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Research Papers

Loudness Functions and Binaural Loudness Summation in Bilateral Cochlear Implant Users

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Loudness functions and binaural loudness summation was investigated in acoustically stimulated bilaterally implanted cochlear implant users. The study was aimed at evaluating growth of loudness functions and binaural loudness summation in cochlear implant subjects as a function of stimulus presentation level at different frequencies. Loudness was assessed using a rating procedure on a scale of 0 to 100. Three experimental conditions were tested: monaural right, monaural left and binaural, each with bands of noise with center frequencies of 0.25, 1, and 4 kHz. Fifteen implanted and five normal-hearing subjects (control group) participated in the experiments. Results demonstrated large variability in the slopes of the loudness functions and the presence of loudness summation in bilateral cochlear implant users, with large individual differences among subjects.

Keywords: loudness; binaural loudness; loudness function; binaural summation; cochlear implant.

1. Introduction

Loudness is one of the basic attributes of hearing that relates the strength of sound percept to sound intensity. The sensation of loudness depends on many factors, such as sound intensity as well as spectral (e.g., Garnier *et al.*, 2000; Anweiler, Verhey, 2006; Leibold, Jesteadt, 2007) and temporal (e.g., Zwislocki, 1969; Florentine *et al.*, 1996; Garnier *et al.*, 1999) properties of the sound.

The traditional loudness scaling methods include magnitude estimation procedure, magnitude production procedure and cross-modality matching procedure (e.g., Marks, Florentine, 2011). In magnitude estimation procedure, the subject is asked to rate the loudness on a continuous and unlimited scale; the subject may use any number that is larger than zero. Magnitude production procedure, in turn, requires the subject to adjust sound intensity to achieve the perception of loudness that is proportional to a specific number. In cross-modality matching, the task of the subject is to adjust some physical quantity magnitude in order to match the loudness of the sound.

In normal-hearing listeners, the loudness function (LF) representing the growth of loudness with sound pressure level (SPL) is well represented by the sone scale and by Stevens' power law (Stevens, 1955; 1957), where the growth of loudness with SPL has a power function with an average exponent close to 0.6 for a 1-kHz tone. Loudness growth is a difficult issue to contend with, especially in cochlear implant (CI) processing. CI amplitude-mapping functions (MAPs) deal with a very complex relationship between perceived loudness and sound intensity, as loudness depends not only on physiological properties and constraints of the auditory system but also on numerous implant parameters. For example, there may be contributions from the microphones, fast and slow acting front-end auto-

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matic gain controls (AGCs), the functions and methods used to "map" audio signal levels to electrical stimulation currents, and the electrical stimulation parameters that affect individual listeners differently.

Various studies have investigated LFs among CI recipients using electrical implant stimulation, i.e. electrical loudness functions (ELFs). These studies demonstrated that the exponent of nonlinear stimulus-current function in the ELFs ranges from 2 to 3.5 (Zeng, Shannon, 1994; 1997; 1999). Thus, it appears that in electric CI stimulation, the ELF grows much more rapidly when compared with the LF known to occur with acoustic stimulation in normal-hearing (NH) listeners. In fact, loudness growth in CI listeners, whether through electrical stimulation or with acoustic stimulation through the listeners' microphones, depends on such implant parameters as electric current and stimulation rate (Zeng, Shannon, 1994), pulse duration, mode of stimulation (monopolar or bipolar), and electrode configuration (separation) (Fu, 2005; McKay, McDermott, 1998; McKay *et al.*, 1994; 2001, 2003; Tong, Clark, 1986; Zeng, Shannon, 1994).

The goal of a CI is to capture sounds from the listener's environment and convert them from an acoustic signal into an electrical signal for the brain to process through electrical stimulation. Basic acoustical parameters of sound relevant for sound perception are: amplitude, frequency (spectral) and change in time (temporal). Cochlear implants must code all these three dimensions of sound: amplitude (intensity) is coded by electric current level of stimulation to create loudness, frequency – by the place of stimulation (different electrode segments) to make a sense of the pitch, and time – by rate of stimulation to create a time-varying sound percept.

CI implants, unlike hearing aids, do not amplify sound, but electrically stimulate the auditory nerve inside the cochlea to provide a sense of sound to a person who is profoundly deaf or has a severe hearing loss. Due to the physiological and technical limitations, cochlear implants cannot reproduce or imitate all the details of signal processing in the auditory system of an otologically normal person so implant users cannot perceive sounds in the same way as normal-hearing listeners. There are four major stages in current generation CI systems that can be identified in an acoustical, digital and electrical signal processing path (Vaerenberg, 2014). In the first stage, the signal is captured by the sound processor (sound window). In the second stage, the signal is decomposed into spectral components (frequency mapping). In the third stage, the signal is detailed into temporal components (temporal waveform analysis). Finally, in the fourth stage, the signal is delivered to the implant/nerve (stimulation rate, waveform, sources). In general, the capture includes speech pre-emphasis, which attenuates strong spectral speech components to help weak consonants to successfully compete with vowels. The decomposition stage relies on band-pass filtering with suitable center frequency and bandwidth to create frequency channels to stimulate various parts of the cochlea in a way that is more or less similar to natural stimulation. Waveform detectors analyze the time waveforms at the filter outputs and deliver current to the electrodes; the log-frequency scale is mapped onto electrode segments.

Loudness perception using a CI is governed by perceptual effects and by three main technical factors: automatic gain control (AGC), logarithmic mapping and conversion to current units (μA). The AGC operates in the same way as in a hearing aid and is more or less linear on a short time scale. In contrast, the logarithmic mapping is instantaneous (and therefore nonlinear) and its I/O function depends on parameters T-level (threshold of perception), C-level (comfort of loudness perception), and Q-factor, which are set during implant fitting. The coding, following the logarithmic mapping, are expressed in the so-called clinical units. These units are converted to μA in a nonlinear way by the implanted stimulator. The combination of logarithmic mapping and conversion to current units is used to model the compression occurring in the cochlea. The parameter values used, however, are determined by the optimization of performance in speech perception, which is not identical to the optimization of linearity in loudness perception which occurs in NH persons. Therefore, the entire loudness processing chain is in most cases nonlinear and different from NH.

Manufacturers (such as *Cochlear* and *Advanced Bionics*) have adopted different approaches in addressing several of the above mentioned processes. They use different microphones and front-end compression schemes. The *Advanced Bionics* system has a wide input dynamic range that is usually spanned by two linear compression slopes, with the second being shallower than the first. In that system, infinite compression (zero slope) is only applied for very high presentation levels. In contrast, the *Cochlear* system has a narrower dynamic range and applies infinite compression at lower levels, typically between 65 and 70 dB for many broadband signals at default sensitivity. When the *Cochlear* system operates below the onset of infinite compression, levels vary much more rapidly than for similar SPL levels with the *Advanced Bionics* system, but once the level is above the compression threshold, the converse is true. The differences in the results reported for the two types of devices largely reflect this basic difference in their engineering design.

The purpose of the present study was to investigate growth of loudness functions (LFs) and binaural loudness summation in bilaterally implanted CI users with acoustical stimuli presented in the sound field that covered a wide frequency range. The loudness functions

(LFs) at low-, mid- and high-frequency regions were measured (0.25, 1, and 4 kHz) in monaural and binaural listening conditions. The comparison of monaural and binaural LFs provided information on binaural loudness summation in CI users. Furthermore, data describing CI users were compared with LFs obtained from a control group of normal-hearing (NH) persons in similar experimental conditions. Fifteen CI users participated in this study, of which seven were using devices manufactured by the *Cochlear Corp.* and eight by the *Advanced Bionics LLC*. Further in the text, the data are grouped according to the device used by the subject, with no intention of making any comparison concerning the quality of the devices.

2. Methods

2.1. Subjects

Fifteen adult CI users participated in this study: twelve females and three males. All the CI users had a postlingual onset of deafness and received minimal or no benefit from hearing aids prior to implantation. None of them had a residual hearing in the implanted ear. Their ages ranged from 36 to 67 years and the duration of their deafness – from 12 to 25 years. Details of the subjects' demographics, etiology, speech processor types and speech processing strategies are presented in Tables 1 and 2. All subjects were bilaterally implanted; eleven received implant devices in the

Table 1. Demographic information for *Cochlear* implant users.

Subject	Implanted	Gender	Etiology	Processor	Internal device	Strategy	Number of channels	Pulse rate [Hz]
S1	simultaneous	F	Hereditary	Freedom	Cochlear CI24RE Contour Advance	ACE	22^a, 21^b	900
S2	simultaneous	F	Unknown	Freedom	Cochlear CI24RE Contour Advance	ACE	22	1200
S3	simultaneous	F	Unknown	Sprit	Cochlear CI24M	ACE	22	1200
S4	simultaneous	F	Unknown	Freedom	Cochlear CI24M	ACE	22^a, 20^b	1200
S5	simultaneous	M	Coggan's Syndrome	ESPrit 3G	Cochlear	ACE	19^a, 20^b	1200
S6	sequential	F	Unknown	Freedom	Cochlear CI24R Contour^a Cochlear CI24M^b	ACE	22	1200
S7	simultaneous	F	Unknown	Freedom	Cochlear CI24R Contour^a Cochlear Nucleus 5^b	ACE	20	900

^{a} right ear, ^{b} left ear.

Table 2. Demographic information for *Advanced Bionics* implant users.

Subject	Implanted	Gender	Etiology	Processor	Internal device	Strategy	Number of channels	Pulse rate [Hz]
S8	simultaneous	F	Hereditary	Harmony	Clarion CII/Hi Focus	HiRes-S with Fidelity 120	16	3712
S9	simultaneous	M	Meningitis	Harmony	ClarionHiRes 90K	HiRes-S with Fidelity 120	16	3712
S10	simultaneous	M	Noise Exposure	Harmony	Clarion CII/Hi Focus	HiRes-S with Fidelity 120	16	3228
S11	simultaneous	F	Unknown	Harmony	ClarionHiRes 90K/HiFocus	HiRes-S	8	829
S12	sequential	F	Otosclerosis	Clarion Auria^a Med-el-cis link+^{b}	Ineraid^a ClarionHiRes 90K^b	HiRes-S^a CIS^b	6	2020
S13	simultaneous	F	Unknown	Harmony	Clarion CII/Hi Focus	HiRes-S with Fidelity 120	16	3712
S14	sequential	F	Unknown	Platinum BTE^a Auria^b	ClarionHiRes 90K^a ClarionHiRes 90K/HiFocus^b	CIS^a HiRes-P^b	8^a, 14^b	406^a 3458^b
S15	sequential	F	Unknown	Harmony	ClarionHiRes 90K/HiFocus	HiRes-S with Fidelity 120^a HiRes-P^b	9^a, 14^b	1904^a 5156^b

^{a} right ear, ^{b} left ear.

left and right ears during a single surgical procedure (simultaneous implantation), four were implanted in two separate procedures (sequential implantation).

2.2. Stimuli and procedure

Loudness functions were determined for octave bands of noise centered at 0.25, 1, and 4 kHz. All stimuli were generated in a MATLAB programming environment, prerecorded and presented via a loudspeaker (Lifeline Amplification System) in a double-walled sound-attenuated chamber. Stimulus presentation level was controlled by the Clinical Audiometer (type GSI 61) and incremented in 5 or 10 dB steps. Stimuli were calibrated such that the level (dB SPL) of signals at the subject's head was 5 dB higher at 0.25 and 4 kHz, and 10 dB higher at 1 kHz relative to HLs set in the audiometer. Subjects were tested while seated in the sound-attenuated chamber facing the loudspeaker at a distance of 1.65 m.

Unbiased loudness scaling can be obtained using open ended scale such as in the absolute magnitude estimation (AME) (Stevens, 1955; 1957). The CI subjects found it hard to use the open-ended scale because all of them had previous experience in different types of experiments with the loudness rating procedure, in which they had been asked to rate the loudness on a scale from 0 to 100. Thus even when asked to use an unlimited scale, the subjects chose the scale they were familiar with. For this reason, it was decided to assess LFs using a rating procedure (presence of possible non-linear effects resulting from using rating procedure are tested in Subsec. 3.2.2). Stimulus presentation levels were selected such that they ranged from levels corresponding to subject's hearing threshold to levels corresponding to the subject's most comfortable listening level.

Subjects were tested using an identical order of listening conditions: right ear, both ears, left ear. Monaural conditions were created by disconnecting the left or right cochlear implant device. The order of stimulus presentation levels was randomized, with the same sequence of levels used for all the subjects. Twenty-one responses were collected at each level; the first three responses were discarded from the LFs calculations.

3. Results

3.1. Loudness functions

Figures 1 and 2 show loudness functions (LFs) measured for *Cochlear* and *Advanced Bionics* implant users, respectively. The successive rows present data for noise bands at three center frequencies: 0.25 kHz (upper row), 1 kHz (middle row), and 4 kHz (lower row). Columns represent individual subjects' data.

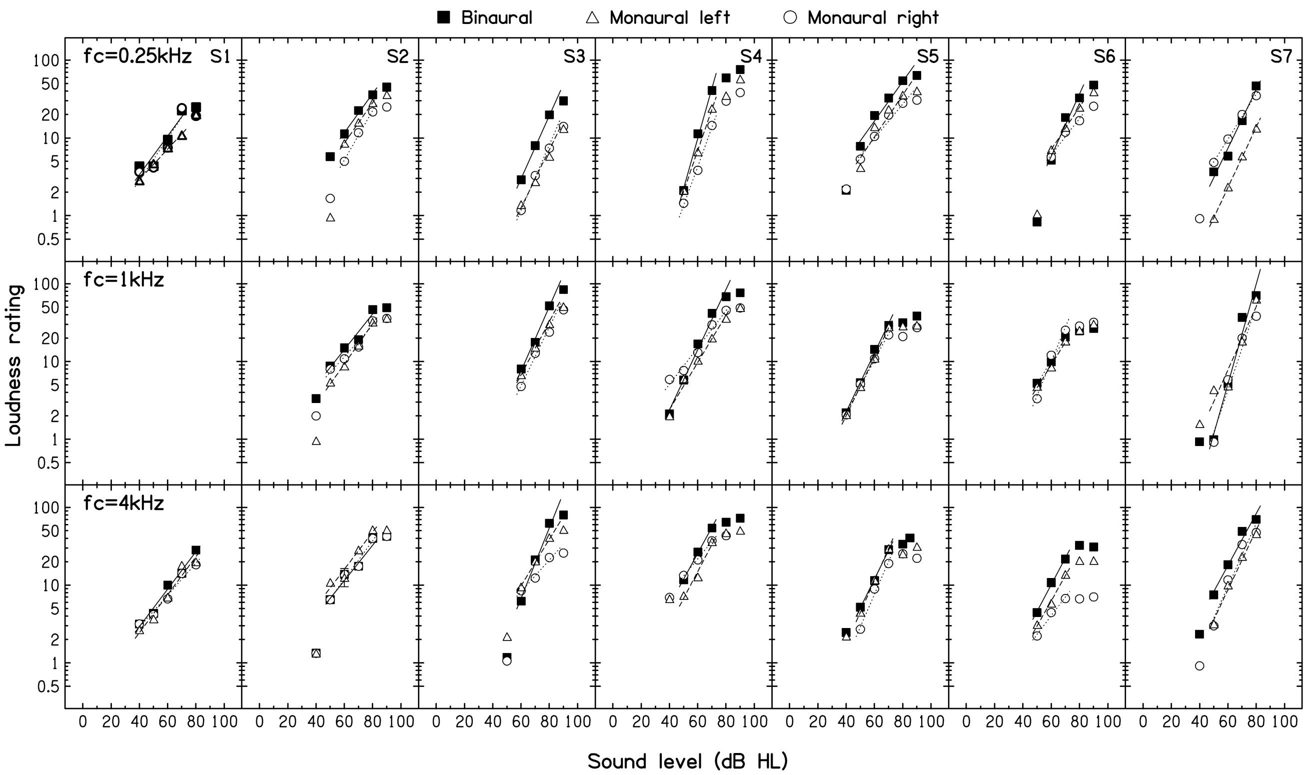

Fig. 1. Loudness functions (LFs) for seven *Cochlear* implant users in binaural (filled squares), monaural-left (open triangles), and monaural-right listening conditions (open circles). Points represent averages of subjects' 18 loudness ratings. The average standard error (SE) is 1.84, 1.52, and 1.43 rating units for center frequencies $f_c = 0.25$, 1, and 4 kHz, respectively. The total average SE = 1.60 rating units. Dispersion measures are not shown in the figure for clarity.

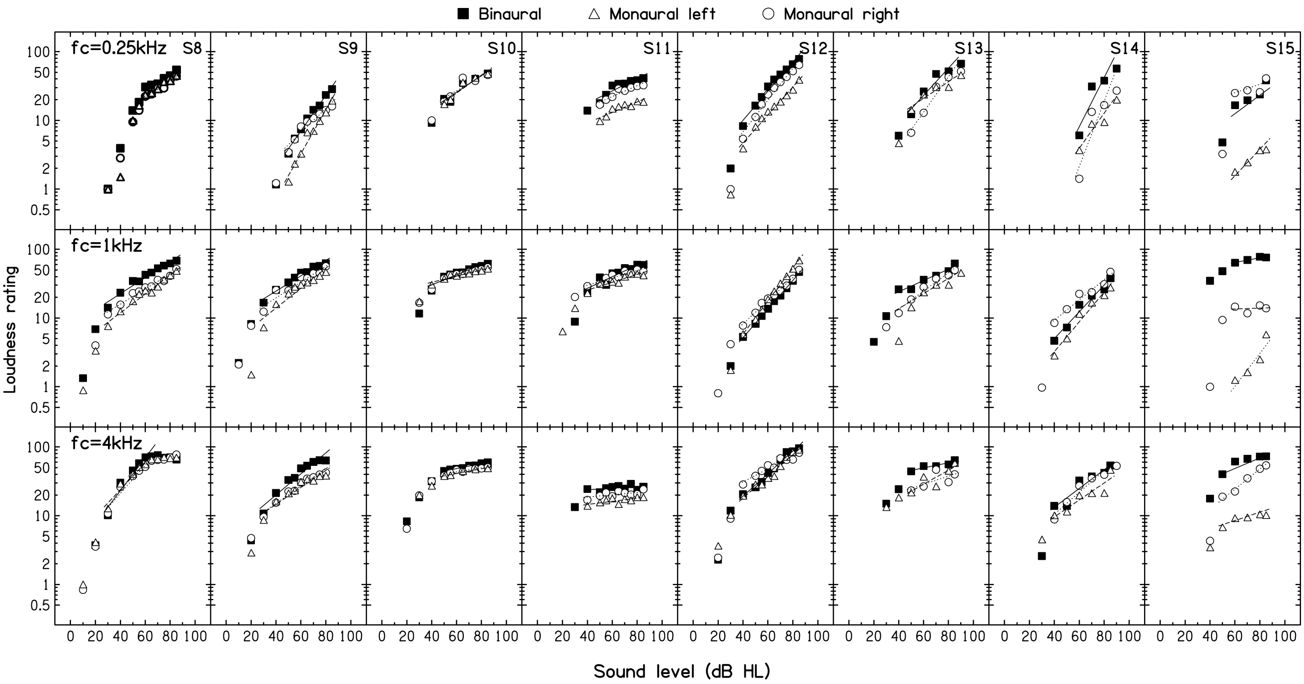

Fig. 2. The loudness functions (LFs) for eight *Advanced Bionics* implant users in binaural (filled squares), monaural-left (open triangles), and monaural-right listening conditions (open circles). Points represent averages of subjects' 18 loudness ratings. The average standard error (SE) is 2.17, 1.74, and 1.97 rating units for center frequencies $f_c = 0.25$, 1, and 4 kHz, respectively. The total average SE = 1.97 rating units. Dispersion measures are not shown in the figure for clarity.

Different symbols correspond to different listening conditions: binaural (filled squares), monaural-left (open triangles), and monaural-right (open circles). The LFs presented in Figs. 1 and 2 were calculated as the arithmetic mean of 18 subjects' loudness ratings obtained for each condition and presentation level. For clarity of figures, Figs. 1 and 2 do not show dispersion measures. For all LFs, the standard error (SE) was smaller than 3 rating units for levels of 40 dB and below, and only very exceptionally exceeded 6 units for levels of and above 50 dB. The average SE = 1.60 and 1.96 for data in Figs. 1 and 2, respectively. This SE corresponds to an average standard deviation of 6.8 and 8.3 rating units, respectively.

Results showed large variability in loudness rating curves and no systematic differences in thresholds across the conditions tested. The systematic trends were not apparent in either the shapes or slopes of the LFs, thresholds, or binaural loudness summation values. The hearing threshold, assumed to be the initial point at which any rating is assigned to the loudness of a stimulus, varied among subjects from 20 dB HL to 60 dB HL. This value was typically lower for the *Advanced Bionics* (Fig. 2) than for the *Cochlear* implant users (Fig. 1). Considering all conditions, among *Cochlear* implant users, the hearing threshold occurred at 40 dB HL (ten cases), 50 dB HL (nine cases), or in one case at 60 dB HL (Subject S2 at 0.25 kHz). For the *Advanced Bionics* implant users the threshold for sound sensation was about 20 dB lower than for the *Cochlear* implant users. In seven cases, thresholds were as low as 10 or 20 dB HL. For five subjects at 1 or 4 kHz, eight thresholds were at 30 dB HL and seven at 40 dB HL. In two conditions, thresholds were at 50 or 60 dB (Subjects S14 and S15 at 0.25 kHz).

The differences in the LFs between the *Cochlear* and *Advanced Bionics* users were notable at the threshold of comfort, i.e., level at which the LFs began to saturate. For the *Cochlear* implant users (Fig. 1), the plateau was generally observed at 70 or 80 dB HL. In contrast, for the *Advanced Bionics* users (Fig. 2) the possible effect of compression was less visible, as the LFs did not reach plateau at high HLs, but rather their slope decreased.

Parameters such as current values, pulse rate and compression activation point are generally common among devices; hence, the two CI types are likely to affect the shapes of the LFs in a similar way. Subjects S7 (Fig. 1) and S12 (Fig. 2) did not reach plateau at high intensity levels at any frequency, which may be due to effects of device compression. Four other subjects (S1, S3, and S6 in Fig. 1 and S9 in Fig. 2) did not show compression effects at certain noise-band center frequencies. Results also differed in the slope of the LF functions. The LFs among the *Cochlear* users (Fig. 1) typically showed steep slopes in the range of intensity levels below plateau seen above 70 or 80 dB HL, due to device compression. The LFs obtained from the *Advanced Bionics* users (Fig. 2) were often two-staged, being steep over a limited level range from hearing threshold to about 40 dB HL and less steep but not flat in the wider range of levels above 40–60 dB HL. In some cases, such as for Subject S11 (Fig. 2), the LF was almost flat or flat (at 4 kHz) at levels between 30 to 80 dB HL. Large flat segments of the LFs were also observed for Subjects S10 (at 1 and 4 kHz), S13 (at 4 kHz), and S15 (at 1 and 4 kHz). In summary, even though CI programming is intended to provide comparable listening conditions across electrodes and stimuli, the loudness functions in Figs. 1 and 2 show that the measured perceptual loudness proves notably different among subjects and across frequency bands.

3.2. Slopes of loudness function in cochlear implant users and normal hearing subjects

3.2.1. Slopes of loudness function in cochlear implant subjects

The simplest way of describing loudness growth shown by the LFs in Figs. 1 and 2 is to fit straight lines to the data in log-log coordinates and calculate the lines' slope coefficients. Lines were fitted to the largest linear segments of LF-consistent growth in the mid- to high-level ranges, approximately larger than 50 dB HL, to represent most common listening conditions. Due to large intersubject variability seen in data in Figs. 1 and 2, where subjects' thresholds occurred at levels between 40 and 60 dB HL (*Cochlear*), or between 10 and 60 dB HL (*Advanced Bionics*), it was not possible to set identical level ranges for slope determination for all subjects. Lines were fitted from levels starting at 40 to 60 dB HL (for *Advanced Bionics* also 30 dB in four cases), and ending at 70 or 80 dB HL. On average, the slope of LFs was determined within the ranges of 53–77 dB HL (250 Hz), 48–75 dB HL (1000 Hz), and 48–76 dB HL) for *Cochlear* implant users, and within 53–80 dB HL (250 Hz), 40–80 dB HL (1000 Hz), and 41–78 dB HL (4000 Hz) for *Advanced Bionics* implant users. Thus, despite certain differences, the fitting procedure covered most the important level range for speech (between 50 and 80 dB HL).

For *Cochlear* users (Fig. 1), lines representing loudness growth were fitted with the exclusion of the short plateaus observed above 70 or 80 dB HL. For *Advanced Bionics* users (Fig. 2), lines representing loudness growth were fitted excluding the drop in ratings close to the hearing threshold observed below 40 dB HL seen in the data for the majority of subjects. Fitted lines representing slope of LFs are drawn in Figs. 1 and 2 for a clear reference showing the level range in which the slope coefficients were calculated for each condition and the data points included in these calculations.

Slope coefficients corresponding to lines plotted in Figs. 1 and 2 are shown in the upper (*Cochlear*) and lower (*Advanced Bionics*) panels of Fig. 3, respectively. For each subject, symbols in the middle refer to the 1-kHz noise band center frequency, whereas symbols on the left and right refer to 0.25- and 4-kHz band noise, respectively. As in Figs. 1 and 2, filled squares, open triangles and circles correspond to the binaural, monaural-left, and monaural-right listening conditions. Slope coefficients were calculated using scaling allowing for a direct comparison to the sone scale. Therefore, a value of 0.6 is in agreement with average slope coefficient known for the sone scale at 1 kHz (Canévet *et al.*, 1986).

For *Cochlear* implant users (Fig. 3, upper panel), considerably large slope coefficient variability across frequencies was observed for Subjects S4 and S7, espe-

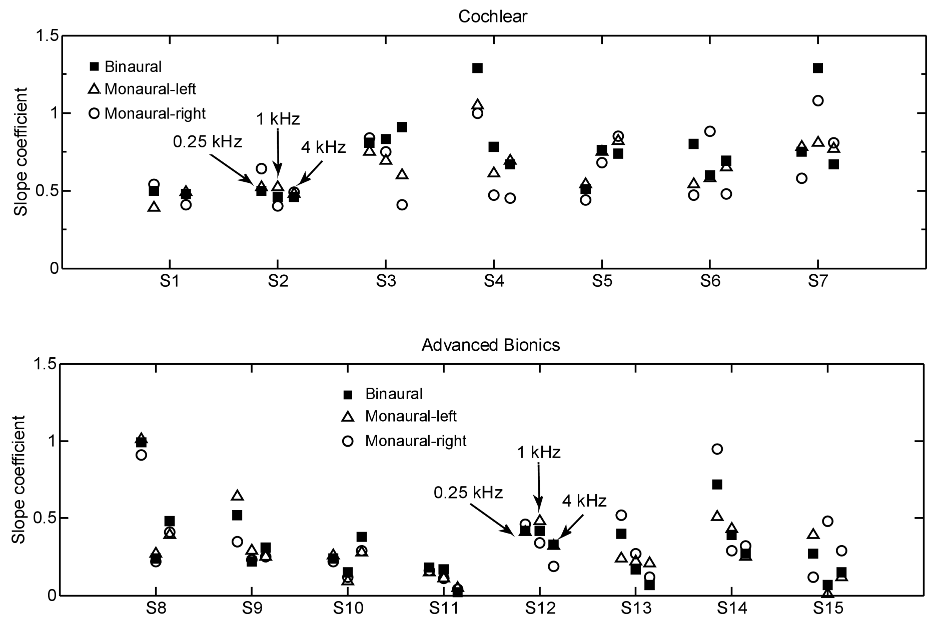

Fig. 3. Slope coefficients in binaural (filled squares), monaural left (open triangles), and monaural right (open circles) conditions at 0.25-, 1-, and 4-kHz stimulus frequencies.

cially in the binaural condition. By contrast, it can be noted that the data for Subjects S3, S5 and S6 at all frequencies showed a smaller effect on the slope in the binaural listening condition than that in monaural conditions. Thus, the data of these subjects provide some evidence to suggest that binaural listening seems to create a more stable perception of loudness than monaural conditions. It is not true, however, for Subjects S4 and S7.

For *Advanced Bionics* implant users (Fig. 3, lower panel), slope coefficients were usually smaller than for *Cochlear* implant users, which corresponds to the level seen in Figs. 1 and 2 with less steep growth of loudness. Within this group, similar slopes were observed in monaural and binaural conditions for Subjects S10–S13. Slope coefficients, however, were generally lower than 0.4–0.5 and for Subject S11 lower than 0.2, which corresponds to flat LFs shown in column four in Fig. 2. For example, for this subject, at 4 kHz within the HL range of 40–85 dB the increase in ratings only ranged from 24 to 26, 15 to 19, or 19 to 24 points for binaural, monaural left and monaural right conditions, respectively. By comparison, within the same HL range, the average increase in ratings by Subject S8 at 4 kHz varied between 20 and 40 rating points depending on listening condition (Fig. 2, column one, lower panel). Therefore, there was essentially no increase in loudness for all listening conditions reported by Subject S11. This is possibly related to the specific electrode configuration of the cochlear implant worn by Subject S11, with only eight electrodes active within the frequency range from 372 to 6299 Hz, and only two electrodes (tuned to 3330 and 6299 Hz) in the high frequency range. This may also explain why the LFs of this subject were flat at 4 kHz, whereas they displayed small but larger growth at 4 kHz than at 0.25 and 1 kHz. However, it cannot be excluded that Subject S11 simply did not do very inefficiently in the rating procedure.

3.2.2. Comparison of slopes of loudness functions in cochlear implant users and in normal hearing subjects

This section examine the extent to which the LFs of CI users correspond to loudness growth in NH listeners, such as represented by the sone scale. Therefore, the LFs' slope coefficients seen in Fig. 3 were compared with slopes obtained from the control group of five NH subjects, and with data available in the literature. The NH subjects (females, ages 37 to 59 years) had absolute thresholds of less than 10 dB HL at all audiometric frequencies and no history of hearing disorders.

Another problem investigated in this section is that the rating procedure used in this study might be affected by certain nonlinearity due to the rating range restricted to 0–100 points (compressed and distorted shape of the LFs, floor and ceiling distortions). To validate the loudness rating method two testing procedures were used in the NH group: absolute magnitude estimation (AME) of loudness (Stevens, 1955; 1957), and loudness rating identical to that used for the CI users. The AME procedure was implemented in order to standardize the rating method conducted on a fixed scale of 0–100 points following the method commonly accepted for loudness scaling in which positive numbers are freely assigned to perceived loudness. Subjects were specifically instructed to use any numbers that were deemed appropriate (whole numbers, decimals, fractions) and were informed that the set of numbers they are allowed to use has an unlimited range. Other than the numbers being greater than zero, no further constraints were imposed. In each subject's measurement session, the AME scaling was conducted before the loudness rating to avoid any possible influence of using fixed 0–100 scale on the responses in the AME procedure. Loudness assessment with NH subjects was carried out for HLs ranging from 10 to 90 dB.

Measurements were conducted binaurally at the noise-band center frequencies of 0.25, 1, and 4 kHz. Each measurement was repeated nine times, and the LFs were calculated from the subject's last six responses. Thus, the results of scaling by normal hearing individuals shown in Fig. 4 are averaged over 30 measurements per point.

For all noise-band center frequencies, the LFs in Fig. 4 are approximately linear at HLs above 30 or 40 dB and display a drop at the hearing threshold. The slope coefficients of lines fit to the data (fitting range 20–90 dB HL) are 1.08 and 0.73 (at a 0.25-kHz noise-band center frequency), 0.96 and 0.56 (at 1 kHz), and 0.84 and 0.50 (at 4 kHz), for AME and the rating procedures, respectively. Thus, as might be expected scaling with the AME provided steeper LFs than the rating procedure with slope coefficients larger by 50–70%. Despite the difference in slope coefficients, the LF shapes obtained by the AME and loudness ratings were similar. The difference in slope between the AME and the rating procedures is relatively small given the intersubject variability of LF slopes, a finding that has been previously observed (e.g., Stevens, Guirao, 1964; de Barbenza *et al.*, 1970; Hellman, 1981; Hellman, Meiselman, 1988; Zwicker *et al.*, 1957; Zwicker, Zwicker, 1991). Thus, the rating procedure used for the *Cochlear* implant users is a reasonable measure of loudness growth, fairly analogous to the sone scale.

As the control group in the study was limited to five NH subjects, the slope coefficients of the LFs obtained for the cochlear implant subjects were also compared with data available from other studies. Figure 5 shows distributions of LF slope coefficients of *Cochlear* and *Advanced Bionics* implant users compared with distributions of slope coefficients for NH persons pooled together from Stevens and Guirao (1964), de Barbenza *et al.* (1970), Hellman (1981), and Hellman and Meiselman (1988). Data from implant users for

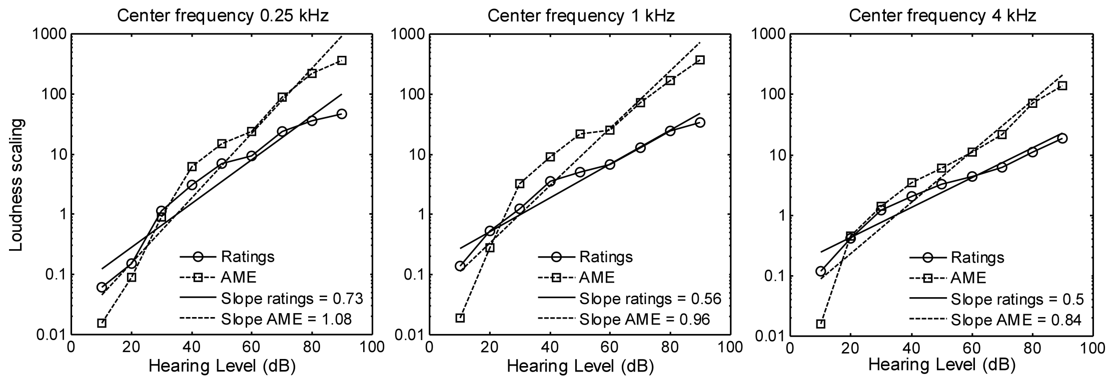

Fig. 4. Results of absolute magnitude estimation (AME) and rating procedures for five normal-hearing persons. Binaural presentation. Data to calibrate rating procedure.

0.25-, 1-, and 4-kHz noise-band center frequencies from the present study were pooled together as the data reported by Hellman (1981) showed that LFs determined at those frequencies for NH listeners are similar. A comparison of the results reveals that the distribution of slope coefficients from the *Cochlear* implant users is generally aligned with data from NH listeners. The LF slope coefficients for the *Advanced Bionics* implant users are in many instances smaller than that reported in the literature for NH persons. It should be recalled here that within the NH subjects (Fig. 4) slope coefficients were observed smaller in rating procedure than in AME procedure. This means that differences of *Advanced Bionics* implant users and NH persons are actually smaller than that seen in Fig. 5.

As it was earlier mentioned, slope coefficients were calculated for a HL range of approximately 50–80 dB. It is sensible to make a comparison with slopes of NH subjects in that the overall policy is to design implants that would restore as much as possible, normal hearing properties. Thus, selecting this level range was in agreement with the linear part (in log-log coordinates) of Stevens' power law, with no nonlinearity seen for at threshold levels. It has to be stressed here that the *Advanced Bionics* LFs resemble the sone scale quite well, due to fact that their ratings drop at threshold.

With no further experiments, it is now difficult to judge how the differences in slopes of LFs discussed in this section affect hearing ability of cochlear implant users in practice. The primary optimization of the devices is stressed for speech. If so, the plateau at highest levels corresponding to the comfort level seen in *Cochlear* implant users in 50% of cases has likely no meaning for speech perception (further increase in loudness is not important for speech). Formally, it would be accounted for by smaller slope coefficients than those presented in Fig. 5 for *Cochlear* implant users (as fitting would include plateaus).

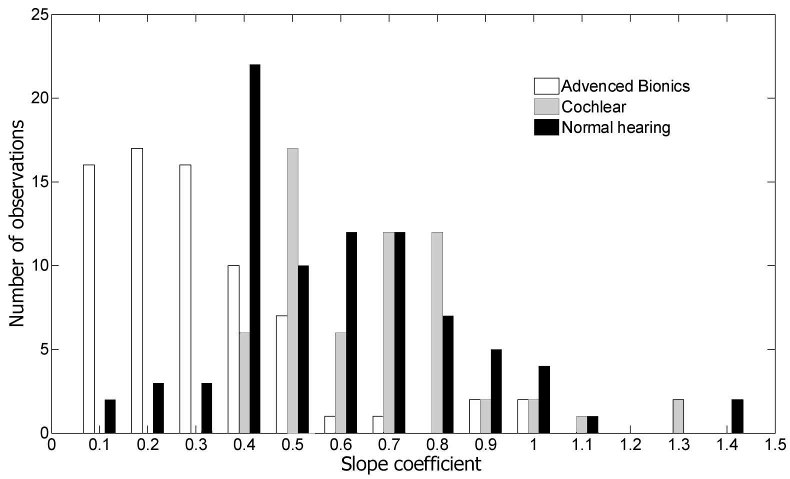

Fig. 5. Distribution of slope coefficients for LFs of *Cochlear* and *Advanced Bionics* implant users in comparison with distribution of slope coefficients in normal-hearing subjects. Data from the present study include all tested frequencies and listening conditions. Empty symbols represent slopes for the AME procedure; closed symbols represent slopes for rating procedure obtained from the control group of normal-hearing persons. Data from literature are pooled results from Stevens and Guirao (1964), de Barbenza *et al.* (1970), Hellman (1981), and Hellman and Meiselman (1988) studies.

Another issue is that rating procedure is a rather difficult task for subjects and one cannot be sure whether flat LFs seen in some cases for the *Advanced Bionics* subjects (Subject S10, S11 and S15) really represent their loudness percept or perhaps result from their inability to provide correct judgments.

3.3. Binaural loudness summation in bilateral cochlear implant users

The effect of binaural loudness summation among cochlear implant users can be estimated based on the fit to the LFs obtained in binaural and monaural listening conditions. To assess the gain obtained in the binaural condition, differences in HLs between monaural-left or monaural-right and binaural listening conditions were calculated for identical loudness ratings. This approach assumes rating stability regardless of the mode of signal presentation. It limits the influence of inter-subject differences and possible nonlinearities in how the ratings are assigned. Some problems in using differences in HLs arise for shallow LFs. In such cases, calculations became inaccurate and led to somewhat inconsistent results.

Lines fitted to the LFs shown in Figs. 1 and 2 in different listening conditions often show a clear effect of binaural loudness summation but in many instances are not parallel to each other. Clear examples of loudness summation can be noticed in LFs obtained for Subjects S2–S5 at 0.25 or 1 kHz, S8 at 1 kHz or S9 and S11 at 4 kHz. Dominance by one ear was seen for some subjects (e.g. S1, S7 and S9 at 0.25 kHz); other subjects showed little or no binaural summation (e.g. S5, S6, and S10 at 1 kHz). In some cases (S2 at 4 kHz and S4 at 1 kHz), the binaural condition led to lower loudness ratings than did monaural conditions. Because of such complexity in the results, the differences in level among fitted lines were calculated both in low and high HLs for each subject and each noise-band center frequency. The results are summarized in Tables 3 and 4 for *Cochlear* and *Advanced Bionics* devices, respectively. Particular columns (from the left

Table 3. Binaural to monaural-left and binaural to monaural-right level differences showing effect of binaural loudness summation in the *Cochlear* implant users. 'Low HL' refers to low hearing levels and 'High HL' to high hearing levels. 'Average' represents average level gain for low HL and high HL.

Subject	Low HL		High HL		Average		Left–Right Ear
	Bin – Left	Bin – Right	Bin – Left	Bin – Right	Bin – Left	Bin – Right	
	[dB]						
	250 Hz						
S1	3.2	2.7	11.4	0.2	7.3	1.5	5.8
S2	5.2	11.1	4.4	6.9	4.8	9.0	−4.2
S3	10.0	9.6	12.3	8.8	11.1	9.2	1.9
S4	0.7	4.4	5.2	10.1	2.9	7.2	−4.3
S5	8.1	8.4	6.1	14.1	7.1	11.2	−4.2
S6	−3.3	−0.8	6.2	13.4	1.4	6.3	−4.9
S7	13.3	−7.1	12.4	2.0	12.9	−2.5	15.4
	1000 Hz						
S2	8.3	3.2	4.8	7.3	6.5	5.2	1.3
S3	1.4	5.0	6.7	7.6	4.0	6.3	−2.2
S4	−0.2	−13.9	11.0	11.7	5.4	−1.1	6.5
S5	1.3	0.4	1.5	3.8	1.4	2.1	−0.7
S6	1.6	3.4	2.0	−3.1	1.8	0.2	1.6
S7	−10.3	3.2	7.1	5.2	−1.6	2.4	−4.0
	4000 Hz						
S1	3.1	0.2	2.7	7.2	2.9	3.7	−0.9
S2	−45.7	0.0	−7.1	−2.3	−6.4	−1.1	−5.3
S3	−5.4	−4.2	7.6	26.5	1.1	11.1	−10.0
S4	7.1	−1.8	6.4	8.1	6.8	3.2	3.6
S5	1.3	5.7	−0.6	3.3	0.4	4.5	−4.1
S6	5.8	12.3	7.0	20.8	6.4	16.5	−10.2
S7	9.4	8.4	5.5	3.1	7.4	5.7	1.7

Table 4. Binaural to monaural-left and binaural to monaural-right level differences showing effect of binaural loudness summation in the *Advanced Bionics* implant users. Dashes represent conditions for which level difference calculations were not reliable due to small slope coefficients in lines fitted to the LFs. Other details as in Table 3.

Subject	Low HL		High HL		Average		Left–Right Ear
	Bin – Left	Bin – Right	Bin – Left	Bin – Right	Bin – Left	Bin – Right	
	[dB]						
	250 Hz						
S8	9.4	11.0	5.8	6.7	7.6	8.8	−1.2
S9	12.2	−3.3	5.7	13.2	8.9	5.0	3.9
S10	1.7	−3.1	−0.9	1.0	0.4	−1.0	1.4
S11	–	8.9	–	13.8	–	11.3	–
S12	16.2	7.3	18.1	3.6	17.2	5.5	11.7
S13	−2.5	11.8	19.3	3.6	8.4	7.7	0.7
S14	12.2	12.2	22.4	6.1	17.3	9.2	8.1
S15	–	–	–	–	–	–	–
	1000 Hz						
S8	21.4	phantom.13.0	13.8	16.9	17.6	15.0	2.6
S9	19.3	8.8	8.0	7.4	13.6	8.1	5.5
S10	−8.6	−1.1	20.1	9.7	5.7	4.3	1.4
S11	5.0	−8.6	26.1	18.0	15.5	4.7	10.8
S12	−2.1	−12.0	−8.2	−1.1	−5.2	−6.5	1.3
S13	22.1	19.3	11.0	2.2	16.5	10.7	5.8
S14	8.8	−18.1	4.4	−3.5	6.6	−10.8	17.3
S15	–	–	–	–	–	–	–
	4000 Hz						
S8	−3.1	0.3	4.9	5.9	0.9	3.1	2.2
S9	8.7	7.8	20.7	17.8	14.7	12.8	1.9
S10	20.0	13.9	22.4	21.4	21.2	17.7	3.5
S11	–	–	–	–	–	–	–
S12	1.8	−22.0	3.8	11.7	2.8	−5.2	8.0
S13	–	–	18.5	–	–	–	–
S14	11.5	6.4	16.1	−0.2	13.8	3.1	10.7
S15	–	25.3	–	8.6	–	17.0	–

to the right) show differences in level between binaural and either monaural conditions at low HLs, high HLs, as well as the average line shift in level.

For the *Cochlear* implant users binaural loudness summation is seen either at low or at high HLs in a total of about half the cases listed in Table 3. These are all conditions in which differences between binaural and monaural-left or monaural-right conditions are similar. In certain cases, the gain resulting from binaural listening can be as large as 8–12 dB (Subjects S3 and S5 at 0.25 kHz, S4 at high HL, 1 kHz, and 4 kHz, and Subject S7 at low HL, 4 kHz). It should be noted, however, that binaural loudness summation is best considered when the monaural loudness is balanced across ears, that is, when the binaural condition versus either monaural-right or monaural-left yields similar level differences. In all cases in which an imbalance occurred between monaural-left and -right conditions, it is likely that subjects used their better ear in the binaural condition. In such a situation, the difference in level between the binaural condition and the better ear likely represents the gain obtained from binaural listening. Thus, to assess gain in binaural listening, the smaller level difference in the respective columns of Tables 3 and 4 should be considered. Average shift in LFs (right column of Table 3) shows loudness summation in which gain level ranged from 3 to 9 dB for eight conditions. In another eight conditions, the effect of binaural loudness summation was weak (level gain 0–2 dB), and in four conditions the effect was negative. Regardless of the variability in Table 3, an overall conclusion can be drawn, that information from the two ears is combined and resulted in binaural summation.

Data for *Advanced Bionics* users (Table 4) show somewhat mixed results because of the small slope coefficients of the LFs (see Figs. 2, 3, and 5). For this reason, level differences larger than 25 dB were considered outliers and were not taken into account. The arbitrary limit of 25 dB was selected considering that gain level in binaural listening in normal-hearing subjects does not exceed 10 dB at levels of 80–90 dB SPL, and equals only a few dB at low levels of 20–30 dB SPL. Among *Advanced Bionics* users (Table 4) a strong effect of binaural loudness summation was observed for Subjects S8 and S9. Some effect was seen for Subjects S11 and S13. A negative effect was observed for Subject S10 especially at 0.25 kHz, and for Subjects S12 and S14 at 1 kHz. Within this group only Subjects S8 and S10 displayed fully balanced LFs between left and right ear monaural conditions.

In conclusion, for the majority of subjects and in most conditions considered either strong or some effect of binaural summation was observed. It may be expected that proper and clearly seen binaural summation is greatly dependent on balanced programming of cochlear implants in bilaterally implanted listeners.

4. Discusion

The purpose of this study was to investigate monaural and binaural loudness growth and the effects of binaural loudness summation in adult bilateral CI recipients. It was found that LFs in acoustic stimulation varied among the tested group of subjects who used *Cochlear* or *Advanced Bionics* implants. Binaural loudness summation depends on stimulus presentation level, noise-band center frequency, and the cochlear implant used. A substantial spread of LF slopes is seen among the subjects. In *Cochlear* implant users the slope coefficient values are within the range of the slopes observed for NH listeners whereas for *Advanced Bionics* implant users the LFs are more shallow (Fig. 5).

Data shown in Figs. 1 and 2 should be discussed in relation to the implant settings for particular subjects. As listed in Table 1 for most subjects with the *Cochlear* device, the Advanced Combination Encoder (ACE) sound coding strategy was used in combination with the Freedom processor (except for Subjects S3 and S5, who used SPrit and ESPrit 3G processors). All subjects used default factory sensitivity (a value of 10) and similar overall volume settings. The so-called T-SPL and C-SPL parameters were set to default values, which define the T (threshold) and C (comfort) levels, respectively, at 25 dB and 65 dB SPL. Due to the calibration of the loudspeaker system used for the study these values corresponded to about 20 and 60 dB HL at 0.25- and 4-kHz noise-band center frequencies, or to 15 and 55 dB HL at a 1-kHz center frequency. Thus, the acoustical dynamic range of the implants was set to about 40 dB. This is generally consistent with the dynamic range of loudness growth seen in Fig. 1, extending from the level associated with the minimum loudness rating close to the hearing threshold up to the activation of compression. This dynamic range is about 30–40 dB for almost all subjects.

Another issue is that the ACE devices used on-line determined spectral maxima to stimulate electrodes of corresponding channel frequency. The number of spectral maxima used varied between 5 and 12, depending on the subject. However, when narrow noise bands were applied, the processor likely selected channel frequencies within particular noise bands. The octave-wide bands of noise extended from 165–350 Hz, 680–1280 Hz, and 2700–5230 Hz, for band center frequencies of 0.25, 1, and 4 kHz, respectively. This most likely resulted in activation of two electrodes within the CI electrode array (188–313 Hz and 313–438 Hz) by the low-frequency noise band of 0.25 kHz. By contrast, for the higher frequencies of 1 kHz and 4 kHz it is likely that up to five electrodes were activated (covering the frequency range of 688–1313 Hz or 2688–5313 Hz). Furthermore, in this high-frequency range, there may also be activation of two more electrodes (below and above) resulting from high- and low-frequency slopes of filtered noise, leading to seven consecutive electrodes being activated. As can be seen in Fig. 1 there is no apparent systematic effect of stimulus type, thus, activation of a lower vs larger number of electrodes does not seem to have an effect on LFs and binaural summation as measured here.

All the *Advanced Bionics* implants worn by Subjects S8–S15 were equipped with *Harmony* processors (see Table 2) and programmed according to the Auria +AGC2 scheme, which involved strict compression after exceeding a specific input sound pressure level. The linear dynamic range below the compression activation point depended on the value of the Input Dynamic Range (IDR) parameter, which was set to 60–80, creating usable stimulation growth in the range of about 50 dB. As the LFs shown in Fig. 2 extend over a wider dynamic range characterized by a flatter slope and less noticeable compression than in the case of *Cochlear* implant users presented in Fig. 1, the LFs was also calculated as a function of electrode current. In the vast majority of cases, it was found that the LF shape expressed as a function of the current resembles that obtained for the sound pressure levels.

While in normal-hearing listeners loudness is often interpreted in terms of auditory filter bank and excitation along the basilar membrane (Fletcher, 1953; Zwicker, Scharf, 1965; Zwicker, 1958), in CI recipients this mechanism is obviously different because the cochlea is bypassed and the electrical charge directly stimulates the auditory nerve. In normal-hearing persons, loudness growth depends on the degree of neural activity in the auditory system. The electric charge resulting from the electrode current amplitude and pulse width is equivalent to that in cochlear implant users. Widening this pulse for the same current amplitude results in more electrical charge delivered to the electrode and increases the perceived loudness. Although the perceived loudness depends on the purely technical parameters of the cochlear implant, the pattern and variability of nerve survival play a critical role. Stimulation rate may also influence and explain the difference in loudness judgments obtained from the CI recipients. Certain studies (e.g. Fu, 2005; Sanpetrino, Smith, 2006) have suggested that optimal speech recognition performance requires restoring normal-hearing LF in cochlear implant users. Results of this study provided information about the growth of LFs with free-field acoustical stimulation, which shows that slopes of LFs are close to slopes common in normal-hearing subjects.

The effect of binaural loudness summation varies across subjects and differs at low and high stimulus intensity levels. For binaural loudness summation, differences in the pattern and viability of nerve survival between the left and right ear are likely to be significant. For example, Subject S4 (Fig. 1) not only expressed significant difference in loudness judgments between monaural and binaural conditions but also difference between monaural right and left conditions. In Subject S4's case, identical hardware and exactly the same parameter settings in both implants resulted in a large difference in loudness judgment between ears, which might be also caused by a difference in electrode insertion depth in the left and right ear. Another factor influencing binaural loudness summation is possibly the auditory deprivation introduced by a profound hearing loss lasting for many years. Long-term hearing loss may interfere with central mechanisms integrating information coming from the left and right ears (Litovsky *et al.*, 2010).

It is important to recognize that current signal processing in cochlear implants and device fitting were developed for monaural applications, with the assumption that CI devices work independently of each other. From a technical point of view, asynchronous stimulation in two devices may result in uncorrelated neural pulses which are unlikely to provide a useful cue such as interaural differences needed for localization but also influencing binaural loudness summation. Variability in the results presented in this study is not unusual for research projects involving cochlear implant users. However, it suggests that improvement in bilateral fitting of cochlear implants may require a signal processing stage devoted to simultaneous control of implants in the two ears.

Another factor causing difference in loudness perception between the left and right ear is electrode configuration and location. For an implant system, bandpass filters assigned to different locations along the electrode array do not process the same frequency ranges in all CI recipients. The CI is not able to stimulate the cochlea, preserving the original monotonic tonotopic frequency organization of the normally functioning ear. For example, Fu (2005) have shown that low-frequency stimulation may be processed differently at different electrode locations.

Despite the intersubject differences, the described experiment showed that loudness perceived by the CI recipients depends both on sound pressure level and noise-band center frequency, which confirms the need for frequency-specific mapping functions postulated by Hoth (2007). At present, the loudness mapping function implemented in implant devices is set between the threshold (T) and the comfortable loudness (C) levels. Speech level is targeted to fall within this range of T and C levels, over the entire frequency spectrum. Typically, mapping processes do not take into account the varying temporal and spatial patterns of the speech processor output, nor do they take into account whether devices are unilaterally or bilaterally implanted. This may lead to uncontrolled loudness balance.

5. Conclusions

The loudness functions (LFs) obtained in this study with free-field acoustical stimulation for fifteen bilaterally implanted subjects can be summarized by the following conclusions:

1) Monaural and binaural LFs show large variability among implant users in slope and binaural loudness summation.
2) A large difference was seen in the effect of stimulus frequency on LF slopes. While for many subjects the slopes were similar across stimulus frequencies, for some subjects they differed significantly. Binaural conditions were more likely to produce more similar LF slopes at different stimulus frequencies than monaural conditions.
3) Binaural listening summation was observed at different stimulus frequencies. However, in many cases, when listeners had a dominant ear there was no clear loudness summation; in fact, subjects may have assigned higher loudness ratings in the monaural condition than in the binaural condition.
4) Variability in the LF slopes and inconsistent binaural loudness summation clearly suggest that device mapping for bilateral implantation should include careful consideration of binaural effects, such as loudness summation.

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References

1. Anweiler A.K., Verhey J.L. (2006), *Spectral loudness summation for short and long signals as a function of level*, J. Acoust. Soc. Am., **119**, 5 Pt 1, 2919–2928.
2. de Barbenza C.M., Bryan M.E., Tempest W. (1970), *Individual loudness functions*, J. Sound Vib. **11**, 4, 399–410.
3. Canévet G., Hellman R.P., Scharf B. (1986), *Group estimation of loudness in sound fields*, Acustica, **60**, 4, 277–282.
4. Fletcher H. (1953), *Speech and Hearing in Communication*, Van Nostrand Co., New York, 2nd Ed.
5. Florentine M., Buus S., Poulsen T. (1996), *Temporal integration of loudness as a in function of level*, J. Acoust. Soc. Am., **99**, 3, 1633–1644.
6. Fu Q.-J. (2005), *Loudness growth in cochlear implants: effect of stimulation and electrode configuration*, Hear. Res., **202**, 55–62.
7. Garnier S., Micheyl C., Arthaud P., Berger-Vachon C., Collet L. (1999), T*emporal loudness integration and spectral loudness summation in normal-hearing and hearing- impaired listeners*, Acta Otolaryngol., **119**, 2, 154–157.
8. Garnier S., Micheyl C., Arthaud P., Collet L. (2000), *Effect of frequency content on in categorical loudness normalization*, Scand Audiol., **29**, 4, 253–259.
9. Hellman R.P. (1981), *Stability of individual loudness functions obtained by magnitude estimation and production*, Perc. Psych., **29**, 1, 63–70.
10. Hellman R.P., Meiselman C. (1988), *Prediction of individual loudness exponents from cross-modality matching*, J. Speech. Hear. Res., **31**, 4, 605–615.
11. Hoth S. (2007), *Indication for the need of flexible and frequency specyfic mapping functions in cochlear implant speech processors*, Eur. Arch. Otorhinolaryngol., **264**, 2, 129–138.
12. Leibold L.J., Jesteadt W. (2007), *Use of perceptual weights to test a model of loudness summation*, J. Acoust. Soc. Am., **122**, 3, EL69.
13. Litovsky R.Y., Jones G.L., Agrawal S., van Hoesel R. (2010), *Effect of age at onset of deafness on binaural sensitivity in electric hearing in humans*, J. Acoust. Soc. Am., **127**, 1, 400–414.
14. Marks L.E. (1978), Binaural sumation of the loudness of pure tones, J. Acoust. Soc. Am., **64**(1), 107–113.
15. Marks L.E., Florentine M. (2011), *Measurment of loudness. Part I. Methods, problems and pitfalls*, [in:] *Loudness*, Florentine M., Popper A.N., Fay R.R. [Eds.], pp. 57–88, Springer, New York.
16. McKay C.M., Henshall K.R., McDermott H.J. (2003), *A practical method of predicting the loudness of complex electrical stimuli*, J. Acoust. Soc. Am., **113**, 4 Pt 1, 2054–2063.
17. McKay C.M., McDermott H.J. (1998), *Loudness perception with pulsatile electrical stimulation: the effect of interpulse interval*, J. Acoust. Soc. Am., **104**, 2 Pt 1, 1061–1074.
18. McKay C.M., McDermott H.J., Clark G.M. (1994), *Pitch percepts associated with amplitude-modulated current pulse trains in cochlear implantees*, J. Acoust. Soc. Am., **96**, 5 Pt 1, 2664–2673.
19. McKay C.M., Remine M.D., McDermott H.J. (2001), *Loudness summation for pulsatile electrical stimulation of the cochlea: effects of rate, electrode separation, level and mode of stimulation*, J. Acoust. Soc. Am., **110**, 3 Pt 1, 1514–1524.
20. Sanpetrino N.M., Smith R.L. (2006), *The growth of loudness functions measured in cochlear implant listeners using absolute magnitude estimation and compared using Akaike's information criterion*, Conf. Proc. IEEE Eng. Med. Biol. Soc., **1**, 164–1644.
21. Stevens J.C., Guirao M. (1964), *Individual loudness functions*, J. Acoust. Soc. Am., **36**, 11, 2210–2213.
22. Stevens S.S. (1955), *The measurement of loudness*, J. Acoust. Soc. Am., **27**, 5, 815–829.
23. Stevens S.S. (1957), *The direct Estimation of Sensory Magnitudes – Loudness*, Am. J. Psychol. **69**, 1, 1–25.
24. Tong Y.C., Clark G.M. (1986), *Loudness summation, masking and temporal interaction for sensations produced by electrical stimulation of two sites in the human cochlea*, J. Acoust. Soc. Am., **79**, 6, 1958–1966.
25. Vaerenberg B., De Ceulaer G., Szlavik Z., Mancini P., Buechner A., Govaerts P.J. (2014), *Setting and reaching targets with computer-assisted cochlear implant fitting.* Hindawi Publishing Corporation, The Scientific World Journal, **2014**, article ID 646590, doi: 10.1155720147646590.
26. Zeng F.-G., Shannon R.V. (1994), *Loudness-coding mechanisms inferred from electric stimulation of the human auditory system*, Science, **264**, 564–566.

27. Zeng F.-G., Shannon R.V. (1997), *Loudness of simple and complex stimuli in electric hearing*, Annals of Otology, Rhinology & Laryngology, Suppl., **166**, 235–238.

28. Zeng F.-G., Shannon R.V. (1999), *Psychophysical laws revealed by electric hearing*, Neuro Report, **10**, 9, 1931–1935.

29. Zwicker E. (1958), *On psychological and methodological bases of loudness* [in German: *Über psychologische und metodische Grundlagen der Lautheit*], Acoustica, **8**, 1, 237–258.

30. Zwicker E., Flottorp G., Stevens S.S. (1957), *Critical band width in loudness summation*, J. Acoust. Soc. Am., **29**, 5, 548–557.

31. Zwicker E., Scharf B. (1965), *A model of loudness summation*, Psychol. Rev., **72**, 1, 3–26.

32. Zwicker E., Zwicker U.T. (1991), *Dependence of binaural loudness summation on interaural level differences, spectral distribution, and temporal distribution*, J. Acoust. Soc. Am., **89**, 2, 756–764.

33. Zwislocki J.J. (1969), *Temporal summation of loudness: an analysis*, J. Acoust. Soc. Am., **46**, 2, 431–441.