hearing ear were presented alternating between both sides. The subject had to turn an adjusting knob to change the frequency of the acoustic stimulus. The matching task was terminated when the patient pressed a key indicating that the stimulus in the hearing ear had the same pitch as the stimulus in the implanted ear. The subjects could listen to the stimuli without a time limit. For five of the subjects (S1, S2, S7, S13, S14), ten fixed start frequencies were chosen randomly between 125 and 1000 Hz. For each electrode and each start frequency two estimates were collected. The average adjustment for each electrode was calculated as the median of 20 estimates. Due to the limited range of residual hearing, for subject S4, a reduced set of seven start frequencies was chosen randomly between 75 and 300 Hz resulting in 14 estimates for each electrode.

The electric stimuli consisted of biphasic current pulses with pulse duration of 26.7 µs per phase. The stimuli had duration of 500 ms; a constant stimulation rate of 800 pps was used. Depending on subject, three to six apical electrodes (E1 to E3-E6, see Fig. 1, page 3) were stimulated. For subject S4 the number of electrodes depended on the range of residual hearing (hearing sensation up to 400 Hz). Subject S13 performed the test only at the four most apical electrodes because the stimulation of electrodes E5 and E6 evoked a stimulation of the facial nerve. The current amplitude was adjusted to the perception of comfortable loudness for each stimulated electrode (see chapter I.3., page 14).

The acoustic stimuli consisted of pure tones with a 25-ms rise/fall time and were digitally generated on an IBM-compatible PC using ‘Matlab’® software. The signals were delivered via D/A converter and amplifier and were presented over headphones (HDA 200, Sennheiser). The frequency of the sinusoids could be adjusted between 125 and 1000 Hz (75 to 400 Hz for S4) in 1-Hz steps. The stimuli’s amplitudes were determined via ‘Matlab’® software as following: prior to testing the amplitude of the pure tones to achieve comfortable loudness were determined at 125, 250, 500, 750 and 1000 Hz (additionally 75 Hz for S4). The amplitudes during the experimental run were then interpolated according to these measurements depending on test frequency. All stimuli had duration of 500 ms; the interstimulus gap between the electric and acoustic stimulus was 500 ms.

c) Results

The individual results of the pitch matching task are shown in Fig. 20. The median and the twenty single estimates are plotted for each electrode and subject. Although the adjustments for each electrode are varying in a wide range, the average adjusted frequency of the acoustic stimulus is increasing with increasing electrode number in each subject. Most of the subjects adjusted the frequency of the pure tone for the two most apical electrodes E1 and E2 as equal. Two subjects do not show differences in the adjustments between more basal electrodes: S1 between E4 and E5, S15 between E3 and E4.

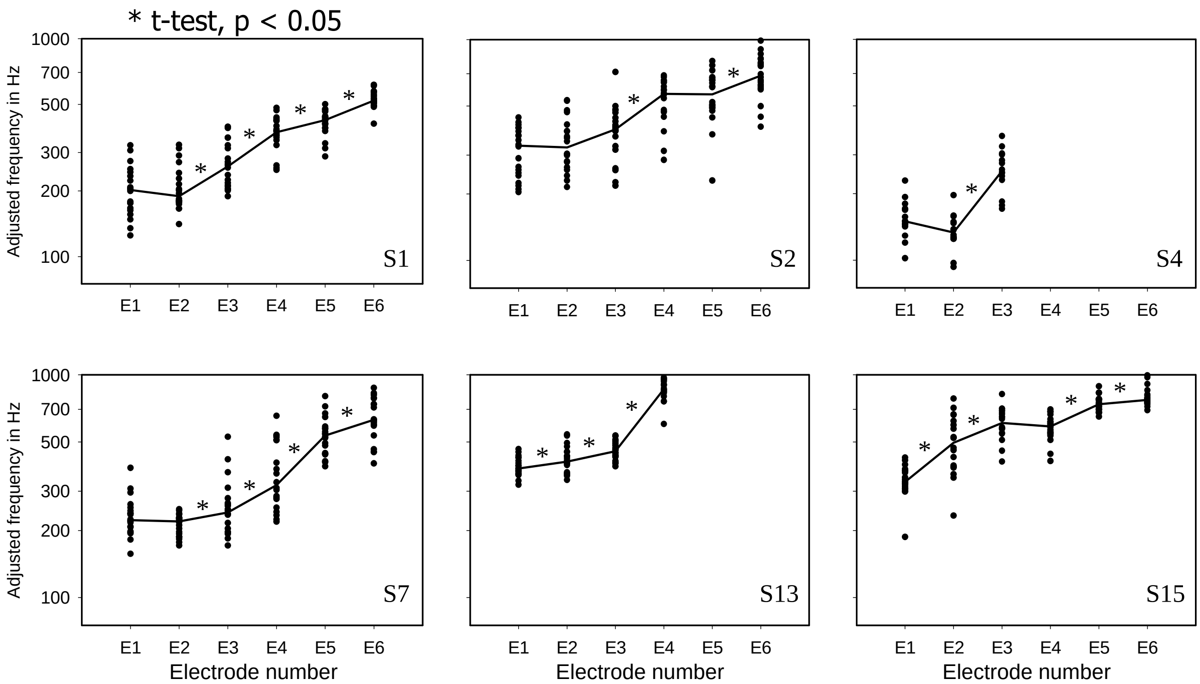

FIGURE 20. *Individual results for experiment 6. Adjusted frequencies of pure tones matching with electrical stimulation are plotted as a function of electrode number. Average depicted with continuous line. Significant differences between two neighbored electrodes are indicated with a star.*

Figure 21(a) shows that the results vary considerably between subjects. The average adjustment corresponding to E1 was between 150 Hz (S4) and 380 Hz (S13). The adjusted frequency corresponding to E6 varies between 520 Hz (S1) and 780 Hz (S15). The results of the subjects S1, S2, S7 and S15 show a similar ascending slope on a logarithmic frequency scale for those electrodes which elicit a significantly different pitch perception.

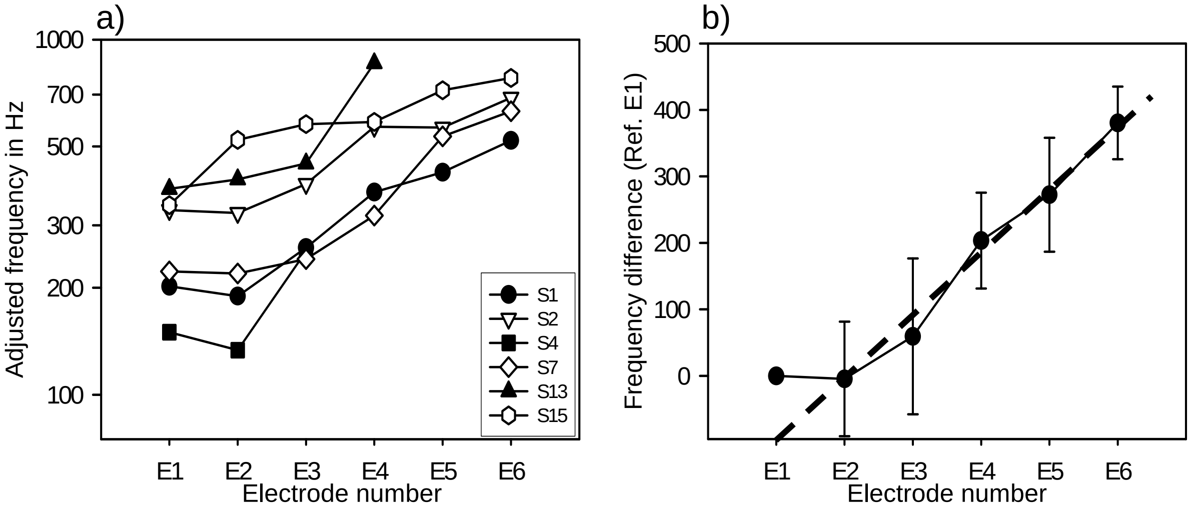

FIGURE 21. (a) *Individual averaged frequency adjustments replotted from Fig. 20.* (b) *Average adjustments with standard deviation of four subjects who completed the test for all six electrodes. Results are calculated in relation to the average adjustment of E1. Dashed line: linear regression between E2 and E6 (R^2=0.98).*

To circumvent the influence of between subject insertion depth variations (see Fig. 22 for Stenvers' view x-ray scans), Fig. 21(b) shows the averaged adjustments (median) related on E1 for four subjects (S1, S2, S7, S15) who performed the test at all six electrodes. The average data shows no statistical significant difference ($p<0.05$) in frequency adjustment between the two most apical electrodes. Presumably due to the perception of an increased pitch height the average adjustment starts to increase at E2. A linear regression analysis of the adjustments between E2 and E6 shows a high correlation (R^2 = 0.98). The electrode distance of 2.4 mm corresponds on average to a difference in the adjusted frequency of 98 Hz.

d) Discussion

Assuming that the participating subjects in experiment 6 with remaining hearing at the non-implanted ear are able to estimate a difference in pitch height without confusing it with a difference in sound quality, an increasing pitch height with increasing electrode number could be proved. However, four of six subjects estimated the pitch of the most apical electrodes E1 and E2 at the same level. Between E2 and E6 the pitch is increasing with a slope of 40 Hz per mm from apical to basal.

The frequency adjustments show a large within as well as between subject variability. The estimated frequency for the most apical electrode lies in between 150 and 380 Hz. Figure 22 shows the Stenvers' view x-ray scans of the position of the electrode array in the cochlea for each subject.

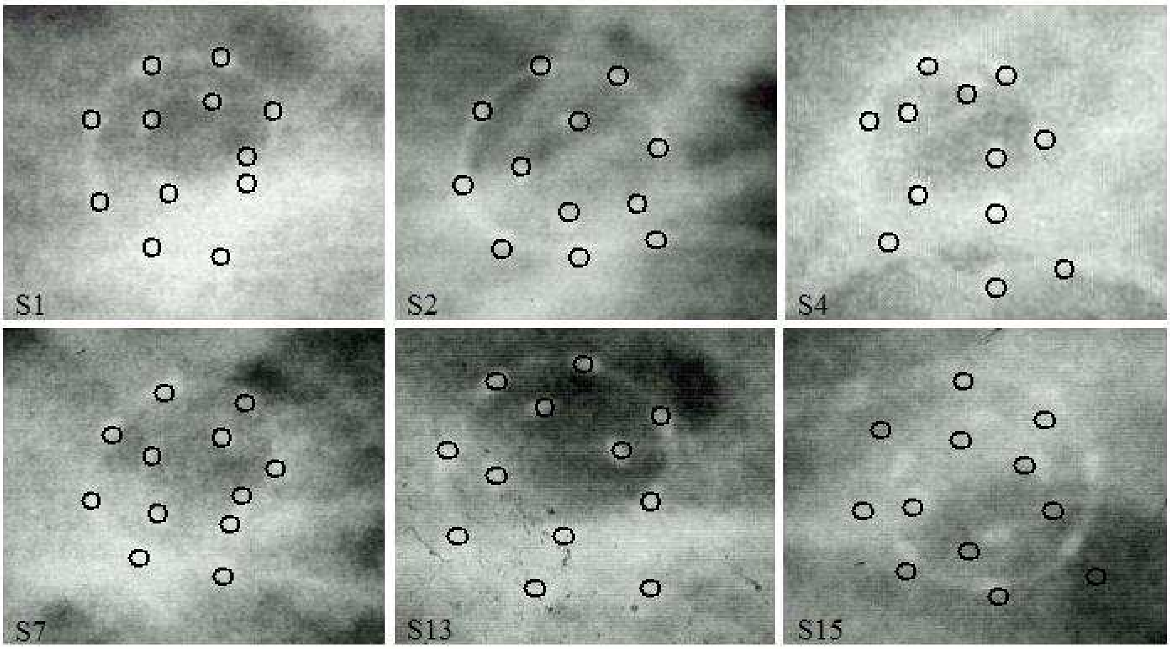

FIGURE 22. *Stenvers' view x-ray scans of the electrode arrays of the subjects performing the binaural hearing experiment 6. Electrodes are highlighted with black circles for better visibility.*

The electrode arrays of subjects S4 (E1: 156 Hz) and S7 (E1: 234 Hz) are inserted especially deep up to two complete turns. The electrode array of subject S1 (E1: 205 Hz) is inserted only a little less deep. The arrays of subjects S13 (E1: 387 Hz) and S15 (E1: 337 Hz) and especially of S2 (E1: 318 Hz), however, are comparably less deep inserted into the cochlea. That means that the interindividual differences for the estimated frequency of E1 might be contributed to the different positions of E1 in the cochlea and therefore the different places of electrical stimulation.

Dorman et al. (1994) have conducted an experiment comparable to the present study with a single subject provided with an Ineraid prosthesis (Eddington, 1980). They indicated that the most apical electrode of this device was located at 22 mm from the round window and the electrodes are spaced at 4-mm intervals, although the insertion depth was not controlled by Stenvers' view x-ray scans. They investigated the adjusted acoustic frequency at the non-implanted ear for different rates of electrical stimulation at the implanted ear. Two adjustments were collected for each condition which differed by maximally 80 Hz. As the average pitch perception changes with changing stimulation rate up to about 300 pps with large differences between subjects (see experiment 2, page 24), their data for the 400-pps condition can roughly be compared with the results of experiment 6. The frequency for the matching acoustic stimulus at the most apical electrode was adjusted to 380 Hz, the frequency for the next more apical electrode (18 mm from the round window) was adjusted to 460 Hz. In comparison with the results of experiment 6 – E4, 23.1 mm from the round window: 465 Hz, E6, 18.3 mm from the round window: 666 Hz – the frequency adjustments according to Dorman et al. (1994) are lower. As the pulse rate for pitch saturation is varying considerably between subjects (see experiment 2, page 24), it is not sure whether the pitch perception above 400 pps would not further increase with increasing pulse rate for this subject. Therefore, the relatively low pulse rate might still influence the pitch perception depending on electrode location. Furthermore, the different results might also be contributed to the characteristic of the hearing loss of the subject in Dorman et al. (1994). The subject shows a ski-sloping audiogram with nearly normal threshold at 250 Hz (25 dB HL), moderate loss at 500 Hz (50 dB HL) and deafness to test tones with frequencies of 1 kHz and above. The subject might have adjusted the frequency of the matching acoustic stimuli according to a pitch perception one or two octaves below the pitch height of the electric stimulus due to the well known effect of octave confusion (Terhardt & Gruber, 1986) and because he could only adjust the frequency of the acoustic stimulus in a limited range up to less than 1 kHz.

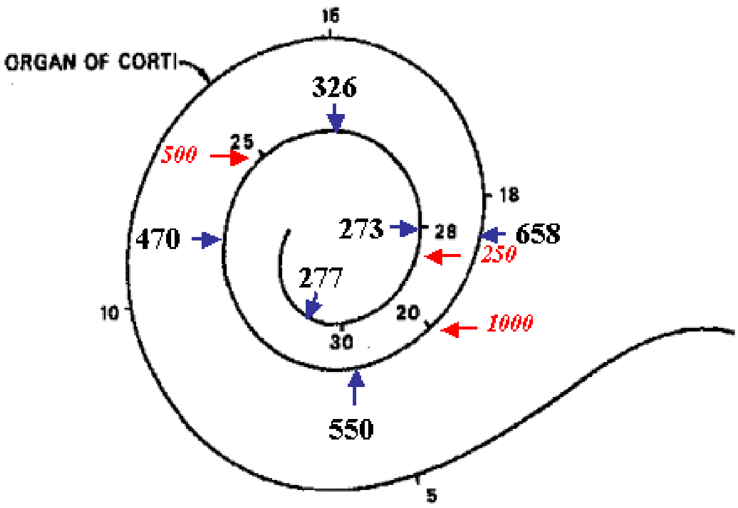

FIGURE 23. *Schematic drawing of the cochlea with the frequency-place allocation for normal hearing (italic numbers) after Otte et al. (1978). The frequency-place allocation according to the position of electrodes and the average adjusted frequencies in experiment 6 are indicated with the numbers in bold font type. Small numbers indicate the distance from the round window.*

Figure 23 shows the allocation of estimated frequencies at the estimated positions of the six most apical electrodes (maximal insertion) in the cochlea. In comparison the frequency-place allocation for oscillating best frequencies in normal hearing after Otte et al. (1978) is indicated. The averaged frequency adjustments of experiment 6 differ from the best frequencies in normal hearing at fixed places in the cochlea. The pitch of the most apical electrode E1 (277 Hz) is matched with a higher acoustic frequency as it was expected from the frequency-place allocation (a 30.4 mm distance to the round window corresponds to a best frequency of about 200 Hz in normal hearing, Zwicker & Fastl, 1999). The pitch of E2 (272 Hz) was estimated at nearly the same frequency as E1 although E2 is located 2.4 mm more basally. The adjusted acoustic frequency for E2, however, corresponds to the best frequency in normal hearing at this place of the cochlea. The matched frequencies elicited by E3 (326 Hz) and E4 (470 Hz) are lower compared to the frequency-place allocation in normal hearing where the place of a 500 Hz pure tone is located between E3 and E4. The same effect occurs for the frequency adjustments for E5 (550 Hz) and E6 (657 Hz) which should stimulate a region in the cochlea with a best frequency of about 1000 Hz in normal hearing. The main outcome of experiment 6 is that there are major differences in terms of the frequency-place allocation when electrical and acoustical stimulation are compared. First, the increase of the pitch perception from apex to base is different from normal hearing: The linear regression of the estimates between E2 and E6 shows a slope of about 40 Hz per mm. In normal hearing, the best frequency increases by 70 Hz per mm in the apical region (Zwicker & Fastl, 1999). Second, the adjustments of the most apical electrodes show no changes in pitch height although located at a 2.4-mm distance.

The differences between the electric/acoustic frequency matches of the present study and the frequency-place allocation in normal hearing might be caused by different influences. First, it is possible that the participants were subject to octave confusions in their adjustments of the acoustic frequencies. The range of remaining hearing and therefore the range of adjustable frequencies for the acoustic stimulus were limited. As a consequence the subjects had to find matching frequencies within the limited range. This effect is therefore more likely to occur at more basal electrodes because the acoustic range of the apical electrodes should surely correspond to the given frequency range. However, the distributions of the single adjustments for each subject and electrode do not show two centers. The amount of the variance and the standard deviation do not change depending on electrode position. During the experimental runs the subjects hardly ever reached the upper limit of the adjusting knob. This is also visible in the single estimates which are hardly reaching the region of 1000 Hz.

The great variance of the individual estimates (standard deviation in average 20%) indicates that the subject's task to adjust the acoustic frequency was not easy to solve. The subjects had participated in most of the described experiments and were therefore somehow trained to judge the perceived pitch but the variance of their results differs from the variance of normal hearing subjects for a similar task. This effect can be traced back to the high level of hearing impairment at the non-implanted ear. During the experimental runs subjects often reported that the perception of the acoustic and electric stimuli differed in a way that it was hard to compare the pitch. The electrical stimulation evoked a clear and pleasant pitch sensation; the acoustic stimulus was perceived as buzzy and was often accompanied by a feeling of uncomfortable non-auditory sensations. This is possibly due to the high stimulus amplitudes that were necessary to make the stimuli audible. Subjects also reported that the pitch of the acoustic stimulus hardly changed in the frequency range higher than about 500 Hz. It is imaginable that the individual characteristics of the hearing loss are responsible for this observation. Four out of six subjects showed a steep ski-sloping hearing loss. Especially in this region of high hearing loss so called 'dead regions' might appear (Moore et al., 2000; Moore & Alcantara, 2001). In these regions of the cochlea, the hair cells as well as spiral ganglion cells are damaged to a very high amount or completely destroyed. If the basilar membrane is excited with a best frequency which is located in a dead region, then only the edge of the stimulus activates residual spiral ganglion cells in the surrounding regions. The pitch of the stimulus will be perceived corresponding to the position of the residual spiral ganglion cells. That means that depending on the amount of damaged or destroyed spiral ganglion cells, a certain incoming frequency range elicit the same pitch perception. This effect might influence the estimates in experiment 6. Different frequencies at the acoustically stimulated ear might evoke the same pitch perception and consequently the individual estimates for one electrode spread. Considering the individual results, the estimates of S7 for E4 to E6 might be contributed to this effect. The standard deviation for these electrodes is increased in comparison to electrodes E1 to E3. The subject shows 85 dB to 100 dB hearing loss in the range of the estimated frequencies for E4 to E6. However, the estimated frequencies for E4 to E6 are still significantly increasing. Subject S15 shows a larger standard deviation for estimates at E2 and E3. This subject has an especially profound hearing loss over the whole frequency range (85 to 105 dB). It is possible that the thresholds in the pure tone audiogram are also influenced by the effect of dead regions. S15 shows no significant difference between E3 and E4, whereas all other estimated frequencies for neighbored electrodes are significantly different.

It is also possible that those subjects who can not profit of a hearing aid at the non-implanted ear are not used to the sensation of acoustic hearing after months of electric hearing. The standard deviation of the individual estimates however does not seem to differ between the subjects with and without hearing aid.

The standard deviation is also independent of the level of hearing loss. The subject with the fewest hearing loss (S2), together with the subject with the most severe hearing loss, S15, show the greatest standard deviation. For subjects S1, S4 and S13 the standard deviation is smallest.

Another reason for the differences between electric/acoustic frequency matches for fixed stimulated places in the cochlea and the frequency-place allocation in normal hearing might be due to the rate of stimulation for the electric stimuli. The stimulation rate was fixed at 800 pps independent of place of stimulation. The mismatch between the considered best frequency at the place of stimulation and the effective stimulation rate might influence the perceived pitch of the electric stimulus. Experiment 2 (page 24), however, has shown that the place of stimulation dominates the pitch sensation for electrical stimulation. Especially for subjects S1 and S2 the pitch sensation hardly changed with changing stimulation rate in experiment 2. The mismatch between place and rate of stimulation might influence the amount of the estimated frequency but it does not explain the lack of pitch difference between E1 and E2 neither the reduced increase of estimated frequency with stimulated place (40Hz/mm) in comparison to normal hearing (70Hz/mm).

III. SIGNAL PROCESSING

1. Background

a) The MED-EL CIS strategy

The speech processor Tempo+ (MED-EL, Innsbruck) employs the CIS (continuous interleaved sampling) strategy according to Wilson et al. (1991). An implementation of the CIS strategy in ‘Matlab’® is provided as a demonstration for the Research Interface Box (University of Innsbruck, Austria, see also page 12). This implementation allows the processing of digitized sound files (wav-format, Microsoft, Redmond, United States of America) with a resolution of 16 bit and a sampling frequency of 44.1 kHz. The signal processing part of this implementation only differs in terms of the envelope extraction method from the strategy which is used in the Tempo+. The general conversion of incoming analogue signals in the Tempo+ is shown as part of the complete signal processing path in Fig. 24 although the analogue section (A) is not part of the ‘Matlab’® CIS implementation.

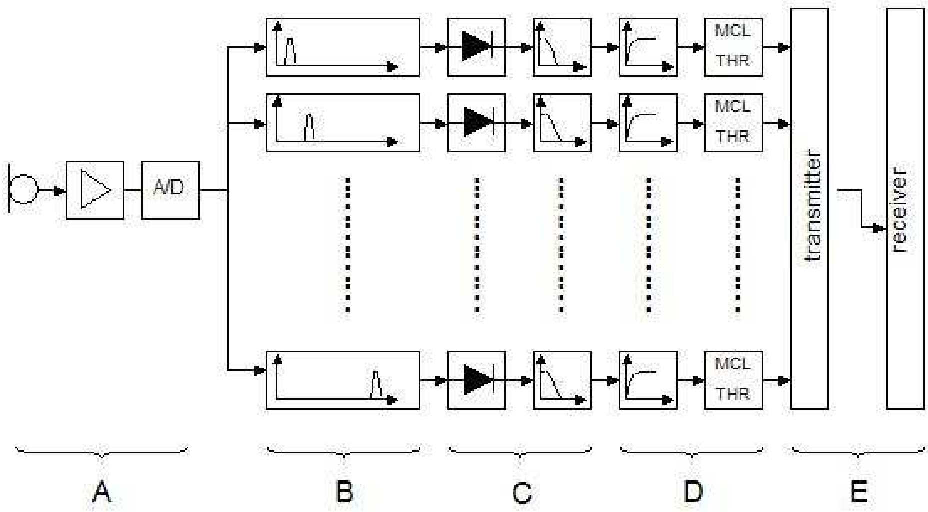

FIGURE 24. *Block diagram of the CIS signal processing strategy (Wilson et al., 1991). (A) analogue part, (B) filter bank, (C) envelope extraction, (D) current mapping, (E) transcutaneous radiofrequency transmission line to the receiver.*

The digital signal is first preemphasized by means of a first order digital filter (see attachment 2 for transfer function). Subsequently, the signal is filtered by a 12 channel band pass filter bank (section B and Fig. 25(a)). The default setting of the filter bank distributes the lower and higher cutoff frequencies of each channel logarithmically between 300 and 7000 Hz. After the band pass filter section an envelope extraction is realized by means of a two way rectifier and a subsequent low pass filter (higher cutoff frequency 400 Hz) in each band. All filters are of type Butterworth (section C). The envelope serves as amplitude modulator source for a pulse train carrier whereby the amplitude is defined according to the predetermined individual dynamic range of the patient and a logarithmic transformation (section D). Electrical threshold (THR) and most comfortable loudness level (MCL) are adjusted individually and are necessary to map the stimulation into the individual dynamic range. The information is then transmitted to the receiver's coil which is located inside the implant. The electrodes are stimulated non-simultaneous in an interleaved order optimized to avoid interactions among channels (stimulation order: E1, E7, E2, E8, E3, E9, E4, E10, E5, E11, E6, E12). The default pulse rate for each electrode is 1515 pps which is consequently in each filter bank channel the sampling rate of the signal processing. The trains of biphasic pulses are delivered with temporal offsets that should eliminate any residual overlap across channels. This is also supported by the short pulse duration of 26.7 µs.

b) Consequences of conducted pitch perception experiments

The MED-EL COMBI 40+ provides 12 electrodes with a relatively wide electrode spacing of 2.4 mm compared to other electrode array designs. Experiment 1 (page 16) has shown that this spacing is wide enough for the average cochlear implant user to discriminate between adjacent electrodes. This was found for three reference electrodes positioned at the apical, medial and basal region of the cochlea. Experiment 6 (page 51) was analyzing in a more detailed way the pitch perceptions evoked by the stimulation of different apical electrodes. The results show that the two most apical electrodes are indiscriminable in terms of elicited pitch. Consequently, the pitch information underlying in the two lowest band pass filter sections, which is transmitted by E1 and E2, can not be differentiated. In agreement with experiment 1 (page 16), the difference in the perceived pitch between electrodes E1 and E3 is significant. That means that only the information, which is transmitted by electrodes E1/E2 to E12 evokes a different pitch sensation in the brain. However, a higher number of independent channels in pitch would improve the transmission of spectral fine structures in speech and music signals.

As the COMBI 40+ electrode array design provides only 12 electrodes, different methods of increasing the number of independent channels - which is equal to the number of different pitch percepts – were examined. The pitch sensation evoked by electrical stimulation is either influenced by place or rate of stimulation as it was observed in experiment 2. The results of this experiment have shown that an increasing stimulation rate up to about 300 pps evokes an increasing pitch sensation. However, the changes in pitch height are accompanied by changes in sound quality up to 566 pps (experiment 3, page 30). Furthermore, experiment 4 (page 36) reveals that base rate changes of more than 25% are necessary to elicit a just noticeable pitch change for stimulation rates below 300 pps.

2. Design of the RateCIS strategy

The results of experiments 1 to 6 were applied to modify the CIS strategy, which will be referred to as the RateCIS strategy. The main goal was to increase the number of independent pitch channels. This was realized by switching the stimulation rate for a selection of electrodes between two fixed pulse rates: the normal CIS stimulation rate of 1515 pps and a lower stimulation rate of 252 pps. The lower stimulation rate was chosen according to the results of experiments 2 (page 24) and 3 (page 30) with the goal to effectively change the perceived pitch elicited by the selected electrode but not to reduce the sound quality by a distinct amount. This way, it can be assumed that a kind of increasing pitch perception between the electrodes could be determined. Namely, the stimulation of electrodes with increasing electrode number evokes an ascending pitch perception; for each of the selected electrodes when stimulated at the low stimulation rate, an additional pitch height might be evoked. Ideally, the so created pitch height would be in the range of the pitch heights in between this electrode and its apical neighbor.

As the results of experiment 6 (page 51) show that electrode E1 and E2 evoke the same pitch perception for five out of six subjects, the stimulation rate of E1 was fixed at 252 pps and the stimulation rate of E2 was fixed at 1515 pps in order to allow a different pitch perception for those electrodes. The difference in pitch between E1/E2 and E3 was small in comparison to the pitch difference between the other electrodes used in experiment 2. Therefore, electrode E3 was also stimulated at the fixed rate of 1515 pps.

The low stimulation rate of 252 pps was achieved by down-sampling. Within a CIS cycle the selected electrode was not activated for a fixed number of times. To reduce the stimulation rate from 1515 pps to 252 pps, the selected electrode is only stimulated in every sixth cycle. If all available electrodes were selected to switch between the low and high stimulation rate, a situation could occur where for five CIS cycles no electrode would be stimulated. This effect should be avoided because it might result in an audible break of the signal followed by a sensation comparable to the switch-on of a signal within the transmission of a stimulus. To prevent this undesired interruption of stimulation only six out of twelve electrodes were selected for pulse rate switching in the RateCIS strategy, namely E4, E5, E6, E7, E8 and E9. Considering the stimulation order of the classical CIS strategy (see chapter III.1., page 64) and the fact that the stimulation rate of electrode E1 is fixed to the low pulse rate, this means that maximally two adjacent electrodes are not stimulated for five consecutive CIS cycles (E1 and E7, E9 and E4).

Concerning the analysis of the signal prior to the stimulation of electrodes, only a few changes were introduced into the classical CIS processing in order to point out mainly the effect of pulse rate switching. The main change was to increase the number of band pass filters. In the CIS strategy employed for the COMBI 40+, there are 12 band pass filter which are associated with 12 electrodes. In the new RateCIS strategy the number of band pass filters is increased to 18 (overview over new filter bank in Fig. 25(b)). This way, each of the electrodes with switching pulse rate is associated with two of the band pass filters. The association of electrodes, stimulation rate and band pass filter is shown in Table II.

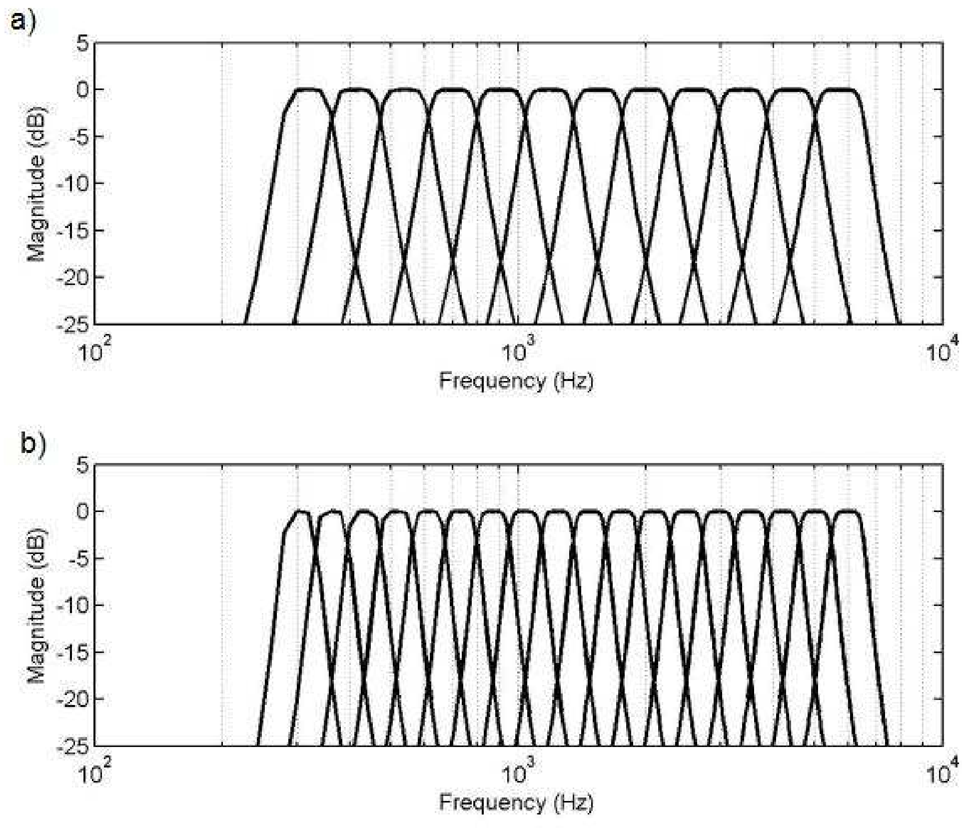

FIGURE 25. *Arrangement of band pass filters for the two different speech coding strategies as a function of frequency.* (a) *Filter bank with 12 band pass filters of the CIS*, (b) *filter bank with 18 band pass filters of the RateCIS strategy.*

The results of the experiments 2 (page 24) to 4 (page 36) on rate pitch perception have shown that the loudness of the stimuli changes with stimulation rate. Therefore, the amplitude of the current pulse is calculated based on two different measurements of maximum comfortable level and threshold level namely for the high stimulation rate and for the low stimulation rate.

TABLE II. *Allocation of band pass filters to electrodes. Line 1 represents the 12 available electrodes, line 2 the band pass filter. In line 3 it is noted, whether the electrode has a switching (‘?<>?’ notes a decision whether the energy in one ore the other band pass filter is higher) or fixed stimulation rate. Line 4 shows the stimulation rate for each electrode depending on the decision in line 3.*

Electrode	**E1**	**E2**	**E3**	**E4**		**E5**		**E6**	
Filter Nr	1	2	3	4	5	6	7	8	9
Decision	no	No	no	?<>?		?<>?		?<>?	
Rate	252	1515	1515	252	1515	252	1515	252	1515

Electrode	**E7**		**E8**		**E9**		**E10**	**E11**	**E12**
Filter Nr	10	11	12	13	14	15	16	17	18
Decision	?<>?		?<>?		?<>?		no	No	no
Rate	252	1515	252	1515	252	1515	1515	1515	1515

The stimulation rate for each electrode is analyzed for any stimulation in the CIS cycle. For example, E4 is associated with filter number 4 and 5. For any stimulation in the stimulation cycle, the signal energy in band pass filter 4 is compared with the signal energy in band pass filter 5. If the signal energy in band pass filter 4 is higher, E4 is stimulated at the low pulse rate; if the signal energy in band pass filter 5 is higher, E4 is stimulated at the high pulse rate. As this analysis is done after each finished stimulation cycle (for the high rate in the consecutive cycle, for the low rate in the sixth consecutive cycle as it remains inactivated for six cycles), spectral changes in the band which is including band pass filter number 4 and 5 are transmitted to the electrode. This way, the information of 18 band pass filters is transmitted to 12 electrodes and small changes in the signal which occur between two adjacent filter bands over time are transmitted.

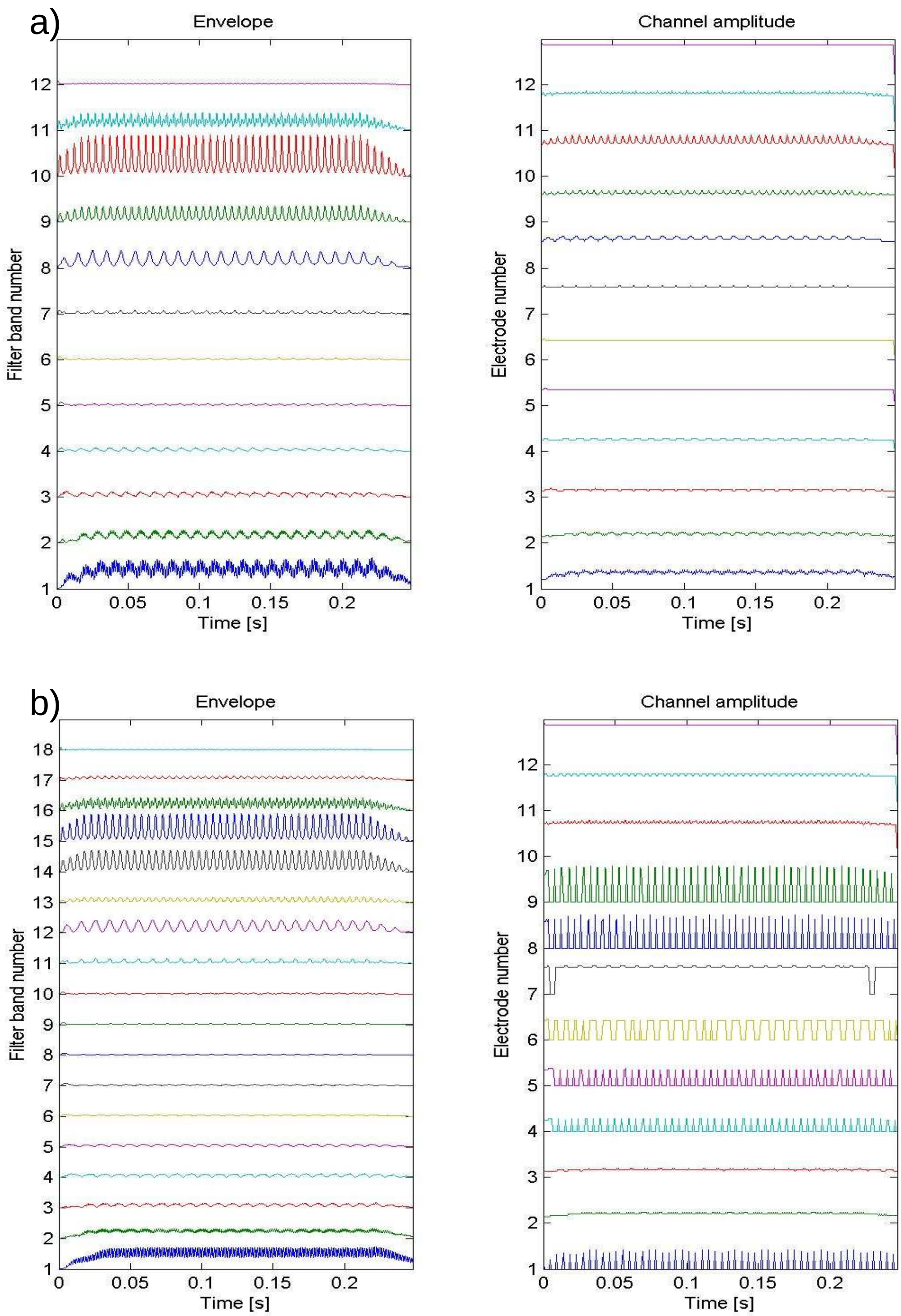

FIGURE 26. *The synthesized vowel 'i' with a fundamental frequency of 100 Hz according to Fant (1970).* (a) *Amplitudes in the 18 band pass filters of the RateCIS strategy*, (b) *resulting stimulation of the 12 electrodes with the current mapped for subject S7.*

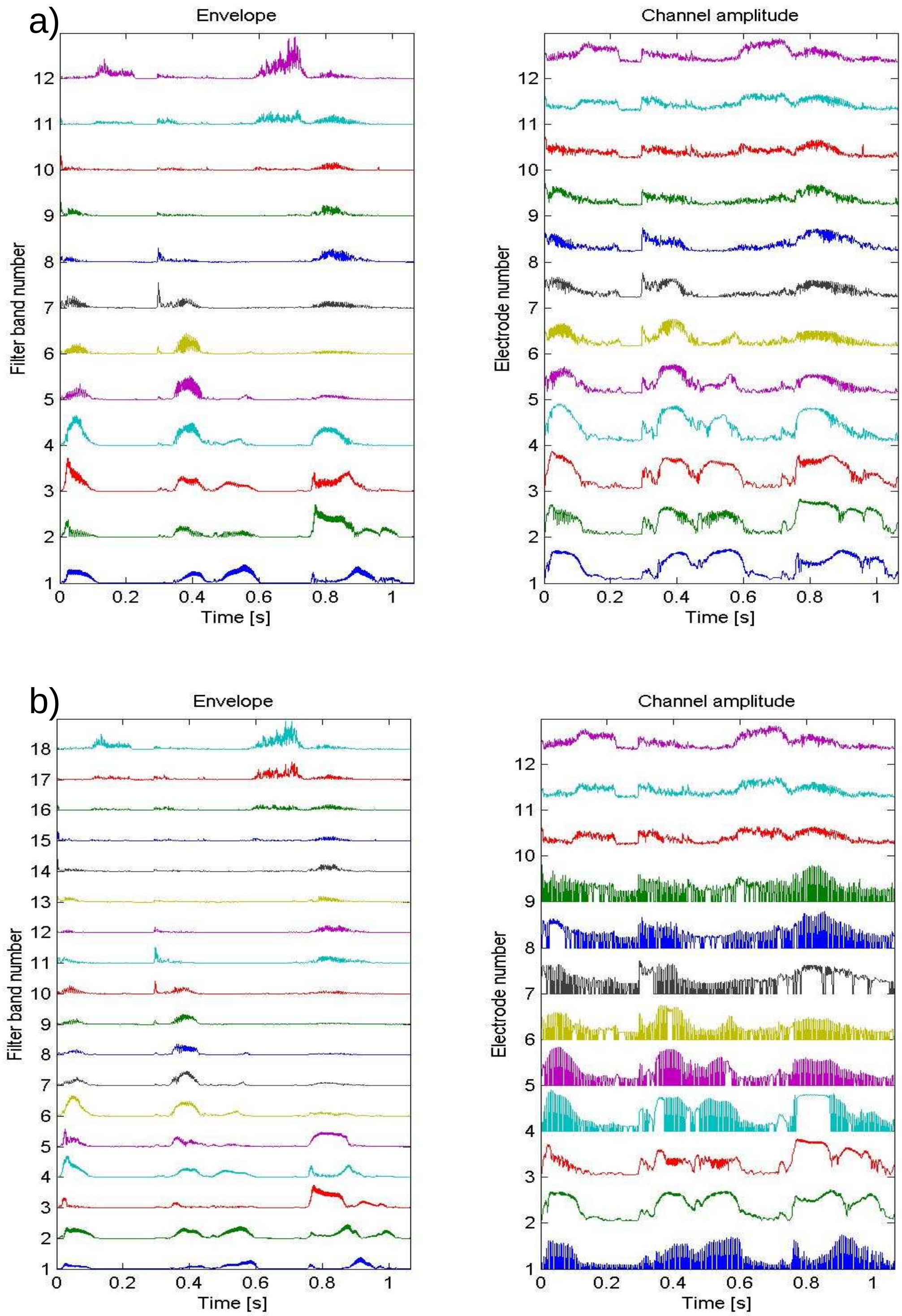

FIGURE 27. *The German word 'Karussel' by a female speaker over time.* (a) *Amplitudes in the 18 band pass filters of the RateCIS strategy*, (b) *resulting stimulation of the 12 electrodes with the current mapped for subject S7.*

Figures 26 and 27 show a band pass filtered signal and the resulting channel amplitudes with the CIS (Fig. 26/27(a)) and the RateCIS strategy (Fig. 26/27(b)). The incoming signal in Fig. 26 is the vowel ‘i’ synthesized by a model according to Fant (1970) with three formant frequencies (220 Hz, 2200 Hz, 3300 Hz) and a fundamental frequency of 100 Hz. The incoming signal in Fig. 27 is the German word ‘Karussel’ pronounced by a female speaker. In both figures, there is a clear difference in channel 1 which is constantly stimulated at the low stimulation rate with RateCIS. The formant frequencies of the vowel ‘i’ are analyzed by different filters for CIS and RateCIS. The second formant frequency (2200 Hz) is analyzed by the filter allocated to E8 with CIS, the third formant frequency is analyzed by the filter allocated to E10 (Fig. 26(a) left side). Therefore, the channel amplitudes of electrodes E8 and E10 are higher in comparison to the other channel amplitudes. With the RateCIS strategy, the second formant frequency of the vowel ‘i’ (2200 Hz) is analyzed by the lower filter of E8 (filter band 12, see also Table III, page 88) and causes a low stimulation rate for E8 with RateCIS. Consequently, the second formant frequency also causes a higher energy level in the higher filter allocated to E7, and therefore E7 is stimulated at the high rate. The third formant frequency is analyzed by the lower filter of E9 (filter band 14) and causes a low stimulation rate for E9 with RateCIS. In Fig. 27(b) it can be observed that within one pronounced word (the German word ‘Karussel’) all of the selected electrodes for pulse rate switching (E4, E5, E6, E7, E8, E9) are stimulated at the two stimulation rates in contrast to the constant stimulation rate for those channels which have fixed stimulation rate (E1, E2, E3, E10, E11, E12).

IV. TESTS WITH THE RateCIS STRATEGY

The RateCIS strategy was designed in order to enlarge the presentation of the spectral information of the incoming signal. This was achieved by allowing a pulse rate switching for several electrodes. The more detailed representation of spectral information might improve the speech perception, the detection of melody contours and the sound quality of speech and music. The effect of the introduced pulse rate switching was tested in a preliminary experiment on the discrimination of pure tones with different frequencies. The RateCIS strategy was then tested for speech perception and speech and music appreciation.

1. Participants

In order to create comparable conditions on what concerns the number of stimulated electrodes, the subjects were chosen based on the fact that all 12 electrodes were available for stimulation and placed intracochlear. That way it was possible to test the complete idea of the RateCIS strategy which is based on band pass filtering with 18 channels combined with a stimulation of 12 electrodes. Out of the sixteen subjects who participated at all hearing experiments, nine subjects (S1, S5, S6, S7, S10, S12, S15, S16, S17) fulfilled the selection criteria of 12 electrodes available for stimulation and were available for testing the RateCIS strategy. Furthermore, their stimulation parameters allowed a fixed minimal pulse width of 26.7 µs for all subjects and electrodes.

Concerning the rest of the subject group, S2 had a short circuit between electrodes S7 and S8 which were therefore evoking the same pitch perception; S3 and S11 had electrodes placed outside the cochlear or evoking unpleasant pitch perceptions; S13 had two electrodes in the middle of the electrode array switched off because they were evoking facial stimulation. S4, S8 and S14 were no longer available for testing.

2. Verification of the pulse rate switching

In order to test the idea that a pulse rate switching at one electrode would result in a recognizable pitch difference, a preliminary experiment tested the discrimination of two sequentially presented pure tones. The pure tones were chosen in a way that their frequencies were the center frequencies of two adjacent bands of the modified RateCIS filter bank belonging to the same electrode. This way, it was tested whether the consequently switching pulse rate at the respective channel was resulting in a discriminable different pitch perception.

a) Method

The RateCIS strategy was tested for the pitch discrimination of two pure tones with different frequencies. The frequencies of the pure tones were chosen to be the centre frequencies of two band pass filters of the RateCIS strategy which are associated with the same electrode (for example the centre frequencies of band pass filters 8 and 9 associated with electrode E6, see Table II., page 69). The two stimuli were presented in a three interval test. The subject's task was to indicate whether the pitch in the second interval was higher or lower than the pitch in the first and third interval by pressing on a button on a touch screen. No feedback was given. Six pairs of pure tones with frequencies which were the centre frequencies of two neighbored band pass filters - associated each with one of six electrodes for switching pulse rate - were tested. For each pair of sinusoids twenty comparisons were made. Attachment 3 shows an example for the electrode stimulation at the time when electrode E6 switches from the low stimulation rate to the high stimulation rate. The attachment is an extract of the control file which is provided by the software in order to verify the stimulation parameters (sequence of stimulated channels, amplitudes, current range, pulse width and minimal pulse distance, see also chapter II.2., page 12).

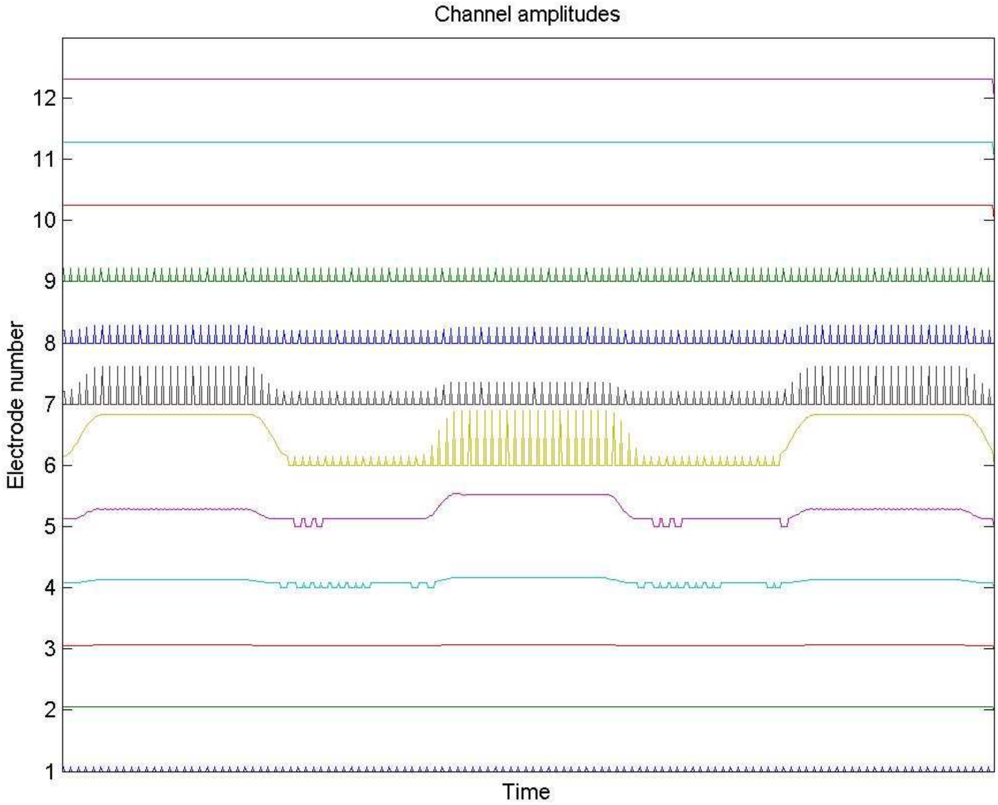

FIGURE 28. *The stimulation of the electrodes according to the RateCIS strategy for the presentation of three sinusoids (high-low-high) of which the frequencies are analyzed by the band pass filters allocated to electrode E6.*

The extract starts at a time when electrodes E1, E7, E8, E9 and E4 are in a low stimulation rate mode (amplitude 0). Electrode 6 is stimulated at a high amplitude in the first presented cycle (pulse number 2423). For the next six CIS cycles the amplitude of E6 is set to 0 (pulses 2435, 2447, 2459, 2471, 2483 and 2495). Then E6 switches to the high stimulation mode (pulse number 2507). Figure 28 shows the resulting channel amplitudes with RateCIS for a sequence of pure tones with frequencies in the centers of filter bands 8 and 9 (high-low-high). The stimulation rate consequently changes in E6. Furthermore, due to the filter band overlap there is a change in amplitude at the adjacent electrodes.

b) Results

Figure 29 shows the results in percentage of correct answers as a function of electrode with switching pulse rate. In nearly all cases the pitch height of the pure tones was judged in the right order.

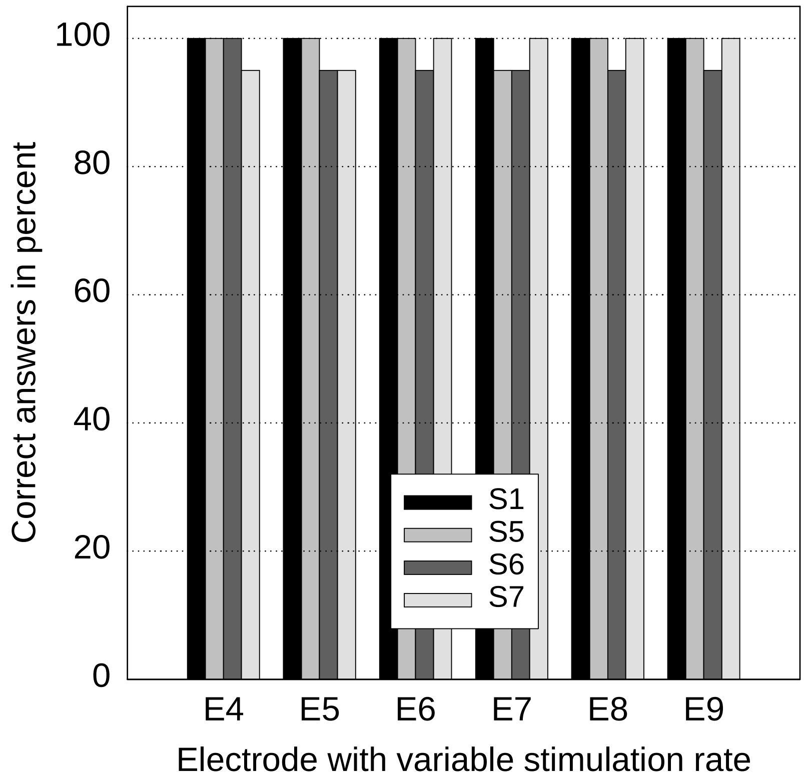

FIGURE 29. *Discrimination of pure tones with the RateCIS strategy. The scores in percent correct for a discrimination of two pure tones with different frequencies are plotted as a function of electrode with switching pulse rate for four subjects.*

Only one incorrect answer per subject was given which means that the pitch height of the presented pure tones was discriminated significantly at each of the six electrodes with pulse rate switching (based on the confidence interval for the binominal distribution for performance better than chance, 50% correct).

c) Discussion

The new RateCIS strategy is analyzing the presented frequencies of two pure tones in a way that the stimulation rate of that electrode switches which is associated with the band pass filters analyzing the two frequencies. The experiment has shown that this pulse rate switching at one electrode causes a discriminable pitch perception and allows the discrimination of the two pure tones. The new RateCIS strategy is analyzing the same band width between 300 and 7000 Hz as the CIS strategy. Divisions of this band width into 18 filter bands for the RateCIS strategy instead of 12 filter bands for the CIS strategy results in a narrower filter band width for the RateCIS strategy. That means that the discrimination of two pure tones with the CIS strategy is possible only for a larger difference between the frequencies of the pure tones if one is assuming that only frequencies which are analyzed by different filter bands result in a discriminable pitch difference. Therefore, the frequency discrimination of pure tones should be improved with the new RateCIS strategy (18 bands) in comparison to the classical CIS strategy (12 bands). This effect enlarges the presented spectral information for all incoming signals as speech and music.

3. Speech recognition tests

a) Method

Different measures of speech recognition were conducted to compare the classical CIS strategy and the new RateCIS strategy. All speech tests were conducted with the Research Interface Box. The test of an online processing and live speech modus was not possible because the design of the Research Interface Box limits the online processing by a slow serial RS232 connection to the host PC. Furthermore, the restricted memory capacity of the Research Interface Box restricts the duration of speech or music signals to maximally 4 s. The speech material was preprocessed using the patients’ fitting maps for both strategies which were adjusted to comfortable loudness for speech presentation and then presented by direct stimulation of the electrodes.

First, Freiburger polysyllables (numbers) were tested in an open-set test (Hahlbrock, 1970). A training list was performed before testing two lists containing 20 numbers for data collection. This procedure was repeated for each strategy. Second, an open-set sentence test (‘HSM’) was tested. The HSM test consists of everyday German sentences like ‘Warum mußt Du immer rauchen’ (‘Why do you always smoke’) or ‘Meine Batterie ist leer’ (‘My battery went empty’) with one list containing 20 sentences and a total of 106 words (Schmid, 1997). Scoring was based on the total number of correctly identified words in two lists. Before testing each strategy, one training list was performed with the same strategy. Both tests were recorded from a trained male speaker. Third, polysyllables were tested in an unofficial open-set test. The polysyllables were collected from the speech training material of ‘Hören-Sehen-Schreiben’ (Träger, 2001) and were spoken by a female speaker. The tested words were randomly chosen out of 40 words. For each speech strategy five words were presented as training before testing and the recognition of 15 words was evaluated for data collection. The first strategy which was tested with the speech tests was chosen randomly for the nine subjects in order to avoid any learning effect. The subjects were not informed about the strategy they were listening to and only got the information that there were two different programs which they should compare.

An additional questionnaire (see attachment 4) was filled out after testing to assess the subjective speech quality of both strategies. The questions included the clarity, sonority, brightness, naturalness, intonation and the general impression. The last question asked which of the strategies the subject would prefer for speech recognition in the daily life.

b) Results

Figure 30 shows the results of the three different speech tests. The scores are given in percentage of correct recognized words and are averaged for all nine subjects. The different colors show the results for the two different strategies. The results for the Freiburger numbers and Polysyllables by a female speaker reach a very high average score of speech recognition of over 90% for both strategies. For the more difficult HSM sentence test, there is a difference between the CIS (92.2% in average) and RateCIS strategy (84.6% in average). The difference is significant (t-test, $p<0.01$).

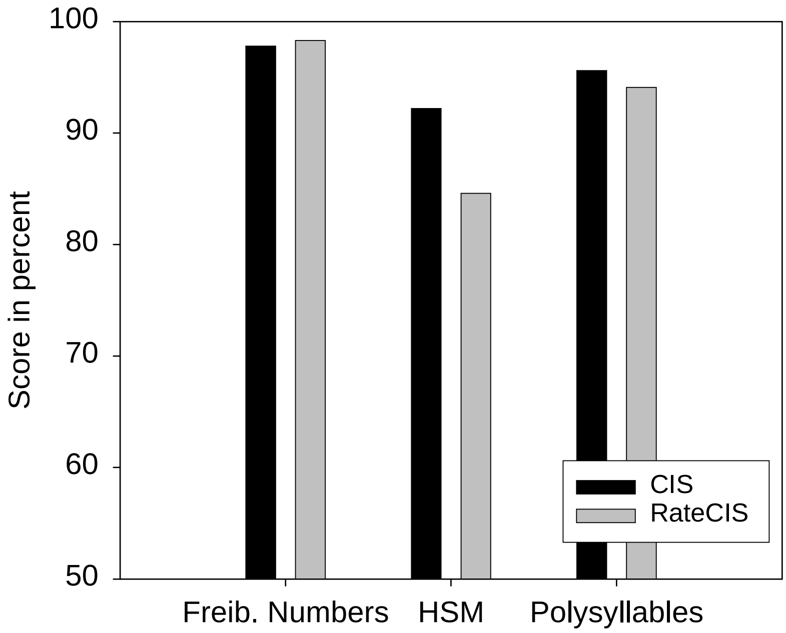

FIGURE 30. *Scores of the speech tests. The percent correct of the Freiburger numbers, HSM sentence test and polysyllables are plotted.*

Figure 31 shows the detailed results of the HSM for all nine subjects. There are six subjects with no significant difference between the two strategies for the HSM (S1, S6, S10, S15, S16, S17) and three subjects with a significant difference between the two strategies (S5, S7, S12). All of those subjects score higher with the CIS strategy than with the RateCIS strategy.

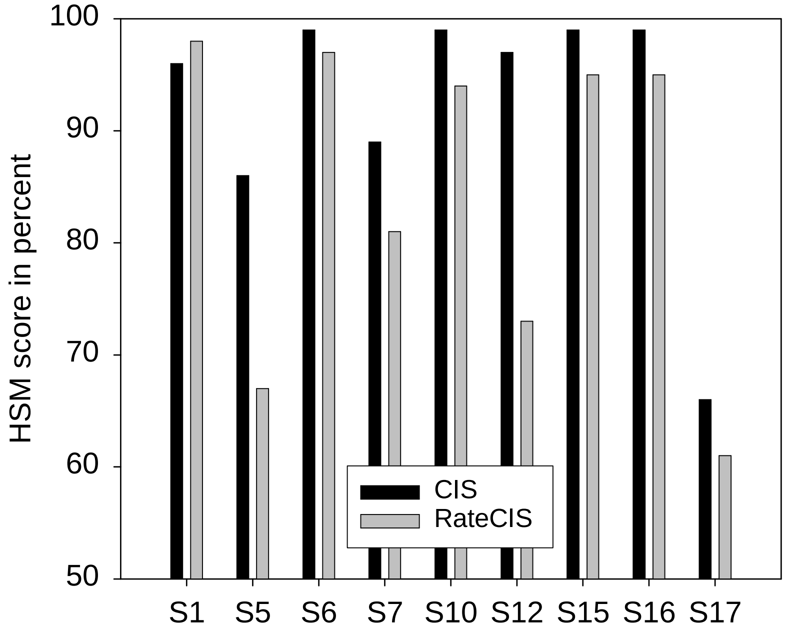

FIGURE 31. *The individual results for the HSM sentence test in percent correct; the parameter is speech strategy.*

Figure 32 shows the averaged ratings for the different test items of the questionnaire. Each of the topics could be scaled between 1 and 7. For all topics ‘1’ indicates a minimal answer, that means for example, no clarity (distortion), no brightness (especially low timbre), no naturalness (unnatural); ‘7’ indicates a maximal answer, that means for example, very pleasant, very good impression, but also very much effort to understand. The different colors show the averaged ratings for the two strategies. For the items *pleasance*, *general impression male speaker*, *brightness male speaker, naturalness* and *effort* there is no difference between the two strategies. However, the subjects rate the sound of the CIS strategy as slightly clearer than the sound of the RateCIS strategy and for a female speaker the general impression is slightly better and the timbre slightly brighter. The CIS strategy seems to transfer more information on the intonation of speech than the RateCIS strategy. Nevertheless, all differences between the ratings of the two strategies are very small and not significant (t-test, $p<0.05$).

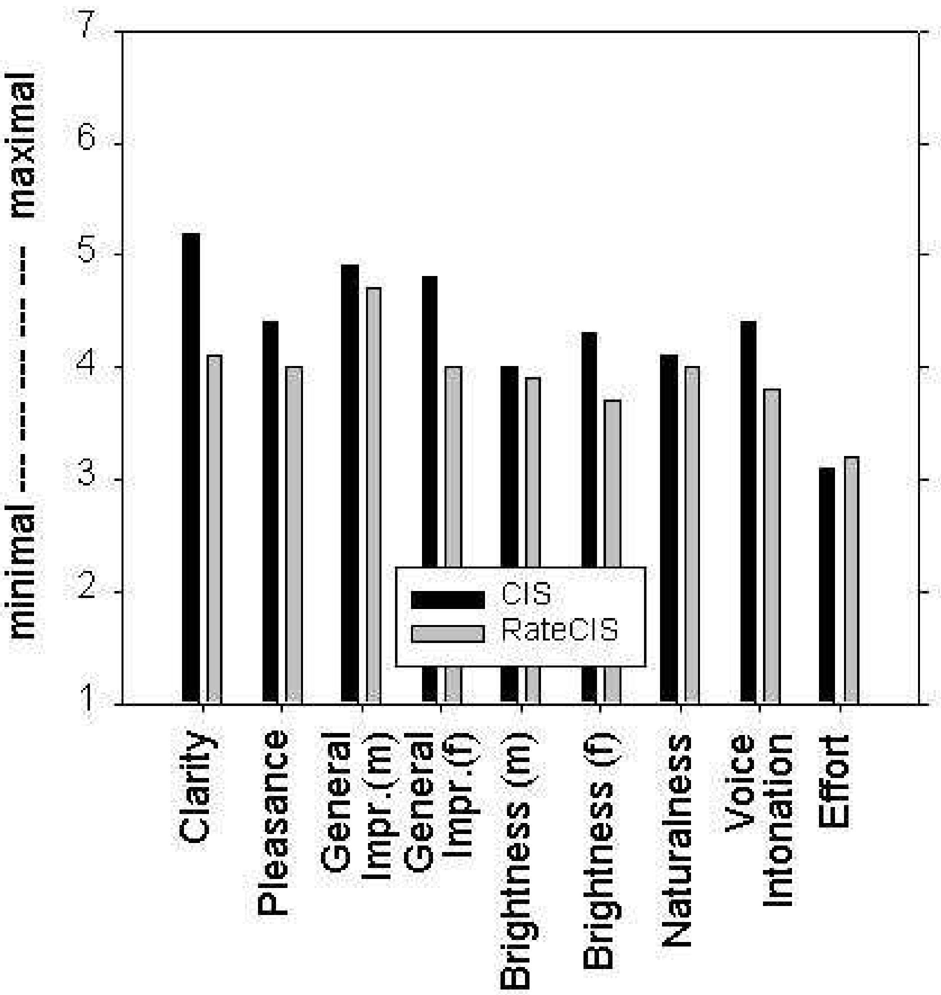

FIGURE 32. *Average subjective rating of the sound quality of speech for different categories between 1 (minimal impression) and 7 (maximal impression).*

4. Music tests

a) Method

The music tests consisted mainly of the presentation of musical sounds. All musical sounds were taken from the listening training material of 'Hören-Sehen-Schreiben' (Träger, 2001) and were recorded live. First the sounds of four different instruments were presented randomly either with the CIS or with the RateCIS strategy. The sounds of the instruments consisted of two tones for flute, piano and xylophone and of a little melody for the organ. The subjects were asked to identify the instrument and to try to recognize sound differences between the two strategies. In addition, the answer of the subject and the described subjective sensation were noted for further evaluation. Then six short melodies (German songs for children) were presented. The melodies had to be shortened to a total duration of 4 s due to hardware restrictions. Therefore only the first notes of each melody could be presented. The melodies were sequentially presented with the two strategies whereby the strategy for the first presentation of each melody was randomized. The subjects were asked to identify the melody which was a very difficult task due to the restricted duration. Additionally, they should try to recognize sound differences between the presentation with the CIS and the RateCIS strategy. After the presentation of the musical sounds with both strategies, a questionnaire was filled out (see attachment 5). The questionnaire included questions about the recognition of melody and rhythm with both strategies and questions about the sound quality like clarity, sonority, brightness, intonation and the general impression.

b) Results

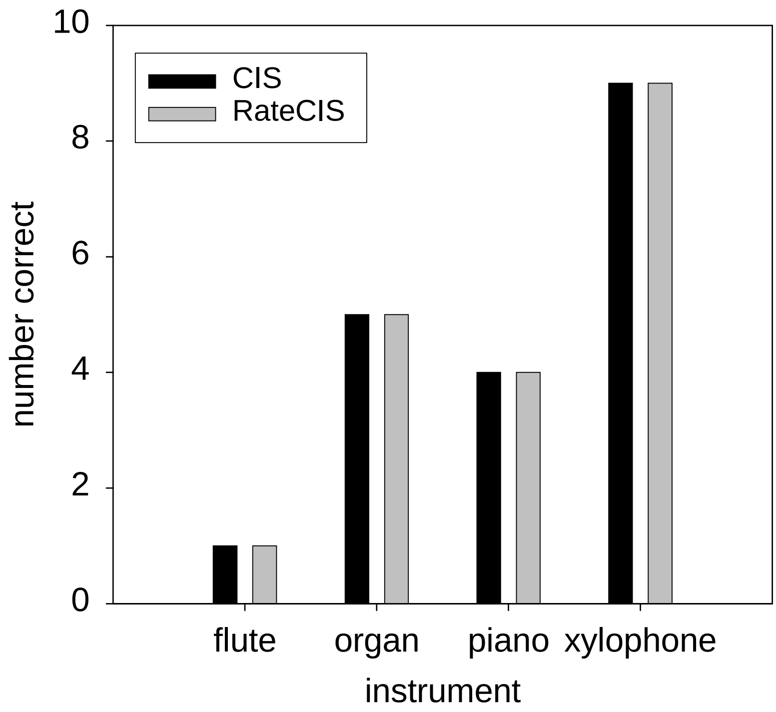

FIGURE 33. *Recognition of instruments. The number of subjects who were able to recognize each of the instruments is given for the CIS and the RateCIS strategy.*

Figure 33 shows the results of the instrument recognition test. The number of subjects which have recognized the instrument correctly is plotted for both strategies and all instruments.

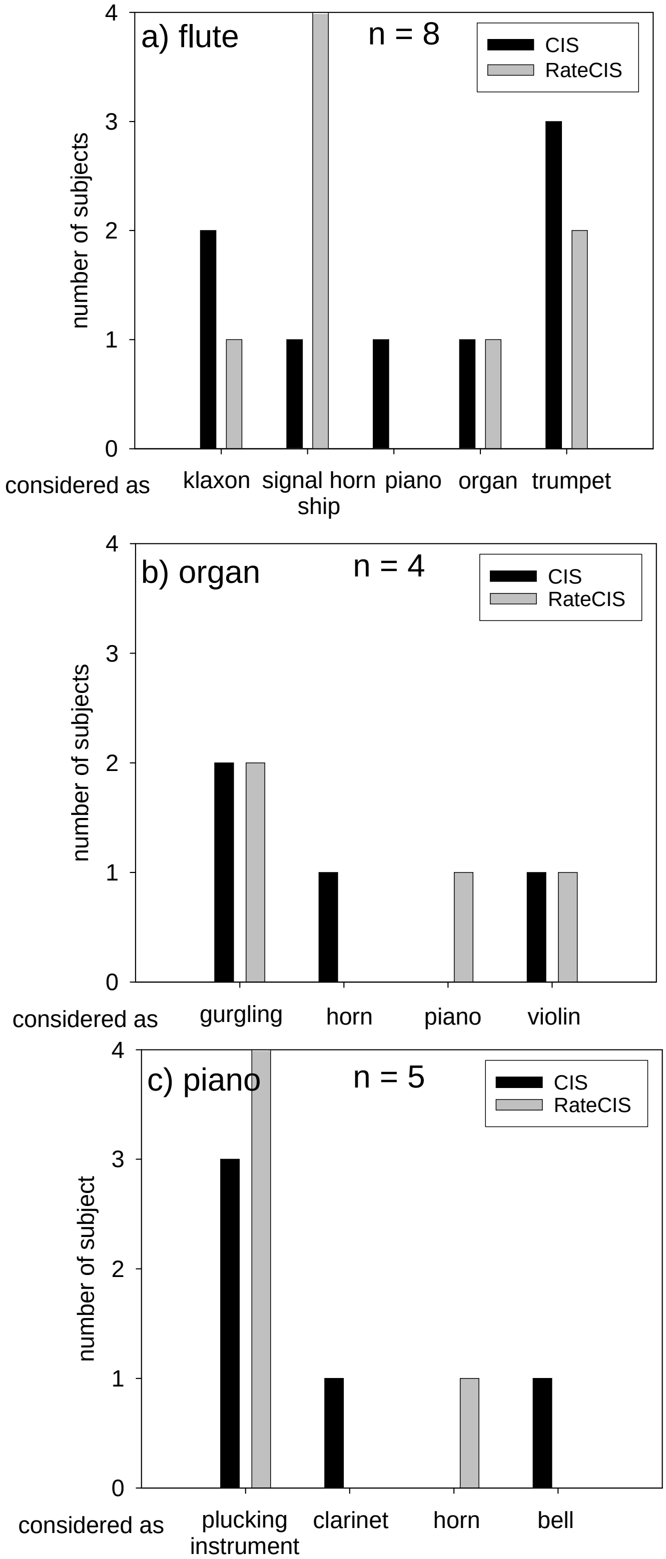

FIGURE 34. *Instrument confusions.*

The xylophone was recognized by all nine subjects, the organ only by five subjects and the piano by four subjects. The flute was the most difficult instrument to recognize and was identified correctly by only one subject. Some of the subjects could not at all imagine what kind of instrument was played and some gave wrong answers.

Figure 34 presents the given wrong answers for those three instruments which were not recognized by some subjects. The flute was considered as a klaxon, signal horn of a ship or trumpet by most of the subjects, some also considered it as piano or organ. A signal horn of a ship was named mostly for the RateCIS strategy, the klaxon and trumpet mostly for the CIS strategy. The organ was mostly considered as a gurgling sound, some subjects thought it would be a horn, piano or violin. Most subjects considered the piano to be any kind of plucking instrument like a guitar or zither, some considered it as clarinet, horn or bell.

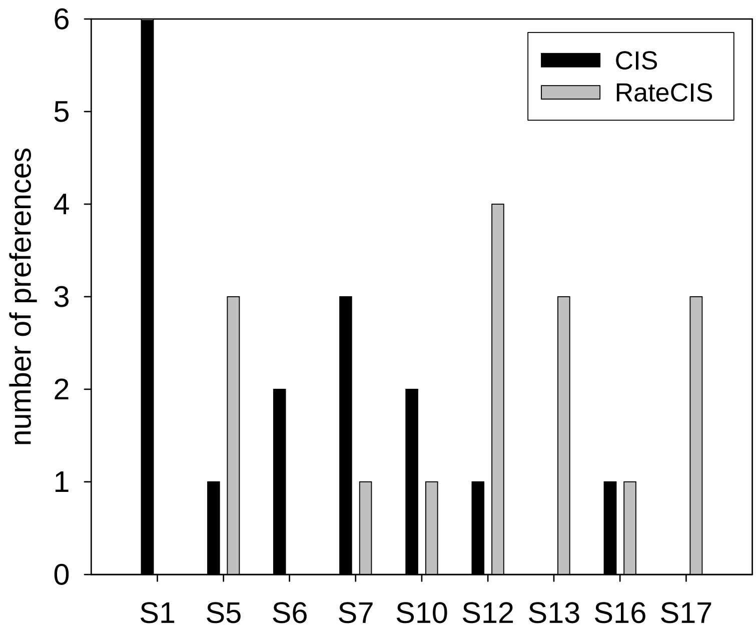

FIGURE 35. *Number of preferences of each subject for the presentation of six short melodies with the CIS and the RateCIS strategy.*

After the presentation of each melody with the two strategies, the subjects were asked whether they preferred the sound of the first or the second strategy. Figure 35 shows the number of preferences for each subject. In some cases, none of the strategies was preferred and therefore only a reduced number of ratings is shown for some subjects. Some subjects had clear preferences as S1 who preferred the CIS strategy for all melodies. Other subjects had only in some cases preferences like S16 who preferred one melody with the CIS strategy and a second melody with the RateCIS strategy, for the other melodies there was no preference.

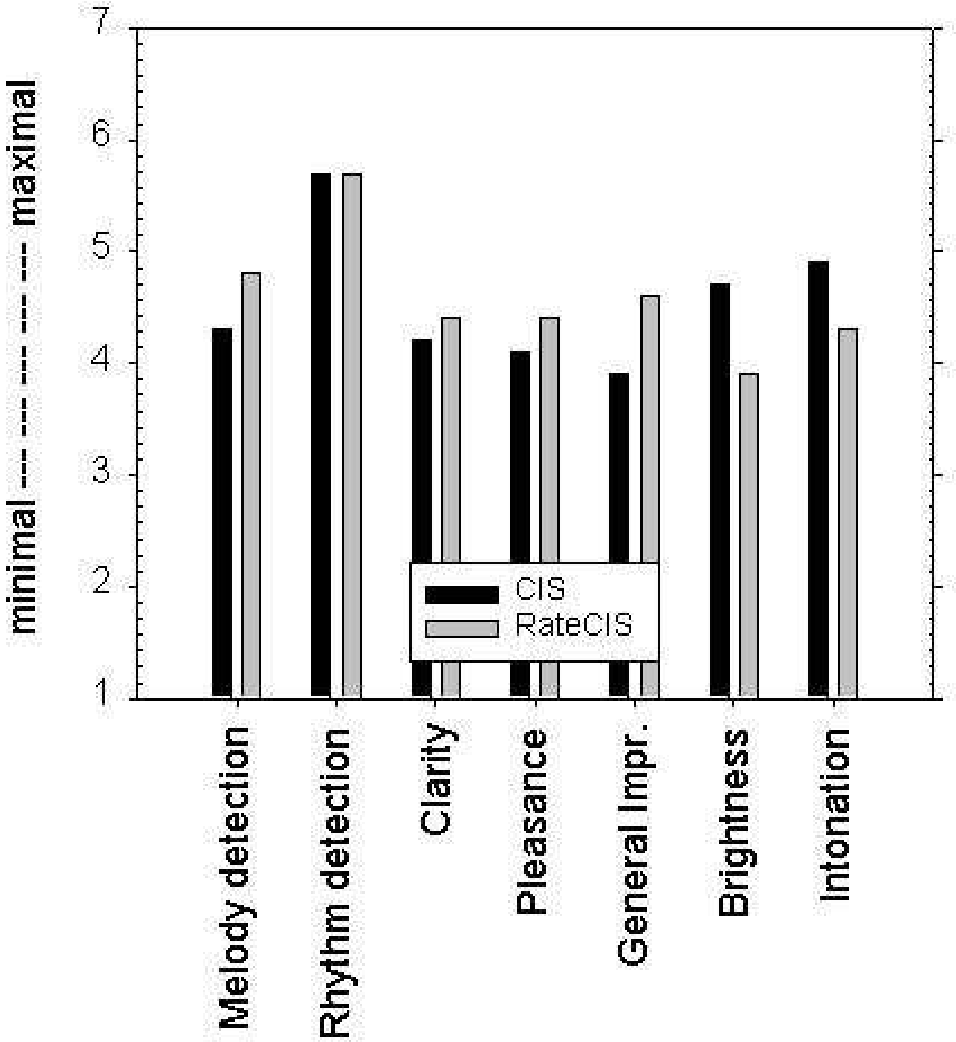

FIGURE 36. *Average subjective rating of the sound quality and impression of music for different categories between 1 (minimal impression) and 7 (maximal impression).*

In total, four subjects preferred more melodies when presented with the CIS strategy, four subjects preferred the melodies when presented with the RateCIS strategy and one subject found no difference.

The ratings of the different items for the appraisal of music in the questionnaire concerning sound quality and preferences are shown in Fig. 36. The ratings are scored from the minimal answer ‘1’ to the maximal answer ‘7’ for each of the items. The averaged ratings over all nine subjects are shown for both strategies. There are only small and not significant differences (t-test, $p<0.05$) in the ratings between the two strategies. For *rhythm detection, clarity* and *pleasance* there is hardly any difference. The RateCIS strategy was preferred for the detection of a melody and for the item of the *general impression* of the sound. However, the CIS strategy was preferred for the intonation of the melody and was judged as having a brighter timbre.

5. Discussion

The new RateCIS strategy was designed in order to give a more detailed representation of the spectral information of the incoming signal. The expectation was that the sound quality of speech and especially music would improve. Figure 37 shows the total preference of the subjects. The subjects were asked which of the strategies they would prefer in daily life considering speech perception and music approval. Concerning speech perception six of nine subjects preferred the CIS strategy; however, for the perception of music six subjects preferred the RateCIS strategy. In summary, the results show that there is hardly any difference between the two strategies but some subjects tend to prefer a certain strategy for a certain sound. It seems as if the preferences can not be generalized for all subjects but are very individual.

The stimulation of some electrodes with a lower stimulation rate than the normally used CIS-stimulation rate results mainly in a lower timbre of speech and music. This was reported by the subjects during the experiments and is also shown in the ratings of the questionnaires. Interestingly, this effect is not present for the perception of musical sounds where no difference in terms of clarity between CIS and RateCIS was observed. It seems as if the effect of a rate reduction is less disturbing for the perception of music than for speech perception. As a result of experiment 3 (page 30), the stimulation of electrodes with a lower stimulation rate below 566 Hz results also in a more buzzing sound quality. This is noted in the result of the questionnaire where speech is rated much clearer with the CIS than with the RateCIS strategy.

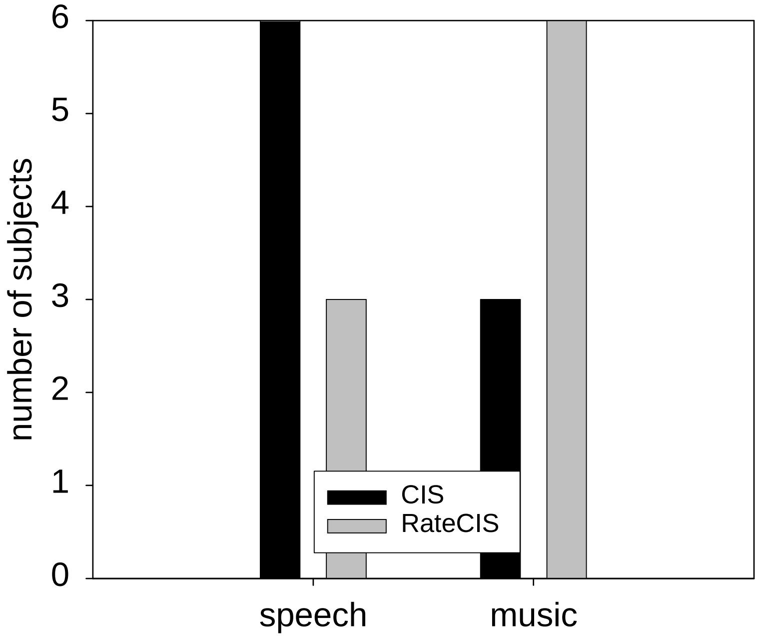

FIGURE 37. *Number of subjects which preferred the CIS or RateCIS strategy for the perception of speech and music.*

Concerning the general impression, there is a preference for the CIS strategy when processing speech uttered from a female speaker. Whereas speech uttered from a male speaker, shows no difference in terms of the general impression. It seems as if mainly the brightness of the speech causes a difference in the ratings of the general impression for the female speaker utterances because for the male speaker, there is no difference in the brightness and general impression between the two strategies and all other items were rated independently of the gender of the speaker. It seems as if the different allocation of the band pass filters and the smaller bandwidth per band in the RateCIS strategy is affecting the sound of speech signals and that this effect is more noticeable for female than for a male voices. Increasing the number of band pass filters for a fixed band width of the incoming signal from 12 to 18 results in a displacement of the frequency to electrode allocation (see Table III).

TABLE III. *Cut off frequencies of the band pass filters for the CIS and RateCIS strategy.*

	Electrode	**E1**	**E2**	**E3**	**E4**		**E5**		**E6**	
CIS	lower cutoff	300	390	507	659		857		1115	
	higher cutoff	390	507	659	857		1115		1449	
RateCIS	lower cutoff	300	357	426	507	604	720	857	1021	1217
	higher cutoff	357	426	507	604	720	857	1021	1217	1449

	Electrode	**E7**		**E8**		**E9**		**E10**	**E11**	**E12**
CIS	lower cutoff	1449		1884		2450		3185	4141	5384
	higher cutoff	1884		2450		3185		4141	5384	7000
RateCIS	lower cutoff	1449	1726	2056	2450	2918	3476	4141	4933	5876
	higher cutoff	1726	2056	2450	2918	3476	4141	4933	5876	7000

This is mainly noticeable for the most basal electrodes E10 to E12 which are stimulated at a fixed stimulation rate of 1515 pps in either the classical CIS or the RateCIS strategy. The frequencies analyzed by E10 are shifted up for the RateCIS strategy (filter band with cutoff frequencies 4141 and 4933 Hz) in comparison to the CIS strategy (filter band with cutoff frequencies 3185 and 4141Hz). This shift causes a presentation of signal components at the same electrode place with a lower frequency for the CIS strategy than for the RateCIS strategy which results in the perception of a brighter timbre.

The effect of a perception of brighter timbre with the CIS strategy is also observed for music perception. In contrast to the rating of the general impression of female voices, this test item is considered better for the RateCIS than for the CIS strategy. Probably, a lower timbre is perceived as being more natural for music than for the perception of female voices.

In addition to the brightness of the sound there is one other aspect which is reported for both speech and music perception. The intonation of speech and music is not increased by the adaptive pulse rate switching in the RateCIS strategy. On average, speech and music were rated as being more intonated when presented with the CIS strategy. Some of the subjects had a problem with the content of this question for intonation of speech or music. It is possible that those subjects were rating the amount of emphasizes more than the intonation in terms of the amount of melodic contours in speech or a higher amount of different tones for music. If this is the case, then the difference between the two strategies can be explained by this means: in the CIS strategy, there is a constant stimulation of each electrode and therefore a very fast and constant transmission of amplitudes in each band pass filter. With the RateCIS strategy, some electrodes are stimulated at a lower stimulation rate depending on the incoming signal. The consequently longer time distance between stimulations might result in a softer perception of the sound because there is less superposition of the stimulation amplitudes in the CIS cycle. This might result in a weak transmission of the loudness when several channels are stimulated at a lower pulse rate and cause a less emphasized speech and music perception.

The individual ratings of speech and music presented in attachments 6 and 7 support the impression that the preference for one or the other strategy is very individual. For speech perception and quality subjects S1, S7, S10, S12, S16 and S17 preferred CIS and subjects S5, S6 and S15 preferred RateCIS. Concerning the appraisal of music, subjects S1, S7 and S10 preferred CIS and subjects S5, S6, S12, S15, S16 and S17 preferred RateCIS. The preference of CIS for speech sounds can be explained by the fact that all of the subjects were well experienced users of the CIS strategy. Although the processing of the CIS strategy with the Research Interface Box and the ‘Matlab’® code differs in some points from the actual CIS strategy used in the most recent version of the Tempo+ speech processor, most subjects did not experience extremely large differences to their normal listening program. And this despite the fact that they were not used to the fitting as a daily used listening program because the adjustments of the single electrodes were only based on comfortable loudness and threshold levels.

TABLE IV. *Disabled electrodes and total band width of the subjects' regular speech processors fitting.*

Index	Electrodes switched off for normal use	Analyzed bandwidth for normal use
S1	E1, E2	350-5500
S2	E7	300-7000
S3	E12	300-8500
S4	E1, E2, E3	300-8500
S5	E1, E2	300-5500
S6	-	300-7000
S7	-	300-7000
S8	-	300-5500
S10	-	300-7000
S11	E2, E3, E12	300-7000
S12	-	300-7000
S13	E5, E6	300-8500
S14	-	300-7000
S15	-	300-7000
S16	-	300-7000
S17	E2	300-5500

Furthermore, differences occurred for those subjects for whom mostly one or two of the most apical electrodes were disabled in their regular processor fitting due to an unpleasant sound quality (S1, S5, S17, see Table IV). Differences also occurred for those subjects who were using a different totally analyzed band width (S1, S5, S17, see Table IV). It seems certain that for those three subjects the sound differences in comparison to their daily used speech processor fitting were noticeable for both strategies. The individual preferences however show that these effects were not dominating the decision criteria for one or the other strategy as the ratings of those three subjects are very different.

Three subjects (S1, S7, S10) preferred the CIS strategy for both speech and music. It is possible that for those subjects the pulse rate switching at a selection of electrodes was perceived as an extremely changing pitch perception or as an extremely reduced sound quality. Regarding the influence of pulse rate changes on pitch perception (experiment 2,

Fig. 8, page 27) there is a large influence of pulse rate on perceived pitch height for subjects S7 and S10, whereas for S1 there is only a small change of perceived pitch height depending on stimulation rate. Concerning the influence of pulse rate switching on sound quality (experiment 3, Fig. 11, page 32), there is a change of sound quality with stimulation rate at the basal electrodes for S1 but only up to a stimulation rate of 238 pps which is not influencing sound perception for a stimulation rate of 252 pps in the RateCIS strategy. For subject S7, there is no change in the sound quality with changing stimulation rate; for subject S10, however, there is a significant change of sound quality with stimulation rate up to 566 pps for E1, E3 and E7 whereby E1 and E7 are electrodes with switching stimulation rate in RateCIS. That means that the results of experiments 2 and 3 can not explain the preferences of subjects S1, S7 and S10 for the CIS strategy because they are not consistent for those subjects.

The RateCIS strategy was tested in comparison with the CIS strategy for speech and music perception. There were preferences for both strategies which can only partly be explained by the effects of pulse rate switching, the different set of band pass filters or the previously achieved results for pulse rate changes or the individual experience of the listeners who tested the strategies. In summary, the RateCIS strategy should be further evaluated because no time of adaptation could be given to the subjects. It is possible that the subjects will rate the two strategies very differently when using it in daily life. The RateCIS strategy was successful mainly concerning the appraisal of music which is an interesting step towards an especially designed music program.

V. GENERAL DISCUSSION

1. Hearing experiments

This thesis describes different experiments which were conducted in order to investigate parameters and effects of pitch perception elicited by electrical stimulation of the auditory nerve in cochlear implants. The perceived pitch height - and the way to influence it - is an important topic for the cochlear implant technique because current cochlear implant devices only provide a limited representation of the spectral fine structure in the incoming signal. Nevertheless, cochlear implants allow most patients to understand speech without lip reading. However, there are some restrictions in noisy environments and losses concerning the quality of speech sounds and especially the sound of music. Therefore, the experiments described in this thesis were conducted with the objective to find parameters to enlarge the representation of the spectral fine structure in the incoming signal.

The hearing experiments in this thesis were performed with subjects implanted with the MED-EL COMBI 40+. This device has an electrode array which allows an especially deep insertion of the stimulating electrodes into the cochlea and provides a large distance between neighbored electrodes of 2.4 mm. The first approach in this thesis was to investigate whether this large electrode distance provides discriminable pitch perceptions for all neighbored electrodes over the whole range of the electrode array. The hearing experiment 1 (page 16) showed that for the average user, the electrode distance of 2.4 mm of this device is wide enough to elicit discriminable pitch perceptions. This result corresponds to previous electrode discrimination experiments conducted with a device with a smaller electrode distance (Busby & Clark, 1996; Collins et al., 1997; Nelson et al., 1995; Pfingst et al., 1999; Tong & Clark, 1985). Those experiments showed that there are some cochlear implant users who can discriminate neighbored electrodes with an electrode distance as small as 0.75 mm (Nelson et al., 1995). For the average user, however, the discrimination performance improved with increasing distance between the compared electrodes. Therefore, an electrode distance of about 2 mm seems sufficient in order to allow the average cochlear implant user to discriminate between all neighbored electrodes.

Besides the place of stimulation – that means the location of the electrode in the cochlear - a different parameter to change the pitch perception is the stimulation rate. In experiment 2 (page 24) a scaling of the pitch height depending on rate and place of stimulation was performed. This experiment showed that changes in the stimulation rate up to 283 pps are influencing the pitch perception. The elicited pitch height increases at all tested electrodes with increasing pulse rate. Previous research reported varying upper limits of the stimulation rate to influence the pitch perception up to 1000 pps (Hochmair-Desoyer et al., 1983; Shannon, 1983; Tong & Clark, 1985; Pijl & Schwarz, 1995; Fearn & Wolfe, 2000; Zeng, 2002). For the average user however, experiment 2 confirms a limit of about 300 pps.

During the experimental run of experiment 2 a changing sound quality with changing stimulation rate was additionally reported by the participating subjects. Therefore in experiment 3 (page 30), the effect of the stimulation rate on the sound quality of the stimulus was investigated. The scaling of the sound quality in experiment 3 showed that the sound quality increases with increasing stimulation rate up to about 566 pps. In the literature there are hardly any reports about this effect. Some investigators noted that there was a reduction of the sound quality when using extremely low stimulation rates (Fearn & Wolfe, 2000; McDermott & McKay, 1997).

The scaling experiments on pitch height and sound quality confirmed in some aspects the expectations based on psychoacoustic ideas. The pitch perception in normal hearing is composed of place and temporal coding. First, depending on the frequency of the incoming signal, the traveling wave of the fluid inside the cochlea has different oscillation maxima. Those maxima occur at the apex of the cochlea for low frequencies and at the base of the cochlea for high frequencies. This oscillation activates spiral ganglion cells which are associated with the inner hair cells at different locations along the cochlea and which represent the spectral information of the signal in the auditory brainstem. A second coding mechanism consists of temporal cues. The activation of the neurons occurs in combination with the oscillating frequency of the basilar membrane. Therefore the neurons deliver information to the brain in a resembling temporal pattern as it consists in the incoming signal. The combination of both, place and temporal coding is further processed in the auditory nuclei of the brainstem.

Due to this psychoacoustic background the expectation for the experiments with changing electrical stimulation rate was that there would be differences in the influence on the pitch perception depending on the place of stimulation. It was expected that low stimulation rates would reduce the sound quality less at apical places of stimulation because the activated neurons and the related auditory nerve fibers would be tuned to low frequencies. Furthermore, it was expected that there would be less influence of stimulation rate on pitch height in the basal region of the cochlea than in the apical region of the cochlea because the neurons and related auditory nerve fibers would not be tuned to low frequencies. The results of experiments 2 and 3 have shown that the pitch height increases along the same slope at all four different places of stimulation. That means that the temporal coding of pitch height works independent of cochlea region and that there is no tuning of neurons or auditory nerve fibers related to a distinct frequency range influencing the processing of temporal information. Furthermore, the results of the sound quality scaling (experiment 3) have shown that sound quality is reduced for low stimulation rates at more basal places of stimulation in the cochlea. That means that concerning the sound quality, there is a kind of tuning of the auditory nerve fibers and the central processing in the brain because a shift of low stimulation rates to basal places causes a reduced sound quality. However, this effect is not as strong as expected because at all stimulated cochlear places, the slope of the perceived sound quality up to 566 pps was similar.

The small range of perceived pitch height when changing the stimulation rate led to experiment 4 (page 36), which investigated the just noticeable difference in stimulation rate. The results of experiment 4 showed that in comparison to normal hearing subjects, the changes in stimulation rate must be considerably larger in order to evoke a different pitch sensation. In the most interesting range where the pitch percept also changes on the subjective scale, the pulse rate difference limen amounted to about 25% of the base rate (200 pps). This might be due to different influences like the limited range of stimulation in the cochlea, the mismatch of rate and place of stimulation and the high level of hearing loss before implantation.

Furthermore, the discrepancy in the pulse rate difference limen between acoustic and electrical stimulation might be caused by the different excitation pattern which is evoked in the auditory nerve fibers (Hartmann et al., 1998). The response of the nerve fibers to an acoustic stimulus is much more stochastic than the response to an electric stimulus. This difference might influence the detection of small stimulation rate changes. Consequently, in experiment 5 (page 47) a more stochastic excitation pattern was evoked using amplitude modulated stimuli. The high carrier rate of 5081 pps was expected to cause a more stochastic response of the auditory nerve and to decrease the difference limen for amplitude modulated stimuli. The results of experiment 5 however show that the amplitude modulation difference limen was significantly higher than the pulse rate difference limen for the same base rates. That means that the stimuli used in this experiment could not support the theory that the more stochastic excitation pattern would cause a better detection of temporal changes.

In experiment 6 (page 51) the topic of the thesis returned to the place of stimulation. Experiment 1 (page 16) had shown that the electrode distance of 2.4 mm was wide enough to provide discriminable electrodes on what concerns the perceived pitch height. However, the amount of the perceived difference in pitch height corresponding to a shift in electrode place remained unclear as well as the range of the perceived pitch height when stimulating the electrodes along the cochlea. Therefore, in experiment 6 the task of six subjects with residual hearing in the low frequency region at the non-implanted ear was to adjust the frequency of an acoustic stimulus in a way to perceive the same pitch height as it was elicited by electrical stimulation of one electrode at the implanted ear. The experiment showed that the pitch perception at the electrically stimulated side could hardly be compared with the pitch perception at the acoustically stimulated side due to the great hearing loss. The adjusted frequencies for the most apical electrode varied between 150 and 350 Hz. Between the two most apical electrodes there was no significant difference in the adjusted frequencies. The increase in adjusted frequency with increasing electrode number from apex to base up to electrode E6 was linear. This is in contrast to the estimated frequency-place allocation for normal hearing where there is a logarithmical increase from apex to base (Zwicker & Fastl, 1999). However, the results should be considered carefully due to the high level of hearing loss in the non-implanted ears of the participating subjects.

2. New speech processing strategies

The aim of the actual thesis was to implement the results of the hearing experiments into a transformed speech processing strategy in order to enlarge the representation of the fine spectral structure in the incoming signal. The RateCIS strategy presented in this work (chapter III and IV) involves a pulse rate switching at a selection of electrodes. It was tested for speech perception and music appraisal in nine subjects. Despite the fact that all subjects were used to the classical CIS strategy, the RateCIS strategy was very well accepted by all listeners and yielded comparable results for the tests of numbers and polysyllables. For the average listener, there was a slightly better score of correctly recognized words with the classical CIS strategy only for the sentence test. An additionally filled out questionnaire revealed that six out of nine subjects preferred the classical CIS strategy and three out of nine subjects preferred the RateCIS strategy for the transmission of speech signals. Concerning the appraisal of music, there was a clear advantage for the RateCIS strategy which was preferred by six out of nine subjects. Due to hardware limitations, the RateCIS strategy could not be tested in live speech modus. Therefore, only a very limited acclimatization time could be given to the subjects. Despite this, it seems yet that in the actual implementation, the RateCIS strategy is a reasonable alternative to the classical CIS strategy and might be preferred especially for the presentation of music.

There are a lot of other approaches of investigators who try to find new ways of signal processing or stimulation techniques in order to enhance the speech understanding in noise or the sound quality of music. In the following, the most promising topics are presented.

a) Reducing the neural spread of excitation

i) New electrode array design

One of the most important topics concerning a better presentation of spectral information is the matter of discriminable electrodes. In all current cochlear implant systems the incoming signal is band pass filtered and the information of each band pass filter is transmitted to a different electrode. The representation of the incoming signal by a fixed number of band pass filters and allocated electrodes (in the current systems there are between 12 and 22 channels) is only highly effective if all electrodes evoke a different pitch perception. There are different approaches in order to provide a better discrimination for the average user especially for cochlear implant systems with an electrode spacing of 0.75 mm (CI24 by Cochlear) or 1.1 mm (HiRes90K by Advanced Bionics).

It is considered that indiscriminable electrode are caused by a large spread of the electric field when stimulating a single electrode. The electric fields of two adjacent electrodes may overlap and that way elicit the same pitch perception. One approach in order to avoid a broad spread of neural excitation consists of the idea to move the electrode array closer to the modiolar wall in the cochlea where the fibers of the auditory nerve are located. This approach has been realized for the Cochlear CI24 system and the Advanced Bionics CII Bionic Ear and HiRes90K system. The precurved electrode array of the CI24 Contour is held straight by a stylet which is withdrawn after or during the insertion of the array into the cochlea (Cohen et al., 2001). The array then curves in a way to be close to the modiolus. The studies of Tykoncinski et al (2001) for adults and Parkinson et al. (2002) for kids show that lower thresholds can be reached with the Cochlear CI24 Contour electrode. The Advanced Bionics' approach for the CII Bionic Ear was the insertion of an additional positioner into the cochlea in order to locate the electrode array closer to the modiolus (Kuzma & Balkany, 1999; Fayad et al., 2000). The most current implant by Advanced Bionics, the HiRes90K uses the HiFocus electrode which is also precurved in order to be located closer to the modiolus after the insertion like the Cochlear CI24 Contour. The studies of Donaldson et al. (2001) and Young & Grohne (2001) describe lower threshold for the CII system with positioner. This means that the electrodes are located closer to the auditory nerve because less current is necessary to reach the same perception of loudness as with the normal electrode array. One can suppose that the described lower thresholds and the closer position of the electrodes to the auditory nerve are related with a better discrimination of adjacent electrodes. However, a very recent work by Boëx et al. (2003) reveals that there is no difference in the electrode interaction between the Ineraid system and the Clarion system with positioner. In the future multicentre studies will address the question if subjects implanted with electrode arrays which are positioned closer to the modiolar wall will achieve better speech perception especially in noise compared to subjects implanted with straight electrode arrays.

ii) Different shape of the electric field

A different approach in order to decrease channel interaction is to minimize the spread of the electric field and therefore to reduce the spread of neural excitation. Early cochlear implant systems used a monopolar stimulation mode whereby the electric field spreads between one active electrode on the intracochlear array and a reference electrode which is located outside the cochlea. Other cochlear implant systems use bipolar stimulation. With bipolar stimulation the electric field spreads between two electrodes on the array inside the cochlea. Busby et al. (1994) showed in a subject group of nine postlingually deafened users of the Cochlear CI22M that higher thresholds are needed with bipolar stimulation compared to monopolar stimulation. However, they also reported that there are more pitch reversals and a reduced range of pitch perceptions with monopolar stimulation compared to bipolar stimulation. Kral et al. (1998) have shown that a high spatial resolution can be reached with tripolar stimulation (the electric field then spreads between three electrodes on the array) when measuring electric potential distributions. The study was conducted with the Cochlear CI22M electrode array in a tank filled with saline solution but also in cat cadaver cochleae and living cat cochleae. Jolly et al. (1996) suggest a quadrupolar stimulation mode. Their models show a highly focusing action and a greater choice of parameters to shape the electric field with quadrupolar stimulation. However, Miller et al. (2003) argue that monopolar stimulation needs less current and that the larger spread of the electric field means that a larger number of fibers remain unsaturated; this would correspond to a greater stochastic behavior of the fibers which is more similar to the excitation pattern in normal hearing. In current cochlear implant systems, different solutions are provided. The devices by Cochlear (Melbourne, Australia) allow different ways of stimulation depending on the implant type and the strategy which is chosen. The most current system, the CI24M, works mostly with bipolar stimulation. The Advanced Bionics system (Sylmar, United States of America) also allows different speech processing strategies using a monopolar or bipolar stimulation mode. The MED-EL system applies monopolar stimulation without the possibility to modify the wiring of the electrodes. One can assume that with the large electrode spacing of 2.4 mm there is much less spread of the electric field. Therefore this system uses the monopolar stimulation mode in order to reduce the consumption of electric current. Besides, the system works according to the idea of a more stochastic stimulation as suggested by Miller et al. (2003) described above. In the implementation of the transformed CIS strategy, the RateCIS strategy, the stimulation mode was monopolar due to manufactural restrictions. As the results of the actual thesis have shown that the electrode distance of 2.4 mm is wide enough for the average user to evoke discriminable pitch perceptions when stimulating neighbored electrodes, it seems reasonable to keep the actual monopolar stimulation mode. It can hardly be assumed that additional effects of the RateCIS strategy can be expected when using a bipolar stimulation mode.

b) Desynchronization of the neural response

The fact of a stochastic behavior of the auditory nerve fibers was already observed by Hartmann et al. (1984) who found a more stochastic excitation pattern of the auditory nerve fibers in cats as a response to an acoustic than to an electric stimulus. Rubinstein et al. (1999) have developed a model in order to achieve more spontaneous like activity of the auditory nerve fibers. They used a high rate pulse train (5000 pps) with fixed current amplitude of sufficient magnitude to evoke highly synchronous spikes in all 300 axons and showed that the representation of the electric stimulus waveforms in the temporal discharge pattern of the auditory nerve fibers improved. The spontaneous like activity of the auditory nerve fibers leads to a better temporal resolution. This was proven by Litvak et al. (2003) who measured the activity of auditory nerve fibers in cats when switching a sinusoidal modulator on and off. The unmodulated responses showed a hyper synchronization and a narrow dynamic range whereas the fibers were sensitive to modulation depths of 0.25% for a modulation rate of 417 Hz and a larger dynamic range could be measured. Over a 20 dB range of modulation depth there were resembling responses to normal hearing. The fact of a larger dynamic range is correlated with better speech perception (Loizou et al., 2000a) and especially a better speech perception in noise (Boike & Souza, 2000). The effect of the larger dynamic range was also investigated by Hong & Rubinstein (2003). They did the first hearing experiments with a conditioning pulse train in human cochlear implanted subjects. The amplitude of the conditioning pulse train was set at a level at which it was not perceptible after five minutes. The levels of additional presented sinusoids were measured at different rates (202, 515 and 1031 Hz). A larger dynamic range with the conditioning pulse train was found with variations of the location of the electrode. They therefore proposed the implementation of a conditioning pulse train at different levels for different electrode pairs because different electrode pairs may generate different dynamic range patterns. Meyer et al. (2003) have tested the frequency discrimination of sinusoid burst at single electrodes with and without an additional conditioning pulse train. They found a better discrimination at 200 Hz by 13.5% for the average data of 13 subjects. This means that the addition of a conditioning pulse train not only decreases the thresholds but it also enhances the temporal and spectral resolution which might altogether lead to a better speech perception. The approach of a more spontaneous activation of the auditory nerve fibers in the actual thesis was investigated by running a hearing experiment with amplitude modulated stimuli. In this case the expected stochastic excitation pattern did not decrease the discrimination of the modulation rate. Therefore, in the new RateCIS strategy, the idea of a stochastic excitation pattern was not implemented. However, the detection of differences in the modulation rate differs from the idea of a conditioning pulse train and the so far reported effects of a larger dynamic range and better frequency discrimination seem to be promising. Further work will have to prove the thesis of a better speech perception with conditioning pulse trains in cochlear implanted subjects.

The effect of spontaneous like activity of the auditory nerve is somehow incorporated in the idea of Benham & Zeng (2003) who claimed that the presentation of an additional noise would enhance the spectral resolution in cochlear implant systems. They measured the discrimination of vowel like sinusoids with an additional Gaussian noise at single electrodes and found a better discrimination than without additional noise. A similar model was already proposed by Morse & Evans (1999) who observed the presentation of the first formant seen in amplitude changes of the simulated discharges in a model of the auditory nerve. They therefore expected that the coding of temporal information would enhance using an additive noise.

A different idea leading to the same effect of spontaneous like activity of the auditory nerve fibers was investigated by Loizou et al. (2000b). They found a better word recognition when stimulating the electrodes with a very high pulse rate of 2100 pps compared with a stimulation rate of 800 pps in the MED-EL system. Further experiments have to be conducted in order to prove whether a conditioning pulse train, an additional noise or a higher stimulation rate are increasing the speech perception in cochlear implant subjects.

c) Reducing the mismatch of frequency-place allocation

Another often discussed approach in the signal processing and electrode allocation is the question of frequency-place mismatch. Figure 1 shows that the most apical electrode of the Nucleus and Clarion system is located in an area in the cochlea where a frequency of 600 Hz is having its maximal oscillation frequency in normal hearing. The most apical electrode of the MED-EL system is located in the area of the maximal oscillation frequency of 200 Hz. Nevertheless the signal processing of all systems consists of a band pass filtering in a spectral region minimally 180 and maximally 10800 Hz. This means that spectral information of the signal is contributed to electrodes which are located in a region of the cochlea where different frequencies are located in normal hearing and this results in a frequency-place mismatch. Baskent & Shannon (2002) have investigated the effect of a matching frequency-place allocation in comparison with a compression and expansion in normal hearing listeners using a noise band vocoder simulating different insertion depth of the electrode array and different numbers of electrodes. They found that speech perception was best for the matched condition. The MED-EL device which was used for the hearing experiments in this thesis is probably offering the best match of frequency-place allocation. Additionally, in order to keep the number of changed parameters concise, the filter bank used for the implementation of the RateCIS strategy was maintained as it is proposed for the CIS strategy. Furthermore, the results of experiment 6 (page 51) have shown that the average frequency for the most apical electrode was adjusted to 277 Hz. Therefore a filter bank between 300 and 7000 Hz seems reasonable for this implant type. However, the cut off frequencies of the filter bank should be generally adjusted individually for this implant type according to the electrode position of the most apical electrode in order to provide an optimal match of the frequency-place allocation.

d) Better transmission of the fundamental frequency

Geurts & Wouters (2004) propose a different transformation of the signal processing. They think that it is very important to better transmit the fundamental frequency of complex sounds in order to achieve an enhanced discrimination of complex sounds with different fundamental frequencies. Therefore they implemented a new filter bank in the signal processing of the LAURA implant. The new filter bank is designed in order to analyze the first harmonic of a complex sound in two adjacent filter bands. It is implemented using a so-called tree structure resulting in 12 filters. As the LAURA implant only provides eight electrodes, the eight highest filter bands are combined to four broader and flat filter bands. In the frequency region below 450 Hz, there are four filter bands. All filters are overlapping and based on a simple loudness model. They are called triangle filter due to their frequency response which is approximated with a 16th order infinite impulse response (IIR) filter. The new filter bank was tested for the discrimination of synthetic vowels in four subjects. The results with the new filter bank were significantly better than the results with the classical filter bank. That means that the new filter bank can provide information about the fundamental frequency of a complex sound in addition to the information which is coded by temporal envelope fluctuations. This idea seems very reasonable and should be investigated for different cochlear implant systems with different electrode numbers. The filter types used in the work of Geurts & Wouters (2004) should additionally be tested in comparison with the actual filter types in all cochlear implant systems in order to find the best combination of electrode array and signal processing.

e) Integrating rate information

Lan et al. (2004) also incorporated the information of the fundamental frequency in the signal processing and proposed a dynamic modulation of frequency and amplitude. They presented frequency modulated pulses whereby the modulation frequency was chosen according to the fundamental frequency in each channel. The signal processing was especially designed to enhance the speech perception of tonal languages which include more tonal codes than western languages. This signal processing strategy was tested in comparison with the signal processing of the classical CIS strategy in normal hearing subjects. In both cases the filter bank consisted of only four channels. With this signal processing strategy a better perception of Chinese tones, phrases and sentences was achieved. The approach of Lan et al. (2004) is based on the representation of the fundamental frequency in each channel as it was analyzed for the respective channel. It seems as if the transmission of spectral information in their signals when presented to normal hearing subjects was improved. For cochlear implant users, however, only changes in the modulation rate of 25% of the base rate could be detected in experiment 6 of the actual thesis. It is possible that modulation rate changes in all channels simultaneously might increase the ability of cochlear implant users to detect changes in the modulation rate. Furthermore, it might be the case that the tonal information in their speech stimuli can be transmitted by changes of 25% of the base rate.

Another approach with a redesigned filter bank was presented by Fearn (1999) in his doctoral thesis for the Cochlear CI22 implant. He implemented ten channels analyzing frequencies below 1000 Hz. For his InstrumentL strategy the electrodes allocated to those ten channels were additionally stimulated at varying stimulation rates according to the analyzed frequency in the according band. In a second approach he used the classical filter bank but for his VocL strategy there were five channels below 1000 Hz which were stimulated at varying stimulation rates according to the analyzed frequency in the associated band. The new strategies were tested in live speech modus with a clinical speech processor over a testing period of a few months in four subjects. In one subject the pitch discrimination ability for half tones was tested and showed an improvement with the InstrumentL strategy after one month. The second subject immediately performed better with the VocL strategy than with the ACE strategy concerning the determination whether two tones were the same or different. All four subjects reported that music had a better sound with InstrumentL and VocL, speech however had a better sound with the classical ACE strategy. According to the hearing experiments on stimulation rate conducted in this thesis it seems not reasonable to change the stimulation rate in a range up to 1000 Hz or to allow very fine changes in the stimulation rate at each electrode. Experiment 2 (page 24) has shown that stimulation rate changes are only noticeable up to 300 pps for the average user. Furthermore, in experiment 4 (page 36) a pulse rate difference limen of 25% of the base rate was determined. Therefore, in contrast to Fearn (1999) the basic idea of RateCIS was not to transmit the exact frequency analyzed within a filter band to the allocated electrode but to enlarge the spectral information in a way to create pitch perception between two adjacent electrodes with a fixed change in stimulation rate.

3. Prospects

Although the proposed RateCIS strategy yielded promising results especially for the perception of music, additional work is necessary in order to further enhance the quality and perception of speech and music.

First, the varying stimulation rate in each channel should be fitted individually in order to create a type of tonal scale for all combinations of place and rate of stimulation.

Second, a new filter bank should be tested in the low frequency range in order to reduce the frequency-place mismatch. The new filter bank could also be fitted individually according to the electrode allocation controlled via Stenvers' view x-ray scans. A different filter bank should be tested according to the estimated pitch for the different electrodes in experiment 6 (page 51).

Third, the fitting of the new RateCIS strategy should be optimized by means of a loudness growth control with small band noises of different frequency regions in order to find a more equalized and individually adopted fitting which might cause additional differences in the sound perception when compared with the classical CIS.

Fourth, the RateCIS strategy should be tested for a longer period in live speech modus in order to get repeatable or improved results after a time of acclimatization. This way, more difficult speech recognition tasks might be tested and allow a comparison with CIS in a situation with additional background noise.

Fifth, the RateCIS strategy should be tested during a first fit session with cochlear implant patients in order to test whether the lower sound of the RateCIS strategy is easier to adopt after implantation and whether it sounds more natural than the CIS strategy to patients without experience with the CIS strategy.

SUMMARY

A study comprised of six hearing experiments was conducted in order to investigate parameters to influence the pitch perception elicited by direct electric stimulation of the auditory nerve. In addition, a new stimulation strategy for the cochlear implant COMBI 40+ (MED-EL, Innsbruck, Austria) was developed and tested.

The results derived from a total number of 16 subjects reveal a dominating influence of the place of stimulation in contrast to the rate of stimulation on pitch perception. It was shown that the electrode distance of 2.4 mm for this device is sufficient to allow discriminable electrodes in pitch along the whole array. The influence of stimulation rate on pitch is limited to pulse rates up to about 300 pps. Within this range, the just noticeable change of pitch elicited by pulse rate as well as modulation rate amounts to about 25% of the base rate. In addition it was observed that the sound quality increases with increasing pulse rate up to about 566 pps independent of electrode location. Subjects with residual hearing at the non-implanted ear revealed that the pitch elicited by the most apical electrode depends on the insertion depth of the array and is linearly increasing with electrode location (40 Hz/mm).

The results of the hearing experiments were implemented to modify the well known CIS strategy. The new development (termed RateCIS) was designed in order to increase the amount of transmitted spectral information, thus the number of effective channels. Six electrodes were selected to switch adaptively between a high stimulation rate (1515 pps) and a low stimulation rate (252 pps). A test of the RateCIS strategy showed that results for speech recognition are comparable to the CIS strategy. The RateCIS strategy was subjectively preferred by some of the subjects although the majority preferred the CIS strategy for speech recognition and sound quality. Concerning the recognition and appraisal of music however, the RateCIS strategy was preferred by the majority of subjects. Regarding the fact, that the tests were conducted during one day without time for adaptation to the new signal processing, the RateCIS strategy could serve as an interesting option especially for music appraisal.

ZUSAMMENFASSUNG

Sechs Hörversuche wurden durchgeführt, um herauszufinden, welche Parameter die Tonhöhenwahrnehmung bei direkter elektrischer Stimulation des Hörnervs beeinflussen. Außerdem wurde eine neue Signalverarbeitung für die Ansteuerung des COMBI 40+ Implantats (MED-EL, Innsbruck, Österreich) entwickelt und getestet.

Die Hörversuche zeigen einen dominierenden Einfluss des Stimulationsorts über die Stimulationsrate auf die Tonhöhenwahrnehmung. Es konnte gezeigt werden, dass ein Elektrodenabstand von 2.4 mm ausreicht die Elektroden entlang der gesamten Cochlea anhand der Tonhöhe zu unterscheiden. Der Einfluss der Stimulationsrate auf die Tonhöhe ist limitiert in einem Bereich bis 300 pps. In diesem Bereich beträgt der gerade wahrnehmbare Pulsratenunterschied ca. 25%. Zusätzlich sinkt die Klangqualität ab einer Rate von 566 pps. Bei Versuchspersonen mit Restgehör auf dem nicht implantierten Ohr konnte gezeigt werden, dass die Tonhöhe bei Stimulation der apikalsten Elektrode von der Einführtiefe des Elektrodenträgers abhängt und linear mit dem Elektrodenort ansteigt (40Hz/mm).

Die Ergebnisse aus den Hörversuchen wurden in einer umgewandelten CIS-Sprachstrategie (genannt RateCIS) verarbeitet. Diese Strategie wurde entwickelt um die Information an spektralen Eigenschaften und somit die Anzahl an effektiven Kanälen zu erhöhen. Sechs Elektroden wurden ausgewählt um adaptiv zwischen einer hohen (1515 pps) und einer niedrigen (252 pps) Stimulationsrate zu wechseln. Ein Test der neuen RateCIS-Strategie zeigt vergleichbare Ergebnisse im Sprachtest mit der CIS-Strategie. Die RateCIS-Strategie wurde subjektiv für die Wahrnehmung von Sprache von weniger Versuchspersonen bevorzugt als die CIS-Strategie. Für die Wahrnehmung von Musik jedoch wurde die RateCIS-Strategie von der Mehrheit der Versuchspersonen bevorzugt. Wenn man berücksichtigt, dass die Evaluierung der RateCIS-Strategie an einem Tag ohne Möglichkeit zur Eingewöhnung durchgeführt wurde, könnte die RateCIS-Strategie als ein interessantes zusätzliches Programm speziell für die Wahrnehmung von Musik verwendet werden.

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ATTACHMENTS

Attachment 1: Stimulation matrix of three stimuli with gaps

First stimulus (pulse1) channel 10 with high stimulation rate (736 pps)
Second stimulus (pulse 18) channel 10 with high stimulation rate (736 pps)
Third stimulus (pulse 35) channel 10 with low stimulation rate (453 pps, every 2nd stimulation for a stimulation rate of 906 pps)

download and stimulation tool V.2.05 for the RIB
>>>>>>>> SERIAL PORT: 2
>>>>>>>> BAUD RATE: 115200

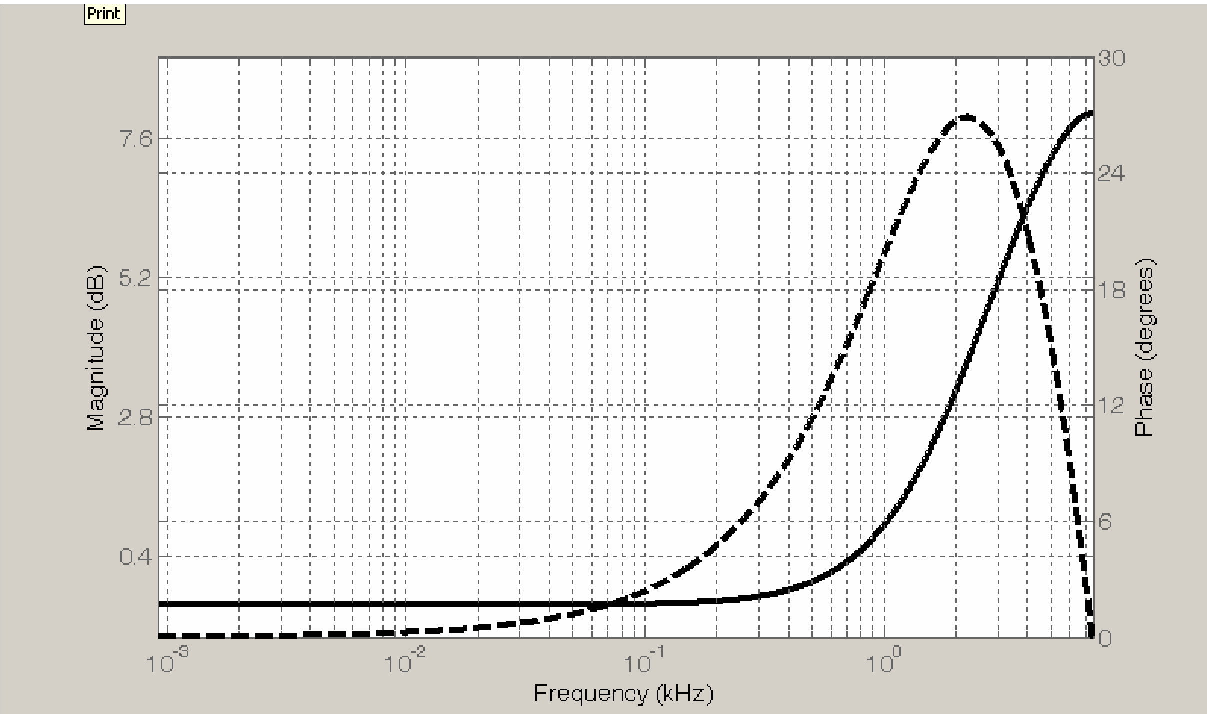
44 pulses in 33 words:

1	0.000 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
2	1.358 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
3	2.717 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
4	4.075 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
5	5.433 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
6	6.792 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
7	8.150 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
8	9.508 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
9	10.612 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
10	11.715 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
11	12.818 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
12	13.922 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
13	15.025 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
14	16.128 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
15	17.232 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
16	18.335 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
17	19.438 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
18	20.542 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
19	21.900 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
20	23.258 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
21	24.617 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
22	25.975 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
23	27.333 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
24	28.692 ms:	ch 10	amp 104	rng 1	wid 16	md 815	
25	30.050 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
26	31.153 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
27	32.257 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
28	33.360 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
29	34.463 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
30	35.567 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
31	36.670 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
32	37.773 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
33	38.877 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
34	39.980 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
35	41.083 ms:	ch 10	amp 112	rng 1	wid 16	md 662	
36	42.187 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
37	43.290 ms:	ch 10	amp 112	rng 1	wid 16	md 662	
38	44.393 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
39	45.497 ms:	ch 10	amp 112	rng 1	wid 16	md 662	
40	46.600 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
41	47.703 ms:	ch 10	amp 112	rng 1	wid 16	md 662	
42	48.807 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros
43	49.910 ms:	ch 10	amp 112	rng 1	wid 16	md 662	
44	51.013 ms:	ch 10	amp 0	rng 1	wid 16	md 662	zeros

duration 31270 bits, 52.117 ms
>>>>>>>> COMMAND: Execute 0 0 1

Attachment 2: Preemphasis digital filter

Transfer function of the first order digital filter applied for preemphasis. Magnitude and phase (dashed line).



Attachment 3: Stimulation matrix for two pure tones

Frequencies between band filters of E6 (low-high)

2413	132.660 ms:	ch 1	amp 0	rng 2	wid 16	md 33
2414	132.715 ms:	ch 7	amp 0	rng 2	wid 16	md 33
2415	132.770 ms:	ch 2	amp 4	rng 2	wid 16	md 33
2416	132.825 ms:	ch 8	amp 0	rng 2	wid 16	md 33
2417	132.880 ms:	ch 3	amp 4	rng 2	wid 16	md 33
2418	132.935 ms:	ch 9	amp 0	rng 2	wid 16	md 33
2419	132.990 ms:	ch 4	amp 0	rng 2	wid 16	md 33
2420	133.045 ms:	ch 10	amp 20	rng 2	wid 16	md 33
2421	133.100 ms:	ch 5	amp 10	rng 2	wid 16	md 33
2422	133.155 ms:	ch 11	amp 22	rng 2	wid 16	md 33
2423	133.210 ms:	ch 6	amp 12	rng 2	wid 16	md 33
2424	133.265 ms:	ch 12	amp 25	rng 2	wid 16	md 33
2425	133.320 ms:	ch 1	amp 0	rng 2	wid 16	md 33
2426	133.375 ms:	ch 7	amp 0	rng 2	wid 16	md 33
2427	133.430 ms:	ch 2	amp 4	rng 2	wid 16	md 33
2428	133.485 ms:	ch 8	amp 0	rng 2	wid 16	md 33
2429	133.540 ms:	ch 3	amp 4	rng 2	wid 16	md 33
2430	133.595 ms:	ch 9	amp 0	rng 2	wid 16	md 33
2431	133.650 ms:	ch 4	amp 0	rng 2	wid 16	md 33
2432	133.705 ms:	ch 10	amp 20	rng 2	wid 16	md 33
2433	133.760 ms:	ch 5	amp 10	rng 2	wid 16	md 33
2434	133.815 ms:	ch 11	amp 22	rng 2	wid 16	md 33
2435	133.870 ms:	ch 6	amp 0	rng 2	wid 16	md 33
2436	133.925 ms:	ch 12	amp 25	rng 2	wid 16	md 33
2437	133.980 ms:	ch 1	amp 0	rng 2	wid 16	md 33
2438	134.035 ms:	ch 7	amp 0	rng 2	wid 16	md 33
2439	134.090 ms:	ch 2	amp 4	rng 2	wid 16	md 33
2440	134.145 ms:	ch 8	amp 0	rng 2	wid 16	md 33
2441	134.200 ms:	ch 3	amp 4	rng 2	wid 16	md 33
2442	134.255 ms:	ch 9	amp 0	rng 2	wid 16	md 33
2443	134.310 ms:	ch 4	amp 0	rng 2	wid 16	md 33
2444	134.365 ms:	ch 10	amp 20	rng 2	wid 16	md 33
2445	134.420 ms:	ch 5	amp 10	rng 2	wid 16	md 33
2446	134.475 ms:	ch 11	amp 22	rng 2	wid 16	md 33
2447	134.530 ms:	ch 6	amp 0	rng 2	wid 16	md 33
2448	134.585 ms:	ch 12	amp 25	rng 2	wid 16	md 33
2449	134.640 ms:	ch 1	amp 6	rng 2	wid 16	md 33
2450	134.695 ms:	ch 7	amp 0	rng 2	wid 16	md 33
2451	134.750 ms:	ch 2	amp 4	rng 2	wid 16	md 33
2452	134.805 ms:	ch 8	amp 17	rng 2	wid 16	md 33
2453	134.860 ms:	ch 3	amp 4	rng 2	wid 16	md 33
2454	134.915 ms:	ch 9	amp 17	rng 2	wid 16	md 33
2455	134.970 ms:	ch 4	amp 0	rng 2	wid 16	md 33
2456	135.025 ms:	ch 10	amp 20	rng 2	wid 16	md 33
2457	135.080 ms:	ch 5	amp 10	rng 2	wid 16	md 33
2458	135.135 ms:	ch 11	amp 22	rng 2	wid 16	md 33
2459	135.190 ms:	ch 6	amp 0	rng 2	wid 16	md 33
2460	135.245 ms:	ch 12	amp 25	rng 2	wid 16	md 33
2461	135.300 ms:	ch 1	amp 0	rng 2	wid 16	md 33
2462	135.355 ms:	ch 7	amp 17	rng 2	wid 16	md 33
2463	135.410 ms:	ch 2	amp 4	rng 2	wid 16	md 33
2464	135.465 ms:	ch 8	amp 0	rng 2	wid 16	md 33
2465	135.520 ms:	ch 3	amp 4	rng 2	wid 16	md 33
2466	135.575 ms:	ch 9	amp 0	rng 2	wid 16	md 33
2467	135.630 ms:	ch 4	amp 0	rng 2	wid 16	md 33
2468	135.685 ms:	ch 10	amp 20	rng 2	wid 16	md 33
2469	135.740 ms:	ch 5	amp 10	rng 2	wid 16	md 33
2470	135.795 ms:	ch 11	amp 22	rng 2	wid 16	md 33
2471	135.850 ms:	ch 6	amp 0	rng 2	wid 16	md 33
2472	135.905 ms:	ch 12	amp 25	rng 2	wid 16	md 33
2473	135.960 ms:	ch 1	amp 0	rng 2	wid 16	md 33
2474	136.015 ms:	ch 7	amp 0	rng 2	wid 16	md 33
2475	136.070 ms:	ch 2	amp 4	rng 2	wid 16	md 33
2476	136.125 ms:	ch 8	amp 0	rng 2	wid 16	md 33
2477	136.180 ms:	ch 3	amp 4	rng 2	wid 16	md 33
2478	136.235 ms:	ch 9	amp 0	rng 2	wid 16	md 33
2479	136.290 ms:	ch 4	amp 7	rng 2	wid 16	md 33
2480	136.345 ms:	ch 10	amp 20	rng 2	wid 16	md 33
2481	136.400 ms:	ch 5	amp 10	rng 2	wid 16	md 33
2482	136.455 ms:	ch 11	amp 22	rng 2	wid 16	md 33
2483	136.510 ms:	ch 6	amp 0	rng 2	wid 16	md 33
2484	136.565 ms:	ch 12	amp 25	rng 2	wid 16	md 33
2485	136.620 ms:	ch 1	amp 0	rng 2	wid 16	md 33
2486	136.675 ms:	ch 7	amp 0	rng 2	wid 16	md 33
2487	136.730 ms:	ch 2	amp 4	rng 2	wid 16	md 33
2488	136.785 ms:	ch 8	amp 0	rng 2	wid 16	md 33
2489	136.840 ms:	ch 3	amp 4	rng 2	wid 16	md 33
2490	136.895 ms:	ch 9	amp 0	rng 2	wid 16	md 33
2491	136.950 ms:	ch 4	amp 7	rng 2	wid 16	md 33
2492	137.005 ms:	ch 10	amp 20	rng 2	wid 16	md 33
2493	137.060 ms:	ch 5	amp 10	rng 2	wid 16	md 33
2494	137.115 ms:	ch 11	amp 22	rng 2	wid 16	md 33
2495	137.170 ms:	ch 6	amp 0	rng 2	wid 16	md 33
2496	137.225 ms:	ch 12	amp 25	rng 2	wid 16	md 33
2497	137.280 ms:	ch 1	amp 0	rng 2	wid 16	md 33
2498	137.335 ms:	ch 7	amp 0	rng 2	wid 16	md 33
2499	137.390 ms:	ch 2	amp 4	rng 2	wid 16	md 33
2500	137.445 ms:	ch 8	amp 0	rng 2	wid 16	md 33
2501	137.500 ms:	ch 3	amp 4	rng 2	wid 16	md 33
2502	137.555 ms:	ch 9	amp 0	rng 2	wid 16	md 33
2503	137.610 ms:	ch 4	amp 7	rng 2	wid 16	md 33
2504	137.665 ms:	ch 10	amp 20	rng 2	wid 16	md 33
2505	137.720 ms:	ch 5	amp 11	rng 2	wid 16	md 33
2506	137.775 ms:	ch 11	amp 22	rng 2	wid 16	md 33
2507	137.830 ms:	ch 6	amp 18	rng 2	wid 16	md 33
2508	137.885 ms:	ch 12	amp 25	rng 2	wid 16	md 33
2509	137.940 ms:	ch 1	amp 0	rng 2	wid 16	md 33
2510	137.995 ms:	ch 7	amp 0	rng 2	wid 16	md 33
2511	138.050 ms:	ch 2	amp 4	rng 2	wid 16	md 33
2512	138.105 ms:	ch 8	amp 0	rng 2	wid 16	md 33
2513	138.160 ms:	ch 3	amp 4	rng 2	wid 16	md 33
2514	138.215 ms:	ch 9	amp 0	rng 2	wid 16	md 33
2515	138.270 ms:	ch 4	amp 7	rng 2	wid 16	md 33
2516	138.325 ms:	ch 10	amp 20	rng 2	wid 16	md 33
2517	138.380 ms:	ch 5	amp 11	rng 2	wid 16	md 33
2518	138.435 ms:	ch 11	amp 22	rng 2	wid 16	md 33
2519	138.490 ms:	ch 6	amp 22	rng 2	wid 16	md 33
2520	138.545 ms:	ch 12	amp 25	rng 2	wid 16	md 33
2521	138.600 ms:	ch 1	amp 0	rng 2	wid 16	md 33
2522	138.655 ms:	ch 7	amp 0	rng 2	wid 16	md 33
2523	138.710 ms:	ch 2	amp 4	rng 2	wid 16	md 33
2524	138.765 ms:	ch 8	amp 0	rng 2	wid 16	md 33
2525	138.820 ms:	ch 3	amp 4	rng 2	wid 16	md 33
2526	138.875 ms:	ch 9	amp 0	rng 2	wid 16	md 33
2527	138.930 ms:	ch 4	amp 7	rng 2	wid 16	md 33
2528	138.985 ms:	ch 10	amp 20	rng 2	wid 16	md 33
2529	139.040 ms:	ch 5	amp 12	rng 2	wid 16	md 33
2530	139.095 ms:	ch 11	amp 22	rng 2	wid 16	md 33
2531	139.150 ms:	ch 6	amp 27	rng 2	wid 16	md 33
2532	139.205 ms:	ch 12	amp 25	rng 2	wid 16	md 33
2533	139.260 ms:	ch 1	amp 6	rng 2	wid 16	md 33
2534	139.315 ms:	ch 7	amp 0	rng 2	wid 16	md 33
2535	139.370 ms:	ch 2	amp 4	rng 2	wid 16	md 33
2536	139.425 ms:	ch 8	amp 18	rng 2	wid 16	md 33
2537	139.480 ms:	ch 3	amp 4	rng 2	wid 16	md 33
2538	139.535 ms:	ch 9	amp 17	rng 2	wid 16	md 33
2539	139.590 ms:	ch 4	amp 7	rng 2	wid 16	md 33
2540	139.645 ms:	ch 10	amp 20	rng 2	wid 16	md 33
2541	139.700 ms:	ch 5	amp 13	rng 2	wid 16	md 33
2542	139.755 ms:	ch 11	amp 22	rng 2	wid 16	md 33
2543	139.810 ms:	ch 6	amp 31	rng 2	wid 16	md 33
2544	139.865 ms:	ch 12	amp 25	rng 2	wid 16	md 33

Attachment 4: Questionnaire for speech perception

Fragebogen Sprache Programm A - Programm B

Datum:_______________ Name:________________________________

1) **Klangfarbe**

a) **Klarheit der Sprache:**

extrem unklar	1	2	3	4	5	6	7	extrem klar
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

b) **Wohlklang:**

extrem unangenehm	1	2	3	4	5	6	7	extrem angenehm
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

c) **Gesamthöreindruck (Männerstimme):**

extrem schlechte	1	2	3	4	5	6	7	extrem gute Qualität
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

d) **Gesamthöreindruck (Frauenstimme:**

extrem schlechte	1	2	3	4	5	6	7	extrem gute Qualität
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

e) **Beurteilen Sie die Helligkeit des Klanges für die Männerstimme:**

extrem dunkel	1	2	3	4	5	6	7	extrem hell
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

f) **Beurteilen Sie die Helligkeit des Klanges für die Frauenstimme:**

extrem dunkel	1	2	3	4	5	6	7	extrem hell
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

g) **Beurteilen Sie die Natürlichkeit des Klanges:**

extrem unnatürlich	1	2	3	4	5	6	7	extrem natürlich
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

h) Beurteilen Sie die Intonation in der Stimme:

extrem eintöniger Klang	1	2	3	4	5	6	7	extrem melodiöser Klang
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

2) Wie anstrengend ist es die Sprache zu verstehen:

extrem wenig anstrengend	1	2	3	4	5	6	7	extrem anstrengend
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

3) Bei welchem Programm ist die Sprache deutlicher zu hören?

Programm A ☐ Programm B ☐

Attachment 5: Questionnaire and evaluation for music perception

Fragebogen Musikhören Programm A – Programm B

Datum:_______________ Name:________________________________

Bitte kreuzen Sie die zutreffenden Antworten an.

4) Melodieerkennung

Keine Melodie erkannt	1	2	3	4	5	6	7	viel Melodie erkannt
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

5) Rhythmuserkennung

Keinen Rhythmus erkannt	1	2	3	4	5	6	7	viel Rhythmus erkannt
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

6) Klangfarbe

a) Klarheit der Musik:

extrem unklar	1	2	3	4	5	6	7	extrem klar
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

b) Wohlklang:

extrem gering	1	2	3	4	5	6	7	extrem hoch
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

c) Gesamthöreindruck:

extrem schlecht	1	2	3	4	5	6	7	extrem gute Qualität
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

d) Beurteilen Sie die Helligkeit des Klanges:

extrem dunkel	1	2	3	4	5	6	7	extrem hell
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

e) Beurteilen Sie die Intonation/Betonung der Melodie:

extrem wenig betont	1	2	3	4	5	6	7	extrem betont
Programm A	☐	☐	☐	☐	☐	☐	☐	
Programm B	☐	☐	☐	☐	☐	☐	☐	

7) Welches Programm würden Sie bevorzugen um Musik zu hören?

Programm A ☐ Programm B ☐

8) Instrumentenerkennung:

4 Instrumente vorspielen mit Programm A und B abwechselnd

Flöte ………………………………………………………………………………

Orgel …………………………………………………………………………..

Piano ……………………………………………………………………………...

Xylophon ………………………………………………………………………

9) Melodieerkennung:

6 Songs vorspielen mit Programm A und B abwechselnd

Entchen ………………………………………………………………………………

Hänschen……………………………………………………………………………..

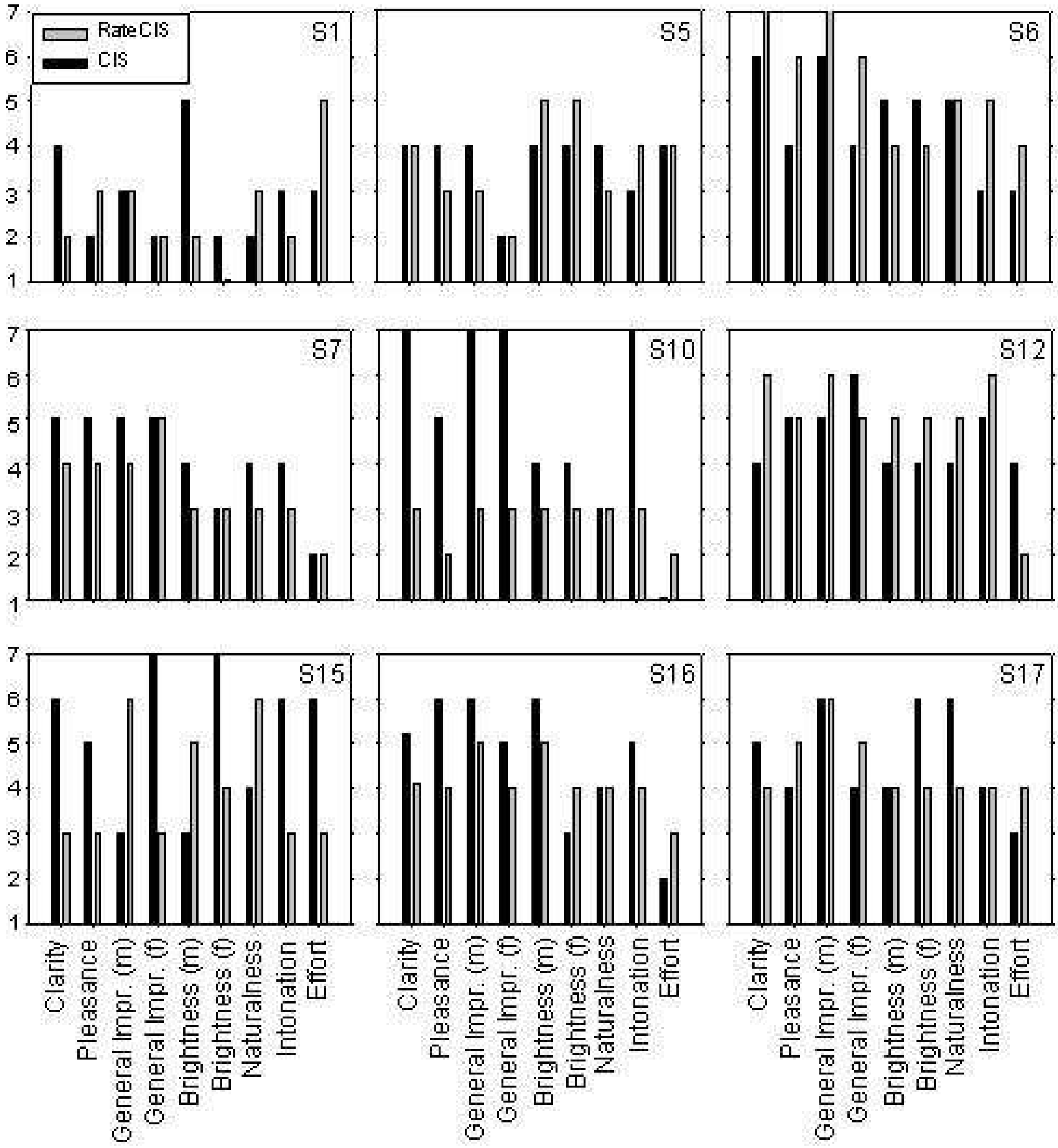
Kuckuck ……………………………………………………………………………...

Männlein…..………………………………..………………………………………

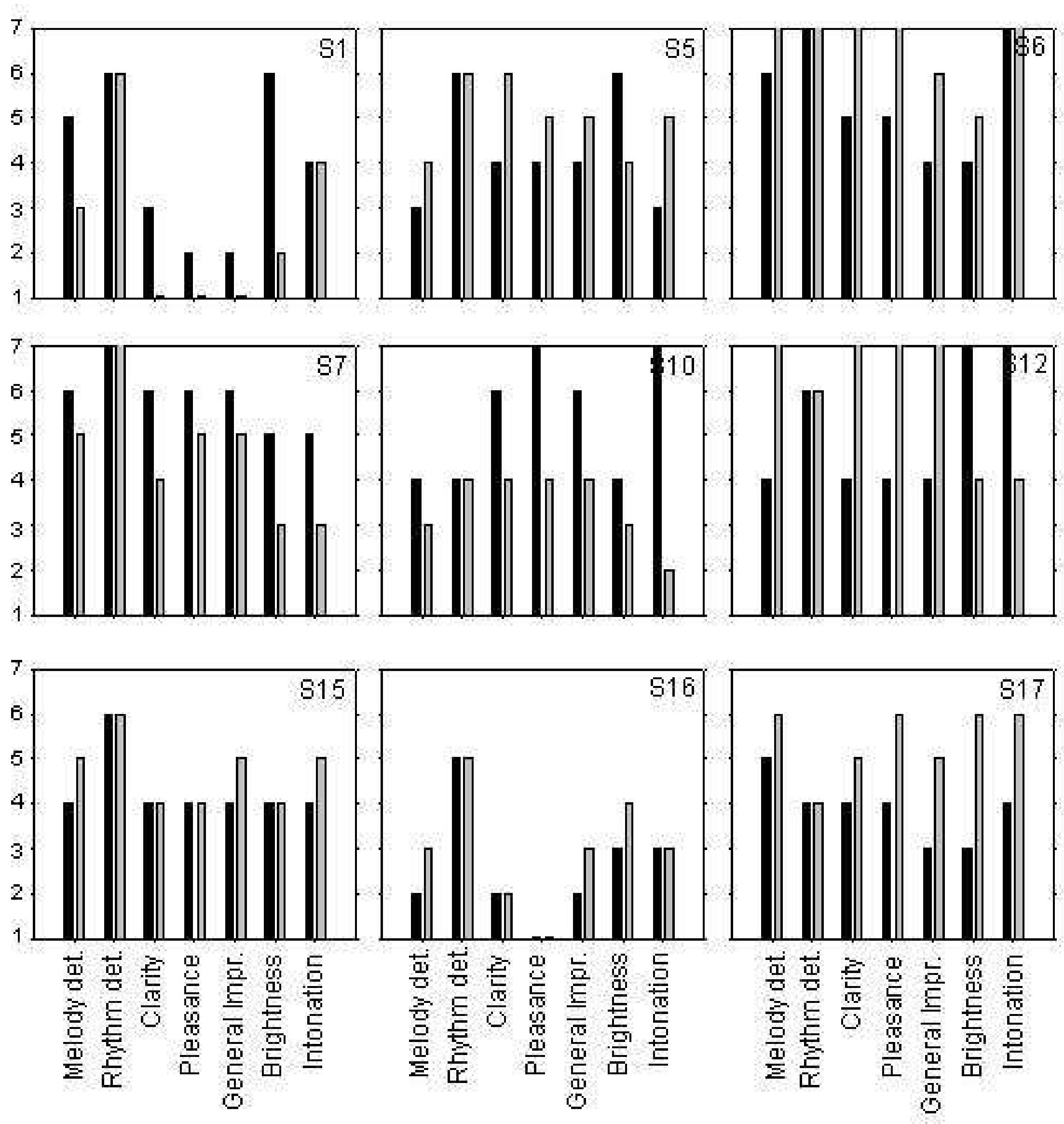
Mühle ………………………………………………………………………………...

Vögel …………………………………………..……………………………………

Attachment 6: Individual ratings for speech perception



Attachment 7: Individual ratings for music appraisal



ACKNOWLEDGEMENTS

Thanks to everyone who contributed to this thesis.

First of all I would like to thank my subjects who supported me over a long time and came to me immediately as soon as I called them because I had programmed some new experiment and needed someone to listen to my new idea. Without them and their untiringly dedication this thesis would not have been possible. I will miss you and our conversations very much!

I would also like to thank my supervisor Dr.-Ing. Uwe Baumann who offered me the possibility to work in the field of cochlear implant research. Our discussions have inspired the topic of this work. He has supported me with his profound knowledge in psychoacoustics and medical physics and has shared his experience with cochlear implants with me.

I would further like to thank all the reviewers of my documents which have tried to make the thesis more readable: Ilona Anderson, Ulrike Neeff, Dr.Peter Nopp and Dr. Andrew J. Oxenham.

I am indebted to the Department of Applied Physics at the University of Innsbruck (Prof. E. Hochmair) for provision of the research interface and programming support.

This research was supported by DFG Grant No. FO 306/2-3 (Research group "Auditory objects").

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Award

Förderpreis of the DGA (German society of audiology) 2004 for the presentation at the annual meeting of the DGA in 2003 ('Just noticeable difference of the rate pitch perception dependent on the place of stimulation')

Hobbies

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