Sensitivity to interaural level difference and loudness growth with bilateral bimodal stimulation

Running title: ILD sensitivity with bilateral bimodal stimulation
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Abstract

The interaural level difference (ILD) is an important cue for the localization of sound sources. The sensitivity to ILD in 10 users of a cochlear implant (CI) in the one ear and a hearing aid (HA) in the other severely impaired ear was measured. For simultaneous presentation of a pulse train on the CI side and a sinusoid on the HA side the just noticeable difference (JND) in ILD and loudness growth functions were measured. The mean JND for pitch-matched electrical and acoustical stimulation was 1.7 dB. A linear fit of the loudness growth functions on a dB versus μA scale shows that the slope depends on the subject’s dynamic ranges.
1 Introduction

The interaural level difference (ILD) is an important cue for the localization of sound sources. The perception of ILDs in normal hearing subjects has been described extensively and recently also results for ILD perception in bilateral cochlear implant users have become available. However, ILD perception in subjects using a contralateral bimodal system, i.e., a cochlear implant (CI) in the one ear and a hearing aid (HA) in the other has not received much attention yet.

Two main factors play a role in assessing the utility of ILDs for sound localization in users of a contralateral bimodal system, namely sensitivity to ILD and bilateral loudness growth. By determining the just noticeable difference (JND) in ILD, we can assess whether sensitivity to ILDs is high enough to interpret real-life ILD cues. By assessing bilateral loudness growth we can assess whether the loudness mapping in current CI speech processors interferes with ILD perception.

Another factor that plays a role is the frequency-to-place mapping. In current CI speech processors, signals are commonly processed in several frequency bands. Each band is then assigned to a certain electrode. In clinical practice the correct tonotopic assignment, which differs per patient, is disregarded. Therefore when a narrow band sound is acoustically presented to a bimodal system user, it is likely to be presented to different places in each cochlea. While it has been shown that ILDs can still be detected when such a frequency mismatch is present [Francart and Wouters, 2007], it does degrade detection performance and may have an adverse effect on the integration of sounds between ears.

While measures of sensitivity to the basic localization cues (interaural level
difference and interaural time difference) are not yet available for users of bilateral bimodal hearing systems, several publications report on localization performance [Ching et al., 2001, Tyler et al., 2002, Seeber et al., 2004, Dunn et al., 2005]. Because measurement methods differ a lot across studies, it is hard to compare or summarize the results. Overall, most subjects can do side discrimination or lateralization using a bimodal system and only a small fraction of the subjects can do more complex localization tasks. Performance using clinically fitted bimodal systems is generally very limited.

Zeng and Shannon [1992] assessed bimodal loudness growth in three auditory brainstem implant subjects. One subject had normal hearing in the non-implanted ear, the other subjects had a 40 to 50 dB flat loss at all audiometric frequencies. Loudness growth was measured by sampling equal loudness points between the left and right ear at regular intervals of the total dynamic range. The acoustic stimulus was presented continuously and the electric stimulus was a series of short bursts presented once a second. The subject had to adjust the loudness of the electrical stimulus to the equal loudness point. When plotted on a dB versus $\mu$A scale, the loudness growth functions for all three subjects were linear and their slope depended on the dynamic range (DR) of both the acoustical and electrical part.

Eddington et al. [1978] also found a linear dB versus $\mu$A relationship with a single subject. Dorman et al. [1993] came to the same conclusion using one CI subject with a pure tone threshold of 25 dBHL at the test-frequency (250 Hz) in the non-implanted ear and a slightly different procedure. The results for CIs therefore seem to correspond to the more extended results for auditory brainstem implants.

Loudness growth using only a CI has been measured by letting subjects estimate the loudness of several stimuli on a scale. Procedures differ between
studies, but commonly the perceived loudness varies exponentially as a
function of linear current. [Zeng and Shannon, 1994, Gallego et al., 1999,
Chatterjee et al., 2000, Fu, 2005]

JNDs in ILD in normal hearing subjects are well known. Mills [1960]
presented 5 subjects with a reference with no ILD, followed by a stimulus with
an ILD introduced. The stimuli were pure tones. Using the method of
constant stimuli, the JND in ILD was determined from half the interquartile
separation of the psychometric curves for each subject. JNDs were around
1 dB for 1000 Hz, somewhat smaller for lower frequencies and around 0.5 dB
for frequencies higher than 1000 Hz.

Yost and Dye [1988] measured JNDs in ILD for pure tones and different
reference signals at 75 % correct, using a linear fit of the psychometric curve.
For the reference at ILD = 0 dB JNDs were found of approximately 0.75, 0.85,
1.20, 0.70 and 0.73 dB for 200 Hz, 500 Hz, 1000 Hz, 2000 Hz and 5000 Hz. In
the 2AFC procedure, subjects perceived one stimulus on the right side and one
on the left side and had to respond which one was on the right.

While there have been few studies on lateralization of simple stimuli by
hearing impaired subjects [Moore, 1995, p133], performance is not closely
related to monaural audiometric thresholds. However, poor performance is
usually related to an asymmetric loss.

JNDs in ILD in bilateral CI users have been measured using the audio input of
the Med-El implant by Senn et al. [2005] and Laback et al. [2004]. JND values
of respectively 1.2 dB and 1.4 – 5 dB difference in electrical voltage at the
audio input were reported. In the latter study, stimuli were chosen such that
the pitch percept of the active electrodes corresponded. Lawson et al. [1998]
and van Hoesel and Tyler [2003] on the other hand used an experimental
processor to directly stimulate for the Nucleus implant, bypassing the clinical speech processor. Lawson et al. [1998] found JNDs of 1-4 current units, which equals $0.09 - 0.35 \text{dB}$ change in electrical current and van Hoesel and Tyler [2003] found JNDs of $<0.17 - 0.68 \text{dB}$ change in electrical current.

In this study, we assessed JNDs in ILD and loudness growth in 10 subjects that used a CI in one ear and were severely hearing impaired in the other ear. To evaluate maximal performance on loudness balancing experiments, first a pitch matching experiment was performed to identify the acoustical sinusoid with the frequency that sounded the most similar to an electrical stimulus presented on the most apical electrode of the CI. Then loudness balancing experiments were done over the entire acoustic dynamic range. From the found crossover points of the psychometric curves, the loudness growth function (LGF) can be determined and from the slopes the JND in ILD can be found. As a worst case scenario, the experiments were repeated with the most basal electrode. In this way the influence of poorly matched systems (CI and HA) can be assessed.

2 Methods

2.1 Apparatus

The subject’s clinical devices were not used. Our test setup consisted of version 3 of the Apex program developed at Lab Exp ORL (K.U.Leuven) [Laneau et al., 2005] connected to an RME Hammerfall DSP soundcard and an L34 device provided by Cochlear Ltd. An interface to the L34 was implemented in Apex and subsequently used to communicate with the L34. The L34 is a modified speech processor that allows arbitrary pulse sequences
to be sent from a personal computer to the implant.

The first channel of the soundcard was used to drive an ERA 3A insert phone and the second channel was used to trigger the start of the electrical stimulation. In this way, the electrical and acoustical signals were exactly synchronized as could be verified by using an implant-in-a-box and an oscilloscope.

The insert phone was calibrated using a 2cc coupler conform to the ISO389 norm. A maximum distortionless sound pressure level of 112 dBSPL could be achieved on all test frequencies. The maximum distortion component we measured for pure tones at 112 dBSPL was 43 dB below the sound pressure level of the main component. The shape of both the electric and acoustic signal were checked using an oscilloscope.

### 2.2 Stimuli

All electrical stimuli were 0.5 s trains of biphasic pulses of 900 pps (pulses per second) with a phase width of 25 µs and an inter phase gap of 8µs. The stimulation mode was monopolar, using both extracochlear reference electrodes in parallel (MP1+2). These parameters correspond to the clinical maps used by the subjects on a daily basis. The pulse train definitions were generated using custom Matlab scripts and saved to disk. The electrical pulse shapes were generated by the subject’s implant and all pulse shape parameters were identical to the settings in the subject’s clinical map. We will report electrode numbers in apex-to-base order, such that electrode 1 is the most apical and electrode 22 is the most basal.

All acoustical stimuli were generated using Matlab and were sinusoids of 0.5 s
long, ramped in and out over 50 ms using a cosine window to avoid clicks at
the beginning and end of the stimulus.

2.3 Procedures

Two sets of data were collected. For the first set, the most apical electrode of
the CI was used. This electrode stimulates the lowest place-frequency that can
be stimulated with the CI. In the second set, the most basal electrode was
used that yielded a clear auditory percept and had a minimum dynamic range
of 30 CU.

Both electrodes were fitted independent of the clinical fitting. The T
(threshold) level was chosen as the just audible level and the C level was the
lowest level that was rated as very loud on a 7-interval loudness scale
(inaudible - very soft - soft - good - loud - very loud - intolerable). Several
parameters for each subject are given in table 1.

All procedures were performed for both the most apical (set 1) and most basal
(set 2) electrodes. First a pitch matching procedure was done to find the
best-matching acoustical pitch for both electrodes. Then the frequency of the
acoustical stimulus was fixed and several loudness balancing experiments were
done to assess loudness growth and JNDs in ILD.

2.3.1 Pitch matching

A pitch matching procedure was used to determine the frequency of the
acoustical sinusoid for which the perceived pitch optimally matched the
perceived pitch of a pulse train of 900 pps on the selected electrode. At these
high rates, the perceived pitch varies only with place and does not depend on
variations in the rate [Shannon, 1983, Zeng et al., 2004]. Pilot testing with 2 subjects indeed revealed no difference in percept or difference in results from the matching procedure for stimulation at 900 pps or at 7200 pps. Also, as the rate was fixed for all experiments, no influence of rate pitch on our results is to be expected.

First the acoustical stimuli were balanced in loudness against an electrical stimulus that sounded comfortably loud (the most comfortable level of the electrical stimulus, corresponding to the label “good” on the loudness scale, was determined during the electrical fitting). If the required loudness could not be achieved for some of the acoustical stimuli, the electrical stimulus was reduced in loudness and the balancing procedure started all over again. In this phase, balancing was done by indicating the perceived loudness on the same loudness scale that was used for the electrical fitting. The balancing serves no other purpose than to avoid loudness cues interfering with pitch cues. It has neither relation with nor influence on the loudness balancing experiments performed later in this study.

Second, pitch matching was done using constant stimuli procedures with 4 presentations per stimulus. The electrical and acoustical stimulus were presented sequentially in random order. Every stimulus was presented twice as the first stimulus. The subject had to indicate whether the first or second stimulus sounded higher in pitch. The electrical stimulus was uniformly roved in loudness over 10% of the electrical dynamic range to avoid subjects using residual loudness cues in spite of the loudness balancing previously performed.

A first rough estimation of the matching acoustical pitch was performed by sampling the acoustic frequencies over 2 octaves, spaced by 1/5 oct, resulting in 11 acoustic frequencies. This sampling corresponds approximately to the sampling used by Boex et al. [2006]. Then finer scale estimation was
performed by using the first obtained rough frequency estimate as the
geometrical mean of 11 frequencies over a range of 0.5 oct. The loudness
balancing procedure was then repeated for each of these frequencies and the
results of both constant stimuli procedures were merged into a single
psychometric curve to obtain the best matching frequency. In total, the
subject had to answer $11 \times 4 = 44$ times in the rough measurement and
another 44 times in the finer scale measurement.

A 2 parameter psychometric function was fitted using a maximum likelihood
method to find the 50% point, besides the slope in the 50% point. An example
psychometric function of the fine scale estimation for the most apical electrode
of subject S4 is shown in figure 1. The first pitch estimate was 280 Hz. Then
11 acoustical frequencies were sampled around this first estimate and a
psychometric function was fit to the results. This resulted in a rough estimate
of 250 Hz. Then 11 acoustical frequencies were sampled on a finer scale around
250 Hz and again a psychometric function was fitted to the results. The latter
function is shown in figure 1. The final matched pitch was thus 250 Hz.

To confirm the pitch matching, subjects were queried whether the acoustical
pitch percept corresponded well to the electrical pitch percept for several
intensities. After confirmation, the found pitch was considered correct and
used in all subsequent experiments.

Finally the acoustical dynamic range for the matching frequency was
determined by finding the acoustical threshold. The upper limit was always
the upper limit of the used transducer (112 dBSPL) and was not perceived as
uncomfortable by any of the subjects. Note that because of this upper limit of
the transducer, the upper limit of the perceptual dynamic range for acoustical
stimulation could not be determined.
Figure 1: Psychometric function for the fine pitch matching experiment for subject S4, set 1.
If no pitch match could be obtained for the most basal electrode, an acoustical frequency was selected for which the dynamic range that could be stimulated was greater than 10 dB and that was subjectively selected as “most similar” to the electrical stimulus. As for most subjects there was no residual hearing at the matching frequency, a lower frequency than the matching frequency was selected. Therefore we can consider set 2 as unmatched, or at least matched worse than set 1.

2.3.2 Loudness growth and JND determination

After the pitch matching procedure, loudness growth and JNDS in ILD were determined by performing several sequential loudness balancing runs. As it is important here to obtain accurate and objective values, a constant stimuli procedure was used instead of the more subjective (but faster) procedure used to balance stimuli before pitch matching.

For set 1 (the most apical electrode), loudness balancing between acoustical and electrical stimuli was done for several electrical levels uniformly spaced over the electrical dynamic range with intervals of 5% of the dynamic range. In most subjects, for the upper part of the electrical dynamic range, no acoustic amplitude could be found that sounded equally loud (possibly due to the acoustical transducer’s upper sound level limit). For the second set (the most basal electrode), loudness balancing was done by sampling the acoustical dynamic range with intervals of at most 5 dB.

If time permitted, more levels were tested, both for set 1 and 2. To avoid subjects being able to answer correctly by using only one ear, different levels for both ears were presented during the same run. The electrical stimuli were mostly varied in steps of 5% of the subject’s electrical dynamic range, the
acoustical stimuli in steps of 2 dB. Step sizes were larger in the first few experiments (to get used to the protocol) and smaller if necessary to find enough points on the slope of the psychometric curve.

In all loudness balancing experiments, the electrical and acoustical stimuli were presented simultaneously, unlike the pitch matching experiments and the loudness growth function experiments for bimodal stimulation found in the literature. The subject was instructed to indicate whether the signal on the left or right hand side was louder. In one run, each stimulus was presented 4 times. When the same stimulus occurred in more than one run, the results of these runs were combined after verifying that they were compatible by overlaying the psychometric curves. There were no disparities within a test session. The subjects performed 2 or 3 sessions of loudness balancing per electrode. For subject S1, the results between the first and second session seemed to differ. Therefore loudness balancing results were only used from the second session with S1 because during the second session much more data was collected than during the first.

To determine the loudness growth function between electrical and acoustical stimulation, psychometric curves were fitted for several fixed levels of either the electrical or acoustical part. Psychometric functions were fitted using the psignifit toolbox version 2.5.6 for Matlab (see http://bootstrap-software.org/psignifit/) which implements the maximum-likelihood method described by Wichmann and Hill [2001]. 68% confidence intervals around the fitted values were found by the BC\(_A\) bootstrap method implemented by psignifit, based on 1999 simulations.

For the measurements of set 1 (most apical electrode), the electrical dynamic range was regularly sampled. The corresponding acoustical value was determined by fitting a psychometric function for each value of the electrical
amplitude.

For set 2 (most basal electrode) on the other hand, the acoustical dynamic range was regularly sampled. Therefore the process was reversed and the electrical value was determined for each sampled value of the acoustical amplitude. All loudness growth functions are shown in figure 3.

If for a certain psychometric function the confidence interval could not be determined by the bootstrap method, the point was discarded from all further analyses. This was the case when no data points were available on the slope of the psychometric function (only at the edges). This occurred for only 24 of 186 fits.

The psychometric function fitting results in several equal-loudness points with error bars. On these points linear regression was performed and $R^2$ was calculated as an error measure.

From the slopes of the psychometric functions, JNDs in ILD were determined as half the difference between the 75% point and the 25% point of the psychometric curve. A 68% confidence interval for the JND was determined by combination of the confidence intervals around these points found by the bootstrap method.

2.4 Subjects

Ten subjects were recruited amongst the clinical population of the university hospital of Maastricht (AZM) and the university hospital of Leuven. All subjects were volunteers and signed an informed consent form. This study was approved by the medical ethical committee. All subjects wore a HA contralaterally to their CI on a daily basis and used a CI of the Nucleus24
type (Cochlear Ltd). S1 and S5 had an electrode array of the Contour Advance type, the other subjects had an array of the Contour type. The clinical processors were of the ESPrit3G type for all but one subject, and of the Freedom type for one subject. All unaided subject audiograms as measured during routine audiometry are shown in figure 2. Demographic information for all subjects is given in table 1.

The subjects came to the hospital for 4 or 5 sessions of about 2 hours with at least one week time and maximally one month time between sessions. As the residual hearing of subject S5 abruptly decreased by 10 dB between two sessions, no measurements were made for set 1 for this subject.

Subject S9 had an incomplete electrode array insertion in the cochlea with two electrodes lying outside of the cochlea. All other subjects had normal electrode insertions. Subject S4 has been re-implanted after failure of his first implant, which was implanted in 2002.

3 Results

3.1 Pitch matching

While for a few subjects some training was needed, pitch matching went smoothly for the experiments with the most apical electrode (set 1). In the next test session, the pitch matching experiment was repeated for verification and the results were always within a few hertz of the previous match. Therefore the result of the first session was used for all subsequent experiments of set 1. The identified frequencies are listed in table 1.

For the most basal electrode (set 2) however, in many cases no clear pitch
Figure 2: Unaided subject audiograms. Note that the vertical axis starts at 50 dBHL. No symbol means no threshold could be found at that frequency.
match could be found. This is probably due to the lack of acoustical residual hearing at higher frequencies (unaided subject audiograms are shown in figure 2). When this was the case, the subject had to select the acoustic frequency that was “most similar” to the electric pulse train. In this case, only acoustic frequencies were presented where the dynamic range was $> 10 \text{ dB}$.

For subject S1 the matched pitch for the set 2 experiments was 1124 Hz, but the acoustic dynamic range at this frequency was only 6 dB. We therefore used 500 Hz instead, where the dynamic range was 30 dB. For subject S3 the matched pitch of electrode 1 was 420 Hz. While no good match could be found for electrode 22, 250 Hz was preferred by the subject. Subject S4 reported that the electrical stimulus sounded higher for all acoustical frequencies that could be tested, therefore 250 Hz was selected in the subsequent tests. Subject S7 reported that the electric stimulus was always higher, but preferred the sinusoid of 370 Hz, because it sounded the most similar to the electrical stimulus. For subject S9, the matched pitch for the 20th electrode was lower than for the first electrode. This may be due to the subject’s partial electrode array insertion, which can cause atypical stimulation patterns when stimulating electrodes on the edge of the cochlea.

Overall, in set 1, the subjects perceived the acoustical and electrical stimuli as very similar and after some exposure most of them reported the stimuli to fuse to a single percept. One subject (S6) reported that the acoustic stimulus sounded somewhat “warmer”. In set 2 however, there was a clear perceptual difference between the stimuli, causing the stimuli not to fuse to a single percept. Therefore, this set should be considered as a worst case scenario, that may occur in practice if the frequency mapping of the CI is very different to the “acoustic” mapping, i.e., when the low frequencies (that can be perceived acoustically) are presented on an electrode that is at a much higher place in
the cochlea than the place in the cochlea that is activated by acoustic stimulation.

3.2 Loudness growth functions and JNDs in ILD

Based on loudness balancing experiments, loudness growth functions (LGFs) between electrical and acoustical stimulation were determined. All LGFs are shown in figure 3. The error bars were determined using the bootstrap method. $R^2$ values are plotted next to each LGF. For each set, a single LGF is plotted per subject based on linear regression.

From the slope of the psychometric functions, JNDs in ILD were determined for various intensities. For simplicity we will specify all JNDs in dB change in the acoustical ear for a fixed electrical current in the other ear. Figure 4 shows an example set of JND values for the entire dynamic range of subject S6. In this case the median JND was 2.0 dB. Figure 5 shows the median of all JNDs for all subjects and both sets. For set 2 the JNDs were converted from current units to dB using the fitted loudness growth function. It can be seen that generally the JND increases when going from set 1 to set 2. The mean JND was 1.7 dB for set 1 and 3.0 dB for set 2. For comparative purposes, some results from the literature on normal hearing subjects and bilateral CIs were also plotted [Mills, 1960, Yost and Dye, 1988, Laback et al., 2004, Senn et al., 2005].

When drawing figures similar to figure 4 for all other subjects, it can be seen that for set 1 (apical electrode), the JNDs are very similar over the measured range of intensities. For set 2 however, for some subjects a falling tendency can be observed, i.e., JNDs decrease with increasing sound intensity.
Figure 3: Loudness growth functions between electrical and acoustical stimulation for set 1 and 2. The error bars were determined using a bootstrap method.
Figure 4: JNDs per electrical intensity for subject S6. The X-axis shows the fraction of the electrical dynamic range in current units. Error bars are 68% confidence intervals determined using a bootstrap method. The dashed line shows the median and thick error bars on the right hand side show the 25% and 75% quantiles.
Figure 5: All JNDs in ILD in dB change in the acoustical ear for a fixed electrical current in the other ear. The 75% and 25% quantiles are indicated. The JND for S10, set 2 is 7.6 dB. Above the label 2CI, the diamonds show data from [Senn et al., 2005] and the plusses show data from [Laback et al., 2004] for bilateral CI users. Above the label NH, the diamonds show data from [Yost and Dye, 1988] and the plusses show data from [Mills, 1960] for normal hearing listeners.
Some subjects found the task subjectively easier for set 2, because they could more easily differentiate between both ears. However, objectively all of them performed better for set 1. For the latter set, when asked afterwards whether they could hear a single fused sound coming from a certain direction instead of just cueing on loudness differences between their ears, 4 subjects answered they could, 2 answered they could not and the 4 other subjects could not answer the question. The fused sound was however not externalized (i.e., was perceived as being located inside the head).

4 Discussion

4.1 Pitch matching

Boex et al. [2006] report results of pitch matching of six subjects using Clarion electrode arrays using a similar procedure as the procedure used in this paper. The matching frequencies for electrode 1 were found to be 460, 100, 290, 260, 288 and 300 Hz. Our results for the most apical electrode (set 1) are in the same range, as can be seen in Table 1. The higher value for subject S9 can be explained by the partial and thus less deep insertion of the electrode array.

For the most basal electrode (set 2) the subjects reported a perceptual difference between the stimuli at both ears and in many cases no clear pitch match could be found. The subjects may have selected the acoustical signal that they could perceive the most clearly, i.e., where the dynamic range was sufficient, instead of the signal that was best matched in pitch. Also, a comparison of the obtained frequencies to values in the literature shows that the latter are on average higher or unmeasurable. Boex et al. [2006] report values of 3050 Hz, 1290 Hz and 1200 Hz for the most basal electrode of Clarion.
electrode arrays and Dorman et al. [2007] report a value of 3400 Hz for the
most basal electrode of a MedEl Combi40+ cochlear implant. Therefore the
pitches of the electrical and acoustical signal of set 2 of the present study
should be considered unmatched. Arguably the JND results of set 2 would not
have been very different if the same acoustical frequency would have been used
as in set 1.

4.2 Loudness growth

Dorman et al. [1993] and Zeng and Shannon [1992] report a linear growth of
the acoustical level in dB versus electrical amplitude in µA. This is not
contradicted by our data. A regression line was fitted through all points per
subject per electrode and drawn in figure 3. $R^2$ values are shown next to each
regression line. However, as the acoustic dynamic range of our subjects is
rather small, it is hard to make a strong statement on this topic.

Visual inspection of the set 1 data for subject S6 reveals that an exponential
transform of the current may provide a better fit. Indeed, when applying
linear regression on the acoustical values in dB versus the electrical values in
clinical current units, $R^2$ increases from 0.91 to 0.95. However, in set 2 for this
subject this effect is not observed, neither is there a clear tendency of increase
or decrease of $R^2$ over the other subjects when applying an exponential
transform of the current level.

The slopes of the regression lines are subject dependent and depend on both
the electrical and acoustical dynamic range [Zeng and Shannon, 1992]. In a CI
processor the subject’s dynamic range is explicitly used when mapping the
output of a signal processing channel to an electrode. In current clinical
practice it is however not as explicitly used in HA fitting, and the HA is fitted
separately from the CI. Therefore a typical fitting of a bimodal system is likely to be suboptimal for ILD perception.

According to Krahe et al. [2000] the use of a ramped acoustical signal and a non-ramped electrical signal could have slightly influenced our results due to possible confounding interaural time difference (ITD) cues. In this case the crossover points we used to determine the LGFs would all have a slight bias in one direction. We are however confident that our subjects could not perceive ITDs in the present signals because: (1) preliminary tests showed that the results were not influenced by a shift in time of either of the signals; (2) while the electrical and acoustical signal were exactly synchronized in time at transducer level, they were probably not psychoacoustically synchronized. In the acoustical path an extra frequency-dependent delay is present because the sound wave travels through the cochlea while the electrical signal arrives instantly. Next to the ramp in the acoustical signal, this extra delay severely degrades ITD cues at the onset of the signals. (3) As a high rate pulse train was used electrically and a pure tone acoustically, there were no clear envelope cues in the signals.

4.3 Just noticeable differences

The mean JND in ILD of set 1 over all subjects was 1.7 dB, the mean of set 2 was 3.0 dB.

JNDs in ILD for low frequency tones in normal hearing subjects are around 1dB [Mills, 1960, Yost and Dye, 1988], thus bimodal system users perform slightly worse, but certainly close enough to normal hearing subjects for normal ILD perception of low-frequency stimuli.
While there is quite some variability amongst subjects, the JND in ILD performance of our bimodal subjects is comparable to the performance of the bilateral CI subjects tested by Senn et al. [2005] and Laback et al. [2004]. Here JND values of respectively 1.2 dB and 1.4 – 5 dB are reported.

A comparison of the JNDs of set 1 and set 2 shows a clear increase in JND for all subjects except S2, S8 and S9. This relates to the fact that for the other subjects the most basal electrode could not as accurately be matched in pitch because of a lack of residual hearing at the high frequencies. Francart and Wouters [2007] show that JNDs in ILD increase in normal hearing listeners with increasing frequency separation between the ears. This result is confirmed for 2 bilateral CI users by Laback et al. [2004].

### 4.4 Relation to localization performance

While ILD sensitivity is high, performance on localization tasks is still poor for many bimodal system users. This is probably due to three main physical problems with current CI speech processors and HAs.

A first problem is the absence in real-life signals of large ILDs at the low frequencies because the head shadow effect is small at larger wavelengths. While ILDs can reach values of 20 dB at higher frequencies, they are only in the order of a few dB in the low frequencies. As most users of a bimodal system do not have residual hearing in the high frequencies, they do not have access to clear ILDs. This could be improved by using a signal processing system to amplify ILDs in the low frequencies.

A second problem is the inability to use fine structure ITD cues because they are not transmitted by the CI speech processor and even if they were, the
latency between CI and HA is not optimized.

A third problem is suboptimal bilateral fitting. A CI and HA are in many cases still fitted separately, without paying much attention to loudness balance between both ears. Moreover, compression characteristics are not matched, resulting in unclear ILDs. When considering the transduction of a clinical speech processor, the resulting loudness growth functions (acoustical input versus electrical output) do not have the same shape as the functions found in this paper. Hoth [2007] measured these loudness growth functions between computer controlled stimulation and stimulation via a speech processor with 15 subjects using clinical fitting software for direct electrical stimulation and the subject’s well-fitted speech processor and acoustical noise bursts for “acoustical” stimulation. He found that the functions are nonlinear, subject dependent and even electrode dependent within one subject.

To physically assess the loudness transfer characteristics of a clinical speech processor, we made simulations of an example SPrint speech processor using the Nucleus Matlab Toolbox. Figure 6 shows the mean output current for a certain acoustical input at the microphone for two hypothetical fittings. It can be seen that the nonlinearity increases when increasing the overall current level. The obtained transfer functions are very similar in shape to the transfer functions found by Hoth [2007]. The stimulus was a sinusoid of 250 Hz, chosen to fall exactly in the middle of the first channel of the subsequent processing. The threshold level was set to 50 CU in a first simulation and to 200 CU in a second simulation. The corresponding comfort levels were set to 80 CU and 230 CU. The ACE strategy was used with 8 maxima, the sensitivity was set to 12 and the Q parameter of the loudness growth processing was set to 20. The resulting current level was calculated as the RMS value of all current levels over all channels. Note that this last step implies an over-simplified loudness
model, that can however be used as a simple approximation. If the same
analysis is done with only one channel selected, it saturates at about
50 dBSPL and does not provide a realistic picture. It is clear that the resulting
transfer function is not linear and will most probably interfere with ILD
perception in the case of bimodal stimulation. The abrupt changes in the
function are due to quantization into current levels, but are on the edge of
most subjects’ loudness sensitivity, both binaurally and monaurally [Zeng
et al., 2004], and thus probably not perceivable. Note that the shape of the
transfer function depends on many parameters of the speech processor that
can be set in the fitting process. A different combination of T-levels,
sensitivity, Q and other parameters will therefore result in either a more linear
or an even less linear transfer function.

4.5 Conclusions

Loudness growth functions and just noticeable differences in interaural level
difference were measured in ten users of a bilateral bimodal hearing system.
The loudness growth functions between electric and acoustic hearing can be
well approximated by a linear relationship between current in µA and
acoustical level in dB. The slope of the line depends on both the electric and
acoustic dynamic range and is thus subject dependent. Current CI speech
processors use a logarithmic or near logarithmic transfer function whose
coefficients depend on various parameters that are set during the fitting to
optimize speech perception. This implies that the clinical fitting of the
combination of CI and HA will in most cases not be optimal for ILD
perception and subsequently binaural lateralization performance.

JNDs in ILD are slightly larger than in normal hearing subjects, but certainly
in a range usable for ILD perception. The mean JND for tonotopically
Figure 6: Simulated transfer function of the SPrint processor for two different fittings. The abrupt changes in the function are due to quantization into current levels and the breakpoint is due to the saturation level implemented in the loudness growth processing.
matched electrical and acoustical stimulation was 1.7 dB. However, as ILDs are small in the low frequencies, for many subjects their use will be limited because of a lack of residual hearing in the low frequencies. For subjects that do have residual hearing at frequencies where ILDs are present in realistic listening situations, a proper balancing between CI and HA will be important as the sensitivity to ILDs is high.

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1 List of Tables

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<table>
<thead>
<tr>
<th>Subject</th>
<th>Age</th>
<th>M of use</th>
<th>CI side</th>
<th>Etiology</th>
<th>Most apical electrode (set 1)</th>
<th>Most basal electrode (set 2)</th>
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<td></td>
<td></td>
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<td>37</td>
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<td>Noise exposure</td>
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<td>R</td>
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</tbody>
</table>

Table 1: Subject information: “Age” is the age in years at the moment of testing. “M of use” is the number of months of implant use at the moment of testing. “CI side” is left (L) or right (R) (the HA was on the other side). “Elec” is the electrode number (numbered from apex to base) and “DR” is the electrical dynamic range in current units. “MF” is the frequency of the pitch matched sinusoid in Hz and “Thr” is the acoustical threshold in dBSPL.