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Psychophysics, fitting, and signal processing for combined hearing aid and cochlear implant stimulation

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Abstract

The addition of acoustic to electric stimulation via a cochlear implant has been shown to be advantageous for speech perception in noise, sound quality, music perception and sound source localization. However, the signal processing and fitting procedures of current cochlear implants and hearing aids were developed independently, precluding several potential advantages of bimodal stimulation, such as improved sound source localization and binaural unmasking of speech in noise. While there is a large and increasing population of implantees that use a hearing aid, there are currently no generally accepted fitting methods for this configuration. It is not practical to fit current commercial devices to achieve optimal binaural loudness balance or optimal binaural cue transmission for arbitrary signals and levels. There are several promising experimental signal processing systems specifically designed for bimodal stimulation. In this article, basic psychophysical studies with electric acoustic stimulation are reviewed, along with the current state of the art in fitting, and experimental signal processing techniques for electric acoustic stimulation.

1 Introduction

In the past, only people who were profoundly deaf in both ears were considered for cochlear implantation. Nowadays, it is clear that many implant users perform better in perceptual tests than severely hearing impaired hearing-aid users. This has led to relaxed implantation criteria (Gifford et al., 2010; Mowry et al., 2012), which means that now there is a large and rapidly expanding population of cochlear implant (CI) users with residual hearing. This is illustrated by Sampaio et al. (2011), and also by Dorman and Gifford (2010), who found that the majority of patients implanted in their clinic during the last 5 years had residual hearing thresholds of 85 dB HL or better at 250 Hz. Given the nature of the residual hearing, it is usually stimulated using a high-power hearing aid (HA). If the residual hearing is in the non-implanted ear, this is called bimodal stimulation. If residual hearing is preserved in the implanted ear, the CI and HA can stimulate the same ear. This combination is called electric-acoustic stimulation (EAS) or hybrid stimulation. In the current article, we will use the term “bimodal stimulation” exclusively for CI and HA in opposite ears, and the term EAS for any combination of electric and acoustic stimulation.

In many patients, residual hearing is only present at low frequencies. In the implanted ear there is often no usable acoustic hearing above about 1-2 kHz (von Ilberg et al., 2011). In non-implanted ears there is much variability, but typical cut-off frequencies are around 1-2 kHz, when the cut-off frequency is defined as the frequency above which unaided thresholds are higher than 90-100 dB HL or immeasurable. In these cases, the cochlea can be considered as having a broad high-frequency dead region (complete loss of inner hair cells over a region of the basilar membrane, Moore et al., 2010).

If only one ear is implanted and there is residual hearing in the non-implanted ear, there is binaural stimulation. Having two ears gives rise to several effects that can benefit speech recognition in noise (Akeroyd, 2006; Bronkhorst, 2000). These include binaural redundancy, head shadow, and squelch. Of these, squelch¹ is widely considered to be the only one based on true binaural cues, i.e., cues that arise solely from differences between the ears, beyond two-separate-ears benefits. Another important benefit of being able to perceive binaural cues is the ability to localise sounds (Blauert, 1997; Moore, 2003).

Performance with commercial devices is well documented. In an extensive review of the literature about bilateral CIs and bimodal stimulation, Ching et al. (2007) found that for bimodal stimulation there was a strong effect of complementarity, which is the use of additional information

¹Binaural squelch is the effect whereby the auditory system suppresses noise by using interaural phase differences in either the target or masker signal (but not both). This effect can be studied psychophysically by measuring binaural masking level differences, defined as the difference in detection threshold of a tone in noise between a condition where the same signal is presented to both ears and a condition where the noise signal in one ear is reversed in phase. Similarly a binaural intelligibility difference can be determined by measuring recognition of speech in noise with either identical noise presented to both ears, or with the noise reversed in phase in one ear.

from the acoustic stimulus that is not available through the CI (such as pitch) and vice versa to achieve better performance. Additionally, about half the subjects showed binaural benefits for speech perception. These were mainly attributed to head diffraction and binaural redundancy, which across studies afforded improvements in speech recognition threshold of around 2 dB. Subjects seemed to make only very limited use of true binaural cues. Schafer et al. (2011) conducted a systematic review of the bilateral CI and bimodal literature and similarly found a significant beneficial effect of binaural summation and the head shadow effect for both bilateral CI and bimodal stimulation, and a significant effect of binaural squelch only for bilateral CI stimulation, all relative to monaural CI conditions.

Besides speech perception, EAS is also found to be advantageous for sound quality (Ching et al., 2007; Sucher and McDermott, 2009) and music and pitch perception, as reviewed by McDermott (2011). In tests of pitch perception and melody recognition, it was found that melody recognition improved when adding acoustic stimulation (Dorman et al., 2007a; Kong et al., 2005). Pitch discrimination was better for EAS subjects than for CI subjects (Gfeller et al., 2007). If median residual hearing thresholds at 125 to 1000 Hz were better than 85 dB HL, song recognition was better with the HA alone than with the CI alone in a no-lyrics condition and better with bimodal than with CI alone with or without lyrics present (El Fata et al., 2009). Generally, for tasks that mainly depended on pitch perception, performance with EAS was the same as with only acoustic stimulation. It is clear that pitch perception mainly occurs through the acoustic stimulation.

Current bimodal listeners are normally fitted with a standard CI sound processor and standard HA, often using the same fitting methods as would be used for the individual devices. While clearly this configuration can be advantageous compared to only the CI, there is very limited evidence of the use of true binaural cues. This is partly due to perceptual issues, but one should also consider technical issues, cue transmission through the devices, and fitting. These factors are often ignored when assessing performance with clinical devices. We believe that performance - especially in situations that require the use of binaural cues - could be much better if the signal processing in the devices was optimized for them to work together and if appropriate fitting procedures were used. There are a few devices specifically developed for hybrid stimulation, but their signal processing was not intrinsically designed for hybrid stimulation, and there are no scientifically validated fitting procedures yet for this configuration.

In what follows, we will highlight some problems with current commercial devices, review psychophysical studies with experimental devices, and provide an overview of possible fitting methods for EAS. Finally, we will review different signal processing schemes specifically developed for EAS and provide some future directions.

2 Current devices

Current commercial CIs and HAs were developed more or less independently, without the possibility of their combined use being explicitly taken into account. Unfortunately, this means that many aspects of their signal processing and especially the way they are commonly fitted are not optimal for combined use. This section mainly concerns sound processors and signal processing. The different implant types for hybrid stimulation are reviewed by von Ilberg et al. (2011).

Today, for bimodal stimulation the only clinical option is a standard CI and sound processor combined with a commercial hearing aid. There are no integrated systems. The three major CI manufacturers are Cochlear, Advanced Bionics and Med-El. For hybrid stimulation there are a few options: the Cochlear HybridTM (Buechner et al., 2009) and the Med-El DUET system (Helbig and Baumann, 2010; Helbig et al., 2008). While the current integrated hybrid systems solve some of the problems associated with separate devices discussed in the following sections, such as providing the option to control the difference in processing delay, there are several remaining issues, including the selection of parameter settings such as the optimal delay between acoustic and electric stimulation, and optimal fitting.

In what follows, we will first discuss a number of basic problems with the signal processing of current commercial bimodal devices. Then we will discuss transmission of temporal information

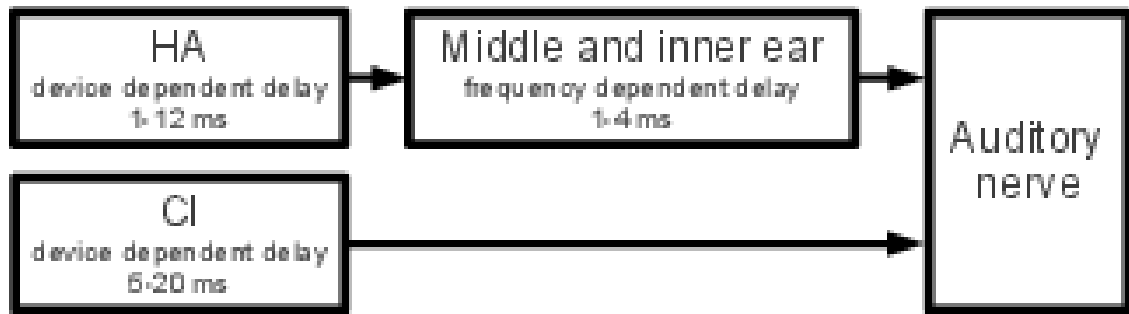


Figure 1: Illustration of the temporal synchronization problem. Adapted from Francart (2008), with permission.

by current sound processing strategies, and finally discuss transmission of binaural cues through bimodal devices.

2.1 Temporal synchronization, place mismatch, and pre-processing

Different devices may have different processing delays, leading to temporal asynchrony between the ears, as illustrated in Figure 1. When a sound arrives at the microphone of the CI sound processor, it is subjected to a device-dependent processing delay on the order of 5-20 ms² (Stone and Moore, 1999). Subsequently, processed signals are decoded by the implanted chip where they may be subjected to an additional short processing delay. Finally, the auditory nerve is directly electrically stimulated. By comparison, when a sound arrives at the microphone of a HA, it also undergoes a processing delay, which in most cases is different from (and often smaller than) the processing delay of the CI sound processor. Then the sound produced by the HA receiver travels through the middle and inner ear before finally stimulating the auditory nerve. The total delay is therefore the sum of the device’s processing delay and the frequency-dependent travelling wave delay. It is clear that in most cases the total delay of the electric and acoustic path will differ, with differences up to tens of milliseconds, such that the neural stimulation occurs first at the side with the shorter processing delay. We will refer to this as the temporal synchronization problem. Technically, it is easily solved by adding an (adjustable) delay to the faster device. While temporal asynchrony on the order of 10 ms is probably not a problem for speech perception (Stone and Moore, 2003), it is most likely a problem for interaural time difference (ITD) perception: while there is some evidence that humans can at least partly adapt to a constant offset up to 684 μ s in one ear (Javer and Schwarz, 1995), it seems unlikely from a physiological and evolutionary perspective that we would be able to adapt to delays of several times the magnitude of naturally occurring ITDs, which range from 0 to around 700 μ s.

Another potential problem is the so-called place mismatch, illustrated in Figure 2. On the acoustically stimulated side, a place in the cochlea will be stimulated that corresponds to the frequency content of the signal. The relationship between place and frequency is estimated by the well-known Greenwood formula (Greenwood, 1990). On the electrically stimulated side, a number of electrodes at certain places in the cochlea will deliver stimulation, based on a frequency-to-electrode (and thus frequency-to-place) mapping that is applied in the sound processor. In current sound processors the frequency-to-electrode mapping is usually left at the default setting without taking into account the electrode insertion depth in each recipient or the “normal” frequency-to-place mapping. It is clear that for most listeners the same sound will result in stimulation of different places in the cochlea for electric and acoustic stimulation. Consider, for instance, a 200-Hz pure tone. Using Cochlear’s ACE strategy, this signal will mainly result in activation of the first filter in the filter bank, corresponding to the most apical electrode. It is known that the

²Processing delay of the CI is defined as the time between the initial deflection of the diaphragm of the microphone of the sound processor and the corresponding first pulse presented on an electrode.

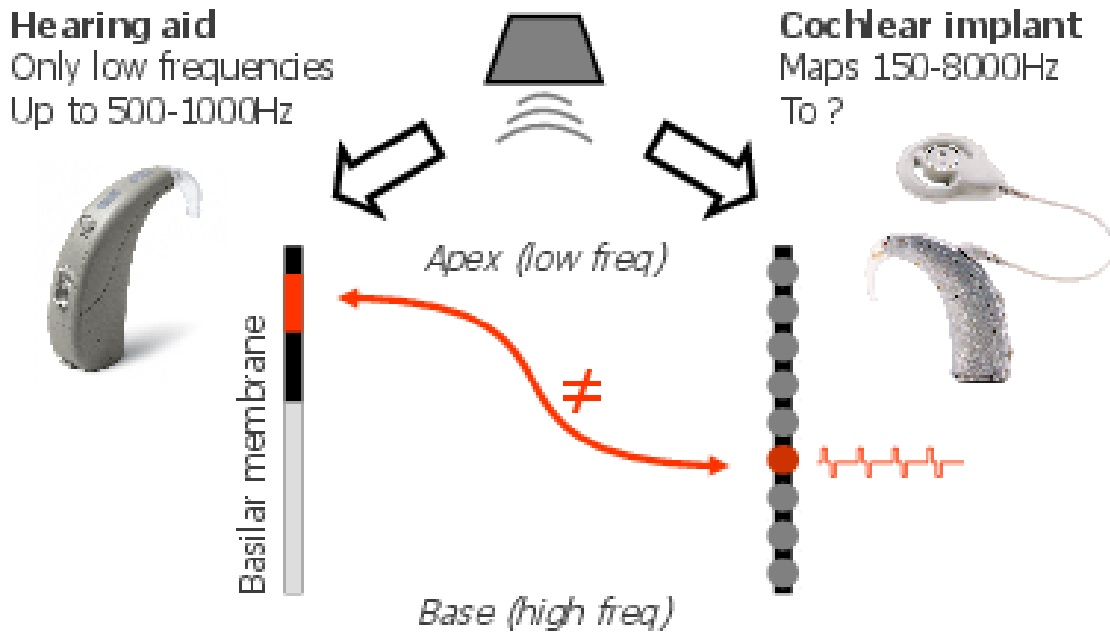


Figure 2: Illustration of place mismatch. The same sound, e.g., a 1-kHz tone, processed by a CI and HA can end up causing neural excitation in different places in the two cochleae. Adapted from Francart (2008), with permission.

cochlear place of the most apical electrode varies widely across subjects. Ketten et al. (1998) found corresponding frequencies ranging from 387 Hz to 2596 Hz with a median of around 1000 Hz for 20 subjects implanted with a Nucleus 22 implant. Data published by Stakhovskaya et al. (2007) indicate a range of approximately 500 Hz to 1000 Hz for typical electrode positions. Consequently, many subjects would be expected to experience a large mismatch between the acoustic and electric place of stimulation. It is currently unclear whether place mismatch is a problem for speech perception (see section 4.1), but large mismatches can be a problem for binaural cue perception (see section 3.3).

CI sound processors and especially HAs nowadays have a large range of pre-processing options, such as compression of the acoustic signal or automatic gain control (Moore, 2008), noise reduction (Bentler and Chiou, 2006), and feedback suppression (Spriet et al., 2010). If there are large across-ear differences in pre-processing, this might have a negative effect on binaural cue transmission and perception. While independent compression systems in HAs in the two ears can dramatically reduce or distort interaural level difference (ILD) cues, such systems do not seem to have a large effect on horizontal localization ability of HA users (Keidser et al., 2006), which is probably due to the redundancy provided by the other localization cues (ITDs and spectral cues). However, for CI users who cannot perceive ITD cues and are presumably less sensitive to spectral cues, the effect might be much larger. Wiggins and Seeber (2011) found in a simulation study with NH listeners that independent compression systems at the two ears influenced lateralisation for several stimulus types, even (but less so) when low-frequency ITD cues were present. Many noise suppression systems can also distort ITD cues (Keidser et al., 2006; Van den Bogaert et al., 2006). For bimodal listeners, who can usually not use ITD cues with their commercial devices (see section 2.3), this probably does not have much effect. However, for listeners with HAs in the two ears (e.g., a hybrid combined with a contralateral HA), or new strategies for bimodal stimulation that enable ITD perception, there could be a serious detrimental effect.

2.2 Transmission of temporal fine structure and envelope timing cues through commercial CI sound processing strategies

In this section we briefly describe the most common commercial CI sound processing strategies and discuss their transmission of temporal fine structure and envelope timing cues. We will come back to this when discussing ITD transmission in the next section. Rosen (1992) defines envelope, periodicity, and temporal fine structure as amplitude modulations between 2 and 50 Hz, between 50 and 500 Hz, and beyond 500 Hz, respectively. Often the periodicity category is left out, considering only envelopes, with rates up to a few hundred hertz, and temporal fine structure beyond that. Here we define temporal fine structure as the fast fluctuations in a signal that can be used by NH listeners to perceive rate pitch, localize low-frequency sounds, and binaurally segregate different sound sources (Akeroyd, 2006). The fine structure is usually amplitude-modulated with a temporal envelope that fluctuates more slowly (up to a few hundred hertz) and contains an important part of the information required to understand speech (Shannon et al., 1995).

While rate pitch perception only requires the overall repetition rate of the fine structure, ITD perception requires the exact fine timing information present in the fine structure and envelope, with a resolution in the range of tens to hundreds of microseconds. Normal-hearing listeners are sensitive to ITDs with thresholds ranging from 10 to 100 μ s, depending on the type of signal (see section 2.3). Some strategies that claim to preserve fine timing cues do preserve the periodicity cue, but do not retain this resolution due to quantization, pulse arbitration, and other factors. Note that while a strategy may transmit the necessary information, this does not necessarily mean that the recipient can perceive this information. E.g., most listeners cannot perceive temporal modulations beyond 300-500 Hz in single-electrode pulse trains (Shannon, 1983).

Most commercial speech processing strategies use a filter bank, followed by envelope detection in each channel, and modulation of fixed-rate pulse trains with the envelopes. Such strategies do not preserve fine structure due to envelope detection with a low cut-off frequency and the use of a fixed-rate carrier. When implemented in a digital system, the sound signal is divided in frames, usually of around 8 ms duration, and the strategy is executed frame by frame. A well-known example of such a strategy is Continuous Interleaved Sampling (CIS) (Wilson et al., 1991).

In most Cochlear devices the Advanced Combination Encoder (ACE) (McDermott et al., 1992; McKay et al., 1994; Vandali et al., 2000) strategy is used. The sound signal is processed by an N-channel filter bank. In each channel the envelope is detected using full-wave rectification, and in each frame only the M channels with maximal envelopes are selected for stimulation. The resulting values are used to modulate high-rate pulse train carriers. Usually N=22 and M=8 with a carrier rate of 900 pps. As in CIS, fine structure is discarded and the non-linear maxima selection process introduces extra temporal distortion. For a carrier rate of 900 pps, the timing information in the envelope is quantized to a multiple of 1.1 ms (the period corresponding to the carrier rate), which is undesirable given a desired ITD resolution on the order of 10-100 μ s.

Advanced Bionics devices commonly use the HiResolution (HiRes) strategy, which is similar to CIS but uses high carrier rates (around 2900 pps). HiRes employs half-wave rectification for envelope detection, but with a fairly high cut-off frequency, thereby retaining part of the temporal fine structure (note that retaining temporal fine structure in the stimulation pattern does not mean it can be perceived). The period of around 300 μ s corresponding to this high carrier rate still introduces undesirable quantization of temporal cues in the envelope. The Fidelity 120 strategy in Advanced Bionics devices is a current-steering strategy (Bonham and Litvak, 2008), which does not have any influence on transmission of timing cues.

Current Med-El devices implement the Fine Structure Processing (FSP) (Hochmair et al., 2006) and FS4 strategies. These strategies consist of standard CIS-like processing for all electrodes except the two to four most apical ones, using a Hilbert transform for envelope detection. On these apical electrodes, the Channel Specific Sampling Sequences (CSSS) strategy (Zierhofer, 2003) is applied. In CSSS a pulse burst is initiated at each positive zero-crossing with amplitudes corresponding to the peak amplitudes of the previous half-wave segment, with repetition rates up to 300-500 Hz for FSP and up to 1000 Hz for FS4. The difference between FSP and FS4 is that in FSP there are usually only one or two CSSS channels and the channel stimulation rate for the remaining channels

is fairly high (around 3000 pps), while in FS4 there are four CSSS channels and the stimulation rate for the remaining channels is lower (around 600 pps). While in the CSSS channels the periodicity of the fine structure is coded, the resulting temporal patterns are quite unlike those available to listeners with normal hearing, and there are pulse arbitration and quantization issues that distort the timing. While this might not be a problem for pitch perception with single or bilateral CIs, this is a problem for binaural cue transmission in a bimodal configuration and for bilateral CIs in noisy or reverberant conditions. Additionally, in the CIS channels, the same envelope temporal cue quantization effect is present as discussed earlier for ACE.

2.3 Binaural cue transmission through bimodal devices

Binaural cues - ITDs and ILDs - are important for sound source localization and are related to speech understanding in noise (Akeroyd, 2006). If a CI recipient uses a HA in the ear opposite to the CI, stimulation is binaural and therefore binaural cues can potentially be perceived. It has been shown that with well-controlled stimuli in the laboratory, bimodal listeners can be sensitive to both binaural cues (see section 3.3). However, with commercial devices bimodal listeners do not appear to use binaural cues (Ching et al., 2007; van Hoesel, 2012). Therefore in this section, we discuss transmission of binaural cues by current commercial devices.

2.3.1 Interaural time differences

NH listeners can use ongoing temporal cues that are present both in the fine structure and the envelope of sound signals (Bernstein and Trahiotis, 1985; Henning, 1974; Yost, 1974). Most current sound processors do not transmit undistorted temporal fine structure (see previous section), and even with specially crafted stimuli listeners can usually not perceive temporal fine structure through electric stimulation, at least not in the sense that NH listeners can. Temporal cues in the envelope are in some cases preserved, depending on the interaction between the spectral shape of the sound and the magnitude and phase response of the CI filter bank, and the level of the signal in each channel.

Besides ongoing cues, NH listeners can also make use of temporal cues at the onset of signals (Henning, 1974). While these are preserved by the CI processing, the time of the first or maximal pulse associated with an onset does not necessarily correspond to the first or peak acoustic stimulation. This is due to quantization effects and other non-linear processing such as maxima selection (see previous section). For instance, with ACE the position of the first pulse is quantized to a multiple of the channel stimulation period, which is typically around 1 ms. Given that realistic ITD cues range from 0 to around 700 μ s, this can result in relatively large temporal distortions.

2.3.2 Interaural level differences

Additionally, good ITD perception depends on consistent ILD cues or at least loudness balance between the ears (Domnitz and Colburn, 1977; Wightman and Kistler, 1992). For ILD cues to be properly transmitted, loudness growth needs to be similar at the two sides. Even if there is acoustic hearing at all frequencies that are associated with stimulation from the CI, and loudness growth through the CI is known and monotonic, configuring a multi-channel compression hearing aid such that loudness growth is the same as in the electrically stimulated ear would still not lead to equal loudness growth in both ears, due to differences in loudness summation across frequencies. Additionally, this would not be practical, because on the one hand it would involve extensive measurements of loudness growth in each subject, which is not feasible in a clinical setting, and on the other hand most CI recipients do not have acoustic hearing at higher frequencies. Therefore, with current devices, if loudness balancing is undertaken at all during fitting, for most patients loudness will only be balanced for a limited range of intensities, and for a limited number of sounds.

Even if loudness growth is similar in the two ears for the signal of interest, for ILD perception there remains the problem of the limited frequency range that can be stimulated acoustically.

ILDs are caused by the head shadow effect, which is the attenuation of sound due to the acoustic properties of the head. Because of the size of the head relative to the wavelength of sounds, ILD cues are mainly present at higher frequencies (> 1500 Hz). Unfortunately, many bimodal listeners only have usable residual hearing up to 1000-2000 Hz (Dorman and Gifford, 2010; Sampaio et al., 2011), and will therefore only have access to small level cues arising from this effect. An additional problem for listeners who do not have access to ITD cues is that the natural ILD-versus-angle function is fairly flat and non-monotonic beyond approximately 45 degrees off-centre (Francart et al., 2011b; Shaw, 1974), which makes it impossible to distinguish between sound directions beyond 45 degrees at one side based on ILD alone.

2.3.3 Binaural cue use with commercial devices

While it is clear that the transmission of binaural cues with current commercial devices is far from optimal, a fair proportion of bimodal listeners is still able to localize sounds to some extent (Ching et al., 2007). In a typical laboratory localization experiment, subjects could theoretically use one or several of the following cues: monaural spectral cues, monaural level cues, ITDs, and ILDs. For normal-hearing listeners, the utility of spectral cues for localisation in the horizontal plane is very limited (Wightman and Kistler, 1997). Additionally, spectral cues are mainly present at frequencies above 6 kHz (Shaw, 1974), so given the typically limited frequency range of the residual acoustic hearing, they could only be potentially used through the CI, and even then they would at most influence the stimulation level on the most basal electrode. It is therefore unlikely that spectral cues contribute much, which is also evidenced by the poor localisation performance seen with monaural CIs (Grantham et al., 2008). In laboratory experiments, monaural level cues are usually eliminated or at least reduced using level roving. While bimodal listeners can be sensitive to ITDs in laboratory experiments with specially crafted stimuli and synchronized experimental processors (see section 3.3), it is unlikely that they can make much use of ITDs with commercial devices, because (1) in current commercial strategies, fine timing cues are either not transmitted, severely quantized, or otherwise distorted, (2) there can be temporal synchronization problems an order of magnitude larger than the largest ITD cues physically available, (3) interaural differences in loudness growth lead to distorted interaural loudness relations, while proper loudness balance is required for good ITD perception, (4) a certain degree of residual hearing is required for good ITD perception, (5) the response patterns in localization experiments tend to follow the shape of the ILD-versus-angle function, i.e., relatively flat for angles larger than approximately 45 degrees (Francart et al., 2011b), and (6) it has been shown that many bilateral CI users do not use ITD cues with commercial devices (Aronoff et al., 2010; Seeber and Fastl, 2008). If spectral cues, monaural level cues and ITDs cannot be used, the remaining cues are ILDs. Therefore by elimination it follows that the (limited) localization ability of bimodal listeners must be based mainly on perception of ILD cues.

Users of a hybrid system and an additional contralateral HA have a distinct advantage: they should be able to perceive ITDs through bilateral acoustic stimulation. Dorman and Gifford (2010) showed that such listeners performed better than bilateral CI users on speech perception tests in spatially separated noise. Dunn et al. (2010) tested eleven bimodal hybrid listeners and found a significant benefit with use of bilateral acoustic stimulation, both for localization and speech perception in noise, presumably due to the use of ITD cues through bilateral acoustic stimulation.

In summary, of all localization cues available to NH listeners (ILDs, ITDs in the fine structure, onset and envelope, and spectral cues) the only ones transmitted by current commercial bimodal systems are partial ILDs, depending on the frequency content of the signal, and limited spectral cues through the CI. This leads to suboptimal localization ability and binaural unmasking.

3 Psychophysical studies

While many authors have studied performance with EAS with commercial devices, the number of fundamental studies that employ controlled stimulation is limited. In what follows, we review

studies that investigate matching pitch and cochlear place of excitation, loudness perception, binaural cue perception, and the characteristics of the residual hearing.

3.1 Matching pitch and cochlear place of excitation

As mentioned in section 2.1, without modifications to the commercial devices, there will often be a mismatch in cochlear place of stimulation between the two modalities for the same acoustic signal. It is currently unknown to what extent place mismatch is problematic. In NH listeners, place mismatch has been found to negatively affect envelope ITD discrimination (Nuetzel and Hafter, 1981) and ILD discrimination (Francart and Wouters, 2007). In EAS users, place mismatch varies, and is difficult to characterize. Attempts to determine degree of mismatch have included behavioural and objective methods.

To address place mismatch, the most straightforward psychophysical method is pitch matching: the subject is presented with stimuli through each type of device, and is instructed to match the pitch of a target stimulus to the pitch of a reference stimulus. Many experimenters have assumed that Greenwood’s formula (Greenwood, 1990) can then be used to convert the pitch (or frequency) estimates to corresponding places of excitation. In early pitch-matching studies, it was found that pitches matched to certain electrodes were much lower (up to two octaves) than would be predicted for those electrodes based on the Greenwood cochlear place-to-frequency map (Blamey et al., 1996; Boex et al., 2006; Dorman et al., 2007b). Later, it was found that the pitch percept associated with an electrode can change during the first few years after implantation, presumably due to brain plasticity (Reiss et al., 2008, 2007), and may ultimately be related to the frequency-to-electrode allocations applied in the sound processor (McDermott et al., 2009). Also, it seems that pitch matching results can be strongly influenced by procedural details. If the procedures are carefully chosen and pitch matching is performed immediately after implantation in subjects with normal (or near-normal) hearing in the non-implanted ear, the disparities with the Greenwood map are small (Carlyon et al., 2010). Green et al. (2012) measured frequency selectivity and pitch matches for nine bimodal listeners. They found that frequency selectivity varied widely and that only the two subjects with measurable frequency selectivity above 500 Hz were able to match pitch consistently. For the other subjects, the matched frequency correlated with the starting frequency. Given all these difficulties, the clinical use of pitch matching procedures does not seem realistic and the results should be interpreted cautiously. It does not seem feasible to use pitch matching to devise an “ideal” frequency-to-place mapping in the CI sound processor. It should be noted that in subjects with single-sided deafness, e.g., patients implanted to treat tinnitus (Buechner et al., 2010; Kleine Punte et al., 2011), it may be possible to use pitch matching to match the filter bank to acoustic stimulation immediately after implantation (before neural plasticity could have influenced the pitch percept associated with the electrodes), taking into account the procedural considerations discussed by Carlyon et al. (2010).

Baumann et al. (2011) performed pitch matching with subjects who had residual hearing both in the implanted and non-implanted ear, and found that pitch matches obtained for either side were similar. They proposed the use of pitch matching to obtain the cross-over frequency between electric and acoustic stimulation for hybrid stimulation. However, this would have to be done before activation of the sound processor, and would be challenging to administer in a clinical environment given procedural difficulties. Moreover, there is no strong evidence that matching the cross-over frequency to the pitch of the most apical electrode is the best approach (see section 4.1).

Two less direct psychophysical methods for place matching have been described in the literature. One is contralateral masking (James et al., 2001), and the other uses sensitivity to ITDs (Francart et al., 2009a, 2011a). In the contralateral masking procedure, a probe stimulus is presented to one ear and is masked with a range of stimuli with different frequency content in the other ear. The masking power of the masker is related to correspondence in place of excitation, assuming that corresponding places in the cochlea lead to increased central masking. James et al. (2001) found consistent threshold elevations for acoustic probes with electric maskers, but less so the other way around. It is currently unclear why electric probes were not efficiently masked acoustically, but it should be noted that the subjects’ residual hearing was worse than that of

typical current bimodal or EAS users. In any case, the place match obtained by this method is not very precise and the procedure is very time consuming.

The ITD sensitivity method is based on the assumption that ITD sensitivity is better for stimuli matched in place. Francart et al. (2009a) measured ITD sensitivity for different permutations of electrodes and acoustic frequency ranges, and found a tendency towards better ITD sensitivity for certain acoustic frequency ranges when combined with a certain electrode. This tendency shifted consistently with electrode number; i.e., more-apical electrodes tended to yield better sensitivity when combined with more apical (lower frequency) acoustic stimuli. Later, Francart et al. (2011a) investigated the issue more thoroughly, but again did not find any statistically significant differences between ITD thresholds, regardless of the combination of electrode and acoustic frequency range. While the procedure yields an indication of a place match, it is not very precise and extremely time consuming.

Smith and Delgutte (2007) measured the binaural interaction component of the EABR (i.e., the difference between the electrical brain stem response for binaural stimulation and the sum of left plus right monaural stimulation) in a cat model and used it to match electrode pairs. It remains to be investigated whether the use of this method is feasible in humans (He et al., 2012).

Other objective methods include imaging (e.g., X-ray and CT scans). The most common method, developed by Cohen et al. (1996), uses X-ray scans to determine electrode position. These positions are then converted to frequencies using the Greenwood frequency-to-place map (Greenwood, 1990). Another method makes use of CT scans (Ketten et al., 1998). Given that recent pitch-matching studies found that pitch matches corresponded well with frequencies predicted from X-ray scans for listeners with no or very limited listening experience, it may be feasible to use scans to match places of excitation.

At present, matching places of excitation between electric and acoustic stimulation is an unsolved problem. Additionally, it is unclear whether any matching between acoustic hearing and the electric filter bank is desirable for optimal speech perception (see section 4.1). As it can take a long time for patients to adapt to a new frequency to place map (Svirsky et al., 2004), it seems therefore best to refrain from trying to match places of excitation in clinical practice until this question is resolved.

3.2 Loudness

In NH listeners, differences in stimulus intensity lead to differences in loudness, where stimulus intensity is a physical measure and loudness a perceptual measure (Moore, 2003). The relation between intensity and loudness can be described using a loudness growth function (LGF), usually plotted on a dB SPL-versus-sone scale. LGFs are normally monotonic, but can vary in shape across frequencies (Moore et al., 1997). In acoustic hearing, LGFs are the same for the two ears, unless there is an asymmetric hearing impairment.

When acoustic stimuli are processed by current CI sound processors, the correct (NH) relative loudness of stimuli is not explicitly coded. All commercial strategies were developed mainly to optimize speech perception. This can lead to large loudness differences between stimuli that would sound equally loud for normally hearing listeners. HAs are also fitted mainly with speech perception in mind, though normal loudness relations are part of the rationales of the major fitting rules (NAL and DSL, Ching et al., 2010). It is clear that when CIs and HAs are combined, loudness relations between the two modalities will often be abnormal and different. In several studies loudness relations between electric and acoustic stimuli have been investigated, either using simple artificially constructed stimuli, or stimuli processed through a sound processor and HA.

Zeng and Shannon (1992) performed binaural loudness balancing experiments between electric and acoustic stimulation in auditory brain stem implant listeners who had substantial acoustic hearing in one ear. They used sinusoidal electric stimuli and pure-tone acoustic stimuli, and showed that the results were well described by a linear relationship between acoustic decibels and electric micro-amperes. Eddington et al. (1978) also found a linear decibel-versus-microampere relationship with a single subject using a CI. Dorman et al. (1993) came to the same conclusion

using a single CI subject with a pure-tone threshold of 25 dB HL at the test frequency (250 Hz) in the non-implanted ear. Francart et al. (2008) confirmed this finding in ten bimodal listeners with modern implants and a severe hearing impairment in the acoustically stimulated ear. It should be noted, however, that, like sound pressure levels in NH listeners, electric currents bear a logarithmic relation to loudness (e.g., McKay and McDermott, 1998). It is not unlikely that a decibel sound pressure level versus decibel current function would have yielded an equally good fit in these four studies. In a reanalysis of the data of Francart et al. (2008), there was no significant difference in goodness of fit for currents expressed on a linear or logarithmic scale. Sucher and McDermott (2009) performed a loudness estimation and loudness balancing task with bimodal listeners using a one-octave wide noise band, filtered between 250 and 500 Hz. They found that acoustic sensation level expressed in percentage dynamic range was linearly related to the loudness-matched electrical level in percentage dynamic range. They concluded that the input dynamic range of the CI sound processor and compression of the HA should be set such that this relation is maintained, so as to achieve optimal loudness balance - at least for low-frequency signals that are audible through the two modalities.

Blamey et al. (2000) measured loudness growth functions and loudness summation for a group of bimodal listeners. The stimuli were 1/3-oct-wide noise bands, processed through a sound processor and through a custom HA. They found a substantial amount of loudness summation for the binaural stimuli. They also found that the shapes of iso-loudness curves were quite different in the two ears, and that dynamic ranges varied considerably. They concluded that standard fitting methods are probably not appropriate for bimodal stimulation, and the levels of the two devices might need to be reduced to compensate for binaural loudness summation.

McDermott and Varsavsky (2009) used electric and acoustic loudness models to predict the loudness growth functions for different stimuli processed by a CI sound processor or a linear HA. They found that for a 250-500 Hz noise band the growth of loudness with level was comparable between devices, and that the main disparities would occur at low input levels. For a 1-kHz tone, however, loudness growth functions were quite different. They suggested the use of numerical loudness models in CIs and HAs to compensate in real-time for differences in loudness growth (see section 5).

In summary, loudness growth in electric and acoustic hearing can be similar or very different, depending on the level and frequency content of the stimulus, the type and degree of hearing impairment, and the signal processing in the CI sound processor and HA. Methods to address these issues during device fitting are discussed in section 4.2.

3.3 Binaural cue perception

Binaural unmasking and sound source localization in NH listeners are governed mainly by the two binaural cues: ILDs and ITDs. NH listeners are sensitive to ILDs with a just noticeable difference on the order of 1 dB (Mills, 1960). They are even sensitive to ILDs in signals with different frequency content in the two ears, but sensitivity decreases with increasing interaural frequency mismatch (Francart and Wouters, 2007). NH listeners are also very sensitive to ITDs in several different features of the acoustic signal: the onset, temporal fine structure (Brown and Yost, 2011; Feddersen et al., 1957; Yost, 1974), and envelope modulations (Henning, 1974; Middlebrooks and Green, 1990), with thresholds ranging from as low as 10 μ s for pure tones, to around 40 μ s for vowels (Akeroyd, 2003), and around 100 μ s for modulated high-frequency tones (Bernstein and Trahiotis, 2002).

In the following we discuss binaural cue sensitivity with electric stimulation in one ear and acoustic stimulation in the other. Bimodal listeners were found to be sensitive to ILDs with a mean just noticeable difference of 1.7 dB for computer-controlled single-electrode stimulation in one ear and a sinusoid delivered acoustically to the other ear (Francart et al., 2008). Given that ILDs can in practice range up to 20 dB, bimodal listeners should be able to use them for localization if the cues are properly transmitted by their devices. This was confirmed by Francart et al. (2011b), who found that bimodal listeners were able to use artificially enhanced ILD cues for localization.

Bimodal listeners with sufficient residual hearing are also sensitive to ITD cues. Francart et al. (2009a) found that the four of eight bimodal listeners who had unaided thresholds better than 100 dB SPL at 1000 and 2000 Hz were sensitive to ITDs in single-electrode low-rate pulse trains presented electrically and filtered click trains presented acoustically, with just noticeable differences on the order of 100-250 μ s (compared with 10-50 μ s for NH listeners, depending on the stimulus). Francart et al. (2011a) extended this study and found that bimodal listeners were sensitive to ITDs in single-electrode transposed signals and in 3-electrode signals processed by a CIS-like strategy. However, the subjects were only sensitive to ITDs for modulation rates up to around 150-200 Hz, which corresponds to the rate limitation for envelope-ITD perception found in NH listeners (Bernstein and Trahiotis, 2002).

Lenssen et al. (2011) measured sensitivity to ITDs between the electric signal and the fine structure of the acoustic signal in a group of listeners who had good sensitivity to envelope ITDs. They found that these listeners were not sensitive to fine-structure ITDs, and hypothesized that this is due to the inability to perceive temporal fine structure through electrical stimulation and a combination of the rate-limitation of envelope-ITD perception and place mismatch.

In section 2.1 we discussed the temporal synchronization problem. After the processing delays of the two devices have been equalized, the electric signal needs to be delayed to compensate for the travelling-wave delay in the acoustically stimulated cochlea. In general, this delay is frequency-dependent and increases with decreasing frequency. From their ITD sensitivity measurement, Francart et al. (2009a) estimated that the median delay of the electric signal relative to the acoustic signal was 1.5 ms. No frequency dependency was found. This is due to the limited sensitivity of the measurement and could also be related to the limited frequency range of the subjects' residual hearing, broadened auditory filters, and the loss of the active mechanism in the cochlea (Robles and Ruggero, 2001). Francart et al. (2009a) compared this value to the difference between average auditory brain stem response (ABR) and Electrical ABR (EABR) wave-V latencies in the literature and found that it was very similar.

In summary, bimodal listeners with sufficient residual hearing are sensitive to ILDs and ITDs in the envelope and transients of acoustic signals, but not to ITDs in the temporal fine structure. Given the technical limitations of current commercial devices, in practice ILDs are only partially perceptible and ITDs are probably not perceptible at all. In section 5.3 we review a number of speech processing algorithms designed to overcome this deficiency.

3.4 Relation between test results and degree of residual hearing

In many publications, a relation has been reported between the degree of residual hearing and the subjects' perceptual performance. For instance, Francart et al. (2009a) found that, out of eight subjects, only the four with an average acoustic hearing threshold at 1000 and 2000 Hz better than 100 dB SPL were sensitive to ITDs. From the discussion about loudness growth and transmission of ILD cues above it also follows that the extent of residual hearing can have a large influence.

Turner et al. (2008) correlated the average low-frequency (125, 250, 500 Hz) hearing thresholds and speech reception thresholds of 19 hybrid subjects. The intersection of the regression line with the average speech reception threshold for electric stimulation suggested that there was an advantage of preserved residual hearing unless the hearing loss approached profound levels. It should be noted though that due to the implantation criteria for hybrid implants, there was only one patient with an average hearing loss greater than 75 dB HL. Buechner et al. (2009) measured recognition of speech masked by a single competing talker in EAS users with the Cochlear Hybrid-L implant. When the acoustic signal was low-pass filtered at 300, 500, or 700 Hz, performance decreased with decreasing cut-off frequency, but significant improvements were found over CI-only stimulation, even for the 500 Hz cut-off frequency. Zhang et al. (2010) tested monosyllabic word recognition in quiet and sentence recognition in noise. The acoustic stimuli presented to the non-implanted ear were either low-pass-filtered at 125, 250, 500, or 750 Hz, or unfiltered (wide band). Adding a low-frequency acoustic signal to the electrical stimulation led to significant improvements in word recognition in quiet and sentence recognition in noise, even when the acoustic bandwidth was limited to 125 Hz. It should be noted, though, that the low-pass filter used had a slope of 90 dB/oct

and therefore did not completely eliminate all information above its cut-off frequency. Cullington and Zeng (2010) tested a CI recipient with normal hearing in the contralateral ear, and measured the effect of low-pass and high-pass filtering the acoustically presented signal. High-frequency acoustic stimulation did not improve performance, but low-frequency acoustic stimulation did, even when unintelligible and limited to frequencies below 150 Hz using a third-order elliptical filter with 80 dB stop band attenuation. Note that when we replicated this filter using Matlab, 80-dB attenuation was only obtained above approximately 1500 Hz, and audible components of a 60-dB SPL speech signal could still have been present up to 500 Hz, where the attenuation was around 40 dB. Brown and Bacon (2009, 2010) and Brown et al. (2010) replaced the acoustic stimulus by a tone modulated in frequency and amplitude, and found significant speech recognition benefits, both for simulated and real EAS, even when the tone was shifted downwards in frequency.

While very limited residual hearing can be sufficient to obtain bimodal benefit, a study by Mok et al. (2006) seems to suggest the contrary. They tested 14 bimodal listeners, of whom 11 showed a significant bimodal benefit, but 2 showed poorer speech perception with CI+HA than with the CI alone. A correlation analysis indicated that participants with poorer aided thresholds at 1000 and 2000 Hz demonstrated greater bimodal benefit. There was no significant correlation for aided thresholds at 250, 500, and 4000 Hz. The authors hypothesized that the mid-to-high frequency information provided by the HA may have conflicted with information provided by the CI. It should be noted, however, that one of the two poorer performers had very low scores in all conditions (<8 % correct), so that result might indicate a “floor effect”, while the other poor performer only showed poorer performance in one test, but better performance in other tests. As aided thresholds are a measure of hearing plus device function, rather than a characteristic exclusively of residual hearing, the positive correlation between aided thresholds and bimodal advantage could also be interpreted as an effect of suboptimal fitting, rather than a perceptual conflict between the two modalities. Dunn et al. (2005) found that 2 of 11 participants performed worse when adding a HA. In this case, suboptimal fitting may also have played a role, but an alternative explanation lies in neural plasticity, because one of the worse performers had only been using his implant for three months: Luntz et al. (2005) and Adunka et al. (2008) showed that there can be significant changes over time in bimodal benefit.

El Fata et al. (2009) tested recognition of songs and found that acoustic stimulation provided an advantage for those subjects who had thresholds better than 80 dB HL between 125 and 1000 Hz. Zhang et al. (2013) measured audiometric thresholds, speech reception scores and spectral-modulation thresholds of 22 bimodal listeners, and found that for all subjects together there was a significant correlation between all three measures and bimodal benefit. However, within the groups of subjects with mild-to-moderate loss and severe-to-profound loss, only spectral-modulation threshold was significantly correlated with bimodal benefit, indicating that it is more sensitive than the other measures. They suggest that spectral-modulation thresholds may be used to identify the ear best suited for implantation in case of symmetric hearing loss.

While the data on the effect of degree of residual hearing is still quite sparse, it seems that a very limited range of low-frequency residual hearing can be useful for speech perception, even if encompassing only the first harmonic of vowel sounds. For binaural cue perception, a wider frequency range is required. Besides the usable frequency range, detection thresholds and other characteristics of the residual hearing can have a significant influence on the benefit of added acoustic stimulation. It should also be noted that the audiogram does not fully characterize the hearing loss. Not only could there be “dead regions” that are not evident from the audiogram, but the degree of hair cell survival can also vary without necessarily affecting the thresholds, but with a significant effect on other auditory characteristics, such as the width of auditory filters.

4 Fitting methods

When attempting to fit a HA and CI for simultaneous use, two important questions are immediately evident. The first is how to set up the frequency-to-electrode allocation in the CI to be compatible with the acoustic stimulation. The second question is how to set the parameters of

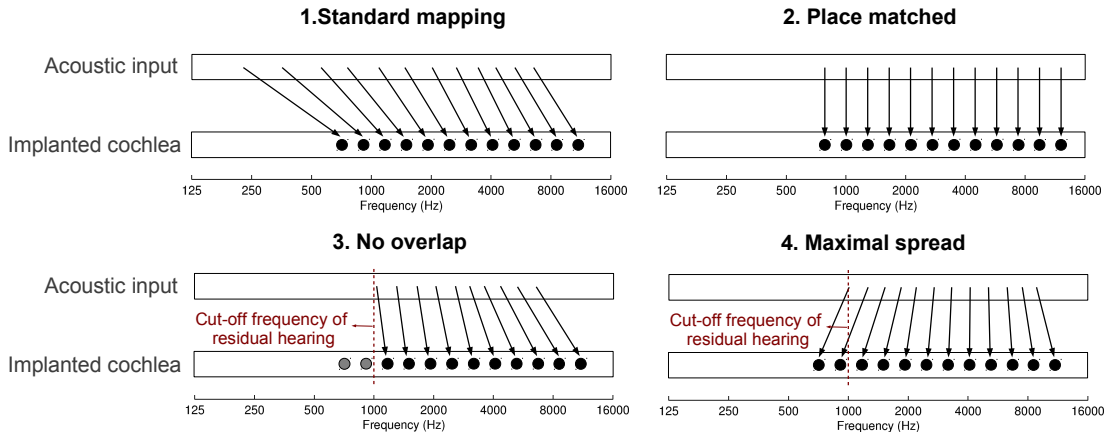


Figure 3: Conceptual diagram of different frequency-to-electrode allocations in the CI sound processor. Note that actual electrode positions in the cochlea are different for each recipient and standard frequency-to-electrode allocations depend on the processor used.

the compression systems in the HA and CI, such that loudness growth is similar for electric and acoustic stimulation for arbitrary stimuli, if this is possible at all.

4.1 Frequency allocation

In commercial CI sound processors, the frequency-to-electrode allocation is usually based on standard tables, only taking into account the number of usable electrodes, but not electrode array insertion depth or the degree of residual hearing. It is assumed that brain plasticity might eventually compensate for any mismatch with the acoustic frequency-to-place mapping. Depending on how well the brain adapts to mismatch, one can envisage several problems when combining the standard mapping with residual hearing. In Figure 3, a conceptual overview is given of four possible frequency-to-electrode allocations for EAS. Panel 1 exemplifies a standard allocation, and the corresponding frequency-to-place mismatch. In this case, the frequency-to-electrode allocation is always the same, regardless of the large inter-subject differences in electrode position that often occur, especially with hybrid (short) arrays. In the place matched case (panel 2 of Figure 3), for each electrode location the corresponding acoustic frequency is determined and the CI filter bank is configured accordingly. Unfortunately, there are currently no simple and scientifically sound methods to do so accurately (see section 3.1). Additionally, depending on insertion depth, frequencies up to 1-2 kHz may not be mapped, which would be detrimental for speech perception if the acoustic stimulus is masked by noise or degraded due to hearing impairment. In panel 3 of Figure 3, an allocation with minimal overlap between electric and acoustic stimulation is shown. Electrodes overlapping in cochlear place with usable acoustic hearing are disabled, and only frequencies above the cut-off frequency of the residual hearing are mapped to electrodes. In panel 4 of Figure 3, only acoustic frequencies beyond the cut-off frequency of the residual hearing are mapped. In this case the frequency resolution of the entire spectrum is maximized, assuming that the residual hearing has good spectral resolution and the electric stimulation has poor spectral resolution due to spread of excitation. When developing a frequency-to-electrode allocation, the main factors to be taken into account are the amount of overlap between acoustic and electric stimulation, and whether places of stimulation are matched across modalities (note that it is currently unknown to what extent the place of stimulation needs to be matched).

von Ilberg et al. (2011) reviewed the fitting methods implemented by the manufacturers of the Med-El DUET processor and the Cochlear Hybrid processor. There are no generally accepted and scientifically validated methods for fitting hybrid devices. Current methods generally fit the amplification at low frequencies (up to 1-2 kHz) in a conventional way and do not amplify at

higher frequencies. The upper cut-off frequency of acoustic stimulation is based on the audiogram of the residual hearing.

The cut-off frequency of residual hearing has been defined in different ways, typically as a frequency beyond which the hearing threshold exceeds a certain level (e.g., James et al., 2006; Kiefer et al., 2005; Vermeire et al., 2008). With unlimited fitting time, and if the residual hearing is good enough to conduct the tests, the best option may be to estimate the lower edge frequency of the high-frequency dead region, assuming one is present. With limited testing time, it is probably best to estimate the cut-off frequency based on the audiogram. For ski-slope audiograms it can be defined as the frequency for which the threshold is 20 dB lower (worse) than the average threshold of the flat part. For gradually sloping audiograms, it can be defined as the frequency at which the threshold is worse than 90 dB HL, although values as low as 65 dB HL have been used in the literature.

Kiefer et al. (2005) asked 13 patients newly implanted with hybrid systems to select one of three frequency-to-electrode allocations, according to personal subjective preference, after 2-3 weeks of using a processor containing the three maps. The maps were (1) the standard map, with a lower frequency boundary of 300 Hz, (2) a map with a lower frequency boundary of 650 Hz, yielding reduced overlap between electric and acoustic stimulation, and (3) a lower frequency boundary of 1000 Hz, presumably yielding a gap between electric and acoustic stimulation. The standard map was selected by 12 patients, and the 650-Hz map by one patient. Note that depending on individual electrode positions, the amount of overlap in the three maps was not necessarily the same for all 13 patients. Also, no behavioural tests were done using the different maps, only using the self-selected best map.

James et al. (2006) asked 10 patients newly implanted with hybrid systems whether they preferred a standard map or a “non-overlapping-shifted” map (Figure 3, panel 3). The latter used the same filter bands as the standard map, but one to three low-frequency bands were deactivated so that there was no overlap for frequencies where hearing thresholds were better than 80 dB HL, and the frequency-to-electrode allocation was shifted apically by the number of deactivated channels. Half of the subjects started with the standard map and the other half with the non-overlapping-shifted map. After 1 month the map was changed over. After 2 months the patients were able to switch between the two maps. Six subjects preferred the non-overlapping-shifted map, and only one the overlapping map. Speech perception results were only reported for the preferred map. Using a very similar protocol, Fraysse et al. (2006) found that 7 of 9 subjects much preferred the non-overlapping-shifted, while 2 preferred the standard map.

Vermeire et al. (2008) conducted sentence tests in noise with four subjects who had acoustic hearing in the implanted ear, using two different CI maps. One was the normal clinical map, with the default frequency-to-electrode allocations, whereas the other was adjusted such that the lowest allocated frequency corresponded to the cut-off frequency of the audiogram. The cut-off frequency was defined as the frequency above which the hearing thresholds became worse than 65 dB HL, based on the observation that the maximal hearing loss attributed to outer hair cells lies in the range of 50-65 dB (Moore et al., 2000). This corresponds to the situation depicted in the fourth panel of Figure 3. This led to shifts of 50-500Hz, relative to the clinical map. Vermeire et al. (2008) found a significant improvement for the shifted filter bank, but there was no CI-only condition to control for the effect of changing the frequency allocation, and it is surprising that a significant effect was found, given the very small difference between the two maps for some patients.

Simpson et al. (2009) used pitch matching between acoustic and electric stimuli to determine the acoustic frequency corresponding to the most apical electrode. They used two different frequency-to-electrode allocations. One was devised such that there presumably was no overlap in cochlear place between electric and acoustic stimulation, corresponding to the situation depicted in the third panel of Figure 3. The other allocation was the conventional one, which led to some overlap. For CI plus acoustic stimulation, frequency allocation was not found to significantly affect speech recognition in listeners who used a HA in the non-implanted ear (three subjects) or both ears (two subjects). In the CI-only condition, however, performance was lower for the non-overlapping map, because there were fewer active electrodes corresponding to the frequency range important for speech perception. This means that the acoustic stimulation in the bimodal

condition was effective in overcoming this deficit. Similarly, Richard et al. (2012) measured speech recognition of five bimodal listeners after disabling electrodes corresponding to acoustically stimulated frequencies. They found that this change caused performance to decrease in the CI-only condition, but not in the bimodal condition.

We are not aware of any studies in which the frequency allocation of electrodes beyond the apical ones was changed to improve place matching with the acoustically stimulated cochlea. In principle, for subjects with sufficient residual hearing at high frequencies, this could improve binaural cue perception, binaural fusion, and speech perception in noise. Also, depending on brain plasticity, large place mismatches at higher frequencies might actually lead to deterioration in speech perception when adding acoustic stimulation, which in current clinical practice would probably be solved by reducing amplification at high frequencies.

It is well known that unilateral CI listeners can adapt to place mismatch (i.e., stimulation of a place in the cochlea not normally activated by the corresponding acoustic frequency). Such adaptation can take a long time (months to years) and might be incomplete, in the sense that performance does not reach that expected for the unshifted condition (e.g., Fu and Shannon, 1999; Fu et al., 2002; Rosen et al., 1999). When adjusting the frequency allocation of the CI filter bank, one should also consider that the residual hearing of most patients is not necessarily stable. Many patients have progressive hearing loss, and it is not uncommon for patients to suddenly lose their residual hearing or even for it to improve unexpectedly. If one were to drastically modify the filter bank to accommodate residual hearing, e.g., by disabling apical electrodes or only mapping frequencies beyond 1-2 kHz to the implant, it is possible that a revision would become necessary in the near future, leading to another long adaptation period, which is very inconvenient for both the CI user and the clinician. Furthermore, such revisions could be required on multiple occasions if the hearing gradually worsens. In the case of short-electrode hybrid implants, this problem can be even more difficult, because a loss of residual hearing could necessitate re-implantation with a conventional long electrode.

The perceptual effects of interaction between electric and acoustic stimulation in the same ear are currently not well understood. Lin et al. (2011) measured simultaneous masking of electric and acoustic stimulation in the same ear, and found that in five short-electrode subjects electric stimulation did not cause masking of an acoustic stimulus, but that acoustic stimulation caused electric threshold elevations at apical and basal electrodes. Vollmer et al. (2010) investigated ipsilateral masking physiologically in cats, and found that electric and acoustic stimulation yielded complex interactions in the region of overlap, strongly dependent on the relative phases of the stimuli. At higher current levels, the masking effect of electrical responses dominated the effect of acoustic responses. The implications for frequency allocation in hybrid devices remain unclear, but when fitting hybrid devices to patients one should be aware of possible interactions. Contralateral masking seems more consistent with phenomena observed in NH listeners (James et al., 2001), so for the bimodal case there is probably less cause for concern.

In summary, the literature is inconclusive about the optimal frequency-to-electrode allocation and HA processing for EAS. While for speech perception neural plasticity might compensate inherently for some suboptimal characteristics of the stimulation, this is not necessarily the case for binaural cue perception, and it is possible that ipsilateral masking has a detrimental effect for some users of hybrid systems.

4.2 Amplification

There are no generally accepted methods for fitting the amplification for EAS. There is, however, one published method that was developed by Ching et al. (2004, 2001). First the CI is fitted according to conventional procedures until a stable map is obtained, and the HA is fitted based on the NAL-RP (Byrne et al., 1990) or NAL-NL1 (Byrne et al., 2001) rule. Subsequently, two fine-tuning steps are performed. In a first step, the HA's frequency response is fine-tuned based on the user's sound quality judgements in a paired-comparisons procedure between three alternative frequency responses, while continuous audiovisual speech is presented through the HA only. In the second step, the overall gain of the HA is adjusted to obtain loudness balance with the CI, again

while presenting running audiovisual speech, but with both the CI and HA enabled simultaneously. For linear HAs the adjustment is done by setting the overall gain to obtain loudness balance at 65 dB SPL only, whereas for non-linear HAs this is done for speech at 55 and 80 dB SPL. Using this type of fitting, speech perception and localization tests have been reported to show significant binaural advantages for adults (Ching et al., 2004) and children (Ching et al., 2001).

Vickers et al. (2001) measured the effect on speech recognition of low-pass filtering speech for hearing-impaired listeners. They found that, for listeners with high-frequency dead regions, speech reception improved with increasing cut-off frequency until the cut-off frequency was 1.5 to 2 times the estimated lower edge frequency of the dead region. With further increases in cut-off frequency, speech reception either remained constant or declined. On average it seems best not to amplify beyond 1.7 times the lower cut-off frequency of an extensive high-frequency dead region, to avoid a decline in speech perception, to save power in the HA, and to avoid acoustic feedback. Note that most bimodal listeners probably do have such a dead region. While the cut-off frequency cannot be readily determined from the audiogram alone, a dead region is likely when the hearing loss at that frequency is 70 dB HL or more (Moore et al., 2010).

While these procedures may be appropriate for current commercial devices, generally loudness growth equalization between devices is not possible to achieve by following a fitting procedure alone. For stimuli that can be perceived through both modalities - usually low-frequency stimuli - it might be possible to make loudness growth similar for the two ears by setting compression parameters in the fitting of current devices, but it could be very time-consuming, and would still not be able to take differences in loudness summation into account. For broadband stimuli a lack of high-frequency residual hearing is another possible complication. The non-linear processing (e.g., maxima selection) in most current sound processors additionally complicates such fitting procedures.

5 Signal processing techniques

In this section we discuss two signal processing techniques that are currently available in commercial devices and seem particularly well-suited for EAS: adaptive dynamic range optimization and frequency lowering. We also discuss three experimental signal processing schemes designed to improve binaural cue perception with bimodal stimulation.

5.1 Adaptive dynamic range optimization

Adaptive dynamic range optimization (ADRO) is an amplification strategy that uses statistical signal analysis to adapt the gain in different frequency bands with long time constants to obtain listener-specific audible and comfortable levels (Blamey, 2005; James et al., 2002). ADRO keeps track of the statistics of the intensity at the *output* of the hearing aid in each frequency band and slowly adjusts the corresponding gain to keep the intensity distribution at the output within a certain range, e.g., such that the 30th and 90th percentiles correspond respectively to soft and comfortable levels. Iwaki et al. (2008) investigated the effects of ADRO processing in six bimodal listeners. Speech recognition thresholds in quiet were significantly lower for two ADRO devices (i.e., a CI and a HA) than two non-ADRO devices. Speech recognition thresholds in noise were significantly lower for ADRO when stimuli were presented from the front or from the CI side. When the noise was presented from the non-implanted side there was no significant effect of ADRO. It should be noted, though, that in the non-ADRO condition the subjects' own HAs were used, with their clinical fitting (after a review), while in the ADRO condition identical HAs were provided by the experimenters. This led to a wide range of aided thresholds in the non-ADRO condition and it means that there may have been differences at the HA side other than non-ADRO versus ADRO which complicate the interpretation of the results.

One disadvantage is that ADRO adjusts the signal levels at the two ears independently, so it would not retain ILD cues in all conditions. Such cues are important for sound source localization, especially if ITD cues are not available.

5.2 Frequency lowering

While one can easily modify the frequency-to-electrode allocation in the CI to suit EAS, a comparable modification has not been possible in commercial HAs until recently, in the form of frequency lowering. Such techniques generally involve shifting selected higher-frequency components in amplified sounds to lower frequencies. In some HAs this is achieved by linear frequency transposition (Kuk et al., 2009), in which a band of high-frequency signals is shifted downwards and mixed with any lower-frequency signals that may be present at the same time. In other HAs, non-linear frequency compression is applied (Glista et al., 2009; Simpson et al., 2005). In the latter scheme, selected high-frequency sound components are compressed in frequency by a progressively increasing factor above a predetermined knee-point, without causing any frequency overlap. Frequency lowering is probably most beneficial for listeners with limited high-frequency hearing, and therefore seems well-suited to EAS users. For example, one potential benefit of frequency compression is that a wider range of acoustic frequencies can be made audible in the limited perceptual bandwidth of the acoustically stimulated ear than with conventional amplification. This might be expected to result in improved speech understanding. McDermott and Henshall (2010) tested speech perception of eight bimodal listeners who were fitted with the Phonak Naida HA with SoundRecover (non-linear frequency compression). Although speech understanding in noise was better, on average, for bimodal listening compared with the use of a CI alone, no benefits associated specifically with activation of the SoundRecover scheme were found. Similarly, Hua et al. (2012) tested nine bimodal listeners with a Widex Mind440 HA utilizing linear frequency transposition, and found no effect of enabling frequency transposition. While there currently do not appear to be any advantages of frequency lowering for speech perception, there could potentially be other perceptual advantages resulting from the audibility improvements provided by frequency-lowering HA schemes. Therefore further investigation into the use of such schemes by bimodal CI users could be valuable.

5.3 Experimental signal processing schemes

To address some of the problems with commercial devices discussed above, a number of signal processing schemes have been proposed that were designed to improve binaural cue transmission with bimodal stimulation. Below we discuss loudness-model based signal processing (SCORE), ILD emphasis, and modulation enhancement (MENs).

Based on work by McDermott et al. (2003) and McDermott and Varsavsky (2009), Varsavsky and McDermott (2012) proposed a loudness normalization strategy for electric stimulation (SCORE) based on loudness models of CI stimulation (McKay et al., 2003) and acoustic hearing (Moore and Glasberg, 1997). A block diagram is shown in Figure 4. SCORE uses a NH loudness model to estimate how a NH listener would experience the loudness of the signal at the microphone of the CI sound processor. It also estimates the loudness of the electrical stimulus at the output of the CI sound processor, and adjusts the overall output level such that the loudness corresponds to the NH loudness estimate. In experiments with unilateral CI listeners, Varsavsky and McDermott (2012) found an improvement for speech perception in quiet at relatively low sound presentation levels with application of the SCORE strategy

Francart and McDermott (2012a,b) recently extended this strategy to bimodal stimulation. In this case similar processing is done in the HA, using a loudness model for impaired hearing (Moore and Glasberg, 1997), fitted to suit the listener’s hearing loss. They evaluated this strategy with six bimodal listeners and found that binaural loudness balance improved, speech perception in quiet remained the same or slightly improved at low levels, speech perception in noise stayed the same, and ILDs were transmitted more faithfully. If this strategy proves successful in more extensive perceptual trials, it will provide loudness normalization, and thus a good amplification/loudness fitting for bimodal and hybrid devices.

The ILD is an important cue for sound source localization, especially if ITD cues cannot be perceived or are not available due to signal processing limitations. However, when using only ILDs for localization, bimodal listeners are faced with three additional problems: (1) the residual

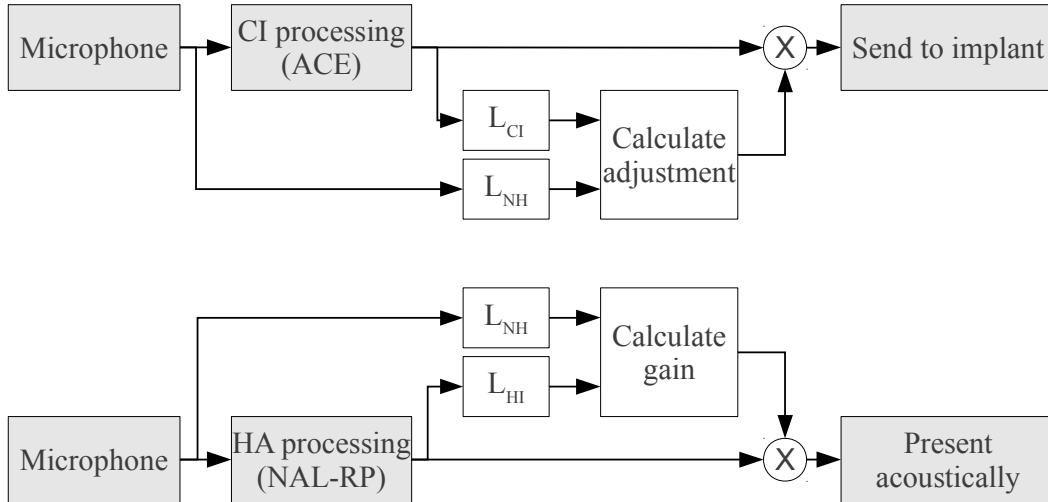


Figure 4: Overview of SCORE signal processing strategy, reproduced from Francart and McDermott (2012b). The grey blocks indicate existing signal processing.

hearing is usually limited to low frequencies, but the head-shadow effect and therefore also the ILD are small at low frequencies, (2) loudness growth functions are generally different for acoustic and electric stimulation, and (3) the ILD-versus-angle function is non-monotonic from approximately 45-60 degrees off-centre (Francart et al., 2011b; Shaw, 1974). While issues (1) and (2) can be solved with application of loudness normalization, issue (3) cannot.

The ILD emphasis algorithm proposed by Francart et al. (2009b, 2011b) solves this by using signal processing to detect the angle of arrival of an incoming sound and imposing an artificial broadband ILD. Tests with six bimodal listeners indicated that, with application of the ILD enhancement algorithm, localization performance improved significantly by 4-10 degrees absolute error, relative to a mean absolute error of 28 degrees in the condition without ILD enhancement.

Bimodal listeners with sufficient residual hearing are sensitive to ITD cues in laboratory experiments (see section 3.3). While several strategies have been developed to improve ITD perception with bilateral CIs (e.g., Smith, 2009; van Hoesel and Tyler, 2003), most are probably not suitable for application with bimodal hearing, because such strategies often introduce features that are not available acoustically. This is appropriate for application with bilateral CIs because the same processing is done for both CIs, but it is a problem for bimodal stimulation. Additionally, the across-channel timing of the electric signal becomes a problem. In acoustic hearing, the outputs of auditory filters with decreasing centre frequencies are increasingly delayed, while for the electric stimulus this is not always the case, and usually not on the same time scale. For instance in the ACE strategy, the basal channels are stimulated first, but the delay between subsequent stimulated channels depends on the total stimulation rate (channel stimulation rate \times number of maxima), and due to the maxima selection process there is no deterministic relation between channel delay and its associated frequency content. Across-channel delays may therefore be quite arbitrary, yielding inconsistent ITDs across place in the cochlea. Also, due to current spread and interaural differences in place of stimulation, the binaural system does not compare the channels of the sound processor's filter bank with the appropriate frequency ranges at the acoustically stimulated side. This means that the binaural system compares channels with different frequency content (acoustic cochlear place versus CI filter bank) and different latency.

Francart et al. (2012) proposed the Modulation Enhancement Strategy (MEoS), which imposes a deeply modulated envelope on all frequency channels simultaneously, similarly to F0 enhancement strategies such as F0Mod (Laneau et al., 2006; Milczynski et al., 2009) or MEM (Vandali et al., 2005). This results in deep and synchronous modulations of the pulse trains delivered to all electrodes. In preliminary experiments, improved ITD detection thresholds were found compared

to the commercial ACE processing in five bimodal listeners.

While first tests with loudness normalization, ILD enhancement and modulation enhancement have yielded promising results, further research is required before these strategies can be implemented in commercial devices. They need to be tested in a real-time system, with a large number of subjects, and combined with each other and the existing processing in commercial processors, such as automatic gain control and noise suppression. The effectiveness of these strategies of course also depends on the patients' perceptive abilities, which vary widely, especially in the case of ITD perception.

6 Summary and future directions

Due mainly to progressive relaxation of implantation criteria there is now a large population of CI users with residual hearing in the implanted and/or non-implanted ear. It has been found that even a very limited range of low-frequency residual hearing can be useful.

Current commercial HAs and CIs were designed for optimal speech perception when used on their own or in conjunction with the same type of device in the other ear. They were not designed specifically to work together. This results in some potential problems: temporal synchronization problems and mismatch in place of stimulation between the two cochleae. While both can be solved technically, this is not implemented in commercial devices, and it is not clear yet how the parameters (delay compensation and frequency-to-electrode allocation) should be set. While many authors have attempted to use pitch matching procedures to determine a good frequency allocation, none has obtained convincing evidence of benefit and there are some procedural issues, so it does not seem appropriate to use these procedures in clinical practice.

Normal-hearing listeners have ILD and ITD thresholds of around 1 dB and 40 μ s and the physical cues in realistic signals range up to 20 dB and 700 μ s. In ideal circumstances, bimodal listeners with sufficient residual hearing can be sensitive to ILDs and ITDs with thresholds of around 1 dB and 150 μ s. It seems a reasonable goal for the sound processing in the devices to preserve these cues with distortion limited to values below these thresholds. However, with current commercial devices this is not the case. Temporal fine structure cannot be perceived through electric stimulation; even purely physically, when assessing the electrical pulse output, the sound processing strategies of all manufacturers have some problems transmitting fine timing information. This means that interaural time differences in the fine structure cannot be perceived with current bimodal devices. While bimodal listeners with sufficient residual hearing are sensitive to ILDs and ITDs in the envelope and transients of acoustic signals, these cues are also distorted in commercial devices, due to poor and asynchronous transmission of temporal modulations on the electric side and differences in loudness growth between the ears. For most CI users, this leads to poorer localization ability and worse speech perception in noise than seems possible in principle.

Next to limited transmission of binaural cues, binaural loudness imbalance is another problem commonly associated with bimodal listening configurations at present. There are no generally accepted fitting methods available for bimodal or hybrid stimulation, and on theoretical grounds it does not seem practical to attain loudness balance with existing commercial devices for arbitrary signals and levels without supplementary signal processing. To fit current commercial devices for subjects with limited residual hearing, the principles given by Ching et al. (2004, 2001) seem reasonable: (1) Make sure the CI on its own is fitted optimally for speech perception and mainly make adjustments to the HA, as most speech recognition will be through the CI, and (2) adjust the overall gain of the HA for binaural balance at average conversational speech levels. Additionally, one should not amplify signal frequencies beyond 1.7 times the estimated lower cut-off frequency of a high-frequency dead region if a dead region is known or suspected to be present in the ear fitted with the HA.

The ideal frequency-to-electrode allocation in the CI sound processor for combination with acoustic stimulation is currently unknown, and given the long adaptation required when modifying the parameters of the filter bank, it is probably best to not change it from the default settings, unless there are compelling reasons (such as electrode faults in the implant). There is an urgent

need to develop, evaluate, and apply fitting procedures for all possible combinations of electric-acoustic stimulation.

Apart from fitting, we have developed some promising new sound-processing strategies specifically to address the problems found with current commercial devices used for EAS, but more research is required before they can be implemented in commercial systems.

In summary, we think the following steps should be undertaken to improve hearing with combined hearing aids and cochlear implants, in particular binaural cue perception, leading to improved sound source localisation and binaural unmasking: (1) further study the delay required in the electrical signal for synchronous neural activation with the acoustic signal and its dependency on frequency and hearing loss, investigate whether the auditory system can adapt to incorrect delays, and design methods to determine the ideal delay for each patient; (2) investigate how to set up the frequency-to-electrode allocation for optimal speech perception; (3) investigate how to optimally set up amplification to achieve the same loudness growth in the two ears; (4) improve the fidelity of envelope timing cues in the electric signal, in a way that is compatible with the acoustic stimulus (i.e., without introducing cues that are not available in the acoustic signal); and (5) design a unified sound processing platform and fitting method for EAS that allows setting of the parameters earlier mentioned and implementation of novel sound processing for bimodal stimulation.

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